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Biodynamic Modeling Techniques for Rotorcraft Comfort Evaluation

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Abstract This work shows how different occupant biodynamic modeling techniques are integrated in a rotorcraft design environment and discusses the resulting differences in comfort assessment. Three modeling techniques, that are used for biodynamic characterization, are considered: lumped parameter, finite element and multibody dynamics. These models are identified for the same gender, age, weight and height and then integrated into a virtual helicopter environment with a seat-cushion interface. A generic helicopter model is used to demonstrate the approach. For each of the three techniques, the vertical acceleration levels at the human-helicopter interface, as required by vibration regulations, and at the head are evaluated up to 30 Hz. At a first glance, it is observed that the lumped parameter is the easiest to implement in terms of model set-up. However, the use of lumped parameter models is limited to the population groups that they are identified from, and thus are not as flexible as the finite element and multibody ones in developing biodynamic models for individuals of an arbitrary population percentile. Furthermore, through numerical analysis it is found that the differences are not very significant in terms of accelerations at the human-seat interface. Therefore, for comfort related issues, the use of more complex models is not justified, unless complicated comfort assessments other than human interface accelerations are required. On the other hand, it is observed that the spine dynamics can play a significant role in estimating the acceleration of head; therefore, the sophisticated finite element and multibody dynamics models redeem their higher modeling cost and computation time when the head-neck health of occupants is considered.

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1 Introduction

Vibrations in rotorcraft are primarily related to the oscillatory response of the airframe to time dependent loads. The predominant sources of vibration are the rotor forces and moments originating from the rotors, fuselage aerodynamics, engine and transmission. The resulting time dependent loads are transmitted to the fuselage, which excites the crew and occupants through their contact with the vehicle, usually the seat surface. In rotorcraft, vibrations can degrade the ride quality of the occupants and crew[1] and might even lead to chronic pain in the long-term [2]. For this reason, the interest on rotorcraft comfort assessment is increasing[3,4].

Helicopter ride-comfort is usually evaluated through flight tests, since measuring vibrations along with the effect of human body mechanical characteristics, i.e. biodynamics, is essential to achieve a realistic comfort assessment. However, this method is not always convenient, since only limited design improvements can be accommodated when the helicopter is ready for flight, and all the flight envelope needs to be analyzed. Therefore, engineers must mainly rely on computational tools to address the limitations of flight tests, when analyzing the potential impact of their design choices on the vibrational level of the helicopter.

Standard methods exist for modeling and analysis of vehicles. Although helicopter analysis requires a multidisciplinary approach, the mechanical properties of the vehicle can be formulated precisely, since engineering materials can be easily tested and extensively categorized. As a result of this deterministic nature of mechanical systems, mature vehicle comprehensive analysis tools exist, which are available and widely used by all the manufacturers [5]. However, considering the bare mechanical properties of the vehicle, one can only estimate the accelerations at selected cabin locations. Neglecting the interaction of the helicopter with the human subjects and design the structure accordingly is the standard practice in industry.

Since the mechanical, physiological and psychological interaction of the human body with the vehicle dynamics may change the magnitude and perception of the accelerations, the resulting effects of vibration on the occupant are not directly correlated to the magnitude of the accelerations. Therefore, comfort assessment should take into account advances in human-machine interaction modeling paradigms, starting from early design stages. However, the mechanical properties of living subjects cannot be standardized and vary within the population; even for the same subject, properties can change with time [6] and according to posture [7]. For this reason, no dedicated technique exists for human biodynamic modeling; the same computational tools used in vehicle analysis are adopted, with averaged and parametrized data [8].

Methods for biomechanical modeling are categorized into Lumped-Parameter (LPM), Finite Element (FEM), and multibody dynamics (MBD) [9]. Lumped parameter modeling (LPM) use basic mechanical elements such as masses, dampers and springs, to formulate the dynamics of the human body. Thanks to its low computational cost and ease of parameter identification, LPM is very common in human biodynamics modeling for comfort assessment. The core of lumped parameter modeling is system identification of human body as an over-simplified mechanical system by experimenting over a target group *in vivo*. As a result, LPM has a weak physical analogy to the human body, there is no single solution, validity of the models is limited to the representation of the subjects in the tested group, and the number of available models is large. The second one, FEM, is particularly useful for the analysis of vibration effects on isolated human organs such as the spine [10], since the flexibility of modeling and the resolution of the output is far richer. However, the computational cost is higher, and identification of human mechanical properties with experiments is more complex as compared to LPM; moreover, handling of rigid body motion is somewhat limited. The last one, multibody dynamics (MBD), is a good alternative for biodynamic analysis, considering its great ability to model joints and nonlinear elements [11]. MBD adds flexibility to LPM with the ease of constraint formulation, and can approach the capabilities of FEM with the formulation of flexible elements. Furthermore, multibody modeling can capture effects related to nonlinearities, especially those originating from 3D geometry, with ease. For both FEM and MBD techniques, the mechanical properties of the building blocks, such as bones and tissues, are required to assemble the whole model representing the human body.

In order to answer the increasing demand for rotorcraft comfort assessment during the design phase, a computational framework is necessary. Since there is no standard for biodynamic modeling, the rotorcraft industry needs guidelines for the proper choice of the biodynamic modeling techniques. Therefore, a comparative study of the biodynamic modeling techniques in the presence of a coupled human-helicopter environment is required. This work addresses such need using a high-fidelity virtual aeroservoelastic modeling environment. The biodynamics along the vertical axis, modeled using lumped parameter, finite element and multibody models, are integrated into a helicopter modeling environment. The acceleration levels resulting from vibrations produced by main rotor vibratory loads in the presence of helicopter aeromechanics are compared.

2 Method

This section describes the aeroservoelastic modeling environment and how human biodynamic and interface models can be integrated to a high-fidelity aeroservoelastic rotorcraft model.

2.1 Virtual Helicopter Model

Analyzing biodynamic models of different origin coupled to helicopter dynamics requires a virtual simulation environment. A successful tool is expected to:

- be flexible in the source of sub-component formulation, to support accurate computation of vibratory loads;
- provide high-fidelity overall virtual modeling through sub-component assembly, hence all typical load-paths between source and human contact (seat, back-rest, foot-rest and hand-grips) can be considered with sufficient level of accuracy;

- have the capability to define forces acting between arbitrary structural points, to input loads calculated by external sources and feedback the biodynamic forces;
- support arbitrary sensor definition compatible for mounting human biodynamic models and interfaces, without the need to reassemble the whole model; therefore, allowing fast adaptation of biodynamic models of different modeling and population origins.

MASST (Modern Aeroservoelastic State Space Tools), a tool developed at Politecnico di Milano, satisfies all the above criteria. It analyzes compact, yet complete modular models of linearized aeroservoelastic systems [12,13]. In MASST, rotorcraft subcomponents are collected from well-known, reliable and state-of-the-art sources, and cast into state-space form using the Craig-Bampton Component Mode Synthesis (CMS) method [14], an effective substructuring approach. This approach is crucial to formulate the helicopter subcomponents (rotors, airframe etc.) in their most suitable platform and compose the overall model. In MASST, the assembled model is cast into a quadruple of matrices **A**, **B**, **C**, **D** that define a state-space system:

$$\dot{\mathbf{x}} = \mathbf{A}\mathbf{x} + \mathbf{B}\mathbf{f} \tag{1a}$$

$$\mathbf{y} = \mathbf{C}\mathbf{x} + \mathbf{D}\mathbf{f} \tag{1b}$$

where vector $\mathbf{x}$ contains the states of the system, $\mathbf{y}$ is the system output, $\mathbf{f}$ includes the inputs. MASST interpolates the state-space model matrices in a generic configuration within the corresponding linear models evaluated in the space of prescribed parameters. In the Laplace domain, the model produces the input-output relationship:

$$\mathbf{y}(s) = \left[\mathbf{C}\left(s\mathbf{I} - \mathbf{A}\right)^{-1}\mathbf{B} + \mathbf{D}\right]\mathbf{f}(s) = \mathbf{G}(s)\mathbf{f}(s). \tag{2}$$

2.2 Coupling Helicopter and Subjects

A virtual helicopter model excluding the human biodynamic response can give the necessary insight into the dynamic behavior of the vehicle itself. However, the interface between the human subjects and the vehicle feeds the subjects' dynamic forces and moments induced by vibrations back into the airframe. This feedback might be significant enough to affect the magnitude of the induced acceleration, which in turn cause a different vibration field on the human body than it would be in the uncoupled case.

The combined effect of human biodynamics, of seat dynamics and helicopter aeromechanics can only be accurately evaluated using a relatively high-fidelity vehicle model. However, since the mechanical characteristics of a human body change significantly from subject to subject and even within a single subject, and biodynamic models show great diversity, it is required to analyze a broad number of models of variable complexity and large population groups. Therefore, the cost associated with re-assembling a detailed model of the entire vehicle with a plethora of human biodynamics models is often not affordable. For this reason, an effective method could take advantage of a platform for high-fidelity aeroservoelastic modeling of rotorcraft, which allows the vibration engineer to modify the dynamics of the baseline plant by adding detailed human feedback models, without the need to re-assemble the coupled model when the biodynamic properties change. Consequently, the common environment supports both fast adaptation of biodynamic models and tracking the effects of design changes on the vibration rating of human occupants.

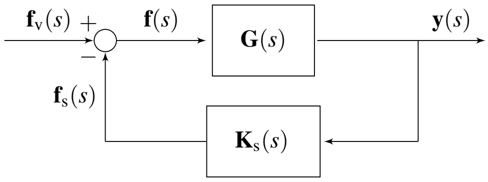

Fig. 1 Block diagram representation of the base vehicle, $\mathbf{G}$, and subject feedback, $\mathbf{K}_\mathrm{s}$.

MASST can export models and proper force-sensor relationships such that any human body can be added as a feedback element that operates from the output of virtual sensors and produces the resulting forces as inputs. For this purpose, it is sufficient to define specific input and output signals in the virtual helicopter model to create the feedback path within the device. According to Fig. 1:

- the input for the virtual helicopter model is defined as the vibratory forces (or moments) $\mathbf{f}_\mathrm{v}$, acting on any airframe point and/or on the rotors;
- the output $\mathbf{y}$ of the virtual helicopter model is chosen as the sensors of position, velocity, and acceleration of any airframe point (or rotor point in multiblade coordinates); thus, it is a (linear) function of the state and input of the model;
- the subjects create a feedback (negative feedback is preferred, to use the same convention of flight control design) loop between the sensors corresponding to the motion and the forces exerted by the subjects, $\mathbf{f}_\mathrm{s}$, at their attachment points,
$$\mathbf{f}_\mathrm{s}(s) = \mathbf{K}_\mathrm{s}(s)\mathbf{y}(s) \tag{3}$$
such that the total force can be expressed as $\mathbf{f} = \mathbf{f}_\mathrm{v} - \mathbf{f}_\mathrm{s}$; both force vectors have the same sequence of elements. The transfer matrix $\mathbf{K}_\mathrm{s}$ represents the synthesis of the human and interface model state-space representation.

Then, the response of the modified system is obtained as:
$$\mathbf{y} = (\mathbf{I} + \mathbf{G}\mathbf{K}_\mathrm{s})^{-1}\,\mathbf{G}\mathbf{f}_\mathrm{v} \tag{4}$$
where matrix $\mathbf{G}$ is the dynamic compliance matrix of the MASST high fidelity tool, ($\mathbf{y} = \mathbf{G}\mathbf{f}_\mathrm{v}$ is the output of the baseline virtual helicopter model, with $\mathbf{K}_\mathrm{s} = \mathbf{0}$). The gain matrix $\mathbf{K}_\mathrm{s}$ can be easily defined using force-response relationships of the attached human vibration or interface model.

Whichever technique is preferred, the human biodynamic and interface models should be put in state-space form in order to be compatible with MASST. In other words:
$$\dot{\mathbf{x}}_\mathrm{s} = \mathbf{A}_\mathrm{s}\mathbf{x}_\mathrm{s} + \mathbf{B}_\mathrm{s}\mathbf{y} \tag{5a}$$
$$\mathbf{f}_\mathrm{s} = \mathbf{C}_\mathrm{s}\mathbf{x}_\mathrm{s} + \mathbf{D}_\mathrm{s}\mathbf{y} \tag{5b}$$

in which vector $\mathbf{x}_\mathrm{s}$ contains the (possibly hidden) internal state of the subjects, $\mathbf{A}_\mathrm{s}$, $\mathbf{B}_\mathrm{s}$, $\mathbf{C}_\mathrm{s}$, $\mathbf{D}_\mathrm{s}$ are the state-space matrices. The state-space form can be made more compact by directly using the transfer functions between the problem-specific inputs and outputs:

$$\mathbf{f}_\mathrm{s} = \mathbf{K}_\mathrm{s}(s)\mathbf{y} = \left[\mathbf{C}_\mathrm{s}(s\mathbf{I}-\mathbf{A}_\mathrm{s})^{-1}\mathbf{B}_\mathrm{s} + \mathbf{D}_\mathrm{s}\right]\mathbf{y}. \tag{6}$$

For the LPM model, the equations of motion were directly written in the state space form given in Eq. 5. The MBD and FEM models input (cushion vertical acceleration) to output (head vertical acceleration) transfer functions were numerically estimated. A continuous-time model identification then allowed to determine the most likely Laplace-domain representation of the input-output relationships, that were transferred to state-space form by means of standard canonical realizations.

3 Occupant Biodynamic Modeling

This section describes the biodynamic modeling techniques, discusses how the mechanical properties of the human body are identified and details the biodynamic models compared in Section 4. The sitting person resting on a seat is preferred, since it is the usual posture of helicopter passengers and crew. For all biodynamic models, a seat and cushion is adapted from a helicopter application [15], in which they are described as a mass suspended by a spring and damper, as sketched in Fig. 2, with data given in Table 1.

Table 1 Numerical values for the seat-cushion model.

	m_i (kg)	c_i (N s m^{-1})	k_i (kN m^{-1})
Seat	13.5[1]	750.00[1]	22.6[1]
Cushion	1.0[2]	159.00[1]	37.7[1]

[1]From Ref. [15]; [2]assumed

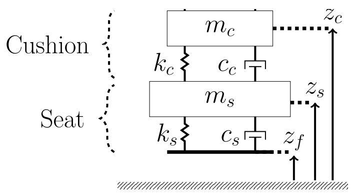

Fig. 2 Cushion and seat model, providing interface between cabin floor and human body

3.1 Lumped Parameter Model

The lumped parameter model (LPM) idealizes the human body as a set of lumped masses connected by springs and dampers. A LPM can range from a single body, representing the mass of the subject, to multidegrees of freedom models including feet and hands. However, increasing the degrees of freedom while remaining within the philosophy of LPM (i.e. fast and easy application using lumped elements to keep the problem size relatively much lower than that of a FEM or MBD model), is of little help to increase accuracy of whole body vibration estimation. For this reason, four degrees of freedom (4DOF) models are found to provide a sufficient number of parameters for effective fitting of a LPM [16].

Among 4DOF LPMs, for the purpose of the present work the apparent masses of six models are compared in Fig. 3, based on the parameters provided in literature [16]. The apparent mass is the ratio of the applied periodic excitation force to the resulting vibration acceleration. It can be observed that the models provide similar levels of apparent mass, and none provides distinctive characteristics. Therefore, all these models are suitable for a LPM biodynamic input. However, among them the Boileau-Rakheja [17] one provides the mass, weight, height, and gender of the group the LPM is defined for. Since this parameterization is necessary for finite element and multibody models, the Boileau-Rakheja model is selected as the LPM human biodynamic model of reference. The Boileau-Rakheja model is composed of four masses with interconnecting spring and dampers as shown in Fig. 4, resting on the previously mentioned seat and cushion model. The average of the Boileau-Rakheja experiment group is considered as the target percentile of the population in this work and are given in Ref. [17] as: age=27.3, height=175.7 cm, total mass=75.4 kg with its 55.5 kg is supported by the seat and remaining rest on feet. The LPM parameters reported in Table 2.

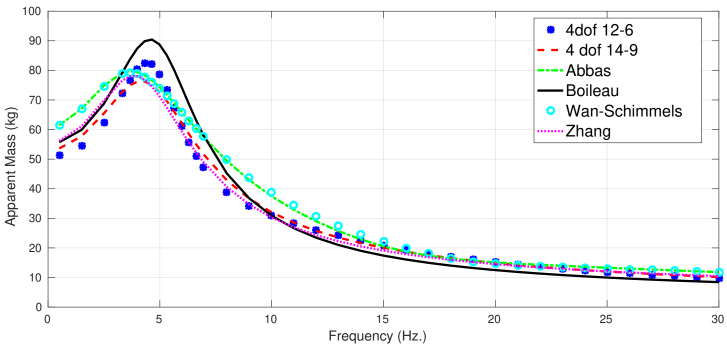

Fig. 3 Apparent Mass comparison of 4 degree of freedom lumped parameter models available in literature[16]

3.2 Finite Element Model

A finite element model of a sitting human has been originally developed, following the works of Kitazaki and Griffin [10, 18], which in turn was based on one by Belytschko [19]. The dynamic behavior of the spine is represented section-wise, each

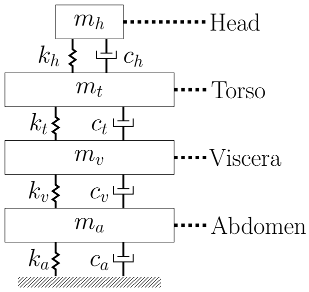

Fig. 4 Boileau-Rakheja lumped pilot model [17].

Table 2 Numerical values for the Boileau-Rakheja Model [17]. The data reflect a population with following average values: age=27.3, height=175.7 cm, total mass=75.4 kg with its 55.5 kg is supported by the seat and remaining rest on the floor

Index	m_i (kg)	c_i ($\mathrm{N\,s\,m^{-1}}$)	k_i ($\mathrm{kN\,m^{-1}}$)
i=h	5.31	400	310
i=t	28.49	4750	183
i=v	8.62	4585	162.8
i=a	12.78	2064	90

section consisting of the corresponding vertebra. In total, 25 vertebral components are taken into account. To them, elements representing the head, buttocks, visceral masses and pelvic masses, including a portion of the mass of the thighs, are added. The original model of Kitazaki and Griffin is limited to the planar behavior in the sagittal plane, whereas the present model, developed in NASTRAN, has been extended to comprehend the complete 3D behavior of the spine. Each vertebral section is modeled by a rigid body, freely allowed to move relative to the other vertebrae. Viscoelastic 6D elements connect the vertebræ nodes, following an approach suggested by Valentini and Pennestrì[20]. Eight visceral masses are connected to the corresponding vertebrae, in sections from T11 to S1. Only relative displacement degrees of freedom are allowed between visceræ and the corresponding vertebræ, since the former are represented by point masses. To prepare the model for the extraction of CMS elements, the masses of the upper limbs are concentrated in two body elements connected to T1. A snapshot of the spine finite element model is provided in Fig. 5.

The last vertebra (S1) of the isolated spine is connected to the cushion surface by viscoelastic elements representing the buttocks tissue. The relative vertical displacement and rotations in the sagittal and coronal planes are allowed with respect to S1. NASTRAN, the FEM tool used in this analysis, allows to directly extract the FRF at the frequencies of interest. Therefore, the FEM model is expressed in the form of Eq. 5. The same human parameters that of LPM is used for scaling mechanical properties: age=27.3, height=175.7 cm, total mass=75.4 kg, with its 55.5 kg is supported by the seat and remaining rest on feet. The total number of degrees of freedom of the model is 153.

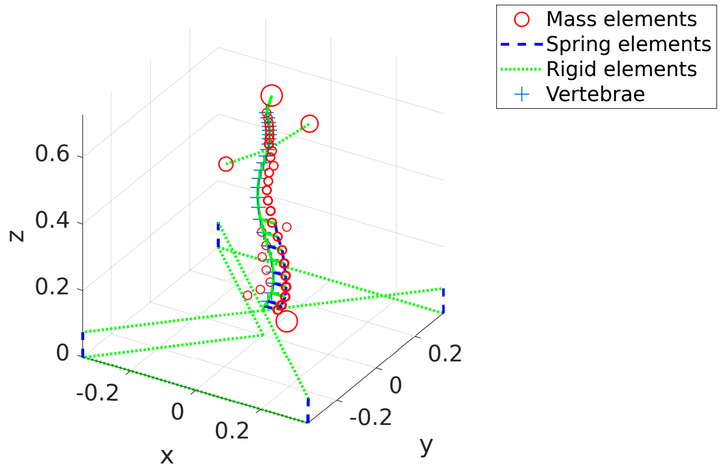

Fig. 5 Discretization of FEM biodynamic model.

3.3 Multibody Dynamics Model

The multibody model is structured in a way similar to that of the FEM, as shown in Fig. 6. The MBD model has been built using MBDyn [21], a free, general-purpose multibody solver developed at Politecnico di Milano[1]. It incorporates concepts first reported in the works of Kitazaki and Griffin [10], Belytschko [19], and Valentini and Pennestrì [22,23,20]. It was initially utilized for rotorcraft-pilot coupling analysis [24]. The model includes 34 rigid bodies associated with the sections of the trunk corresponding to each vertebra from C1 to S1, and to 8 visceral masses. Relative displacements between each vertebral node are allowed only in the local z direction, assumed to lie in the local tangent direction to the spine axis. Relative displacement in the x direction, i.e. the anatomical antero-posterior direction, and in the y direction, corresponding to the anatomical medio-lateral direction, are constrained. These modeling choices are motivated by the anatomical traits of the intevertebral joints. The superior and inferior faces of the vertebræ bodies are interconnected through a symphysis joint: in between the bones faces lies the intervertebral disc, a fibrocartilaginous cushion-like structure that allows for (small) relative rotations and (small) relative displacements. The posterior processes of the vertebræ are instead connected via the *facet joint*, a synovial plane joint that allow for almost null relative displacement between the facet of the joint, thus contraining almost completely the vertebræ relative displacements in the transverse plane.

Vertebræ are interconnected by linear viscoelastic elements, acting on all the remaining, unconstrained degrees of freedom. Visceral masses are also connected to the corresponding vertebræ from T11 to S1, and between them, through linear viscoelastic elements. Detailed modeling of the muscular fascicles for both the MBD and the

[1] `http://www.mbdyn.org/`, last retrieved in October 2018.

FEM model has been deemed unnecessary for the purpose of this work since, while it would allow to extend greatly the range of validity of the models in terms of the extent of deviation of the spinal pose with respect to the reference one, it would have added a significant amount of modeling effort. Furthermore, all the articular joints in the models present a certain degree of muscluar redundancy, i.e. they are actuated by multiple agonist/antagonist pairs: therefore, the resulting model would be underdetermined because of overactuation, and special techniques would have to be applied to be estimate the equivalent impedance at each joint. While similar experiences have been successfully carried out by the authors, regarding in particular the upper limbs' impedance at the control inceptors [11,25], it was concluded that for this particular case the more simplified approach relying on generic linear viscoelastic elements is sufficient for comfort related analysis.

Other lumped masses are placed in correspondence to centers of the shoulder girdles, of the head and of the pelvis. The latter comprises also a third of the mass of the thighs. The pelvic area modeling is completed by the introduction of a mass and a viscoelastic element representing the buttocks. As in the FEM model, last vertebra (S1) of the isolated spine is connected to the cushion surface by viscoelastic elements representing the buttocks tissue. The node representing the buttock degrees of freedom is constrained as to allow only the vertical relative displacement with respect to S1 and the rotations in the sagittal and the coronal plane. The total number of degrees of freedom of the system, before constraints are enforced, is 228. After constraint enforcement, 103 degrees of freedom remain.

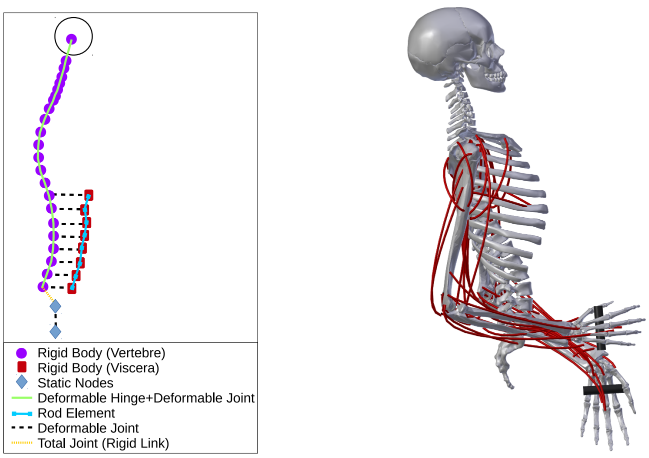

Fig. 6 Schematic representation of the spine multibody model (left) and a CAD representation of the coupled multibody models of the spine and of the upper limbs (right).

The nonlinear MBDyn model is transformed in the form of Eq. (5) by performing a direct time integration while excited by a pseudo-random acceleration signal with band-limited Power Spectral Density (PSD) in the frequency interval of interest. The signal is imposed to the floor node for a simulated experiment and converted to the frequency domain after applying a Fast Fourier Transform. The same human parameters that of LPM is used for scaling mechanical properties: age=27.3, height=175.7 cm, total mass=75.4 kg, with its 55.5 kg is supported by the seat and remaining rest on feet.

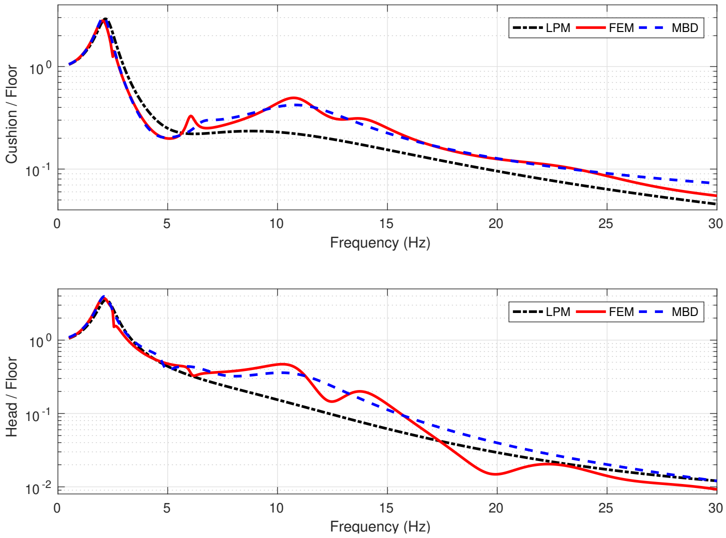

Fig. 7 Frequency Response Function of three biodynamic modeling techniques coupled with seat.

3.4 Scaling of model parameters

The parameters of the LPM are identified based on the results of an average of a given population [17]. Since the LPM is the fitting of a given model structure from experimental data, there is no alternative way to characterize it. However, for FEM and MBD techniques, the body parts, especially the spine, are built from basic elements representing bones and fleshes. Therefore, for FEM and MBD, the structural properties of the building blocks of the body can be determined and used to construct the global human biodynamics model. However, since the mechanical properties of these building blocks vary from person to person, FEM and MBD techniques still require a statistical parametrization. The required mechanical properties of the human body are usually obtained from corpses [26,27] or occasionally from *in vivo* using non-invasive measurements and imaging techniques [28,29].

Since the focus of this work is to compare the numerical biodynamic modeling techniques, available data from the literature is used. The age, height, weight and gender values of LPM is considered as the target percentile, since it is based on experimental results and could not be changed, as opposed to the MBD and FEM models, which offer the possibility to be generated with parameters tailored to a specific subject, as was explained in the previous sections. Reference values of the model inertial and viscoelastic parameters are taken from Kitazaki and Griffin[10] and Valentini and Pennestrì[20]. In particular, values of the intervertebral and vertebra-viscera stiffnesses in the sagittal plane are taken from the former work, while reference values for stiffness in the other direction are taken from the latter one. The damping values are obtained from Valentini and Pennestrì for the intervertebral elements, while for elements connecting visceræ to vertebræ and visceræ to visceræ the damping values are considered directly proportional, with a coefficient of 0.1, to the corresponding stiffnesses. These latter are also taken from Kitazaki and Griffin, together with the reference inertial parameters. The only relevant difference with respect to the cited works resides in the buttocks vertical stiffnesses and dampings since they are absent in Kitazaki and Griffin: the reference vertical stiffness used in this work for LPM is 58.8 kN/m and a proportional damping, with factor 0.025, is introduced. The resulting damping is 1.47 kNs/m. The rotational reference stiffness is 7.40 kNm/rad in the two allowed directions (about the local x axis, i.e. in the coronal plane, and about the local y axis, i.e. in the sagittal plane) and the same proportional damping factor used for the vertical direction is applied, resulting in an isotropic rotational damping of 0.185 kNms/rad.

To adapt the FEM and the MBD models to represent subjects with different anthropometric characteristics, a scaling procedure has been implemented [30]. It is based on a parametric ribcage model published by Shi et al. [31], able to estimate the most plausible geometry of the ribcage taking as input the generic anthropometric parameters age, gender, height, and weight. It has been built identifying the position of 464 landmarks along the ribs of 89 subjects and applying a Principal Component Analysis (PCA) to the resulting dataset.

A parametric NURBS curve representing the spine axis is then fitted, in the thoracic part, to the ribcage model, using the estimated locations of the ribs heads as control points. The remaining parts of curve are adapted by simply scaling the reference shape, identified using the vertebræ positions of the *erect* pose of the Kitazaki and Griffin model [10].

An estimated ribcage geometry has been fitted to the geometry of the Kitazaki and Griffin model, identifying the corresponding most probable anthropometric dataset of the reference subject, i.e. a 34 years old male, 1.78 m tall weighting 84 kilograms, for a Body Mass Index (BMI) of approximately 26.5. The identification procedure was necessary since in the original work of Kitazaki and Griffin no reference is made to the biometrics of the subject. Comparing the estimated ribcage dimensions with the one of the reference subject, scaling factors along the three dimensions $\lambda_x, \lambda_y, \lambda_z$ are calculated as the ratios between the dimensions of the ribcage model bounding box. They are subsequently used to estimate the variation of the model parameters (for both the MBD and the FEM model) with respect to the reference values. The simple procedure employed is here exemplified taking into account the axial stiffness, i.e. considering its order 0 representation and scaling it through simple dimensional analysis as follows:

$$K'_a \sim \frac{EA'}{L'} = \frac{EA}{L} \cdot \frac{\lambda_x \lambda_y}{\lambda_z} = K_a \frac{\lambda_x \lambda_y}{\lambda_z} \tag{7}$$

where K_a' represents the value of the axial stiffness of the subject to be modeled, while K_a represents the reference value. Other parameters are scaled following similar considerations.

4 Results and Discussion

This section presents the results of the isolated human-seat-cushion model first. Then, the vibrational level is presented for the coupled human-interface-helicopter high-fidelity model. For both the isolated and the coupled analysis, two criteria are used. The first one is the accelerations at the interface, i.e. the cushion surface, which are used for comfort assessment standards, such as two commonly used ISO-2631 (1997) [32] and BS-6841 [33]. Although there are differences between the two, the frequency weights of vertical vibration for both standards overlap within the instrumentation tolerances, thus likely to provide the same result in a real-world measurement [34]. Therefore, this selection is not crucial within the scope of this work and ISO-2631 (1997) is selected for frequency weighing due to its international acceptance. The other important comfort criteria is the head accelerations, which is an important health concern for the crew, especially considering that helmets are becoming heavier and heavier due to the installation of vision enhancement equipment [4]. Frequency weighing is not available for the head of a seated person in the mentioned standards; therefore multiplying factors are not used for the head accelerations.

4.1 Isolated Interface-Human

First the LPM, FEM and MBD models of human biodynamics are compared for the isolated seat-cushion and human system without the effect of helicopter dynamics. Fig. 7 presents the response of cushion and head as a result of a vertical excitation coming through the floor. It can be observed that the general trend is the same and all the three techniques capture the largest peak near 2.5 Hz. Additionally, MBD induces a smooth gain, whereas FEM induces several more peaks. As compared to LPM, the MBD model has a larger gain except under 5 Hz. FEM shows the same behavior for the cushion acceleration; however, for the head acceleration, it can result in higher or lower gain depending on the frequency of interest.

It can be noted that the frequency response of the models under 5 Hz is similar while, above that threshold, the response of the more detailed models is richer. This behavior is at least partly supported by experimental data available in literature [35]. Nonetheless, it is a well-known fact that at least for what concerns the overall vibration response of the human body (i.e. disregarding, for example, problems related to acoustics) the response above approximately 10 Hz is of lesser relative importance. This consideration would lead to favoring the significantly simpler LPM models. However, the authors believe that the added complexity of the more detailed models is justified when other important factors are taken into account: it is easier to adapt the FEM and MBD models to subjects of different anthropometric characteristics, for example. In the case of comfort analyses performed for more specific purposes, it might be useful to identify the anatomical regions subjected to the most important stresses: for example, in vibration analyses involving injured passengers, either in HEMS (Helicopter Emergency Medical Service) missions or in possible on-board emergency scenarios.

4.2 Coupled Interface-Human-Helicopter

The high-fidelity baseline helicopter model is built within the framework described in Section 2, based on data representative of a generic, medium weight helicopter with an articulated 5 blade main rotor. A snapshot of the physical kinematic variables of the virtual helicopter model is shown in Fig. 8. The state-space model includes:

- rigid body degrees of freedom;
- flight mechanics derivatives of the airframe, estimated using CAMRAD/JA;
- elastic bending and torsion modes of the airframe extracted from NASTRAN, with 1.5% proportional structural damping superimposed in MASST;
- the first two bending and first torsion modes of the main and tail rotors including aerodynamic matrices in multiblade coordinates obtained using CAMRAD/JA;
- transfer functions of main and tail rotor servo actuators directly formulated in Matlab/Simulink, considering servo-valve dynamics and dynamic compliance [36];
- the nodes and coordinates for the sensors and the forces, directly defined in MASST.

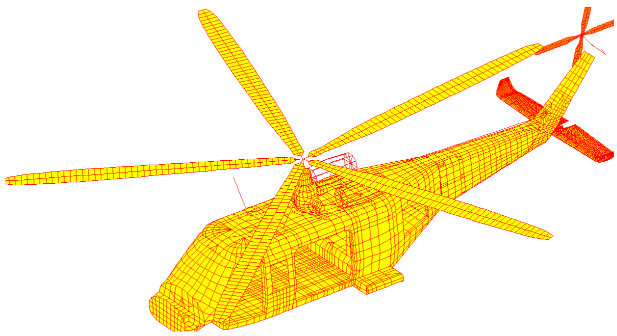

Fig. 8 Snapshot of the baseline virtual helicopter model.

The vibration performance of the coupled human-interface-helicopter model can be evaluated at any point on the cabin floor. The distribution of occupants within the possible seating locations might have an impact on vibration rating when different combinations of seating arrangements are possible if the number of occupants are less than the number of available seats [37]. However, this variability does not effect the comparison of different biodynamic modeling techniques if the occupied

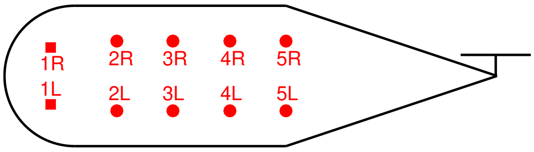

Fig. 9 Distribution and labels of seat attachment points on cabin floor

seats remain at the same location of the cabin floor while comparing different boidynamic models. Therefore, in order to prevent an arbitrary selection and cover the whole cabin floor, 10 seats are assembled into the cabin with a uniform distribution as shown in Fig. 9 using the steps explained in Section 2.2. At these locations, seats and the biodynamic models obtained using the three mentioned techniques are added, representing 2 pilots in the cockpit and 8 crew/passengers in vertical seating posture. Based on the accelerations at these 10 points on the cabin floor, an output vector $\mathbf{y}$ is defined as:

$$\mathbf{y} = \begin{Bmatrix} \ddot{z}_{cockpit,1} \\ \ddot{z}_{cockpit,2} \\ \ddot{z}_{cabin,1} \\ \vdots \\ \ddot{z}_{cabin,n} \\ \vdots \\ \ddot{z}_{cabin,8} \end{Bmatrix} \tag{8}$$

where at each location, $\ddot{z}$ gives the vertical accelerations either of the cushion or of the head. Then, the square of the norm of the accelerations, divided by the number of measurements, is defined as the vibration index, namely:

$$\mathrm{VI} = \frac{\sqrt{\mathbf{y}^T\mathbf{y}}}{10} \tag{9}$$

The biodynamic models obtained using the three techniques are added to the aeroservoelastic helicopter model. At the ten locations on the cabin floor shown in Fig. 9 the accelerations are computed and the vibration index is collected. Fig. 10 shows the results when the acceleration is measured at the cushion surface. All the three models predict the vibrational level within the same order of magnitude, with similar trends. Also, when compared with the isolated response shown in Fig. 7, the peaks other than the first one slightly above 2 Hz, are related to the airframe. Fig. 11 presents the same results for the head acceleration. In this case, there are more differences between the models than in the case of cushion acceleration especially above 5 Hz. Similar to the isolated human body response given in Fig. 7, a probable explanation is that the flexibility of the spine participates in the head response more than it does for the cushion surface. Therefore more sophisticated models with high resolution in the spine dynamics are expected to provide better understanding. Although this is partly supported by experiments [35], further studies are needed to conclude that MBD and FEM predict human head vibration better than LPM. For this reason, a conservative approach can be applied as an alternative to using a single method, by considering the worst case, i.e. selecting the highest amplitude, of all available models for the interested frequency range.

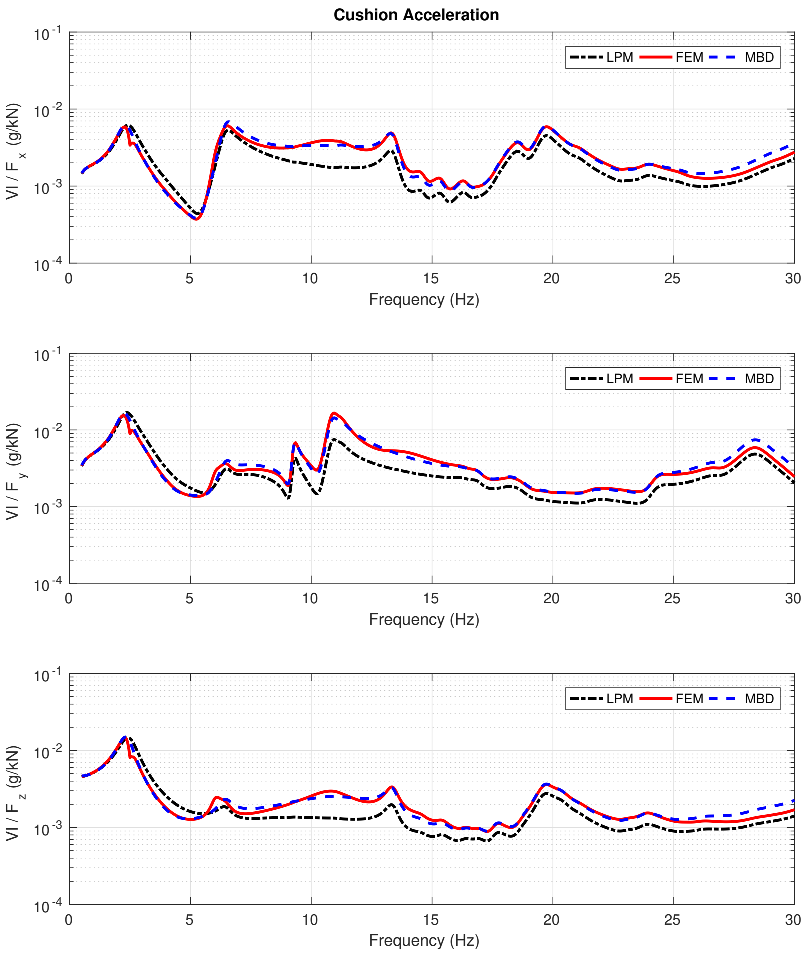

Fig. 10 Averaged Frequency Response Function of three biodynamic modeling techniques coupled with seat and helicopter between longitudinal (F_x), lateral (F_y) and vertical (F_z) unit hub forces and the cushion surface.

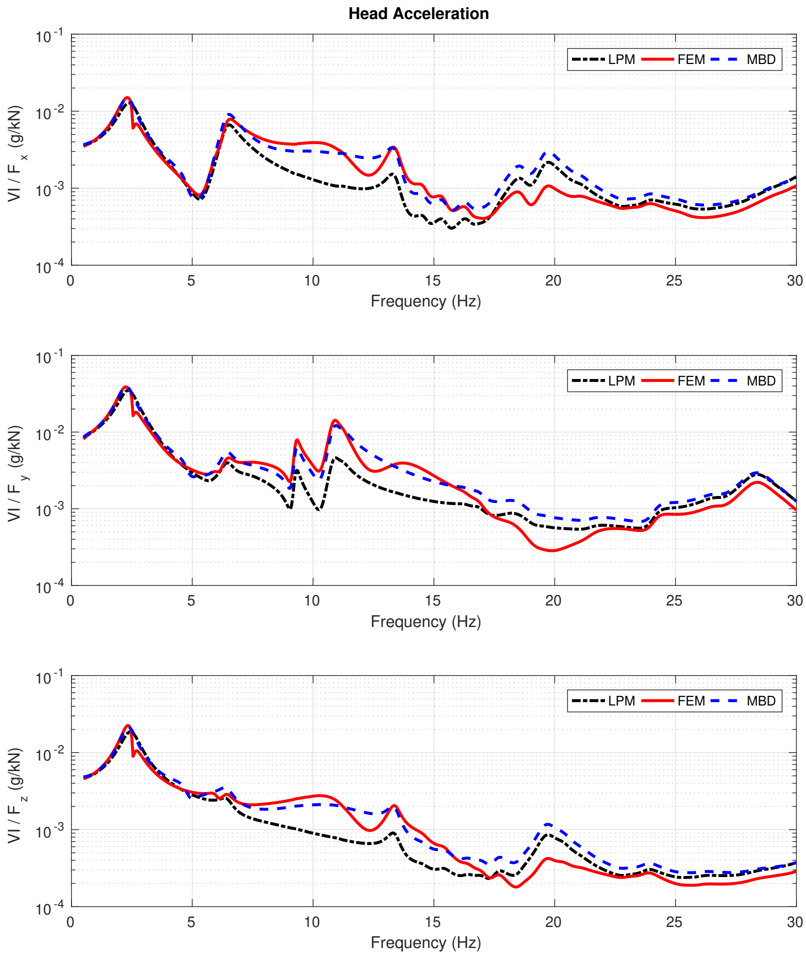

Fig. 11 Averaged Frequency Response Function of three biodynamic modeling techniques coupled with seat and helicopter between longitudinal (F_x), lateral (F_y) and vertical (F_z) unit hub forces and the head.

5 Conclusions

The three techniques of human biodynamic modeling are compared in vertical sitting postures for rotorcraft comfort evaluation, namely lumped parameter (LPM), finite element (FEM) and multibody dynamics (MBD). In brief:

- all the three techniques are determined based on the same gender, age, height, and weight percentile of the population, to make the comparison realistic;
- the biodynamic models are coupled to high-fidelity aeroservoelastic model with a seat-cushion interface;
- LPM relies on experimental data for the identification of the model, therefore it has limited adaptation when the target population digresses from the average of the identified group;

- the LPM is easier to formulate and implement; however LPM cannot provide detailed analysis; such as the strain between two vertebra of the spine. If more detailed information is required in addition to acceleration of major body parts; FEM or MBD should be selected;
- The complexity in using FEM and MBD techniques pays off considering their flexibility in modeling the biodynamics of a target group and formulating more complex comfort merits;
- the acceleration at the cushion shows similar trends; responses are within the same order of magnitude, therefore it is not easy to justify the modeling and computational cost of FEM and MBD models when the aimed point is the human interface surface unless the available LPM does not reflect the anthropometric characteristics of the aimed group;
- the dynamics of the spine plays a more significant role for head accelerations and significant differences are observed. Considering the flexibility in modelling an aimed population and higher output resolution, FEM or MBD is a better choice than LPM when upper body segments are of interest. Alternatively, a more conservative approach can be applied by considering all the available models to obtain a worst case scenario.

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Conflict of Interest Statement

On behalf of all authors, the corresponding author states that there is no conflict of interest.

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