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Computational modelling of the scoliotic spine: A literature review

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Abstract
Scoliosis is a deformity of the spine that in severe cases requires surgical treatment. There is still disagreement among clinicians as to what the aim of such treatment is as well as the optimal surgical technique. Numerical models can aid clinical decision-making by estimating the outcome of a given surgical intervention. This paper provided some background information on the modelling of the healthy spine and a review of the literature on scoliotic spine models, their validation, and their application. An overview of the methods and techniques used to construct scoliotic finite element and multibody models was given as well as the boundary conditions used in the simulations. The current limitations of the models were discussed as well as how such limitations are addressed in non-scoliotic spine models. Finally, future directions for the numerical modelling of scoliosis were addressed.

KEYWORDS
finite element modelling, multibody modelling, musculoskeletal modelling, review, scoliosis, scoliotic spine

1 | INTRODUCTION

Scoliosis is a pathological bending and twisting of the spine[1] in three dimensions.[2,3] It is defined as a spinal curvature in the coronal plane (i.e., the Cobb angle, the angle between the endplate of the two vertebrae either side of the coronal plane curve,[4] see Stokes et al. for further details[4]) exceeding 10°,[1] with recognisable rotation of the vertebrae.[1,5] In adolescents, scoliosis is a common, yet poorly understood, spinal deformity. Reported occurrences normally range between 2% and 5%[1,5,6]; and roughly 80% of cases are idiopathic.[1] Notably, above a Cobb angle of 30°, risks to health seem to increase with the angle.[1,5] These risks include back pain, mental health issues related to appearance, cardiac, respiratory and other physiological problems which in severe cases may lead to death.[1,5,6] Therefore, based on studies of the North American population treatment is suggested for Cobb angles above 20°[5] (i.e. approximately 0.3% of cases).[5] Surgery is only recommended in severe cases of Cobb angles above 50°.[1,7] Surgery is associated with a complication risk of 5%–25% which includes neurological injury, blood loss, injuries due to positioning, infections, flatback deformity, and proximal junctional kyphosis[8] and results in a permanent reduction in mobility when fusion is used.[7] The surgical planning for scoliosis treatment is complex and multi-facetted, thus there is a high variability in the treatment that is considered optimal.[9–11] While there is a clear scope for patient-specific models to predict the outcome of corrective surgery,[12] we are far from a widespread use in the clinical practice. While there is extensive literature on the computer modelling of spine biomechanics, there is the clear need for a systematic review of such literature, specifically targeting the modelling of the scoliotic spine, and its use in the planning of corrective surgery, as it would provide an essential starting point for any computer-aided surgery development in this area.

The biomechanics of the spine can be idealised by modelling the dynamics assuming vertebral bodies as infinitely rigid, muscles as linear actuators, and all the other connective tissues as lumped-parameter idealised joints that link the rigid bodies in a multi-body model (MBM). Under these assumptions the biomechanical behaviour can be described in terms of Ordinary Differential Equations. However, MBMs are not suitable for investigating internal stresses and strains. To do this, we need to assume that all tissues are deformable continua, whose biomechanical behaviour can be described in terms of partial differential equations. Being these boundary-value problems the most convenient numerical integration scheme for such mathematical models is the Finite Element Modelling (FEM) scheme. FEM assumes appropriate constitutive equations for each tissue type to predict the deformability of all different tissues. These two modelling methods can also be combined to produce hybrid models.[13] These models vary in complexity and detail, with differing degrees of personalisation. The importance of model personalisation has been highlighted in many studies; height, weight, muscle and bone morphology, and subject-specific material properties have all been shown to play an important role in simulations.[14,15]

In the last two decades, the interest in computer models of the spine rapidly increased. Several papers and a few reviews were produced on the topic. Driescharf et al.[16] reviewed the literature focusing on the quantification of the loads acting on the lumbar spine. They reviewed both in vivo and modelling studies, focusing for the latter on MBM models, and categorising them on the basis of the approach used to estimate muscle forces. They concluded that the main limitation of current models is the modelling of the intervertebral joint (IVJ) without a translational degree of freedom, and the balancing of net external moments at a single joint.[16] Alizadeh et al.[17] reviewed the literature focusing on studies modelling the cervical spine, which are most commonly employed to study impact conditions. MBMs were the most commonly used, followed by hybrid models and then FEMs. They suggested that in this case, the main limitation was the lack of detail in the musculature modelling.[17] Both reviews identified greater personalisation of models, and the use of patient-specific electromyogram (EMG) and kinematic data, as promising areas of future work.[16,17] Wang et al.[18] reviewed publications studying the biomechanics of scoliosis, but they only considered studies using FEM. To the best of the authors' knowledge the only review on the computational modelling of scoliosis which also addresses some studies using MBM was the paper by Jalalian et al..[19] They reported quite a few FEM-based studies, but a very limited number of MBM-based ones. In the context of scoliotic spine models, MBMs rarely include the facet joints explicitly, but often included the muscles. Conversely as the review by Jalalian et al.[19] noted FEMs often included the facet joints but rarely the muscles. Other reviews on the computational modelling of scoliosis and related surgery are not in English.[20,21]

These papers[18,19] provide systematic reviews of the literature which models the scoliotic spine until 2014; however, none of these papers review the literature published after 2014. Some focus only on specific segments of the spine, and do not specifically target the modelling of scoliosis which typically spans over long spine regions. Some consider only specific modelling methods. The Jalalian et al. review was published in 2013, and since then a considerable number of modelling studies were published, and the methods for patient-specific modelling have significantly improved.

The aim of the present study is to systematically review the literature focusing on the numerical modelling of the scoliotic spine, its use for surgical planning, and understanding the biomechanical properties of the scoliotic spine. Papers will be reviewed in terms of: whether the model is scoliotic, the segment of the spine being modelled, the patient specificity, if the model targets paediatric subjects, and how the model is used in a surgical context. First FEM and then MBM will be discussed; the capabilities of state-of-the-art models and their limitations for both methods will be discussed and avenues for future work will be identified. Since scoliotic spine models (both FEMs and MBMs) are often developed from models of healthy human spines, it seems appropriate to begin with an introduction to healthy spine modelling.

2 | METHOD

A first list of potentially relevant publications was generated by running systematic searches on the PubMed and Scopus indexing services between the period of March 2020 to October 2020. The two searches were conducted using the keywords "scoliosis" AND ("finite element" OR "multibody") and ("normal" OR "healthy") AND "spine" AND ("finite element" OR “multibody”). In addition, an unstructured search was conducted on Google Scholar looking for papers missed in this first search. A second list was generated by scanning the bibliography of the paper in the first list (Figure 1).

Articles in this master list were first filtered by title, if they were animal studies or were not studying the spine (Figure 1). Additionally, they were excluded if they failed to address one of the following categories:

- Scoliosis
- Surgical procedures related to scoliosis
- Biomechanical nature of the spine
- Paediatric spine

Following an initial filtering by title, the studies were filtered by abstract (Figure 1). Studies were excluded if:

- FEM or MBM was not used
- the focus of the study was on bracing treatments
- solely the instrumentation was modelled, or prosthetic device design was the focus
- less than one functional spinal unit was modelled
- the study examined the aetiopathogenesis or progression of scoliosis
- the focus of the study was gait analysis

Searches focused on articles post-2013 as previous review papers have looked at both FEM and MBM of the scoliotic spine until that date, however articles pre-2013 were not excluded.

2.1 | Finite element models of the healthy spine

2.1.1 | 2D FEM models

Two-dimensional (2D) models have been used in preliminary studies and to study large numbers of spine shapes. Galbusera et al. investigated the influence of sagittal balance on lumbar loading. Galbusera et al.[22] simulated many (1000) different spine profiles and included muscle forces . This approach enabled them to investigate trends rather than specific cases, and to showed that the C7 plumb line, the sacral slope, and the lumbar lordosis are all critical factors affecting the lumbar loading. The C7 plumb line is the horizontal distance between a vertical line passing through

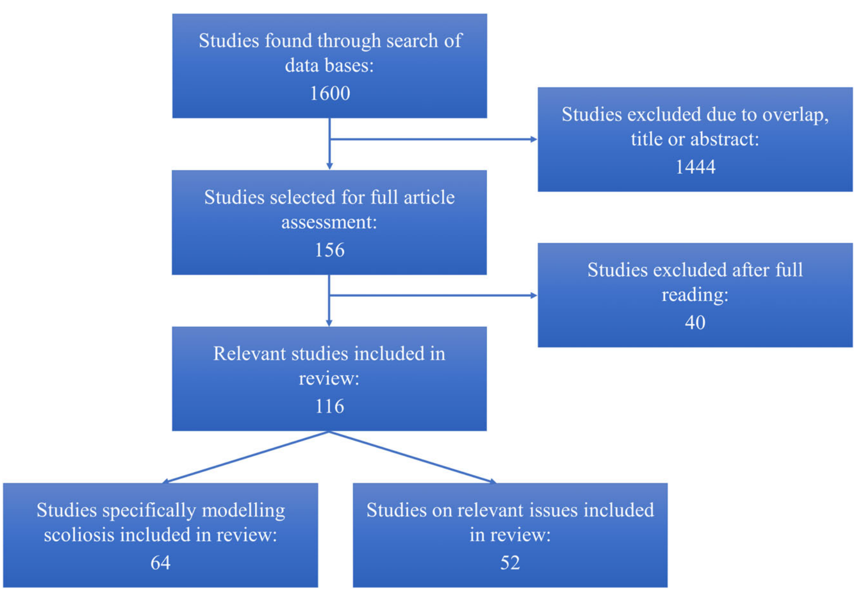

FIGURE 1 Flowchart of article selection

the centre of C7 and S1[23]; the sacral slope is the angle between the upper sacral plate and the horizontal.[24] The author would direct the Reader to Jackson and McManus[23] and Duval-Beaupère et al.[24] for the respective definitions.

Zanjani-Pour et al.[25] developed a 2D model with a poroelastic intervertebral disc (IVD). Their results showed that pore pressure was highly sensitive to the applied vertical translation and marginally sensitive to the Young's modulus of the annulus and the porosity of the nucleus.

2.1.2 | 3D FEM Models

Geometry

Most models are created from computed tomography (CT) scans, either directly or by the scaling of a previous model, however, the musculature is rarely included. A few studies have focused on the creation and validation of models.[26,27] The model developed from CT images by Mills and Sarigul-Klijn is one of the first young (20 years) female models.[27] This provided an initial model for future studies on the young female spine.

The natural curvature of the spine affects its response to loading. Naserkhaki et al.[28] showed the degree of lumbar lordosis substantially affected the load sharing in the facet joints and the ligaments when a follower load (a compressive force which acts along a line which follows the curvature of the spine) and moment were applied. Based on the investigation of 480 different curvatures, increasing lordosis correlated strongly with higher anterior–posterior shear loads in the IVDs, in particular the L5-S1 and L1-L2 IVDs.[29]

Others[30,31] investigated the influence of the IVD geometry on the facet joint forces (FJF), the intradiscal pressure (IDP), range of motion (RoM) and kinematics. These studies highlighted that the FJF, IDP and RoM are most sensitive to the disc height, width and sagittal dimension, the influence of each geometrical parameter depended on the movement.

Constitutive equations

The material properties are an especially important consideration when creating a subject-specific model due to the natural variation between subjects, which will be exacerbated by pathologies; however, these are most commonly taken from literature. Jebaseelan et al.[32] investigated the sensitivity of displacement and deformation of the spinal components to different material parameters under different loading conditions of a L1-S1 model of an eight-year-old. The material parameter the response was most sensitive to depended on the loading condition, for example the model was most sensitive to IVD properties in compression but to ligament properties in flexion. A sensitivity study of the RoM and the IDP to the IVD material model showed less than a 3% difference between linear and non-linear (hyperelastic Mooney–Rivlin for the nucleus pulposus and the annular ground substance and non-linear stress–strain curves from literature for the fibres) models.[26] They found no significant difference between the results for the linear and non-linear models when a compression of 300 N and flexion 7.5 Nm were simultaneously applied.[26] Typically, 300 N is less than 0.5 body weight hence, linear models will only have a limited applicability.

The use of generic material parameters from literature is a common simplification in many lumbar spine models.[33] Schmidt et al. established a method for calibrating the material coefficients of the annulus fibrosus to represent a specific specimen based on the RoM, concluding that only one specific set of parameters would correctly predict the RoM for multiple loading conditions.[33] However, Naserkhaki et al.[34] used eight different databases as the source for the material parameters of ligaments and were unable to predict the RoM, IDP, ligament force/deformation and FJF within one SD of the mean from in vivo and in vitro data. They suggested validation using solely RoM is insufficient because of the large variation of ligament material properties between subjects.

As Fan et al.[35] noted, poroelasticity is an important consideration under vibrational loads in order to observe the time-dependent vibrational characteristics; however, in most studies the porous material properties were not modelled.

Boundary conditions

The term "boundary conditions" refers to the forces and kinematics defined by the researcher for a part or all of the simulation. This should not be confounded with initial conditions which are the forces and kinematics of the model in the first instant of a simulation. Both force and displacement have been used as boundary conditions, force boundary conditions are often applied via follower loads which require careful tuning; while displacement boundary conditions require kinematic data, which can be challenging to gather.

Boundary conditions in terms of displacements has been derived from quantitative fluoroscopy,[36] plane radiographs,[37] stereophotogrammetry[38] and motion capture.[39] Quantitative fluoroscopy allows continuous imaging and is accurate for motions over 3.9° however is limited to 2D imaging[40]; alternatively biplanar fluoroscopy enables three-dimensional (3D) imaging, however, exposure to high radiation dosage is a concern[41]; stereophotogrammetry has been accurate to 0.5° but the subject must be stationary during the imaging process[42]; single plane radiographs have achieved similar accuracy and are less invasive but are limited to static images in 2D[43]; motion capture provides a continuous position, however, suffers from associated soft tissue artefacts.[44] Zanjani-Pour et al.[36] predicted IVD stresses consistent with literature using displacement boundary conditions. However further work is needed to reduce errors in mapping the motion . Dehghan-Hamani et al.[37] used displacement boundary conditions to validate a model against the loading in the ligaments and IVDs. Validation against FJF proved challenging due to lack of data and the difficulty in modelling the facets.[37] Shojaei et al.[45] suggested that displacement boundary conditions could be used in conjunction with optimisation techniques to estimate of muscle forces in FEMs.

Boundary conditions can also be applied in terms of loads, which can represent body weight, muscle forces, and external loads. For simpler representation of the local muscle forces, Rohlmann et al.[46] lumped them together into a single follower load—loads that represent body weight and occasionally muscle forces and intra-abdominal forces.[47] They were then able to include individual loads for a few global muscles and simulated realistic results. However representing the local muscles with a follower load caused the IDP to be slightly larger than it should have been.[46] Investigations have demonstrated alterations to the follower load path substantially influenced the deformation the spinal segment undergoes, thus the path should be optimised in order to achieve more realistic motions and load-carrying capabilities.[48–50] Studies have modified the location of the path to optimise it such that bending moments and shear loads were minimised[50] or the predicted RoM, IDP and disc bulging matched those reported in the literature,[49] both studies concluded the path should be posterior to the centre of the disc. Inclusion of a follower load influenced and limited the possible position of the centre of rotation (CoR),[34] through which the follower load should ideally pass.[50] Foresto et al.[51] found spinal stability was only maintained if the path passed through a specific region for a given posture. In addition to the posture the optimal path was influenced by the magnitude of the follower load.[50] Dreischarf et al.[52] simulated realistic motions by optimising the magnitude to minimise the intervertebral rotations, in another study the magnitude was optimised to minimise the IVJ load.[51] In a number of studies the follower load has been assumed to be task independent.[16,28,53] According to Azari et al.[54] this is not the case. They used a hybrid L4-L5 model to determine the follower load for specific tasks. Hybrid models have used MBM to calculate the muscle forces which are then applied as force vectors at the relevant attachment points in the FEM.[55,56]

Occasionally, the muscles and the intra-abdominal pressure are explicitly modelled, with various degrees of complexity.[29,57,58] One approach is to combine a FEM and an optimization method to calculate muscle forces.[29,39,45,51] In other cases muscles are modelled simply using tension elements[57] while a more sophisticated model has also accounted for the muscle geometry and architecture.[58] Inclusion of the muscles and the intra-abdominal pressure decreased the predicted stress in the IVDs.[57,58]

Non-linearities

It is generally accepted that the biomechanical properties of soft tissues show marked non-linearities, although there has been little investigation into the error caused by neglecting such non-linearities. Linear[36] and non-linear elements[39] have been used to model the IVJ, a lumped parameter which represents some or all of the passive components (IVD, ligaments, cartilages) between two adjacent vertebrae.

Other studies have modelled the individual components of the IVJ. Linear[32] and non-linear models[26–28,34,35,37,49,51] have been used for the IVD, often modelling the nucleus pulposus and the annulus fibrosus separately. Non-linear hyperelastic equations described the RoM much better than linear equations.[59]

The ligaments have been modelled as both linear[32,51] and non-linear.[26–28,35,37,49] Naserkhaki et al.[34] found a piecewise linear model from an in vivo dataset best predicted the mean in vitro rotations, and higher non-linearity caused greater movement of the instantaneous CoR .

Personalisation

Models can be personalised by the geometry, the material properties and the loading conditions; the extent of personalisation of each can vary, both geometric and material personalisation has been shown to be crucial for accurate simulations. As Naserkhaki et al.[28] noted, to fully personalise a model in all of these aspects is nearly impossible. Personalisation of the curvature of the spine was an important factor in the load sharing between the various spinal components and in the spine stiffness as it determined the disc wedging. However, personalisation of the curvature was insufficient to accurately predict the load sharing.[28] Scaling of a model to personalise existing models was not suitable for estimating FJF due to high sensitivity to the facet surface shape and orientation.[37]

Despite personalisation of passive joint properties based on subject sex, age, body weight and height, when simulating motions involving large angles substantial differences in shear and compression loads were still present.[47] Of the parameters personalised, body weight had the greatest affect.[60] Rather, the predictions were most sensitive to trunk flexion angle, load and moment magnitude, and loading asymmetry.[61] Sex and body height had little impact.[61] Thus, to accurately predict spinal loads these subject-specific parameters need to be accurately represented.

However, using the predictions from personalised models to infer the mechanical behaviour of the spine in the wider population is limited as there is much inter-subject variability in such models.[62]

A brief overview of the FEM modelling of the healthy spine has been presented. The present paper aims to address the modelling of the scoliotic spine thus a more in-depth review of healthy FEM spine modelling is outside the present scope. Recently, both Ghezelbash et al.[63] and Dreischarf et al.[16] have conducted literature reviews that focus on the in vitro, in vivo (respectively) and in silico studies (both); the Author recommends referring to these reviews for greater depth.

2.2 | Finite element models of the scoliotic spine

Geometry

The majority of scoliotic spine FEMs are developed from planar and tomographic radiographic images (Table 1) and include the entire spine in the model (Table 2); however, the rib cage was rarely included. The curvature and rib cage have played important roles in the loading of the spine. Development from radiographic images enables a high level of geometrical detail to be captured, including the facets which require manual segmentation as automatic segmentation tools cannot yet detect the gap between articulating cartilages.[99] In the creation of such models, geometric uncertainty due to user identification of landmarks and bony structures needs to be considered. Little and Adam measured intra-observer geometric uncertainty. The most substantial geometrical uncertainty was in the angle of the facet joints in the coronal plane, and consequently the endplate angle, which was comparable to the accepted clinical variation. The effect of the operator-dependent geometrical uncertainty was minimal on all the measured output parameters except the estimated IDP.[83] The geometrical variation of the intravertebral disc space was important in determining the degree of deformity and the extent of possible correction.[84] As the authors stress, these results are intra-observer rather than inter-observer variations and sensitivity of the parameters was only tested for a fulcrum bending test.[83] Inter-observer uncertainty could be greater than the intra-observer uncertainty and the effect could be greater under different loading conditions.

Manual segmentation of the spine is a time consuming process.[76] Hadagali et al.[76] proposed a semi-automatic technique to morph an already segmented spine onto a scoliotic spine based on landmark selection using a dual-kriging method (see References 100,101 for further details). Jobidon-Lavergne et al. have also investigated the used of an automatic segmentation method. Additionally, they investigated the possibility of using the pre-operative shape to enable intra-operative surgery planning, by registration of the pre-operative geometry onto intra-operative images.[80]

Geometries are often constructed from pre-operative standing images, which may result in a different geometry to the intra-operative one, as scoliotic curvature reduces when in a prone position.[74] Accounting for the difference could improve surgical simulation predictions.[74] Simulations have modelled the reduction in curvature due to prone positioning, and patient weight and surgical bed configuration.[72]

Scoliotic spines are categorised by different types of curves. The importance of accounting for the curvature type was demonstrated by Jia et al. They modelled three different curve types, under axial vibrational loading (exposure to vibrations has been associated with spinal problems), there was a different response for each type.[78] Validation proved challenging, and the model excluded any poroelastic effects which as discussed earlier can be important under vibrational loading. Other studies have further evidenced the importance of the curve type. Thoracic curve types have been shown to lead to asymmetric pressure distributions in the lumbar IVD.[91] Curve type and severity influenced the centre of pressure in the S1 endplate; lumbar curves affected the sacral loading more so than other curve types.[87] Scoliotic severity also affected the FJF; under compression and a bending moment higher FJFs were predicted on the convex and concave side when the Cobb angle was above and below 20°.[92]

Very few studies have included the rib cage (Table 2). However Jia et al. showed that the rib cage played an important role in the stabilisation of the spine, especially the scoliotic one, and reduced the maximum Von Mises stresses and

TABLE 1 Degree of personalisation for scoliotic finite element modelling

Study	Models source	Personalisation	
		Geometry	Materials
Agarwal et al.[64]	Adapted a previous model	No	Literature
Agarwal et al.[65]	Previous model adapted to CT scan	No	Literature
Aubin et al.[66]	Coronal and lateral radiographs	Yes	Literature
Berger et al.[67]	Coronal and lateral radiographs	Yes	Literature, personalised with suspension test
Cahill et al.[68]	3D model atlas	No	Literature, adjusted to reproduce in vitro results
Chen et al.[69]	CT scan	Yes	Literature and bone based on CT scan
Clin et al.[70]	Coronal and lateral radiographs	Yes	Literature
Cobetto et al.[71]	Coronal and lateral radiographs	Yes	Literature, personalised with suspension test
Driscoll et al.[72]	Previous model adapted to coronal and lateral radiographs	Yes	Literature, personalised with side bending
Driscoll et al.[73]	Coronal and lateral radiographs	Yes	Literature
Duke et al.[74]	Standing and prone radiographs	Yes	Literature, personalised with side bending
Galbusera et al.[75]	Biplanar radiographs	Yes	Literature
Hadagali et al.[76]	CT scan	Yes	Literature
Haddas et al.[77]	CT scan	Yes	Literature
Jia et al.[78]	CT scan	Yes	Literature
Jia et al.[79]	CT scan	Yes	Literature
Jobidon-Lavergne et al.[80]	Coronal and lateral radiographs	Yes	Literature, personalised with suspension test
Lafage et al.[81]	Stereo-radiograph	Yes	Literature
Li et al.[82]	Previous model developed from a CT scan	Yes	Literature
Little and Adam[83]	CT scan	Yes	Literature
Little et al.[84]	CT scan	Yes	Literature, from multiple sources, derived from adults
Little and Adam[85]	CT scan	Yes	Literature, personalised with fulcrum bending
Little and Adam[15]	CT scan	Yes	Literature
Musapoor et al.[86]	CT scan	Yes	Literature
Pasha et al.[87]	Atlas adapted to coronal and lateral radiographs	Yes	Literature
Pasha et al.[88]	Atlas adapted to coronal and lateral radiographs	Yes	Literature
Rohlmann et al.[89]	X-ray	Yes	Literature
Rohlmann et al.[90]	CT scan	Yes	Literature
Song et al.[91]	CT scan	Yes	Literature
Wang et al.[92]	CT scan	Yes	Literature
Xu et al.[93]	CT scan	Yes	Literature
Xu et al.[94]	CT scan	Yes	Literature
Ye et al.[95]	CT scan	Yes	Literature
Zhang et al.[96]	Previous model developed from CT scan	Yes	Literature, personalised with side bending
Zheng et al.[97]	CT scan	Yes	Literature
Zhou et al.[98]	CT scan	Yes	Literature

Abbreviations: 3D, three dimenisonal; CT, computed tomography.

TABLE 2 Components included in scoliotic finite element modelling[1]

Study	Spinal segment studied	Components modelled				Follower loads
		Rib cage	Facet joints	Intervertebral disc	Ligament	
Agarwal et al.[64]	S1-T1	—	Non-linear Soft contact	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Hydrostatic	Tension only	14% BW at T1 + 2.6% per vertebra
Agarwal et al.[65]	S1-T1	—	—	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Hydrostatic	—	14% BW at T1 + 2.6% per vertebra
Aubin et al.[66]	Pelvis - T1	CVJ – Linear Bone – Linear	Non-linear contact	Lumped linear	Linear Tension only	BW and muscles
Berger et al.[67]	L4 - T3/C5	—	—	Lumped linear	—	—
Cahill et al.[68]	T12-C6	—	—	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Incompressible	Tension only	—
Chen et al.[69]	T4-T12	—	Frictionless Soft contact	Lumper linear	—	—
Clin et al.[70]	Pelvis -T1	—	—	Matrix – Linear Fibres – Linear Nucleus – Linear	Linear	Sectional BW
Cobetto et al.[71]	Pelvis - T1	CVJ - Tension only	—	Lumped linear	Linear Tension only	Segmental BW
Driscoll et al.[72]	S1-L1	CVJ - 3D elastic springs with contact	—	Lumped linear	—	BW
Driscoll et al.[73]	L5-T1	—	Linear	Annulus – Linear Nucleus – Linear	Linear tension only	—
Duke et al.[74]	S1-T1	3D elastic springs	Contact, piecewise linear	3D elastic elements	3D elastic elements	Segmental BW
Galbusera et al.[75]	L5-T1	—	Frictionless Gap 0.4 mm	Matrix – Elastic Fibres – Non-linear Nucleus – Incompressible	Non-linear	—
Hadagali et al.[76]	T12-T1	—	—	—	—	—
Haddas et al.[77]	S1-T11	—	Frictionless contact	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Hyperelastic	Non-linear	BW and muscle optimised path
Jia et al.[78]	S1-L1	—	Frictionless Linear Contact	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Incompressible	Non-linear	400 N optimised path

TABLE 2 (Continued)

Study	Spinal segment studied	Components modelled				
		Rib cage	Facet joints	Intervertebral disc	Ligament	Follower loads
Jia et al.[79]	Sacrum-T1	Linear joint	—	Annulus – Linear Nucleus - Linear	Linear Tension only	BW
Jobidon-Lavergne et al.[80]	Pelvis-T1	Tension only spring	—	Solid elements	Tension only spring	Yes
Lafage et al.[81]	S1-T1	—	Contact	Lumped torsional and bending properties	Linear tension only	—
Li et al.[82]	S1-T1	—	Non-linear Friction	Matrix – Linear Fibres – Linear Nucleus – Incompressible	Linear tension only	BW at T1
Little and Adam[83]	L5-T1	CVJ and ribs linear elastic	Frictionless Soft contact	Matrix – Hyperelastic Fibres – Linear Nucleus – Hydrostatic	Linear and non-linear	—
Little et al.[84]	L5-T1	Non-linear joint	—	Matrix – Hyperelastic Fibres – Linear Nucleus – Incompressible	Linear and non-linear	—
Little and Adam[85]	L5-T1	CVJ and ribs linear elastic	Linear	Matrix – Hyperelastic Fibres – Linear tension only Nucleus – Incompressible	Linear	—
Little and Adam[15]	L5-T1	CVJ and ribs linear elastic	Linear	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Incompressible	Non-linear	—
Musapoor et al.[86]	L5-T1	—	Non-linear, pressure overclosure	Matrix – Linear Fibres – Linear Nucleus – Incompressible	Non-linear Tension only	—
Pasha et al.[87]	S1 - T1	CVJ – Linear tension only springs Ribs - Linear	Non-linear Friction Contact	Lumped linear	Tension only	—
Pasha et al.[88]	S1-T1	Non-linear joint	Contact	Lumped linear	—	Segmental BW
Rohlmann et al.[89]	T12-T1	—	—	Annulus – Linear Nucleus – Linear	—	Local muscles at vertebrae
Rohlmann et al. *(2008)*[90]	L2-T3	—	Linear Compression only	Matrix – Hyperelastic Nucleus – Incompressible	Non-linear Tension only	—

(Continues)

TABLE 2 (Continued)

Study	Spinal segment studied	Components modelled				Follower loads
		Rib cage	Facet joints	Intervertebral disc	Ligament	
Song et al.[91]	S1-T1	—	Non-linear Contact	Matrix – Linear Fibres – Linear Nucleus – Incompressible	Linear Tension only	—
Wang et al.[92]	L5-L1	—	Frictionless Contact Gap 0.5 mm	Annulus – Linear Nucleus – Incompressible	Linear Tension only	400 N at L1
Xu et al.[93]	Sacrum-L5	—	Frictionless Soft contact	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Hyperelastic	Non-linear	BW and muscle optimised path
Xu et al.[94]	S1-T12	—	Non-linear	Matrix – Hyperelastic Fibres – Non-linear Nucleus – Hyperelastic	Non-linear	BW and muscle optimised path
Ye et al.[95]	T12-T1	Ribs linear elastic	—	Lumped linear	—	—
Zhang et al.[96]	Sacrum-T1	Yes	—	Yes	Yes	—
Zheng et al.[97]	S1-T12	—	Contact Friction Non-linear	Matrix – Linear Fibres – Linear Nucleus – Incompressible and viscoelastic	Non-linear	—
Zhou et al.[98]	L5-T12	—	—	Lumped linear	Linear	—

Abbreviations: BW, body weight; CVJ, costovertebral joint.
[1]Muscles were not explicitly modelled in any of the studies.

displacements.[79] The rib cage, ligaments, vertebrae and IVD were examined and the magnitude of the maximum Von Mises stress and displacements reduction ranged between 19.7%–53.4% (in axial compression an increase of 31.5%) and 2.1%–16.9%, respectively, depending on the loading direction.[79]

Constitutive equations

The material properties of both bone and soft tissues are often taken from literature measurements without personalisation (Table 1). In the case of adolescent idiopathic scoliosis (AIS) this presents a twofold problem, firstly there is very little literature data on the mechanical properties of adolescent tissues. Secondly, there has been very little consideration regarding potential differences between the tissue properties of scoliotic and non-scoliotic subjects. Cheuk et al. suggested there could be differences in the stiffness and failure load of bone. Using personalised micro-FE they found the scoliotic subjects had weaker bones; however, this is potentially accounted for by differences in levels of physical activity and calcium intake,[102] regardless of the cause the differences need to be accounted for in models. Although the study was performed on the radii rather than vertebrae, other studies have suggested changes in bone properties of the radii are reflected in changes in the vertebrae.[102] A review paper by Li et al. supported the view that lower bone quality was prevalent in scoliotic subjects,[103] building on this Song et al. investigated the effect of low bone quality. With a geometrically personalised scoliotic model, three different Young's moduli were assigned to the cortical and cancellous bone. They found that lower bone quality elevated the asymmetrical loading affect, already present due to the curvature of the scoliotic spine.[91]

Berger et al. determined personalised IVJ stiffnesses for five AIS subjects from in vivo spinal suspension tests using a FEM. Despite simplifying the joint as a lumped parameter, this still demonstrated a statistically significant difference in subject-specific stiffnesses.[67] Zheng et al. created a geometrically personalised degenerative scoliotic model, they used two sets of material properties one represented normal young subjects and the other represented elderly degenerative scoliotic subjects.[97] They simulated an ex vivo study conducted on young and elderly non-scoliotic spines (data for scoliotic cases was not available) and compared the predicted motion to the measured motion. The model representing elderly degenerative scoliosis predicted a difference in motion of up to 24% compared to ex vivo data of the elderly non-scoliotic spines, a similar difference was observed between the model representing normal young subjects and the ex vivo data of young non-scoliotic spines. Zheng et al. suggested this evidenced the need for pathology-specific material properties.[97]

Boundary conditions

While healthy spine FEM have used displacement and force boundary conditions scoliotic FEM used almost exclusively force boundary condition with the lowermost body completely constrained, often to simulate scoliotic correction or study vibrational effects. To simulate the scoliotic correction of patient-specific models created from CT scans of eight different patients Little et al. used intraoperative force profiles measured in vivo by strain gauge force transducers attached to the tools used to apply a compressive load between the vertebral body screws[104] during the scoliotic correction of 15 different patients. Applying one of three force profiles, the mean or the mean plus/minus one SD of the in vivo data, they were able to predict the correct Cobb angle for seven out of the eight subjects.[84]

Li et al. investigated the effect of vibrational loading, an important consideration as vibrational loading increased the risk of further deformation and has contributed to lower back pain of which scoliotic subjects are at greater risk.[78,79,93] Several differences were noted between their results and similar studies on non-scoliotic spines. The scoliotic spine was less affected by the follower load, but was more sensitive to vibration with lower resonant frequencies and larger amplitudes especially in rotation.[82] These differences can probably be attributed to geometric differences because the study used literature data for the material properties which is based on healthy spines.[82] Other studies have investigated the scoliotic spine under vibrational loading, expanding the knowledge to include the effect in adult degenerative scoliosis cases,[93] the effect of curve type,[78] and of the rib cage.[79] The findings of these studies have supported those of Li et al., that the scoliotic spine has more resonant frequencies and a larger response than a healthy spine. Thus scoliotic subjects are exposed to greater risk of increasing the deformity, lesions at the resonant frequencies,[78,93] and injuries associated with vibrations.[79] Further, sagittal alignment of the scoliotic spine changed the response to be more similar to a healthy spine, which would decrease the risk for scoliotic subjects when exposed to whole body vibrations.[94]

The joints applied between bodies are also an important consideration. Ye et al. investigated the simultaneous correction of scoliosis and pectus excavatum. The type of joint applied between the sternum and the clavicles influenced the behaviour of the model: with the joint fully constrained they showed a simple correction of scoliosis via spine stretching had no effect on the pectus excavatum, however with the joint free in the sagittal plane spine stretching affected the chest deformity.[95] As such, modelling solely the spine can miss important effects scoliotic corrections may have on surrounding bodies.

Haddas et al. used methods they had previously validated[26] to create scoliotic spine models.[77] They investigated the effect of fusion of vertebrae on load transfer. They found fusion increased the FJF and the IDP at the adjacent levels, and decreased them at the fused level.[77]

As with healthy spine models, scoliotic FEMs have included intra-abdominal pressure.[87,88] They found the predicted IDP to be comparable to the IDP of non-scoliotic subjects measured in vitro and in vivo. The predicted vertebral compressive forces were closer to the predictions from other scoliotic models when the intra-abdominal pressure was included, comparison to in vitro or in vivo data was not possible due to a lack of data.[87] Muscles have been represented by a follower load,[93] or force vectors (calculated by MBMs) in hybrid models.[105]

Non-linearities

Some scoliotic models considered the ligaments, the nucleus pulposus, and annulus fibrosus as linear, others as non-linear (Table 2). When the facet joints are modelled, they are often modelled as frictionless with a gap and a spring. In all cases the parameters assigned have been taken from literature. To the best of the author's knowledge there have been no studies investigating the suitability of non-linear compared to linear material properties in the case of the scoliotic spine, whereas such studies have been conducted in health spine FEM.

Personalisation

Personalisation of geometry was common, and though less common material personalisation was also performed (Table 1) personalisation of both has been crucial to accurately predict scoliotic corrections. Kamal et al. compared the spines of a healthy subject and a scoliotic subject. In the subject with a Cobb angle of 24.4° higher muscle forces on the convex side were required for the equilibrium. Increased muscle forces shifted the CoR, which increased the stress in specific parts of the growth plate.[105]

The geometric personalisation alone was not enough to accurately represent the scoliotic spine.[74] Personalised geometry and individually representing all the soft tissue structures and including the rib cage, was not sufficient to consistently predict a scoliotic correction within an acceptable margin.[15] This suggested patient-specific soft tissue properties are also required.[15] Additionally, Duke et al. included the effects of anaesthesia on the muscles and spine stiffness by modifying the stiffness of the IVJ, in doing so they improved the simulation predictions of the intra-operative position.[74] Personalising the spine flexibility and stiffness has substantially improved the predicted axial rotation due to a correction manoeuvre.[81]

Using in vivo side bending tests to personalise the overall stiffness of the spine, some studies have predicted the post-operative curvature to within 5° of clinical measurements[13,106]; however, an accuracy of 10° has been reported.[72] As Little and Adam noted, side bending tests rely on the patient muscle activation. They studied the characterisation of the soft tissue properties of scoliotic subjects using a fulcrum bending test (the patient lies on their side over a rigid cylinder) which has the benefit of using the subjects weight and does not rely on muscle activation.[85] They simulated a fulcrum bending test using literature-reported values for the soft tissue stiffnesses and an imaging-based geometry. For 6 out of 10 cases there was more than a 10% difference between the predicted and clinical flexibility. In these cases, the sensitivity of the predicted flexibility to the costovertebral joint stiffness was investigated. The sensitivity was subject dependent. In addition to supporting the need for patient-specific soft tissue stiffness they suggest finding the stiffness of a single tissue in this way is not possible. Rather several tissues contribute to the stiffness and it is better to consider an overall stiffness.[85] Another method to personalise the soft tissue properties is to use a suspension test.[71,80] The concept for all the personalisation methods is the same, to measure the spine curvature under a certain loading and then to simulate the action. The component material properties or lumped IVJ stiffnesses are then calibrated such that the predicted curvature is similar to the measured curvature.

2.3 | MBMs of the healthy spine

Geometry

MBM are often developed for a particular purpose and then used as an initial model which is modified and personalised in future studies. The curvature, rib cage and to a lesser extent the muscles have to be accounted for in order to obtain realistic results; the most suitable method of positioning the IVJ location is still being studied.

Bruno et al. developed and validated a fully articulated MBM of the entire thoracolumbar spine.[107] Other fully articulated whole spine models exist,[108] as well as ones which assume a rigid thorax.[109–111] Models of just the lumbar spine are most common.[112,113]

A common assumption in spine models is that the thoracic region can be treated as rigid or the rib cage structure can be excluded.[111,113–115] Ignasiak et al. compared the loading and motion for a model with and without a rib cage. The comparison showed the rigid thorax assumption was suitable in activities where there was little thoracic deformation[116] and was only suitable for predictions of the lower lumbar spine.[108] Modelling the articulation of the rib cage improved the accuracy of the results.[108]

Bruno et al. used an MBM of the entire thoracolumbar spine to investigate the effect of lumbar lordosis and thoracic kyphosis on the risk of vertebral fracture. The curvature affected the loading more than the muscle forces,[14] the effect of which is influenced by the muscle cross-sectional areas.[14]

The position of the CoR of the IVJ affected the joint and muscle forces.[114,118] A sensitivity study by Abouhossein et al. showed variations of the CoR location by 1–5 mm lead to differences of 66% in shear load and 10% in the axial force in the IVD.[119] Abouhossein et al. and Senteler et al. postulated that the CoR may migrate during motion,[114,118] potentially to minimise the joint reaction forces.[114] Abouhossein et al. suggested that the migration is necessary to predict the kinematics of the spine.[118] While the literature supports the concept of CoR migration there is a lack of direct evidence.[114,118]

Identification of component parameters

The IVJ has been modelled without stiffness, with linear stiffness and with non-linear stiffness, studies have established the inclusion of a stiffness is essential for accurate simulations, whether modelling the joint as a lumped parameter is suitable or not is still unclear.

In addition to the location of the joint that defines the IVJ, the stiffness plays a critical role in characterising the flexibility of the spine. Simply including the stiffness of the joint reduced the muscle activation and the IDP, improving agreement with in vivo results.[120] Incorrect implementation of the IVD stiffness in OpenSim led to specious moments so Christophy et al. developed a new element which used the change in the relative displacement rather than the relative displacement to calculate the force the IVD would apply.[121]

Often all the passive stiffness components are modelled as a single lumped parameter. The impact of the lumped parameter simplification depended on the direction of motion.[120] Simply modelling the ligaments seemed to improve predictions in flexion.[120] The ligament stiffness also affected the location of the estimated CoR, if it is allowed to migrate.[118]

Muscles played an important role in the stabilisation of the spine. The muscles have been shown to have a notable effect on both the magnitude and the distribution of the intervertebral forces.[119] Han et al. incorporated muscles, ligaments, and rotational disc stiffness into an existing MBM, without any material personalisation. They simulated an upright posture and compared predicted joint reaction forces and muscle activation to in vivo measurements from literature of vertebral body replacement implant forces, IDP and muscle activation. This showed a considerable improvement in muscle force and joint reaction predictions on the base model, which demonstrated the importance of the passive soft tissue structures in spine models.[122]

Boundary conditions

In most models the pelvis or the lowermost vertebra is fixed in all degrees of freedom, and loading is applied at the most superior vertebra in the model[120] and/or the arms if they are included.[107,108] Alternatively, some models specified motions at the bodies based on in vivo or in vitro data.[113,123,124]

Bruno et al. investigated the compressive loads on the vertebrae for many activities represented by a large range of boundary conditions. Substantial variation in the vertebral loading was seen for different activities.[117] In addition to the vertebral loads, the predicted muscle forces were shown to be sensitive to the loading conditions; the muscle forces increased with the introduction of an external moment[111] and the activation patterns changed with variation of the follower loads.[125] Differences noted between the in vivo load data from sensorised telemetric vertebral body replacements and the predicted results were attributed to the differences in the definition of "standing," while well defined in simulations, standing position in vivo is ambiguous without motion capture.[111]

When using EMG data, early lumbar models assumed the muscle force equilibrium could be achieved by considering it at each joint individually.[112] Gagnon et al. challenged this assumption using a model which achieved equilibrium across all joints simultaneously. Solving for equilibrium at the joints individually led to different muscle force estimations and different joint loads.[112] The reliability of EMG signal has presented a challenge as the signals from deep muscle have been difficult to extract, and muscle magnitude intraclass correlation coefficients of 0.41–0.69 have been reported for trunk muscles.[126] The use of EMG data as a boundary condition in MBMs requires careful thought due to the reliability of such data.

Non-linearities

Studies of the spine using MBM have simulated joints with no stiffness[107,111,123] or with linear stiffness.[116,122] The IVJ has often been modelled with a non-linear stiffness.[113,115,118,120,122,124] Byrne et al. investigated the effect of the linear stiffness assumption by comparing the predicted compressive and shear joint reaction forces for no stiffness, linear stiffness and non-linear stiffness for a flexion-extension motion under different external loads (4.5 and 13.6 kg), in different neutral positions (supine and upright), and different kinematics (dynamic stereo X-ray and motion at each joint defined as a fraction of the total motion [rhythm based]).[113] The trends were similar for both neutral positions. The linear model tended to predict larger joint reaction forces than the non-linear model, the difference between the prediction increased towards the end of the motion. Generally, a larger difference was seen in the predicted forces for the 4.5 kg than the 13.6 kg external load. The differences in predicted forces between the non-linear and the linear model was smaller for the rhythm based than the dynamic stereo X-ray-based kinematics.[113] Bryne et al. also found using a non-linear stiffness instead of linear stiffness had a minimal effect on the muscle activation.[113]

Wang et al. introduced a model with a highly sophisticated IVJ that accounted for the non-linearity, the stabilising effect of the rib cage, the stiffening effect due to compression from the follower load, and the multi-segment interaction in the thoracic spine. They used this to simulate in vivo experiments and accurately predicted the RoM.[124] This work demonstrated the numerous factors that need considering when modelling the IVJ.

Personalisation

The spine curvature and the musculature has been personalised in a number of studies and shown to improve predictions. Personalising models by scaling it to a patient's height neglected the patient-specific spinal curvature.[113] This altered predicted compressive joint force reactions by 15%.[113] This finding is supported by the earlier work of Bruno et al.[14]

Bruno et al. found their model to be more sensitive to personalisation of spine curvature than the muscle morphology although both had a substantial influence, the influence of the muscle morphology increased with greater muscle activation.[14] The effect of the personalisation was seen more in the thoracic spine than the lumbar spine.[14]

During posterior spinal surgery the muscles are often damaged, Jamshidnejad and Arjmand investigated the effect of muscle damage on the loading seen in other muscles. Muscle damage led to higher predicted activation in the undamaged muscles.[115]

The work by Schmid et al. was, to the best of the author's knowledge, the only to develop MBMs of the child spine. They developed them through the non-linear scaling of adult models and attempted to validate the models with a number of parameters against experimental data.[123]

This is just a brief overview of healthy spine MBMs, a more in-depth review is outside the scope of this paper. For a more in-depth review of MBM spine models outside of the context of scoliosis, the authors would direct the reader to the review of cervical spine MBMs by Alizadeh et al..[17] Some lumbar MBMs are addressed by Dreischarf et al..[16] To the best of the Authors knowledge there are no other review papers on spine MBMs.

2.4 | MBM of the scoliotic spine

Geometry

Spinal curvature and occasionally the muscles have been included, however the individual component of the IVJ (disc, ligaments and facet joints) are rarely included (Table 3). Bassani et al. created whole, personalised, scoliotic, spine models. Compared to a healthy spine model they predicted differences in the muscle force distribution. Unexpectedly, there were no substantial lateral loads, however this was attributed to modelling mild scoliosis. Larger forces were seen in the sagittal plane and concave side of the curve for larger degrees of kyphosis and axial rotation, respectively.[130] Schmid et al. created and validated subject-specific AIS models. With the introduction of the scoliotic curvature the IVJ compressive load increased. Although, no correlation was observed between curve severity and compressive force, this could be because the curves in the model were mild or moderate.[142] Bassani et al. demonstrated the need to accurately

TABLE 3 Components included scoliotic multi-body model

Study	Spinal segment studied	Components included: Facet joints	Intervertebral disc	Ligament	Notes
Abedrabbo Ode et al.[119]	Pelvis-T1	—	DoF – 6 Elasticity – Linear	—	Muscles – linear spring damper system
Aubin et al.[127]	L3/L2-T4	—	DoF – 3 rotational Elasticity – Linear	—	
Aubin et al.[128]	Pelvis-T1	—	DoF – unclear Elasticity – unclear	—	
Aubin et al.[129]	L5-T1	Non-linear	DoF – unclear Elasticity – Non-linear	Non-linear	Muscles – posterior extensor-flexor at T1 to T3 as spring elements, optimised stiffness. Follower load – Segmental BW
Bassani et al.[130]	S1-T1	—	DoF – 3 rotational Elasticity – Linear	—	Muscles – 89, tension only, no passive stiffness, or muscle wrapping. Follower load – BW
Cammarata et al.[131]	Pelvis – T1	—	DoF – 6 Elasticity – linear	—	Follower load – Segmental BW
Desroches et al.[132]	Sacrum-T1	—	DoF – 6 Elasticity – Linear	—	
Fradet et al.[13]	L5-T1	—	DoF – 6 Elasticity – linear	—	Regional spine stiffness factors included in IVD
Fradet et al.[133]	Pelvis – T1	—	DoF – 6 Elasticity – linear	—	A hybrid model. Follower load – muscle force
Jalalian et al.[134]	L4-T2	—	DoF – 1 rotational Elasticity – NA	—	
Jalalian et al.[135]	L4-T2	—	DoF – 3 rotational Elasticity – Linear	—	
Jalalian et al.[136]	L4-T2	—	DoF – 3 rotational Elasticity – Non-linear	—	
Jalalian et al.[137]	L4-T2	—	DoF – 1 rotational Elasticity – NA	—	
Kamal et al.[105]	S1-T1	—	DoF – 3 rotational Elasticity – Non-linear	—	A hybrid model. Thorax – rigid structure. Muscles – 92, scaled CSA Follower load – gravity at each vertebra
La Barbera et al.[138]	S1-T1	—	DoF – 6 Elasticity – linear	—	
Le Navéaux et al.[106]	Pelvis-T1	—	DoF – 6 Elasticity – Non-Linear	—	
Majdouline et al.[12]	Pelvis -T1	—	DoF – 6 Elasticity – unclear	—	
Majdouline et al.[11]	Pelvis -T1	—	DoF – unclear Elasticity – unclear	—	
Martino et al.[139]	Pelvis-T1	—	DoF – 6 Elasticity – unclear	—	
Petit et al.[140]	L5-T1	—	DoF – 3 rotational Elasticity – Linear	—	

(Continues)

TABLE 3 (Continued)

Study	Spinal segment studied	Components included: Facet joints	Intervertebral disc	Ligament	Notes
Robitaille et al.[141]	L5-T1	—	DoF – 6 Elasticity – Linear	—	
Schmid et al.[142]	S1-T1	—	DoF – 3 rotational Elasticity – Non-linear	—	CVJ modelled with point to point actuators Muscles – MF and ES muscles at each vertebra, scaled CSA
Wang et al.[143]	Pelvis-T1	Yes	DoF – unclear Elasticity – unclear	Yes	
Wang et al.[144]	L5-T1	DoF – 6 Linear	DoF – 6 Elasticity – Linear	Linear	
Wang et al.[145]	L5-T1	Yes	DoF – unclear Elasticity – unclear	Yes	
Wang et al.[146]	Pelvis-T1	DoF – 6 Linear	DoF – 3 rotational Elasticity – Non-linear	Linear tension only	Rib cage stiffening effect included in IVD stiffness
Wang et al.[147]	L5-T1	DoF – 6 Linear	DoF – 3 rotational Elasticity – Non-linear	Linear tension only	Rib cage stiffening effect included in IVD stiffness

Abbreviations: BW, Body weight; CSA, cross-sectional area; CVJ, Costovertebral joints; DoF, degrees of freedom; ES, erector spinae; MF, multifidi.

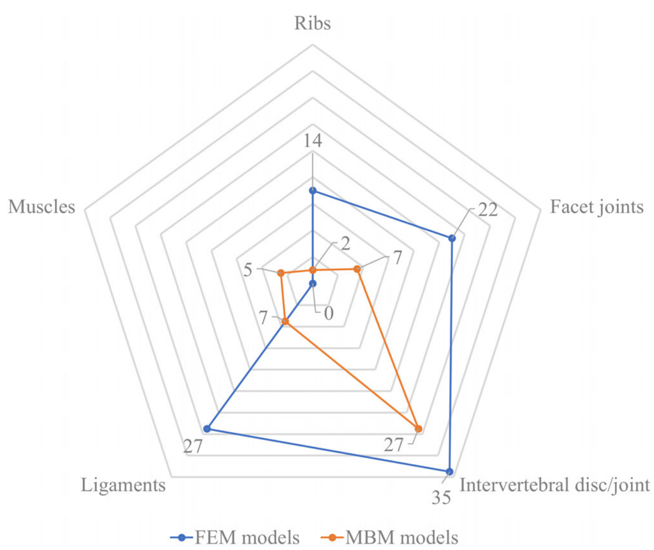

FIGURE 2 Trends of the components included in numerical models of the scoliotic spine with a specific element

capture the spinal curvature, especially as the curvature becomes more severe as it affects both the load magnitude within the discs and the muscle forces.

Schmid et al. also observed muscle geometry and activity asymmetry.[142] They reported the predicted muscle volume was generally in line with the literature; the majority of the models had larger erector spinae volumes on the convex side of the curve, and one third had larger volumes on the concave side. However, for the multifidi muscles Schmid et al. reported larger volumes on the concave side for about 90% of the models. Asymmetric muscle activity partially agreed with the literature, however differences could be attributed to differences between the curvature of the models and the subjects in the literature.[142] Previous studies have also found that the scoliotic spine experienced greater muscle force asymmetry than a healthy spine.[119]

Some MBM unlike FEM of the scoliotic spine incorporated the muscles; however often the IVJs were simplified by lumping the contribution of the facet joints, intervertebral ligaments, and the IVD together (Figure 2.).

Identification of component parameters

The IVJ is most commonly represented as a lumped parameter and personalised based on in vivo side bending tests, doing so has resulted in accurate predictions of scoliotic corrections, however unlike the healthy MBMs CoR migration is rarely considered. Petite et al. used a model with initial, literature based, linear stiffnesses at the joints to develop a method to personalise the IVJ stiffness of a scoliotic spine. They optimised the stiffness to minimise the difference between the predicted and measured Ferguson angle of an in vivo side bending test. The personalisation based on the in vivo side bending test resulted in more than a 40% increase in lumbar and thoracic stiffness compared to cadaver specimens and showed high inter-subject variability.[140] This suggests for scoliotic models to provide accurate predictions the IVJ stiffness needs personalisation.

In the correction of scoliosis, the disc stiffness can increase due to fusion by inserting bone grafts at the IVD.[132] Desroches et al. attempted to account for this. They created a patient-specific model using the personalisation method developed by Petit et al. and expanded it to include six degrees of freedom. Desroches et al. increased the IVD stiffness to represent the bone graft and performed a sensitivity study which showed that the rod shape and implant positioning had a greater impact on the predictions than the IVD stiffness. It is also worth noting that a higher IVD stiffness led to better predicted correction in the sagittal plane but worse in the coronal.[132] This suggests the material properties are highly anisotropic, therefore, to accurately predict the behaviour in all directions this may need to be accounted for. To the best of the author's knowledge there have been no studies examining the effect of the non-uniform spatial distribution of the material properties of the IVD in MBM.

Boundary conditions

There have been few studies on the suitability of the boundary conditions applied and few studies in scenarios not involving a correction manoeuvre. Desroches et al. analysed the impact of constraining the axial rotation of the upper most vertebra, showing it had no impact on the predicted post-operative spinal shape. However, correction of the segments outside of the instrumented region were sensitive to the boundary conditions.[132] Thus, the suitability of the boundary conditions applied is dependent on the spinal region being investigated.

Schmid et al. used their model to investigate the compressive loading at the spinal joints under different loading conditions. They found the compressive load to be sensitive to both load magnitude and direction relative to the scoliotic curvature.[142]

Incorporating gait analysis with an MBM, the intervertebral forces in the lumbar region during gait have been shown to be substantially higher than in simple standing.[119]

Non-linearities

When the IVD stiffness is included, it is often a linear and lumped parameter (Table 3), the impact the non-linearity may have has been studied in greater detail with healthy MBMs. This is only suitable for a limited RoM and could result in inaccurate predictions when simulating surgeries where large rotations and loads are present.[136] Jalalian et al. developed a method to characterise a non-linear stiffness for the IVD using a 2D model. The non-linear stiffness resulted in better shape predictions for all motions, the improvement being more pronounced for larger motions. It was also able to provide accurate predictions for shapes not used in the characterisation of the stiffness. While limited to only being in 2D they suggest the method could easily be expanded to 3D if suitable data is available.[136]

Personalisation

Often the geometry was personalised from radiographs, and in comparison, to the healthy MBM the stiffness of the IVJ has been personalised more frequently. Side bending test and fulcrum bending tests have been used to personalise, the stiffness of the IVJ (Table 4) determining the correct loading directions when simulating these tests was important when personalising the stiffness. Jalalian et al. investigated the direction in which the force in the frontal plane should be applied to simulate lateral bending, specifically how it should be personalised to an individual. Personalising the line of action of the force based on the location of the inflexion points measured from radiographs improved the bending prediction. As they note in the context of lateral bending tests, the line of action is crucial as it determines the loads and moments. Incorrect lines of action would potentially lead to inferior characterisation of the IVJ stiffness.[135]

TABLE 4 Degree of personalisation for scoliotic multi-body model

Study	Model's source	Personalisation	
		Geometry	Materials
Abedrabbo Ode et al.[119]	Coronal and lateral radiographs	Yes	Literature
Aubin et al.[127]	Radiographs	Yes	Unknown
Aubin et al.[128]	Pre-operative standing radiographs	Yes	Literature personalise with side bending tests
Aubin et al.[129]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Bassani et al.[130]	Coronal and lateral radiographs	Yes	Literature
Cammarata et al.[131]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Desroches et al.[132]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Fradet et al.[13]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Fradet et al.[133]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Jalalian et al.[134]	Coronal radiographs	Yes	NA
Jalalian et al.[135]	Coronal radiographs	Yes	Literature personalise with side bending tests
Jalalian et al.[136]	Coronal radiographs	Yes	Literature personalise with side bending tests
Jalalian et al.[137]	Coronal radiographs	Yes	NA – just applied kinematics
Kamal et al.[105]	CT scan and coronal and lateral radiographs	Yes	Literature data
La Barbera et al.[138]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Le Navéaux et al.[106]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Majdouline et al.[12]	Standing radiographs	Yes	Literature personalise with side bending tests
Majdouline et al.[11]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Martino et al.[139]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Petit et al.[140]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Robitaille et al.[141]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Schmid et al.[142]	Previous model adapted to coronal and lateral radiographs	Yes	Literature
Wang et al.[143]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Wang et al.[144]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Wang et al.[145]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Wang et al.[146]	Coronal and lateral radiographs	Yes	Literature personalise with side bending tests
Wang et al.[147]	Coronal and lateral radiographs	Yes	Literature personalise with fulcrum bending tests

Abbreviations: CT, computed tomography; NA, Not applicable.

The study by Jalalian et al. used X-rays of left and right bending and erect positions to personalise their model. They were able to establish a relationship between the bent and erect positions and were able to accurately predict positions which were not used in the characterisation.[137] This is important if models are to be predictive.

2.5 | Validation of scoliotic spine models

Verification and validation are two very important steps in building credibility of numerical models[148,149] Verification, not to be confused with validation, is the process of ensuring the equations of which the computer models comprise are being solved correctly.[149–151] Model verification is in fact the quantification of the error due the numerical approximations introduced by the mathematical model.[148,152] Verification can be achieved through benchmark testing (i.e., solving problems with known solutions), comparisons against results from other software which is known to be valid, and by ensuring physical laws (such as mass conservation) are respected.[148,149,151] In FEMs this may require convergence analysis and in MBMs ensuring that the uncertainty falls within pre-defined thresholds.[149,151,152] Hence verification should be implicit within published studies.

Validation of numerical models is the process of establishing that the model is representative of reality.[148–150] Ideally, a quantitative comparison should be performed between the model prediction and measurements from a dedicated validation experiment.[153,154] Validation is especially important when the models may be used to aid clinical decision-making as the extent of inaccuracy in the model will influence the risk the patient is subject to—less-accurate predictions may lead to poorer clinical decisions.[150] Validation has been very challenging in this area due to a lack of data, validation of the surgical correction is often done by comparing the predicted outcome to the clinical one, however this limits the predictive capability of the models outside of that particular scenario.

Many scoliotic models are validated by simulating the surgical intervention performed and then comparing predicted curvature to the post-operative curvature[13,15,70,72–74,81,84,106,127–129,131–133,139] or by simulating a bending test and comparing the predicted curvature to the measured curvature from radiographic images.[69,85,96,134–137,140] Predicted Cobb angles within 5° of the clinically measured values are often predicted and are considered acceptable as it corresponds to the clinical accuracy.[155]

The downside of this method is that it only offers retrospective validation of the models[132] and in the case of post-operative radiographs only for the procedure performed. In the simulation of scoliotic corrections, if data on the reaction forces at the bone/instrumentation interface was available it could be used for validation[128,132]; however, collection of this data was challenging[70] meaning direct validation for patient-specific models using these parameters was rarely done. Rather, predicted forces can be examined to see if they are in line with the literature data,[128,132] although this is a much weaker validation.

In vivo, in vitro, and/or in silico literature data has been used for validation. Model predictions are compared to the reported RoM,[68,78,79,97,119] muscle activation[142] or force,[130] IDP,[73,105,130] screw pull out forces,[73] loads on the vertebrae,[88] shear stresses within the IVD,[105] and IVD stiffness.[67] These parameters have been compared to the values reported in literature, sometimes to a single data set from literature and the standard deviation on that data if it is available.[68,78,97,130] Some studies have compared to multiple data sets reported in the literature and the range represented by these sets.[79,105,142] In certain cases only a qualitative comparison was performed.[88] Additionally, there is a lack of standardisation in the in vivo and in vitro testing of spines,[156] therefore the variability and lack of data limits the extent of possible validation. Hence, larger sample sizes and more standardised testing methods would enable more robust testing of the accuracy of model predictions. Ideally, the model should be validated under the same conditions in which it is being used: therefore, validation would be performed using data from scoliotic spines.[119] Unfortunately, a lack of in vivo and in vitro data from scoliotic subjects means scoliotic models have only been validated against normal spine data.[79,105]

Alternatively, studies have used models which have been previously validated[12,141,145,147] or developed a new scoliotic spine model using methods validated previously for creating healthy[77,78] or scoliotic spine models.[143,146] While it is not ideal to use a model in a context far from the one for which it was validated, in some cases this is unavoidable due to the lack of in vivo or in vitro data.

When validating against forces acting on the vertebrae and IDP (for subjects between 6 and 18 years old) direct validation was not possible for scoliotic and non-scoliotic alike due to a lack of literature data.[122,123] Further, validation using the FJF was challenging due to the conflicting and limited data available within the literature.[26] Large and conclusive data sets for IDP and FJF would significantly aid the validation of spine models,[26,27] and data sets from scoliotic spines would greatly support the verification of scoliotic spine models.

2.6 | Applications of FEM and MBM to scoliotic spine surgery

Scoliotic models have been widely applied to simulate surgical corrections, and to investigate surgical techniques and instrumentation. Some FEMs with a software platform to aid the development of surgical devices and methods for the correction of scoliosis have been developed.[73,75,83] Aubin et al. developed a MBM spine surgery simulator capable of simulating five different manoeuvres and designed to be used by clinicians.[128] This was based on a geometrically patient-specific model which had been used to compare predicted and clinical outcomes of a Cotrel–Dubousset surgical manoeuvre[127] and incorporated a method to characterise patient-specific flexibility.[140]

Using this simulator, Majdouline et al. conducted a series of studies[11,12] to find the optimal instrumentation configuration which aimed to meet the optimal outcomes recommended by several different surgeons. La Barbera et al. expanded on this work by considering a greater variety of instrumentation options such as different rod contours and screw patterns as well as a more clinically relevant measure of mobility.[138] These studies highlighted the lack of consensus on the optimal outcome of scoliosis surgery as multiple optimal solutions were found.[12,138] The complexity of using such models to find an optimal correction method was demonstrated by Martino et al. who varied five surgical parameters and performed a statistical analysis which found that each parameter had a statistically significant influence on either the implant-vertebra force or a geometrical parameter used to measure the degree of correction.[139]

The spine surgery simulator developed by Aubin et al. has been used to show that pedicle screws allow larger curvature correction than hooks.[141] More recent studies have demonstrated the importance of screw type, with the most suitable screw type being dependent on the individual case.[145] Screw type influences the degree of correction and the loading, with monoaxial screws allowing better correction but experiencing higher post-operative vertebra-implant loads than multiaxial screws.[70,139]

Several studies[69,70,86,98,139,146] have simulated the surgical correction of scoliosis to investigate the effect of increasing screw density, concluding that increasing the number of screws did not necessarily improve the correction. The effect of screw density depended on the side of the curve to which it is applied.[132] Le Navéaux et al. focused on the effect of screw density for the case of a main thoracic curve. Examining changing densities on each side of the curve, they found increased density on the concave side improved the correction, while increasing density on the convex side had little affect.[106]

Some studies have shown lower screw density could achieve lower[86,106,132,139] or similar[70] interoperative loads at the screws as higher density constructs; however, the results from the study by Wang et al. showed lower screw density increased the stress.[146] The literature is more divided over the effect of screw density on the bone-instrumentation stress post-operatively, with some studies showing a reduction in stress with increasing density[70,98] and others an increase in stress with increasing density.[69,106] Clin et al. offered an explanation for this disagreement: while the increase in screw density might be expected to reduce the stress, increased screw density also increased the number of constraints between the rod and the vertebrae, potentially increasing the stress.[70] In addition to the effect density has, the choice of the level at which to attach the screws[96,98,146] and the anterior–posterior position[71,106] played an important role in the stresses and corrections. Increases in correction force was shown to lead to greater Cobb angle correction,[95] until a certain point, when adjacent end plate come into contact.[84,86]

Increasing rod diameter and curvature has been seen to increase the intra-operative screw forces.[89,144] It can also improve the degree of correction in certain planes,[89,144] while simultaneously increasing kyphosis.[144] In another study, changes in rod shape had a significant effect on the degree of correction for one patient but not for another.[132] Choice of rod material has influenced scoliotic correction. Rods made from shape memory materials have achieved similar correction to conventional rods but required lower insertion forces and generated lower post-operative stresses at the bone-instrumentation interface.[147]

In AIS, growing rods have been used to allow growth to continue while reducing scoliotic curvature. This treatment requires multiple interventions to change the length of the rods, which frequently break. More frequent distractions reduced the risk of rod failure, while higher distraction forces increased the risk but resulted in greater curvature correction and greater growth.[64,65] An optimal frequency and distraction force exists which balances growth and the stresses in the rod,[64] which depended on the IVD stiffness as well.[65]Within the context of growing rods, Rohlmann et al. investigated the effect of screw type on the post-operative motion, assuming a perfect correction, that is, a spine in a healthy configuration.[90]

Tethers can be used to correct scoliosis. The effectiveness of anterior tethers and costovertebral tethers has been compared, showing the anterior tethers provided better correction of the coronal deformity and axial rotation.[66] Increasing the cable tension reduced the thoracic Cobb angle and increased the stresses in the growth plate.[71]

Fusion is used in the correction of severe scoliosis. Haddas et al. investigated the effect of fusion on the RoM, the IDP and the FJF at the fused and adjacent levels.[77] Similarly, Pasha et al. studied the effect of fusion in adolescents and looked at the stress distribution and the centre of pressure in the endplates.[88] The levels selected for fusion had a substantial effect on the curvature correction and the specifics of which were unique to each case and are highly dependent to the instrumentation strategy and curvature.[141]

Differences between the simulated corrections and the clinically observed corrections can be due to assumptions such as determining rod shapes from post-operative scans.[96,131,139]

Scoliotic models have also been used to investigate the effect of different surgical techniques on known post-operative complications such as proximal junctional kyphosis[68,129,131,133] and in cases when pectus excavatum is also present.[95]

3 | DISCUSSION

This review presented the current state-of-the-art modelling techniques for the scoliotic spine and its uses. Models have been used to investigate the effect of scoliotic curvature on the spinal loads. Vibrational loading as well as material parameters have been investigated. Some MBMs considered the anatomical and functional muscle asymmetries in the scoliotic spine. Both linear and non-linear material properties were used; however, there seems to be no clear consensus as to the most appropriate material properties for the soft tissues. There has been extensive use of numerical models to simulate corrections, investigate the effect of different correction manoeuvres and instrumentation configurations, which rely on using retrospectively validated models. In the simulation of corrections quantification of the boundary conditions affects could be an interesting research avenue.

Personalisation of model geometry is common; however, personalising the material properties is more challenging and is mostly used to characterise the stiffness of the IVJ as a lumped parameter. There has been very little research into the modelling of the bone and the soft tissues (except for the IVJ as a lumped parameter) for scoliotic cases. Investigations have suggested the effect of material property personalisation may be substantial thus further research is needed in this area, especially to model the individual components of the IVJ. The implementation of the joint model needs further consideration. Studies have suggested CoR migration may occur during motion; more research is needed to determine if this does occur and if so how to incorporate this migration into scoliotic models. In the case of scoliosis, the lack of in vivo and in vitro data makes material characterisation especially challenging. The lack of kinematic and load data is also a major obstacle to the validation of scoliotic models.

Muscle geometry is rarely included, although some studies try to account for the effect of the muscles through follower loads. However, this does not allow for investigations into the effects that surgery may have on the muscles or muscles characteristics in scoliosis. The effect of the rib cage and the costovertebral joint has been shown to affect the spine behaviour. However, the rib cage, like the muscles is not often included in models of the spine; however, its affect is sometimes accounted for by varying the stiffness parameters of the IVJs. More models investigating the role of the rib cage, especially in the case of scoliosis would be beneficial. Additionally, the biomechanical nature of the paediatric spine (both healthy and scoliotic) has been the subject of very few studies, despite the fact it is expected to behave quite differently to the adult spine.

To the best of the author's knowledge another aspect not yet considered in biomechanical models is the change in volume of the nerve canal and the risk of critical strains in the nerve and spine due to correction.

Although rare, scoliosis can occur in the cervical spine, to the best of the author's knowledge this has not been modelled.

Numerical models have been extensively used to investigate surgical corrections and the effect of different correction methods. They are a promising research avenue to better understand scoliosis and optimise its treatment.

CONFLICT OF INTEREST

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