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TECHNICAL PAPER

A survey on static and quasi-static finite element models of the human cervical spine

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Abstract

Finite element analyses are an important source of information on the biomechanical behaviour of the cervical spine; as well as an important tool in the design and evaluation of spinal instrumentation. This article presents a comprehensive survey of the finite element models of the cervical spine that have been used to study its pathological/nonpathological biomechanics under static/quasi-static loading conditions. Publications that met the inclusion criteria were analysed to extract parameters relative to model identification (e.g., spine segment, population, utility, limitations), model structure (e.g., loading/boundary conditions, anatomical structures, constitutive representation), simulation structure (e.g., software), verification (e.g., convergence) and validation (e.g., validated procedure/output, assumptions). Besides summarizing different modelling approaches with their associated parameters, this article outlines generalities and issues related to the obtainment of such models. The survey shows that authors often fail to report parameters that are critical for the reproducibility of results and that, even with fully reported parameters, these models are inherently difficult to replicate because they generally are patient-specific with their geometry based on data from in-house specimens/subjects. Overall, while the survey contributes to an understanding of the implications of following different modelling approaches and allows to take advantage of previously developed models, further research is required to improve the accuracy and utility of these models.

Keywords Biomechanics · Computational model · Material properties · Finite element method · Human cervical spine

1 Introduction

The human cervical spine (CS) is an intricate multi-articular system that supports the head and protects the spinal cord, as well as other important tissues including nerve roots and arteries. This well-engineered system is both strong and flexible, providing mobility to the head/neck. With its inherent complexity and its constant exposure to harmful stresses and forces both through trauma or simple daily activities, the CS is always at risk for developing a number of painful or even life-threatening conditions. Understanding the biomechanics of this part of the body is, thus, crucial not only to prevent injuries but also for the development of appropriate treatments to manage such injuries/diseases and, ultimately, restore the normal function of the spine.

Due to its kinematic and geometric complexity, however, understanding the biomechanics of the CS remains a challenging task in the associated fields of study. Traditional clinical studies based on *in vitro* and/or *in vivo* approaches are limited by (i) the unavailability of proper samples representative of the target individual, (ii) the ethical, regulatory and procedural complexity of research protocols involving such samples, (iii) the large variability between specimens, and (iv) the inability to provide direct measures of internal responses (e.g., stress and strain) [1]. Computational analyses by means of finite element models (FEMs), on the contrary, favour the execution of multiple experiments on the same 'specimen' without the limitations of *in vitro* or *in vivo* models and allow not only for the evaluation of external responses of the spine under mechanical loading, but also for the evaluation of stresses and deformations of internal structures.

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Following the notorious improvements in computational speed and the larger availability of accurate modelling software, FEMs are currently an important source of information on the biomechanical behaviour of the healthy, injured (disease, trauma, aging) and instrumented CS. These models are also providing support in the *interactive* (to a limited degree) design and evaluation of spinal instrumentation (e.g. interbody replacement systems, spinal-plating, etc.) [2]. The Interactive Design Method, a key trend in the domain of product design and manufacturing, is based on the principle of virtual exploration of solution spaces (mostly via the use of virtual prototyping and other interactive simulation tools) as a means to support decision-making during preliminary stages of the design process [3]. Although the interactivity of FEM is limited because it requires post-processing of results and simulation re-runs whenever parameters need to be adjusted [4], FEM models do support decision-making during the early phases of the design process: (i) by allowing for new designs to be rigorously tested in terms of the interactions that exist between the product and the human body before a prototype is even manufactured, and (ii) by enabling modifications to prototypes without the costs and time associated to manufacturing processes. Also, while fully interactive simulation tools for this particular field and application are still to be further explored in order to enhance the productivity and efficiency pursued by the Interactive Design Method, obtaining proper FEM models remains key towards acquiring the essential knowledge on the product-user-environment interactions that can ultimately lead to the attainment of those real time, reduced, realistic, and qualified virtual models that reflect natural interactions [3] and that are central to interactive design. In FEM, relative to the objective of a given study, different modelling approaches may be followed to yield adequate results. Indeed, considerable differences are noted among the multiple models that have been reported in scientific literature. To the authors' knowledge, however, there is not an up-to-date revision of these models that allows the reader to compare and understand the different modelling approaches, as well as to identify the main achievements in the field, the primary areas of debate, and the unresolved research questions. The available reviews are either outdated [5] or lack detail relative to specific characteristics of the models [2,6].

This article provides a comprehensive survey of the existing FEMs of the CS that have been used to study the normal and pathological biomechanics of this part of the body under static or quasi-static loading conditions. The survey is structured based on the parameters required for model development and dissemination [7] and, thus, provides data relative to model identification (e.g., spine segment modelled, represented population, state of represented subject/specimen, utility, limitations), model structure (e.g., type of model, loading/boundary conditions, mesh density, anatomical structures, constitutive representation—formulation, coefficients, source—, interactions), simulation structure (e.g., name/version of software), verification (e.g., mesh convergence) and validation (e.g., validated output, modelling assumptions, validation procedure, studies used for validation). In addition, generalities and issues related to the obtainment of a FEM of the human CS are discussed and a number of opportunities for further research are outlined. Such an overview will facilitate the development of new and improved FEMs of the CS that will, in turn, broaden our understanding of its biomechanics and further aid in the development of interactive tools that can lead to innovative solutions in the field of spinal instrumentation and other related products.

2 Methods

This section describes the approach used to gather, select and analyse the relevant data for this survey.

2.1 Scope of the survey

A systematic literature search was conducted to identify the articles describing FEMs of the CS. The search was conducted by the two authors using the following electronic databases: PubMed/Medline, Web of Science, Scopus, IEEE Xplore, Google Scholar and ProQuest Research Library. The terms used for the search were (Finite element, AND/OR Numeric* i.e., word combinations beginning with the term "Numeric", AND/OR Computational), AND/OR (Analys*, AND/OR Model*, AND/OR Study) AND (Cervical AND/OR Spine AND/OR Neck). References from selected articles were carefully scanned to identify studies that were not retrieved in any of the databases. Data was collected up to November 2016 and only articles written in the English language were covered.

2.2 Inclusion and exclusion criteria

Articles were considered for revision when: (i) the article described a FEM of the full human CS (i.e., C0/Head/Occiput or C1 to C7 or T1) or at least a functional unit of the same (a functional spinal unit—FSU or motion segment—is the smallest physiological motion unit of the spine to exhibit biomechanical characteristics similar to those of the entire spine), i.e., articles describing the model of a single component (e.g., model of a single vertebra/facet joint/intervertebral disc/spinal cord), or articles describing non-human models (e.g., model of the sheep CS), or articles describing models of the thoracic/lumbar human spine were not included, and (ii) the article described a static or quasi-static analysis, i.e., articles describing dynamic analyses (e.g. vehicle collisions or other impact-related simulations) were not included unless the article also described a static or quasi-static analysis. Relative to the first criterion, note that an FSU is made up of two adjacent vertebrae and the intervertebral disc (IVD) between them. Such a unit is named after the vertebrae that it includes (e.g., the C2–C3 FSU is made up of the second and third cervical vertebrae, and the IVD between them; whereas the C2–C4 motion segment is made up of two adjacent FSUs: C2–C3 and C3–C4).

Articles describing models for any population and articles describing any condition (i.e., normal/pathological, with/without instrumentation) of the human spine were included. Models were considered normal/healthy/non-pathological when the authors explicitly mentioned any of those terms or when the authors referred to the absence of bony (and/or other tissue) abnormalities, significant degenerative disease or trauma. Any means used for such asseveration(s) were considered valid, including physical gross or detailed examination, radiographic studies, computed tomography (CT) scans, dual energy radiograph absorptiometry studies, etc. Contrarily, models were considered pathological when they exhibited any condition relative to a spinal disease/disorder, both pre- and post-treatment. Models modified to study the role of specific anatomical structures were also considered as pathological models. Articles were excluded from this revision when they failed to comply with any of the aforementioned criteria.

2.3 Search outcome and data abstraction

Following the main search and after removing duplicated articles, a total of 332 results were considered to be appropriate for additional scrutiny based on the content of their title and abstract. These articles were revised to investigate their compliance with the inclusion criteria. Following the exclusion of non-relevant results, 122 articles were left for further analysis. These articles were then grouped by common authors under the assumption that such articles would be related to the same FEM or some version of such model (i.e., all articles with any author in common were placed in a folder). This process yielded 44 groups of articles or ‘presumed’ models.

Each group of articles was studied to extract a set of model parameters associated with model identification, model structure, simulation structure, verification, and validation. These parameters are based on the work by Erdemir et al. [7], which describes reporting considerations of finite element studies in biomechanics as a means to establish a guideline for model development and dissemination to enhance their reproducibility and reusability (Fig. 1).

In Fig. 1, direct validation refers to validation in which experimentation data was collected by the model developer such that the output from the experiment directly matches the predicted output from the model. Indirect validation refers to validation which uses data that has been previously reported in scientific literature. Only explicitly mentioned values for the given parameters were extracted from the articles. Illustrations of the models, if provided, were not interpreted to extract any information. The data was recorded in one independent pre-structured form per article and all independent forms were compiled into structured tables.

3 The survey

This section outlines the main findings of the complete survey associated with model identification, model structure, simulation structure, verification, and validation. The section also presents a set of tables (for a selection of the revised articles) with values for parameters associated with model identification and model structure. Tables with data for all the revised articles are provided as supplementary online material (see Online Resource 1). The, so-called, selection of revised articles includes those studies with a citation rate per year greater than or equal to 7.5 (i.e., 7.5 [8], 7.6 [9], 8.6 [10], 8.7 [11], 8.9 [12], 9.2 [13], 9.3 [14], 10 [15], 10.5 [16], 10.7 [17], 11 [18], 12.3 [19], 12.5 [20], 13.1 [21], 14.7 [22]). This rate was calculated by dividing the number of citations of each article (as per Google Scholar on February 21, 2017) by the number of years from its year of publication until 2017. This selection of articles accounts for eight of the ‘presumed’ 44 different models. To account for 12, the four articles with the next highest citation rate per year (6.7 [23,24], 6.3 [25], and 6.1 [26]), excluding those for which there was already an article from the same group within the first eight articles (7.4 [27,28], 7.2 [29], 6.7 [30], 6.6 [31]), are also included. Some models from the 90s are included because they still recurrently serve as basis for newer models. All models published after 2015 (e.g., [32–35]) are included in the tables pertaining model identification because they may not have had enough time to accumulate a significant number of citations and, thus, are not favoured by the selection criteria. Their remaining details, though, are only provided in the supplementary material (see Online Resource 1).

In all tables, the revised models are organized chronologically per year of publication (indicated by the model number) but grouped per related authors (indicated by a slight indentation to the right). This order allows for a direct comparison between related articles—and possibly related models—and clarifies the trends in model development.

3.1 Model identification

Table 1 summarizes parameters for model identification for the selection of articles. Relative to the primary utility of the model, two items are provided per model. The first item

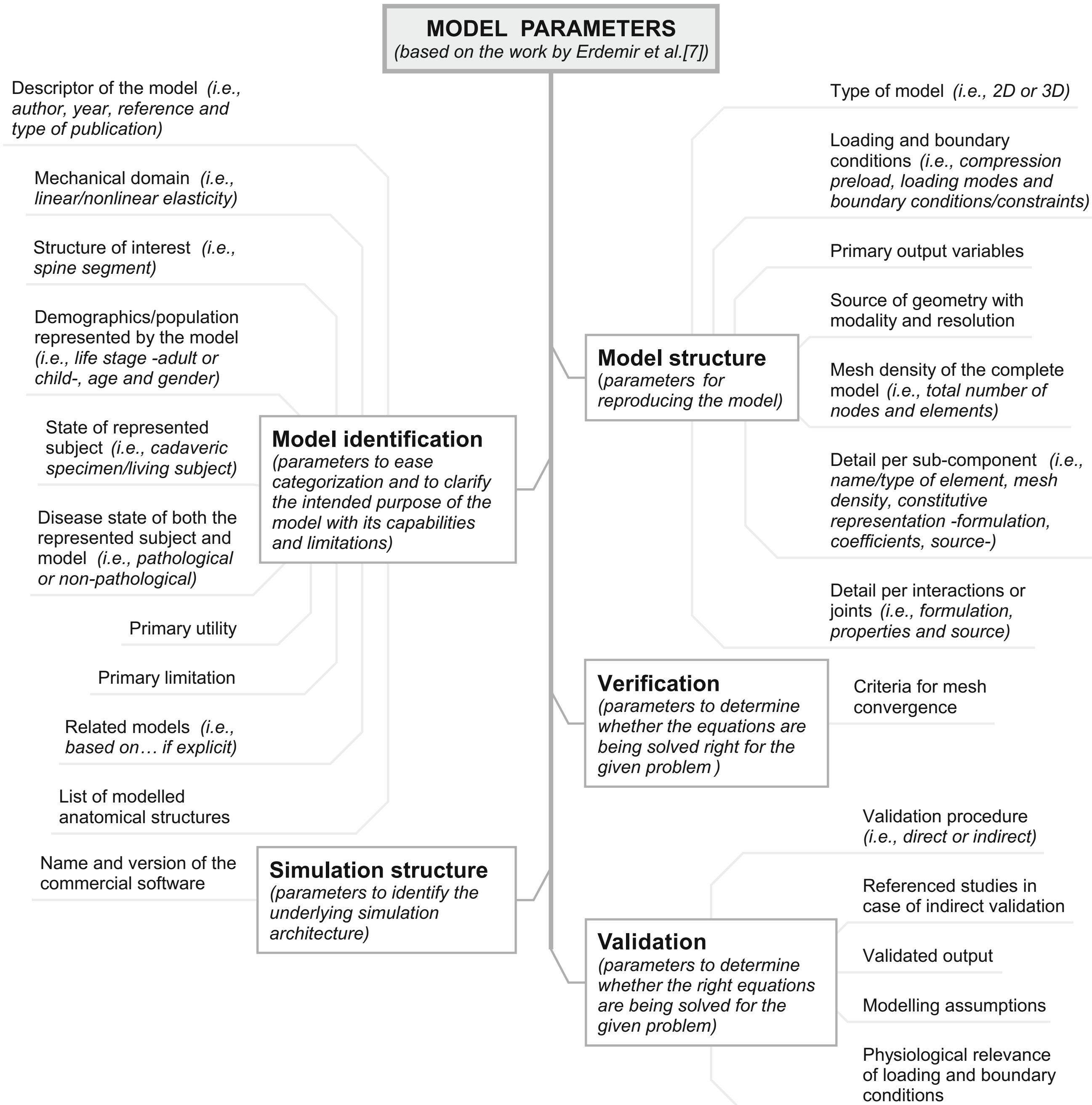

Fig. 1 Model parameters to be extracted per FEM identified in the revised articles

refers to a category (1–3) that differentiates between models used to study the biomechanics (under normal physiological loads) of (1) the injured/instrumented spine and (2) the healthy spine—role of anatomical structures, influence of geometric/material parameters and effects of ageing—. The third group (3) refers to models used to study the biomechanics of the spine in vehicle collisions or similar. This latter group, though out of the scope of this survey, was included because a few of the models within the scope of this survey were also suitable for such types of analyses. The second item provides keywords relative to the exact intended use of the model.

The complete survey revealed that eight of the studies were reported as linear, 55 as nonlinear and 59 were not explicitly described as either by the authors. Despite the importance of the upper CS (C0–C2), only ten articles for C0–C2 were found, whereas 45 articles were found for the segments from C3 or C4 to C5 or C6. Additionally, only 17 articles referring to full CS models (i.e., from C0 or C1 to C7 or any thoracic vertebra) were found. In terms of the population represented

Table 1 Parameters for model identification of FEMs of the cervical spine (generalities and utility)

No.	First author [Ref.]	TP	Mech. domain	Spine segm.	Life stage	Age (in years)	Gender	State of repres. organism	Disease state of subject (source)	Disease state of model(s)	Primary utility (categ.)	Primary utility (key-words)	Based on...
1	Saito [24]	J	L[a]	C0–T2	Ch, A					NP & P	1	Laminectomy,deformities	
9	Goel [22]	J	NL	C5–C6	A	68	M	CS	NP	NP & P	2	Role of facet angle, fibre angle, Luschka's joints, and uncinate processes	
10	Kumaresan [11]	J	L	C4–C6	A	33		CS	NP	NP	2	Material properties	[36]
17	Kumaresan [19]	J	NL	C4–C6				CS		NP & P	2	Degenerative changes, disc, osteophytes	[36,37]
65	Kallemeyn [12]	J		C2–C7	A	74		CS		NP	2	Model validation	[38]
82	Womack [8]	J	NL	C3–C7	A	64	F	CS		NP & P	1	Biomech., disc replacement	[39]
84	Faizan [13]	J		C3–C7						NP & P	1	Biomech., bi-level disc replacement, bi-level fusion, adjoining level disc replacement/fusion	[10,22,40,41]
85	Faizan [10]	J		C3–C7						NP & P	1	Biomech., shape (oval/spherical)/location (superior/inferior ball) of disc replacements	[22,40–42]
6	Maurel [23]	J	NL	C3–C7							2	Biomech., geometry, posterior articular facets, inter-individual variability	
111	Barrey [43][b]	J	NL	C3–C7						NP	2	Biomech., compressive follower preload	[23,44,45]
25	Brolin [21]	J	NL	C0–C3		27	M			NP	2	Model validation, ligaments, material property sensitivity study	[46,47]
51	Hussain [26]	J		C3–T1	A	38	F		NP	NP & P	1	Screw angulations, rigid plate-screw construct	
22	Teo [15]	J	NL	C4–C6	A	68		CS		NP & P	2	Model validation, role of structures (ligaments, facets, nucleus pulposus)	
40	Zhang [20]	J	NL	C0–C7	A	68		CS			2 & 3	Biomech., static loading, dynamic loading	[48,49]
114	Hsu [50][b]	J	NL	C3–T2				LS	NP	P	1	Anterior cervical plate systems, key control variables, fixation stability, Taguchi method	

Table 1 continued

No.	First author [Ref.]	TP	Mech. domain	Spine segm.	Life stage	Age (in years)	Gender	State of repres. organism	Disease state of subject (source)	Disease state of model(s)	Primary utility (categ.)	Primary utility (keywords)	Based on...
45	Galbusera [9]	J	NL	C4–C7						NP & P	1	Biomech., disc arthroplasty	[51]
37	Ha [18]	J	NL	C3–C6	A	69	M	LS		NP & P	1	Biomech., fusion (graft), artificial disc prosthesis (elastomer-type)	
44	del Palomar [4]	J		C0–C7	A	48	M				2	Biomech., disc, quasi-static loading	
46	Greaves [17]	J		C4–C6						P	2	Biomech., spinal cord, injury patterns	
112	Duan [32][b]	J	NL	C3–C7	A	32	M	LS		NP & P	1	Biomech., fixation techniques (pedicle, lateral mass, and transarticular screw)	[52]
78	Lee [16]	J	NL	C2–C7	A	21	M	LS	NP	NP & P	1	Biomech., artificial disc replacement, fixed- and mobile-core prostheses	
113	Ganbat [53][b]	J	L	C5–C6	A	21	M	LS		NP & P	1	Biomech., total disc replacement, mechanical loading, heterotopic ossification formation	[16]
116	Park [54][b]	J		C2–C7	A			LS	NP	NP & P	2	Biomech., disc and facet joint degeneration	[16]
118	Lee [33][b]	J		C2–C7	A	26	M	LS		NP & P	1	Biomech., artificial disc replacement, mobile core artificial disc Baguera C	[16]
120	Mackiewicz [35][b]	J	NL	C2–C6	A	24	M			NP & P	1	Biomech., one-level stabilization, anterior static and dynamic plates	
117	Zhang [34][b]	J	NL	C0–C2	A	31	M	LS	NP	NP & P	1	Biomech., basilar invagination with irreducible atlantoaxial dislocation, Cage + Plate device, Cage + Trans oral atlantoaxial reduction plate device	

Table 1 continued

No.	First author [Ref.]	TP	Mech. domain	Spine segm.	Life stage	Age (in years)	Gender	State of repres. organism	Disease state of subject (source)	Disease state of model(s)	Primary utility (categ.)	Primary utility (key-words)	Based on...
119	Liu [55][b]	J	NL	C0–C2	A	35	M	LS	NP	NP & P	1	Construct stability, occipito-atlanto-axial fixation, occipital plate-rod-screw fixation with or without C1 lateral mass screw	
121	Zafarparandeh [56][b]	J		C2–C7	A	35		LS		NP	2	Model assumptions, symmetry, exact geometry	[57]
122	Zafarparandeh [58][b]	J		C2–C7	A	35		LS	NP	NP	2	Model assumptions, symmetry	[57]
107	Mo [25]	J		C3–C7	A	32	M	LS	NP	NP & P	1	Biomech., arthroplasty, standalone U-shaped disc implant	
115	Li [59][b]	J		C0–C3	A	28	M	LS	NP	NP & P	1	Biomech., stability, transarticular screw and rod fixation techniques	

No. model number; *TP* type of publication; *Mech.* mechanical; *segm* segment; *repres.* represented; *L* Linear; *NL* nonlinear; *Ch* child; *A* adult; *F* female; *M* male; *CS* cadaveric specimen; *LS* living subject; *NP* non-pathological; *P* pathological; *Biomech.* biomechanics; *categ.* category

1, Biomechanics of instrumented spine or post-surgery spine under normal physiological loads; 2, Biomechanics of healthy spine under normal physiological loads, role of any given anatomical structure on the biomechanics of the spine, influence of geometric or material parameters on the biomechanics of the spine, effects of ageing (degeneration) on the biomechanics of the spine; 3, Biomechanics of spine in vehicle collisions or similar

[a] With Displacement Incremental Method to depict the nonlinear behaviour of deformities; [b] Model published after 2015 which may not have had enough time to accumulate a significant number of citations

by the models, 77 articles refer to adult models, whereas only two articles refer to both adult and children models. Only 67 articles report data on the age of the source for the geometry of the model. Of these, sixteen are below 30 years old, thirty-four are between 30 and 40 years old and sixteen are between 48 and 74 years old. The children models are reported by only one article as corresponding to one-, three- and six-year-old children. Relative to gender, only 64 articles provide such information, with 38 describing male subjects/specimens and 26 describing female subjects/specimens. Both cadaveric specimens and living subjects are nearly equally used as the source for geometry with 39 and 42 articles reporting so, respectively. Non-pathological models alone are used in 15 studies and both non-pathological and pathological models are used in 72 studies. Generally speaking, articles that aim to study any given pathology use a non-pathological model (or intact/normal model) as baseline or reference. Only ten studies report the use of a pathological model alone. Detailed descriptions of the specific procedures followed to obtain the pathological models from the nonpathological ones are not within the scope of this article but are available upon request to authors. Relative to the three categories defined for primary utility, the revised articles report models fairly distributed between the first two: 61 studies use models to study the biomechanics of the instrumented/post-surgery spine under normal physiological loads, while 58 studies use models to understand the biomechanics of the healthy spine under normal physiological loads, including the role of anatomical structures, influence of geometric/material parameters, and effects of ageing/degeneration. More specifically, the applications for the reported models include study of: neck biomechanics [1,28,60], effects of ageing/degeneration [19,37,61,62], role of anatomical structures—IVD/facet angle/uncinate processes/Luschka joints—[22,29,40], effects of variations in material properties and/or geometry [11,44], biomechanical effects of surgical procedures—laminectomy with/without facetectomy and/or laminoplasty—[24,36,63], effects of other surgical treatments—discectomy/interbody fusion/other instrumentation procedures—[8,10,64,65], among others.

Table 2 summarizes the anatomical structures per model from the selection of revised articles. Note that, unless explicitly described as cartilaginous, endplates were assumed as bony. The indentations of the headings of this table indicate a substructure of a structure (e.g., annulus fibrosus, nucleus pulposus and cartilaginous endplates are substructures of the IVD, while annulus matrix and fibres are substructures of the annulus fibrosus). The bolded "**X**" indicates that further detail in terms of element type/name and/or constitutive representation is provided in the original article and, thus, in the corresponding table of this survey (see Online Resource 1). On the whole, ligamentous models of the CS usually include the anterior/posterior longitudinal ligaments, ligamentum flavum, interspinous ligament and capsular ligament. These ligaments are included in over 80% of the revised models. The intertransverse ligament and anterior/posterior atlanto-axial membranes, on the contrary, are rarely included in the models with less than 3, 1 and 2% of the models including them, respectively. Muscles are also not generally included in models for studying the static/quasi-static behaviour of the CS. Indeed, they are found in less than 3% of the revised models. Particular posterior elements—pedicles, laminae, etc.—, despite being commonly modelled, are only explicitly mentioned as being included by less than 20% of the articles, and specific modelling details for such structures are only reported in less than 3% of the models.

Not reported in Tables 1 or 2 are the primary limitations of the models reported by the authors. Although nearly 32% of all the revised articles did not include any reference to limitations of their model/study, this parameter is important because it hinders the primary utility of the model and

affects the interpretation of the model's prediction [7]. The two-dimensional model by Saito et al. [24], for instance, is limited to single plane simulations and over-simplifies vertebral geometry and intervertebral joints, most possibly predicting unrealistic load-sharing/stress distributions. Relative to the mechanical domain, despite being a widely-used approach, many authors suggest that the linearity assumption is a limitation to their study that should only be used as a first step in the process because it is well-known that the human cervical response is nonlinear [25,30,61,66]. Kumaresan et al. [11] further suggest that model applicability beyond loading above 1.8 Nm is a limitation of their (linear) analysis and that at higher levels of loading, it may be prudent to incorporate nonlinear and viscoelastic effects of the spinal structures. Other authors similarly suggest that the linearity assumption is valid up to a certain range of loading [15,21,31] and Pitzen et al. [67] claim that this assumption is also valid to simulate performance in an immediate postoperative scenario (i.e., the "initial stability", especially following osteosynthesis), which is mainly a function of bone quality, ligamentous tension, and the implant used.

Another limitation mentioned by some authors is the use of models symmetric about the midsagittal plane, which cannot predict coupled motions in flexion/extension loading modes [15,68]. Regarding modelling details, Yoganandan et al. [37], for example, mention that the lack of ligaments in their model is acceptable as long as the model is exercised under axial compression, where the action of CS ligaments is negligible. Another common limitation with respect to modelling details has to do with the need to model some anatomical structures (e.g., the boundary between annulus and nucleus in the IVD, the geometry of IVDs/facet joints, the thickness of cartilage/endplates) based on qualitative data reported in literature because of the failure of CT scans to capture such

Table 2 Parameters for model identification of FEMs of the cervical spine (anatomical structures included per model)

Model number (No.)	First author [Ref.]	Vertebrae	Body	Cortical bone	Cancellous bone	Bony endplates	Uncinate processes	Posterior elements	Pedicles	Laminae	Articular processes	Transverse processes	Spinous processes	Intervertebral discs	Annulus fibrosus	Annulus matrix	Annulus fibres	Nucleus pulposus	Cartilaginous endplates
1	Saito [24]	X	X	**X**	**X**	**X**		X		X	**X**	X	X	X	**X**			**X**	
9	Goel [22]	**X**	X	**X**	**X**	**X**	X	**X**	X	X	X	X	X	X	X	**X**	**X**	**X**	
10	Kumaresan [11]	X	X	**X**	**X**	**X**		**X**	**X**	**X**			**X**	X	**X**			**X**	
17	Kumaresan [19]	X	X	**X**	**X**	**X**		**X**	X	X		X	X	X	X	**X**	**X**	**X**	
65	Kallemeyn [12]	**X**	X	**X**	**X**	**X**		**X**						X	X	**X**	**X**	**X**	
82	Womack [8]	X	X	**X**	**X**			X						**X**	X	**X**	**X**	**X**	**X**
84	Faizan [13]	X	X	**X**	**X**	X	X	**X**			X					**X**	**X**		
85	Faizan [10]	X	X	**X**	**X**	X	X	**X**						X	X	**X**	**X**	**X**	
6	Maurel [23]	X	**X**	**X**	**X**	**X**		**X**	**X**	**X**	X	X	**X**	**X**	X	X	X		
111	Barrey [43][h]				**X**														
25	Brolin [21]	X	X	**X**	**X**	**X**		X						X	X	**X**	**X**	**X**	
51	Hussain [26]	X	X	**X**	**X**	**X**		**X**						X	**X**			**X**	
22	Teo [15]	**X**	X	**X**	**X**	**X**		**X**						**X**	**X**			**X**	
40	Zhang [20]	X	X	**X**	**X**	**X**		**X**						X	**X**			**X**	
114	Hsu [50][h]	**X**	X	X	**X**			X						**X**	X			X	
45	Galbusera [9]	X	X	**X**	**X**	**X**		X						X	X	**X**	**X**	**X**	
37	Ha [18]	X	X	**X**	**X**	**X**		**X**						X	X	**X**	**X**	**X**	
44	del Palomar [14]	**X**	X					X						**X**	**X**	**X**		**X**	X
46	Greaves [17]	**X**	X					X						**X**				**X**	
112	Duan [32][h]	X	X	**X**	**X**	**X**		**X**						X	**X**			**X**	
78	Lee [16]	X	X	**X**	**X**	**X**		**X**						X	X	**X**	**X**	**X**	
113	Ganbat [53][h]	X	X	**X**	**X**	**X**		**X**						X	X	**X**	**X**	**X**	
116	Park [54][h]	X	X	**X**	**X**	**X**		**X**						X	X	**X**	**X**	**X**	
118	Lee [33][h]	X	X					X						X	X	**X**	**X**	**X**	
120	Mackiewicz [35][h]	X	X	**X**	**X**	X		X						X	X	**X**	**X**	**X**	
117	Zhang [34][h]	X	X	**X**	**X**			X											
119	Liu [55][h]	X	X	**X**	**X**			**X**											
121	Zafarparandeh [56][h]	X	X	**X**	**X**	**X**		**X**						X	X	**X**	**X**	**X**	
122	Zafarparandeh [58][h]	X	X	**X**	**X**			**X**						X	X	**X**	**X**	**X**	
107	Mo [25]	X	X	**X**	**X**	**X**		X						X	X	**X**	**X**	**X**	
115	Li [59][h]	**X**	X		**X**			X											

Model number (No.)	First author [Ref.]	Ligaments	ALL	PLL	LF	ISL	SSL	CL	TL	AL	ApL	VC	AAOM	PAOM	TM	Joints / Contacts	Uncovertebral joints	Facet joints	Synovial fluid	Cartilage	Synovial membrane	Atlanto-axial joint	Medial	Lateral	Atlanto-occipital joint	Other structure/component	Occiput	Disc implant	Cage/spacer implant	Bone graft	Fixation plate	Screws	Other(s)
1	Saito [24]	**X**	X	X	X	X	X																			X	**X**						X[a]
9	Goel [22]	X	**X**	**X**	**X**	**X**		**X**								X	**X**	**X**															
10	Kumaresan [11]	X	**X**	**X**	**X**	**X**		**X**																									
17	Kumaresan [19]	X	**X**	**X**	**X**	**X**		**X**								X	**X**	X	**X**	**X**	**X**												
65	Kallemeyn [12]	X	**X**	**X**	**X**	**X**		**X**																									
82	Womack [8]	X														X		X		**X**						X		**X**					
84	Faizan [13]	X	**X**	**X**	**X**	**X**		**X**								X	X	**X**								X		**X**			**X**	**X**	
85	Faizan [10]	X	**X**	**X**	**X**	**X**		**X**								X	**X**	**X**								X		**X**					
6	Maurel [23]		**X**	**X**	**X**	**X**	**X**	**X**								X		**X**															
111	Barrey [43][h]							**X**																									
25	Brolin [21]	X	**X**	**X**	**X**	**X**		**X**	**X**	**X**	**X**	**X**	**X**	**X**	**X**	**X**		X		**X**		X	**X**	**X**	**X**	X	**X**						**X**[b]
51	Hussain [26]	X	**X**	**X**	**X**	**X**		**X**								X		X		**X**						X				**X**	**X**	**X**	
22	Teo [15]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		**X**															
40	Zhang [20]	X	**X**	**X**	**X**	**X**	**X**	**X**	**X**	**X**	**X**					X		**X**				X	**X**			X	**X**						**X**[c]
114	Hsu [50][h]	X	X	X	X	X		X																		X					**X**	**X**	
45	Galbusera [9]	X	**X**	**X**	**X**	**X**		**X**								X		**X**								X		**X**					
37	Ha [18]	X	**X**	**X**	**X**	**X**		**X**								X		**X**								X		**X**		**X**			
44	del Palomar [14]	X	**X**	**X**	**X**	**X**			**X**	**X**						X		**X**				X	**X**			X	X						
46	Greaves [17]	X	**X**	**X**	**X**	**X**		**X**								**X**										X							**X**[d]
112	Duan [32][h]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		**X**								X						**X**	**X**[e]
78	Lee [16]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		X		**X**						X		**X**				**X**	
113	Ganbat [53][h]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		X		**X**						X		**X**					
116	Park [54][h]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		X		**X**													
118	Lee [33][h]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		X		**X**						X		**X**	**X**		**X**	X	
120	Mackiewicz [35][h]	X	**X**	**X**	**X**	**X**		**X**								X		**X**								X				**X**	**X**	X	
117	Zhang [34][h]	X	**X**					**X**	**X**	**X**	**X**	**X**	**X**	**X**	**X**	**X**		**X**		**X**		X	**X**	**X**	**X**	X	**X**		**X**	**X**		**X**	**X**[f]
119	Liu [55][h]	X	X		X			X	**X**	X	X	X	X	X		**X**		X		**X**						X	X						**X**[g]
121	Zafarparandeh [56][h]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		**X**															
122	Zafarparandeh [58][h]	X	**X**	**X**	**X**	**X**	**X**	**X**								X		**X**		**X**													
107	Mo [25]	X	**X**	**X**	**X**	**X**		**X**								X	**X**	**X**								X		**X**	**X**	**X**	**X**	**X**	
115	Li [59][h]																									X	X					**X**	**X**[e]

The grey scale and indentation of the headings indicate the different levels of the anatomical structures; **X**, More detail is provided in Table 4 for the ISL and in the corresponding tables of the supplementary online material for all other anatomical structures (see Online Resource 1)

ALL anterior longitudinal ligament; *PLL* posterior longitudinal ligament; *LF* ligamentum flavum; *ISL* interspinous ligament; *SSL* supraspinous ligament; CL capsular ligament; *TL* transverse ligament; *AL* alar ligament; *ApL* apical odontoid ligament; *VC* vertical cruciate; *AAOM* anterior Atlanto-occipital membrane; *PAOM* posterior Atlanto-occipital membrane; *TM* tectorial membrane

[a] Intertransverse Ligament (ItL); [b] Interaction between dens and transverse ligament; [c] Ligaments: Anterior membrane and Posterior membrane; [d] Spinal cord, denticulate ligaments, and attachments in the anterior compartment between the dura mater and vertebral bodies; [e] Rods; [f] Posterior Atlanto-Axial Membrane (PAAM) and Transoral Atlantoaxial Reduction Plate (TARP); [g] Occipitoatlantoaxial fixation; [h] Model published after 2015 which may not have had enough time to accumulate a significant number of citations (further detail for this model is not included in the upcoming tables within this article but is provided in the supplementary online material—see Online Resource 1—)

details [49,69]. Furthermore, modelling the nucleus pulposus as an incompressible hydrostatic fluid, a practice commonly observed in the reported models, may not allow studying the mechanics of the degenerated disc in order to address the discogenic pain [29]. Similarly, models of single motion segments do not allow for the analysis of the adjacent segments [61,70].

Another commonly mentioned limitation is the lack of muscles in the models. The majority of analysis do not account for the stabilizing role of muscles in spinal responses [10,12,21,68]. However, authors claim that such lack of muscles is only critical to maintaining stability in dynamic/impact loading situations [30,71]. Goel et al. [22] also suggest that the issue of initial ligament laxity should be considered as it has been suggested that a fairly large neutral zone exists in the CS. The consequence of omitting a neutral zone is that the initial slopes of the load-displacement plots are different for the experimental and theoretical data.

Authors also report limitations relative to validation. Differences between the modelled structure and the specimens tested are an important issue to consider [20,26,71]. Models are normally constructed using geometric data from a single cadaveric specimen or living volunteer, and thus may not represent the range of anatomic variation of the human population [60,62,66,69]. Additionally, computer models provide results that are difficult or even impossible to acquire experimentally and, thus cannot be validated (e.g. stresses/strains). Such results can be used to indicate and compare trends against the validated intersegmental rotational motions and do not necessarily represent correct absolute values [72]. Similarly, while intact models are often validated against experimental data, instrumented models are generally not validated [9]. The nature of the surgery, screw placement, and the properties of implanted devices are known to affect the biomechanics of the spine [13,26].

Limitations regarding material properties are also commonly mentioned. Firstly, properties obtained from the literature may not compare with the *in vitro* model specimen [20,26,73] and it is well-known that material properties vary depending on age, sex, vertebral levels, and other parameters [73]. Indeed, the age of the specimen often has no relation to the mechanical properties of the model because the specimen is used for geometrical purposes only [65]. Secondly, though realistic results can be achieved, modelling ligaments with discrete spring elements is a simplification that prevents unrealistic shear forces in the ligaments and reduces computation time but does not take into account the contact interactions between different ligaments and between ligaments and vertebrae [21]. Thirdly, despite the fact that a degenerative disc segment is often accompanied by ruptures, fissures, cracks, eburnation, tears, osteophytes, endplate calcification, annular fibres laxity, and ligament(s) failure, studies often do not account for such conditions and the intact models are left almost unmodified (both in terms of geometry and material properties) for the analyses of pathological conditions [25,32,66]. Also, degeneration is usually highly variable from patient to patient and, thus, the definition of a standard degenerative state to be used as a basis for the analyses is not straightforward [51]. In models with implanted devices, modelling the bone–implant or bone–screw interface as bonded neglects any possible micro motion, which may not be a realistic simplification, particularly in the immediate postoperative time when osteointegration has not been achieved yet [9,32].

3.2 Model structure

Relative to model structure, few articles describe two-dimensional models [24,74]. The particular detail on loading, boundary conditions and output variables is not included in this article but is available upon request to authors. Generally speaking, pure moments in the range from 0.0 to 3.0 Nm are applied to the superior surface of the superior-most vertebra of the model, which is left unconstrained to accept the external load vector. The moments are applied in flexion/extension and/or left/right lateral bending and/or clockwise/counter clockwise axial torsion with/without an axial compressive preload. Pure compression alone is also applied in some cases. Regarding boundary conditions, the inferior-most nodes of the inferior-most vertebra (or the inferior-most vertebra itself) are usually fully constrained in all degrees of freedom. Output variables include stress patterns and displacements, reaction forces, axial stiffness, angular rotation, etc.

Table 3 summarizes the following parameters for the selection of articles: axial compression preload, loading modes, source of geometry, resolution, and mesh density of the complete model—total number of nodes/elements—. CT scans are normally used for bony structures, whereas cryomicrotome sections and magnetic resonance images (MRI) are used for soft tissue. Anatomic textbooks and other information from the literature are also used to establish the location of ligament insertions and to model soft tissues. Low dose bi-planar X-rays are used to obtain geometrical parameters for parametric models.

Three-dimensional digitizers are also utilized to obtain coordinates and model the geometry of the spine. In-house subjects/specimens are often used as a source for such data. In some cases, medical images and photographs from—for example—the Visible Human Male dataset [88] of the National Library of Medicine are used as the source for the geometric data [17,71,89]. However, although many authors report the means for digitalizing the geometry of the spine, the specific protocols/parameters used for such digitalization are barely reported and, thus, replication of the studies is not a straightforward task. For instance, less than 10% of the revised articles mention details relative to the anatomic position of the digitized spine (e.g., a 5-degree segmental angular measure was used to create the lordotic curve for the C5–C6 model [60]; or the completed C0–C7 FEM was configured to contain a lordosis of about 37 degrees, which is consistent with the neck posture of a seated 50th percentile male [30]; or the extended C4–C7 model had a 12-degree lordosis or an average lordotic curvature of 4 degrees for each motion segment [1]; or lordosis was measured using the four-line Cobb method to be 25 and 22 degrees for the asymmetric and symmetric models, respectively [56]).

Tables 4 summarizes modelling details (i.e., name/type of element and constitutive representation) of the interspinious ligament for the selection of models. Note that this is a sample table. The supplementary online material (see Online Resource 1) contains tables with the modelling details (i.e., name/type of element, mesh density and constitutive representation—formulation, coefficients, source—) for each of the sub-components and for all of the revised models. Additional parameters relative to fibre configura-

Table 3 Parameters relative to model structure of FEMs of the cervical spine (loading modes, source of geometry and mesh density)

No.	First author [Ref.]	Axial compression preload	Loading modes	Source of geometry	Resolution of CT	Total number of nodes	Total number of elements
1	Saito [24]	150 N				Intact model: 863; Post-laminectomy model: 793	Intact model: 1743; Post-laminectomy model: 1607
9	Goel [22]	73.6 N	Axial compression alone or in combination with flexion, extension, lateral bending, or axial rotation (complex loads)	CT	Slice intervals of 1.5 mm	4219	5577
10	Kumaresan [11]		Flexion, extension, lateral bending and axial torsion loading modes	CT and cryomicrotome sections	1.0 mm CT slices		
17	Kumaresan [19]	80 N	Compression mode	CT and cryomicrotome sections	1.0 mm CT slices	15577	12712
65	Kallemeyn [12]		Flexion–extension, lateral bending, and axial rotation	CT for bony structures and MRI for intervertebral discs			130000
82	Womack [8]	40 N	Compression, flexion-extension, lateral bending and axial torsion	CT	0.5 mm CT slices at 1.0 mm intervals [18]	126000	
84	Faizan [13]	Follower load of 73.6 N	Flexion-extension, lateral bending and axial rotation with and without compression preload			Intact model: 24732	Intact model: 21895
85	Faizan [10]	Follower load of 75 N	Extension, flexion, lateral bending and axial rotation			24732	21895
6	Maurel [23]	6 N	Flexion, lateral flexion, axial torque and extension	Geometrical primitives modelled based on three-dimensional coordinates measured using a spatial measuring machine		Each functional unit: 1803	Each functional unit: 1506

Table 3 continued

No.	First author [Ref.]	Axial compression preload	Loading modes	Source of geometry	Resolution of CT	Total number of nodes	Total number of elements
25	Brolin [21]		The model was subjected to axial rotation, flexion, extension, and lateral bending for validation purposes	CT for vertebrae. The soft tissues were added to the model based on data from the literature [75,76,76,78]. Modelling of the intervertebral disc, with the annulus fibrosis surrounding the nucleus pulposus and the facet joints, has been described previously by Halldin [46] and Halldin et al. [47]	Slice intervals of 1.0 mm		
51	Hussain [26]	73.6 N	Flexion/extension, lateral bending and axial rotation	CT. Ligament insertion points and areas of cross section were closely matched with the available published data [79,80]			
22	Teo [15]	73.6 N	Flexion, extension and axial compressive load configurations	Digitized geometrical coordinates obtained using a flexible 3D movement digitizer. An anterior disk height of 5.5 mm, a posterior disk height of 3.5 mm [81], and the thickness of the annulus in the anterior region was taken to be 1.2 times that of the posterior region [82] were adopted		11187	7730
40	Zhang [20]	50 N	Flexion, tension, axial rotation and lateral bending	Geometrical coordinates of vertebrae and skull were measured using a flexible digitizer. In the modelling, the data for the basic geometries of the intervertebral discs were taken from average literature values [81]		28638	22094

Table 3 continued

No.	First author [Ref.]	Axial compression preload	Loading modes	Source of geometry	Resolution of CT	Total number of nodes	Total number of elements
45	Galbusera [9]	100 N	Flexion and extension	CT			
37	Ha [18]	73.6 N	Flexion-extension, axial rotation and lateral bending	CT	Slice intervals of 1.0 mm		38718
44	del Palomar [14]	50 N	Flexion-extension, axial rotation	CT			
46	Greaves [17]			Geometry of the vertebrae and spinal cord was obtained from transverse cryosection images provided by the Visible Human Project, of the National Library of Medicine. Geometries of the remaining model components were based on anatomic descriptions in the literature [80,83–86]	Slice intervals of 1.0 mm		
78	Lee [16]	50 N	Flexion-extension, lateral bending, and axial torsion	CT	Slice intervals of 1.0 mm	Intact model: 38984	Intact model: 77991
107	Mo [25]	73.6 N	Flexion, extension, lateral bending and axial torsion	CT. The intervertebral disc was constructed as a continuum structure that occupied the intervertebral space and was partitioned into annulus fibrosus and nucleus pulposus in approximately 6-to-4 ratio [87]	Slice intervals of 0.5 and 0.6 mm resolution	Intact model: 49916	Intact model: 127148

CT computerized tomography; *MRI* magnetic resonance imaging

Table 4 Parameters relative to model structure of FEMs of the cervical spine (modelling details for the Interspinous Ligament—ISL—)

No.	First author [Ref.]	Mesh	Constitutive representation		
		Element name/type	Formulation	Coefficients	Source
9	Goel [22]	3D nonlinear cable, no compression		$E = 4$ (< 20–40%) 8 ($> 40\%$), $\nu = 0.30$, $a = 13.1$	[90–98]
10	Kumaresan [11]	Two-noded axial elements	Linear, homogeneous and isotropic	$E = 3.4$, $\nu = 0.39$	Mean values: [5,24,28,99,100]; Extreme values: [24,90,97,101–104]
17	Kumaresan [19]	Nonlinear tension-active cable elements		*def.* = 1.3, $F = 16.9$; *def.* = 2.7, $F = 24.4$; *def.* = 4.0, $F = 29.5$; *def.* = 5.4, $F = 32.9$; *def.* = 6.7, $F = 34.9$	[29]
65	Kallemeyn [12]	3D truss elements acting nonlinearly in tension only	Hypoelastic material designation, allowing for the definition of axial stiffness as a function of axial strain	Nonlinear (axial stiffness as a function of axial strain, see [12] for detail), $\nu = 0.30$, $a = 13.1$	[11,18,22,40,42,49,73,105]
84, 85	Faizan [10,13]	T3D2; 3D truss elements	Elastic elements allowed to behave nonlinearly via a "hypoelastic" option. This option also allowed a "neutral zone" to be incorporated in which the ligament provided little stability under minimally applied external loads	$E = 5$ ($< 25\%$) 10 ($> 25\%$), $\nu = 0.30$, $a = 13$	[10,82,91,94]
6	Maurel [23]	Cable elements, tension-only		$E = 3$, Initial slackening = $-0.17 \times 10^2\%$, $a = 3$	
25	Brolin [21]			Initial model (C2–C3): Failure d_f:f_f = 7:37, Neutral zone d_n/d_f = 1/3	[106]
51	Hussain [26]	3D tension truss (two node)		$E = 2$ ($\varepsilon < 40\%$) 8 ($\varepsilon > 40\%$)	[22,107]
22	Teo [15]	3D nonlinear cable, no compression		$E = 1.5$	[22,23,36]
40	Zhang [20]	Two-node nonlinear link elements, permitting tensile axial force only	Linear elastic, homogeneous and isotropic	$\nu = 0.30$; ISL (C1–C2): $E = 10$; ISL (C2–C7): $E = 1.5$; Nuchal ligament: $E = 20$	[49]
45	Galbusera [9]	Nonlinear spring elements	Force-deflection data	Initial average length (mm) = 11.8; $F = 0$, *def.* = 0; $F = 8.5$, *def.* = 1.3; $F = 10$, *def.* = 2.8; $F = 23$, *def.* = 4.1; $F = 28$, *def.* = 5.5; $F = 32$, *def.* = 7	[80,108]

Table 4 continued

No.	First author [Ref.]	Mesh	Constitutive representation		Source
		Element name/type	Formulation	Coefficients	
37	Ha [18]	Nonlinear tension-only-spar-elements		E= 1.5, a= 13.1	[22,24,28,36,63,100,109,110]
44	del Palomar [14]	Nonlinear truss elements that only worked under tension		$E_1 = 4.9$, $E_2 = 3.1$, $\varepsilon_{12} = 26.1$, a= 13.0	[30,111–113]
46	Greaves [17]	Link elements with two nodes and linear shape functions that respond only to tensile loading (LINK 10)	Linear elastic material properties	E= 5	[80,114]
78	Lee [16]	T3D2; 3D linear contact elements applying only to the tension force		E= 1.5, $\nu = 0.30$, a= 10.0	[20,22,107,115,116]
107	Mo [25]	Tension-only axial connector	Nonlinear behaviour assigned with load–deformation curves		[80]

E elastic modulus/young's modulus (MPa); ν Poisson's ratio; a cross-sectional area (mm^2); *def.* deflection (mm); F force (N); ε strain; $d_f : f_f$ point of failure (displacement:force at end of elastic region) (mm:N); $d_n : f_n$ end point of neutral region (displacement:force at end of neutral region) (mm:N); d_n/d_f neutral region length (displacement/displacement)
Similar tables including modelling details for all sub-components and for all the revised models are provided as supplementary online material (see Online Resource 1)

tion are provided for the annulus fibre, including the number of layers and other relevant characteristics, fibre orientation and fibre content. Bony structures are mostly modelled with linear elastic and isotropic materials, though some authors also include poroelastic materials, orthotropic elastic materials or even an isotropic scalar damage material model. For the annulus matrix, both linear elastic and isotropic material and hyperelastic materials are common approaches. Annulus fibres are usually represented by means of rebar elements defined to carry only tensile forces. Both linear and nonlinear properties are found for these elements. Multiple layers of such elements are often defined. Fibres are usually oriented with alternating angles of $\pm 65°$ from the transverse plane, though other orientations are also used. Nearly all authors define the fibre content as 20% of the matrix volume. The nucleus pulposus is represented either with linear elastic and isotropic materials or as an incompressible (or nearly incompressible) fluid. Hyperelastic properties are also found in some models. Ligaments are also defined by means of linear elastic and isotropic materials. Yet, hypoelastic materials and nonlinear stress-strain relationships seem to be the preferred approach. Modelling details of implantable devices and other components (e.g., disc/cage implant, bone graft, fixation plates, screws, rods, spinal cord, etc.) are also provided in the supplementary online material (see Online Resource 1).

Details on the source of geometry and modelling approach for each particular sub-component are not included in the tables but, mostly, facet surfaces (articular processes) are oriented with 45 degrees from the transverse plane [22,40,73] with some variation in the sagittal plane alignment, according to CT geometry [10,65]. The facets are also of varying curvatures from right to left sides, indicating the possibility of varying contact during right/left loading modalities [10]. Kumaresan et al. [61] orient facets at 45 degrees with respect to the vertical axis in the adult model, and at 60, 53 and 47 degrees in the one-, three-, and six-year-old models, respectively. Imajo et al. [68] use orientations of 30, 45, and 60 degrees from the transverse plane. With respect to IVDs, their geometry is often based on mean values found in literature studies as [81,117,118] and on the geometry of adjacent vertebrae [14,23,119]. The annulus pulposus and the nucleus are distinguished based on qualitative anatomical data (from studies including [120–123]) [14,119]. Ligament insertion points and/or cross-sectional areas are chosen to mimic anatomic observations as closely as possible [62,89,115]. The anatomic and histological studies used to obtain such data include [77,80,84,112]. Note that the studies used as reference for geometry/material properties are, primarily, studies published in the 90s and earlier.

Details per interactions/joints—formulation, properties, source— are also provided as part of the supplementary online material (see Online Resource 1). Overall, the most remarkable observation relative to interactions is that most authors fail to report the specific aspects that would allow for the replication of the reported model. Indeed, less than 16% of the articles report modelling details for the uncovertebral joint, and <19 and 5% report details for the atlanto-axial and atlanto-occipital joints, respectively. For the facet joints, more articles (around 70%) provide at least a brief description of the modelling approach. 'Gap' elements and sliding contacts (with and without friction) are common approaches for defining interactions in many models. Incompressible fluid elements are used to represent synovial fluid.

3.3 Simulation structure

Generally, in terms of simulation software and version of the same used by each of the models, ABAQUS (version 1.4–6.13) is reported as being used in nearly 35% of the revised articles, followed by ANSYS (version 4.4A–11.1) and ADINA, which are used in approximately 25 and 11% of the revised articles, respectively. Detail per model is provided in the supplementary material (see Online Resource 1).

3.4 Verification

Only a few of the revised articles mention the criteria used to determine mesh convergence. Epari et al. [74] tested convergence of the model by doubling the number of elements in the mesh until the difference in the average error between strain path plots across the middle of the graft tissue obtained from the two mesh densities was less than 5%. Laville et al. [44] controlled each mesh using commonly described quality criteria (aspect ratio, parallel deviation, maximum corner angle, Jacobian ratio and warping factor) according to Knupp [124]. Galbusera et al. [66] performed a mesh sensitivity analysis in flexion and extension. The selected mesh led to differences lower than 1% in terms of range of motion with respect to a mesh with considerably higher density. Li et al. [71] used the h-convergence test to obtain the final mesh density for the intact model, with the criterion being a change of less than 1.5% in the maximum von Mises stress in each tissue in the model between successive changes in mesh density. Womack et al. [8] tested for convergence within 7% on multiple anatomical structures under a given set of loading conditions (authors provide further detail). Finally, Mo et al. [25] achieved convergence within 1% in the intact model.

3.5 Validation

The detail on validation is not included in this article but is available upon request to authors. On the whole, both direct and indirect validation are used in few of the revised articles [12,22,29]. Indirect validation alone is used in 89 of the articles [18,21,51] and is, by far, the preferred approach for validation. Studies with experimental results used for such validation include [116,118,125,126], among others. The remaining articles are either not clear in their validation procedures or do not offer any information on the topic. Validated output includes force-displacement response [15,19], moment-rotation response and/or range of motion [12,15,44], and stiffness [63].

Relative to assumptions, symmetry about the median plane is a common approach in the reported models [9, 31,67,68]. Authors also assume that material properties of soft and hard tissues of the CS are not much different from those of the lumbar spine [12,49,119] or even other similar (biological) materials [60,73] and that linearity in such properties—homogeneous and isotropic material definition—is valid in some cases despite the common knowledge that spinal materials are inhomogeneous, viscoelastic and have a nonlinear stress-strain response [30,61]. Results obtained from fusion models should be carefully interpreted due to the common assumption that bone grafts are of uniform section and are rigidly connected to the endplates, which may not be the case at the immediate completion of the surgical procedure [26,71]. The thickness of the cortical bone is often assumed to be of a certain average value although it varies depending on the location (anterior/posterior) [18].

Finally, regarding the physiological relevance of loading and boundary conditions, few articles provide such detail. Kumaresan et al. [19] indicate that the selected load magnitude is in close agreement with a previous study by Goel, et al. [22] wherein a compressive load of 73.6 N was applied. In addition, this load magnitude falls within the range of compressive load (47–276 N) calculated by measuring the intradiscal pressure and disc area. Natarajan et al. [73] mention that the 105 N preload was based on the average weight of the human head (65 N) plus a load of 40 N (assumed) to represent the initial tight fit between endplate and graft. These authors also claim that a moment load of 0.5 Nm was chosen because after surgery the motion segment may not carry large loads due to neck bracing. For Li et al. [71], the applied loading is physiologically relevant because the axial compressive preload simulates the weight of the head of an adult male and the magnitudes of each of the moments and the axial compression force are within the ranges telemetrically measured by Rohlman et al. [127] during a variety of activities of daily living in a group of patients. According to Mesfar et al. [89] the mean head weight was measured to be 40 N [128] and the considered head weight (30–57 N) corresponds to the upper and lower limit of the head weight of a sample of the population given that the weight of a normal head can be greater than 40 N when the head supports a supplementary weight (e.g., helmet).

4 Towards an effective FEM of the CS

In the foregoing, we presented a survey on the FEMs of the human CS that have been reported in the literature. Although the models discussed in this paper do not represent the complete range of CS models because those used for dynamic analyses were left aside, the current article is representative in that it provides a comprehensive overview of the available models in terms of characteristics/parameters that contribute to reproducibility and/or reusability of the models and, at the same time, can be used by researchers to ease the decision-making process during modelling/simulation. From the survey, it is clear that the use of FEMs to study complex problems such as the biomechanics of the spine, inherently involves assumptions/simplifications as means to obtain computationally efficient models. However, despite not being consistent with the real-life conditions, simplifications actually yield adequate results as long as the researcher understands their implications/limitations relative to the target application. In this section, we summarize basic considerations that can ease the implementation of effective FEMs of the CS.

Overall, the key aspects of obtaining an adequate FEM of the CS include the: (i) selection of an appropriate solution technique, (ii) definition of proper geometry, (iii) determination of boundary conditions representative of the conditions of biological models, (iv) correct use of material models, and (v) use of appropriate procedures for verification and/or validation [18,129,130].

4.1 Solution technique

Selecting an appropriate solution technique is related to the characteristics of the problem itself. The models reported in the scientific literature describe both static (or quasi-static) and dynamic simulations, as well as both linear and nonlinear solutions. A static analysis assumes approximate conditions in which the applied load does not vary with time. This assumption implies ignoring the inertial and damping forces that may result from the loading conditions. A linear analysis assumes that the relationship between the load applied to the spinal structure and its response (e.g., displacement, stress) is linear. This assumption indicates that: (i) the material follows Hooke's Law within the elastic region, (ii) the deformation is small enough to ignore changes in structural stiffness due to deformation, and (iii) the boundary conditions—magnitude, orientation, distribution of loads—do not change while the load is being applied and the corresponding structural deformation is occurring. Nonlinearity, be it geometric-, material- and/or contact-related, is natural in problems relative to the biomechanics of the human body and, thus, is the most accurate approach to follow. Nonetheless, linear solutions are widely used because they require fewer computational resources while still leading to results within an acceptable amount of error for cases in which the associated assumptions are close to the real scenario. These means that, for example, in a study to evaluate small deformations of a single vertebral body, a linear static solution may lead to adequate results as long as the vertebra is loaded in its linear elastic, small deflection range such that the slight nonlinearity does not affect the results and the difference between a linear and nonlinear solution is negligible. However, for simulations of multiple segments involving super-physiological velocities (e.g. traffic accidents) only a nonlinear dynamic approach will provide desirable results [129,130] because, most possibly, (i) displacements/rotations will become sufficiently large such that equilibrium equations for the deformed configuration are required, (ii) large rotations will produce a change in direction/magnitude of pressure loads resulting from a change in the area to which they are applied, (iii) elastic materials will become plastic and will no longer have a linear stress-strain relation at any stress level, (iv) some structures (e.g. ligaments) will lose stiffness because of buckling or material failure, and/or (v) adjacent components will make/break contact with the contact area as the loads change (e.g., posterior elements of the vertebra).

4.2 Proper geometry

Defining proper geometry involves making multiple decisions, including: two-dimensional versus three-dimensional model, subject-specific versus generalized model (or parametric), anatomically accurate versus simplified geometry, model representative of healthy subject versus pathological subject or subject that has undergone a certain treatment, specific versus non-specific age/gender of the subject, anatomical structures/level of detail to be included in the model, etc. There is not a single set of correct choices because there is no unique/generic best geometric model. Instead, the correct decisions depend upon the objective of the study. For instance, as mentioned by Zafarparandeh et al. [58], it has been proven that women have a slender neck compared to men and, thus, symmetric assumption about the mid-sagittal plane can affect the motion response of their neck more extensively. Consequently, for a study evaluating the effect of symmetric assumption in the motion response of the human CS, the use of a female source for geometry may yield results for the most critical scenario.

Moreover, geometry may be obtained from both living subjects or cadaveric specimens. The process for digitalizing geometry and the modelling approach vary to a certain extent based on the state of the source. The specifics of these approaches are, however, beyond the scope of this article. Also, given the inherent variability between humans, obtaining adequate geometry is a challenge particularly for creating models meant to obtain generalized conclusions [44]. In terms of subject-specific models, which only seek to draw conclusions for the subject tested, the difficulty associated with the geometry is related to limitations such as the time-consuming task of accurately modelling all the structures or the need to determine those structures that are essential [129]. The latter is not a simple task because, like most structures in the human body, the anatomical structures of the CS do not function in isolation. Instead, they achieve their mechanical stability partly from the neighbouring components. Further understanding of these relations between components is needed in order to facilitate the selection process.

Relative to the anatomical structures included, the reported FEMs of the CS may be divided into two major categories: (i) full models or those representing the full CS (i.e., the seven cervical vertebrae—C1 to C7—and/or the head—C0—and/or the first/second thoracic vertebrae—T1 or T2, respectively—), and (ii) partial models or those representing a segment of the CS (e.g., C2–C4) [130] (see Fig. 2). Usually, the partial models include the ligaments relevant to the modelled segment. Partial models also tend to involve a lot of detail (both in terms of geometry and material properties) to ensure an accurate response at the local tissue level. Full models, on the contrary, are typically used for analyses in which the resulting global motion of the head is of greater interest than the responses at the local tissue level and, thus, some level of detail can be sacrificed. Muscles, on the other hand, are generally included only in models representing the full CS because they play a major role in the stabilization of the full CS and head complex. At the segment level, ligaments are more critical for stabilization than muscles.

Aside from the joints listed in Fig. 2, other contacts defined in some models include the one between the transverse ligament and the base of the posterior aspect of the odontoid process.

4.3 Boundary conditions

Representing boundary conditions in biological models is problematic because these conditions continually change due to natural processes such as respiration, blood circulation and hydration level [129]. Additionally, the dependency between anatomical structures and the fact that the whole body is rarely, if ever, modelled adds to the complexity of defining accurate restrictions to the model. CS models are usually fully constrained at the inferior nodes of the lower-most vertebral body while the loading is applied on the upper-most body to simulate the effect of the head. This may result in unrealistic stress distributions on the lower-most vertebra.

4.4 Material models

Decisions regarding material models have an impact on the accuracy of the model. These decisions are associated with the type of element used for each anatomical structure, the material model assigned to such elements (i.e., linear—e.g., isotropic, orthotropic, anisotropic—vs. nonlinear material—e.g., viscoelastic, hyperelastic Mooney-Rivlin—), the specific values used to define the properties of such material models, and the types of contacts used to define the interaction between the different anatomical structures of the model. The models reported in literature use multiple types of elements, including—for three-dimensional models—one-dimensional (e.g., truss, beam), two-dimensional (e.g., shell, membrane), three-dimensional (e.g., solid 4-noded/10-noded tetrahedron, solid 8-noded/20-noded hexahedron), and miscellaneous (e.g., spring, gap, mass) elements. The decision on which type of element to use depends, among others, on the geometry of the anatomical structure. Generally, (i) one-dimensional elements can be used for structures in which one dimension is very large in comparison with the other two (e.g. ligaments), (ii) two-dimensional elements can be used for structures in which two of the dimensions are very large in comparison to the third (e.g. endplates), and (iii) three-dimensional elements can be used for structures in which all three dimensions are comparable (e.g. vertebrae). Given that the software requires the definition of three dimensions in order to do the respective calculations, when one-dimensional elements are used, the remaining two dimensions—area of the cross section—must be defined as additional input. Similarly, for two-dimensional elements, the third dimension—thickness—must be defined as additional input. Further detail on the implications of using a particular approach for a given anatomical structure is beyond the scope of this article.

Relative to the material models, it is clear that biological tissues are typically heterogeneous materials that behave as anisotropic, nonlinear and viscoelastic structures [5,130]. However, in some cases, the consideration of materials with linear properties may be adequate to solve certain type of problems. For example, a linear model can effectively determine the crack initiation of a vertebral fracture, whereas a model considering nonlinear material allows to fully simulate the complete fracture process [129]. Finally, the interaction between structural parts can have a great influence on the results. When defining contacts/interactions, it is important to determine whether friction or sliding are to be considered or not. Also, contact surfaces need to be well defined such that the software is capable of solving the problem. Particularly for studies simulating surgical procedures with implantable devices, understanding the implications of defining—for example—a bonded contact between the endplates and an implanted cage is of great importance. Such assumption

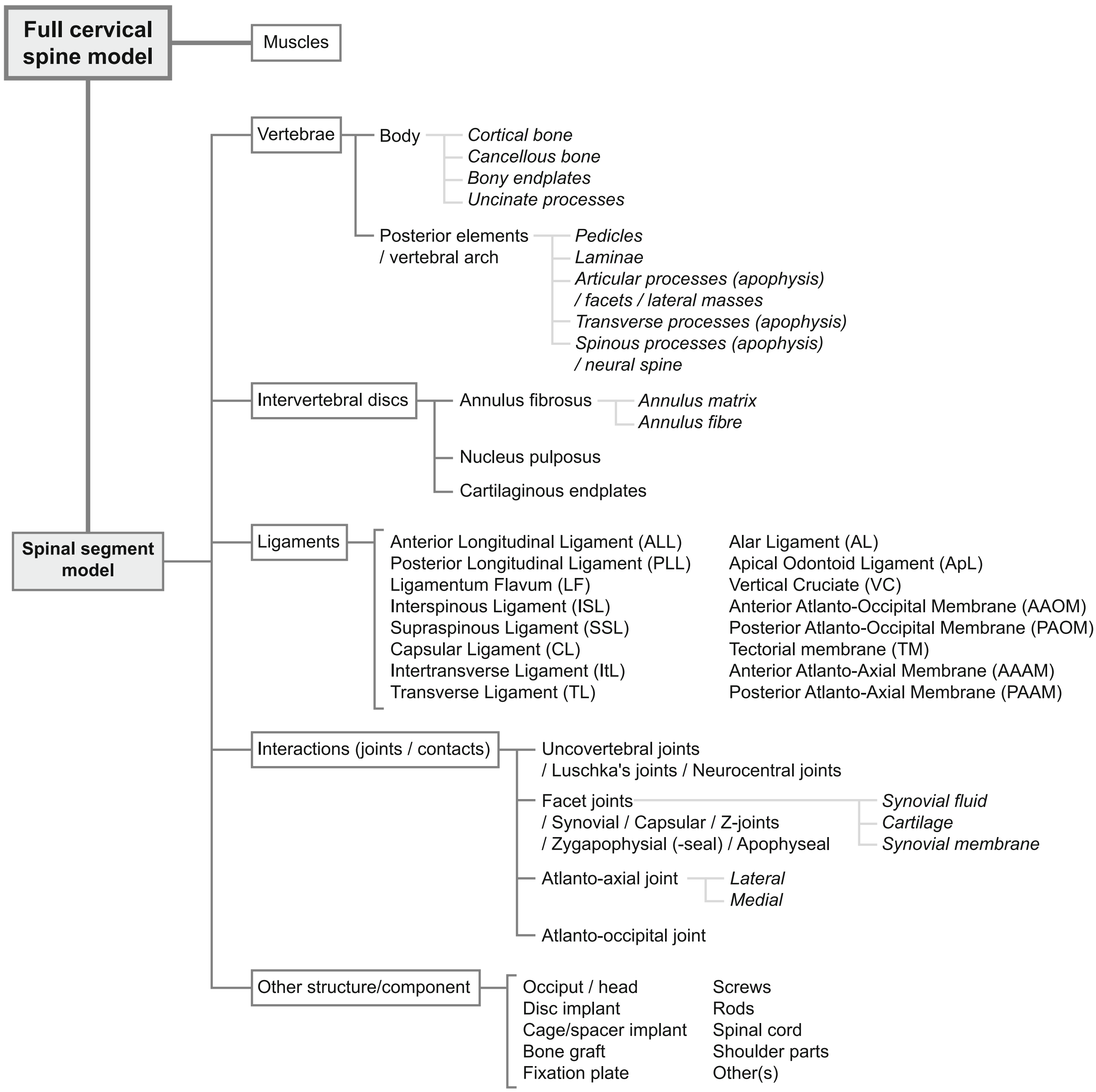

Fig. 2 Anatomical structures of FEMs of the cervical spine

implies that the results are not representative of the immediate post-surgical condition but rather represent a case in which complete fusion between the implanted device and the patient's bone has already occurred (i.e., months post-surgery).

4.5 Verification and/or validation

Two important aspects of the development of FEMs are verification and validation. Although these terms are often used interchangeably, they actually have different meanings. Verification is about solving the equations right, whilst validation is about solving the right equations. In this field of interest, verification is usually limited to the selection of a proper mesh density and the convergence of the solution. The fact that simulations are mostly executed using commercial software that has already undergone code verification eliminates the need to verify the degree to which the output of the computational model represents the solutions to the fundamental mathematical equations. Validation, differently, intends to prove that the output of the model accurately corresponds to the real-world scenario that is being simulated. Hence, the validation procedure is directly related to the scope of the intended application [i.e., variable(s) of interest] and is mostly performed via comparison of the results against those of experimental tests. In FEMs of the CS, validation is frequently done by comparison against *in vitro* studies and, in some cases, against previously validated FEMs. On one hand, validating against *in vitro* results does not fundamentally prove that the model accurately represents the *in vivo* conditions. Yet, it is accepted as an adequate first step towards obtaining an accurate model because it is assumed that if a model cannot reproduce the *in vitro* conditions, it will certainly not reproduce the far more complex *in vivo* conditions. On the other hand, validation against other FEMs is risky because, inherent to the finite element method itself, there is already an additional source of error in the chain.

Another matter to note is that both indirect and direct validation are possible. In the former, the researcher has no control on the experimental protocol and, thus, the data is subject to the intrinsic differences associated with the large inter-human variability and may not actually represent the population of the model to be validated; whereas in the latter, researchers can purposely perform experiments that closely match the scenario to be simulated. During validation, it is also important to take into account that experimental data for some parameters (e.g., internal stresses) is not readily available and, thus, validation is not a straightforward task.

Moreover, a major issue with verification and/or validation is that these terms are often mistakenly used to describe models that have been calibrated to a set of experimental results. While validation compares both sets of results—model and experimental data—to assess whether there is congruency between them, calibration involves adjustment of parameters—geometry, material properties, boundary conditions—to improve the agreement between the model and the experimental data. The risk with calibration is that the calibrated model will eventually yield the desired results under the loading mode for which it was calibrated. However, under different loading conditions or even when the same calibrated properties are applied to a model corresponding to a different geometric source (i.e., a subject with different anatomical characteristics), the analysis may yield results that are far too deviated from reality.

5 Conclusions

The purpose of this article was to present a comprehensive survey of the FEMs of the CS that have been used to study its pathological/nonpathological biomechanics under static/quasi-static loading conditions. The systematic literature search resulted in 332 publications of which 122 (or 44 'presumed' models) were selected based on the inclusion criteria. These publications were analysed to extract parameters relative to model identification, model structure, simulation structure, verification and validation. Besides summarizing different modelling approaches with their associated parameters, this article outlined generalities and issues associated with the obtainment of such models. Overall, there is not a single best modelling approach. Rather, relative to the objective of a given study, different modelling approaches may be followed to yield adequate results. Hence, understanding the implications of following different modelling approaches and taking advantage of previously developed models is critical for the development of new models. We are confident that this survey is a comprehensive referential document that will help to appraise previous solutions and select the best of them to facilitate the advancement in the field. This advancement will be similarly beneficial for those interested in the classical finite element analyses and their applications, as well as for those who aim to develop (or improve) tools for virtual prototyping—based on the concept of interaction—for supporting decision-making in the domain of product design and manufacturing.

In summary, the main conclusion drawn from this survey is that, for the majority of the revised articles, it is not readily possible to replicate the reported studies. Despite replicability being among the major goals of scientific publications, authors in this field often fail to report parameters that are critical for such purpose. Adequately reporting model parameters is not only important to enhance reproducibility but also contributes to a better understanding of the model's value and its potential for reusability in a given scenario. The considerations for reporting finite element analysis studies in biomechanics, published by Erdemir et al. [7], should be used as a first step towards improving this issue. However, even if parameters are clearly and fully reported, FEMs of the human CS are inherently difficult to replicate because they are, for the most part, patient-specific with their geometry based on data from in-house specimens/subjects. More studies to determine the influence of geometric data on the overall biomechanics of both the pathological and nonpathological spine are needed in order to establish the extent to which the (source of) geometry is a critical issue in the reproducibility of studies.

The construction of three-dimensional FEMs of the CS is almost prohibitively time consuming, particularly for models based on living subjects, owing to the need to acquire geometry and generate meshes based on medical image data (with fuzzy borders between different anatomical structures) and referential anatomical textbooks. Improvements in equipment, software and protocols to acquire geometric data as well as to produce the respective meshes would greatly contribute to the full implementation of FEMs in clinical evaluation of patients and in product development scenarios. Studies could also aim at identifying how to overcome the problem of generalization of results obtained by means of patient-specific models. Similarly, approaches to model verification and validation should be further explored for both generalized and patient-specific models. The authors decidedly advise that researchers in the field critically report data relative to model verification and validation because the long-term successful implementation of FEMs in clinical and industrial applications greatly depends on the accurateness of such models.

Further (updated and CS-specific) studies on the constitutive representation of the different anatomical structures as well as the interactions between them can contribute to the development of more accurate models. Calibration of material properties to improve such accuracy should be avoided unless multiple geometrically different models, multiple loading cases, and multiple output variables are used for validation purposes. Children CS models are scarce and, thus, understanding the biomechanical responses of human paediatric spines by incorporating their unique developmental anatomical features is an interesting topic to explore. For studies involving pathological conditions, incorporating the often-present associated degenerative characteristics [e.g., ruptures, fissures, osteophytes, ligament(s) failure] could lead to more accurate results. Bone remodelling after placement of implants, as well as for the study of fractures or the such, is also a topic needing additional study. Further understanding the role of different anatomical structures on the biomechanical behaviour of both the pathological and nonpathological spine under different loading conditions is needed to establish the level of detail required for a given study.

Supplementary material (see Online Resource 1)

The supplementary online material consists of an Excel file containing 47 tables with data relative to model identification, model structure, simulation structure, and validation, for all the models that were revised (122 articles). The file also contains a reference list, table of contents and legend (independent from the article).

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