

A Novel Optimized Finite Element Model of Lenke 1 Adolescent Idiopathic Scoliosis Based on Dynamic Flexibility in Vivo

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Abstract

Background: It is of great significance to optimize the finite element model by spinal flexibility of adolescent idiopathic scoliosis (AIS) patients. The elastic modulus of the intervertebral disc is of critical importance in determining the overall flexibility of the spine. The aim of the present study was to optimize the finite element model of Lenke 1 AIS based on the dynamic flexibility in vivo by matching the optimal elastic modulus of the intervertebral disc.

Methods: The Cobb angles under different longitudinal traction loads of one patient with Lenke 1 AIS were dynamically measured by using a spine morphometer with a posture sensor to plot the Cobb angle-longitudinal traction load characteristic curve. A 3D finite element model of the patient was established. The patient's Cobb angle-longitudinal traction load characteristic curve was used as the dynamic flexibility in vivo to determine the optimal intervertebral disc elastic modulus of the model.

Results: The dynamic flexibility curve in vivo of one Lenke 1 AIS patient was successfully obtained, and the patient's optimal elastic modulus of the intervertebral disc for the finite element model was 5 MPa according to the dynamic flexibility curve in vivo.

Conclusions: The use of dynamic flexibility in vivo to optimize the finite element model can provide a new perspective and approach for model optimization, which can reproduce the biomechanical characteristics in vivo of AIS patients.

Introduction

Adolescent idiopathic scoliosis (AIS) is the most common form of scoliosis with a prevalence of 0.47–5.2% and a female/male ratio that increases substantially with increasing age from 1.5:1 to 3:1[1, 2]. Changes in the physical appearance of AIS patients would affect their self-efficacy and confidence[3], and those with severe syndromes may also experience negative impact on their cardiopulmonary function and body balance, thus reducing the quality of their life. With the in-depth understanding of the pathophysiology and clinical treatment of AIS and the development of spinal orthopedic technology and internal fixation equipment, there has been a consensus that brace treatment is suitable for mild and moderate AIS patients, and surgical treatment is suitable for patients with progressive or severe deformities[4–6]. Body mass index (BMI), scoliosis type, surgical strategy and other factors would affect the surgical outcome[7]. An ideal surgical strategy should include inherent characteristics(i.e. curve flexibility), the complexity of specific correction maneuvers, selection of fusion levels, instrumentation parameters, surgeon experience and correction objectives[8–10]. Biomechanical mechanisms may also have impact on the occurrence and progression of AIS. The biomechanical characteristics (especially spinal flexibility) of each individual patient are an important premise for formulating an appropriate surgical strategy. The finite element model can simulate the spine anatomical structure and biomechanical characteristics of AIS patients in a reverse manner, and therefore has been widely used in the study of the etiology of scoliosis[11, 12], brace development and treatment[13], biomechanical

analysis for surgical strategy formulation and efficacy prediction[14]. The finite element model simplifications and assumptions are the key to make the model suitable to a research. The authenticity of the model is the basis of the research application, and even a slight change in the model parameters would cause obvious deviation in the research results. In order to maximize the simulation of the biomechanical characteristics of the spine in vivo, it is essential to optimize the finite element model.

At present, the finite element model is commonly optimized through the spinal flexibility of AIS patients in a specific state to present their inherent biomechanical characteristics[15–18]. We believe that the spinal flexibility in a specific state is a kind of static flexibility and cannot fully reflect the overall flexibility of the deformed spine, knowing that the dynamic flexibility in vivo of AIS patients is a comprehensive reflection of the biomechanical characteristics of the spines. To the best of our knowledge, there is no research to optimize the model basing on the dynamic flexibility in vivo so far. The aim of the present study is to use the dynamic flexibility in vivo to optimize the Lenke 1 AIS finite element model.

Methods And Materials

Acquisition of dynamic flexibility in vivo

We selected one Lenke 1 AIS patient(patient A) without tumors, infections, rheumatic immune diseases, cardiopuimmonary dysfunction, old spinal injury and other diseases that cannot withstand the subsequent trial. Before the initiation of the trial, the patient and the families were fully informed and signed the medical ethics informed consent forms. This research project was approved by the ethics committee of Shanghai Changhai Hospital (Shanghai, China). The patient took a sitting position, with the upper body upright and the lumbosacral part fixed. The traction occipital band was fixed to the patient's mandible, and the vertical traction force was applied to the patient through the occipital band. The traction force gradually increased from 0N to 160N (about 30% of the patient's weight). The Cobb angles were measured every 20N with a posture sensor-based spine morphometer (Fig. 1). The scoliosis longitudinal traction device was used to control the traction force through a highly sensitive mechanical sensor (Fig. 1).

The trial was conducted by two experienced orthopedic spine surgeons. The patient was measured three times by the two surgeons and the interclass correlation coefficient (ICC) for Cobb angles was 0.93($p < 0.05$). The mean value of the Cobb angles was used as the final result. The characteristic curve of Cobb angle-longitudinal traction load, also known as the dynamic flexibility curve in vivo, was drawn according to the measurement result(Fig. 5).

Establishment of the finite element model

Several simplifications and assumptions were necessary for this fine element model. a. The mandible, cervical spine, ligamentum flavum, supraspinous ligament, interspinous ligament, facet joint and muscles were not included in the model. b. The model did not discriminate the nucleus pulposus from the annulus fibrosus, made of ground substance and layers of fibers. c. The intervertebral disc was assigned a value

mainly based on the annulus fibrosus and modelled as isotropic with the same mechanical properties. d. The tractional load should counterbalance for the gravity loads and muscle forces, which are applied on the spine in vivo. However, it is difficult to determine the actual traction load in the finite element model test. The study neglected the gravity loads and muscle forces. Therefore, most of the other parameters cannot be validated and many results of the finite element analysis only indicate trends and do not necessarily represent the correct absolute values. Furthermore, patient A had surgical indications, a CT scan was needed to provide a reference for pedicle screw placement. CT DICOM format data of T1-L4 spinal segments of patient A were obtained, and the details are shown in Table 1. Referring to the material properties reported in the previous literature, the cortical bone, cancellous bone, intervertebral disc and anteroposterior longitudinal ligaments of the model were assigned an initial value[19–22]. The material properties for the model were extracted from the literature available (Table 2). Using Mimics 17, Geomagic Studio 2014, Hypermesh 14.0, the 3-D finite element model A was established for patient A (Fig. 2). The test of model A was conducted by Abaqus 2018 by setting the model boundary conditions with the bottom surface of L4 being constrained and the top surface of T1 bound by the vertical upward coupling constraint (Fig. 3) to simulate the situation that the patient's waist was fixed and head was pulled by a vertically upward traction force.

Table 1
Characteristics of Patient A

Age(Y)	Gender	Height(cm)	Weight(kg)	EPCA(°)	BPCA(°)	BCR(%)	AV	EV
14	Female	163	51	50	26	48.0	T10	T7, T12
<i>Note:</i> EPCA: Erecting Position Cobb Angle, BPCA□Bending Position Cobb Angle□BCR□Bending correction rate, AV:Apical Vertebra, EV:End Vertebra								

Table 2
Material properties used in the finite element model

Material	Young's modulus(MPa)	Poisson's ratio	Reference
Cortical bone	12000	0.3	Schmidt et al. [18] Nie et al. [19] Ruberté et al. [20]
Cancellous bone	100	0.2	Lu et al. [17] Schmidt et al. [18] Nie et al. [19]
Annulus fibrosus	4.2	0.45	Clin et al. [11] Schmidt et al. [18] Nie et al. [19] Schmidt et al. [24]
ALL	20	0.3	Lu et al. [17] Schmidt et al. [18] Nie et al. [19]
PLL	20	0.3	Lu et al. [17] Schmidt et al. [18] Nie et al. [19]
<i>Note:</i> ALL: anterior longitudinal ligament, PLL: posterior longitudinal ligament.			

Optimization of the finite element model

Calculation of Cobb angle in the finite element model

Taking the coronal axis of the model as a horizontal straight line, two points of the superior endplate of the upper end vertebrae were selected and their coordinates were recorded as (X_1, Z_1) , (X_2, Z_2) , which were connected to obtain a straight line a , with the slope being k_1 ; then the two final points of the inferior endplate of the lower end vertebrae were selected and their coordinates were recorded as (X_3, Z_3) , (X_4, Z_4) , which were connected to get a straight line b , with the slope being k_2 . The angle between the two straight lines was the Cobb angle and calculated by using the following equation(Fig. 4):

$$k_1 = \frac{Z_1 - Z_2}{X_1 - X_2}, k_2 = \frac{Z_3 - Z_4}{X_3 - X_4}, \text{Cobb angle} = \arctan \left| \frac{k_2 - k_1}{1 + k_2 \times k_1} \right|$$

Optimization of the intervertebral disc elastic modulus of the model

The intervertebral discs are of critical importance in determining spinal flexibility[16]. This study mainly optimized the finite element model by matching the elastic modulus (E) of the intervertebral disc. The intervertebral disc was not divided into the nucleus pulposus and annulus fibrosus, but was assigned a value mainly based on the annulus fibrosus. By referring to the literature, E of the annulus fibrosus was 4.2 MPa[14, 23]. Based on the results of the pre-experiment, the disc E of model A was set to 3MPa-7MPa in sequence to facilitate calculation. The traction loads were set to 0N, 20N, 40N, 60N, 80N, 100N, 120N, 140N and 160N, and then a dynamic flexibility test was performed on model A. Using Matlab2017a, the characteristic curve of Cobb angle-longitudinal traction load, the fit curves and equations of patient A and model A with different E were obtained. The goodness of fit was expressed by the coefficient of determination(R^2). The sum of squared differences (SSD) of Cobb angles under the same traction loads between patient A and model A with different E were also calculated. The minimum SSD value indicated that the two curves were closest, and the E of that curve was the optimal intervertebral disc E of model A.

Results

As shown in Fig. 5, the dynamic flexibility curve in vivo of patient A was obtained successfully. The 3D finite element model A of patient A was established (Figs. 2 and 3), and the characteristics of patient A are shown in Table 1. Five dynamic flexibility curves of model A with different disc elastic moduli and the dynamic flexibility curve of patient A were obtained (Fig. 5, Table 3). The fit curves (with all the R^2 greater than 0.9) and equations between Cobb angles and longitudinal traction loads of model A with different E and patient A were obtained (Fig. 6, Table 4). The SSD value of Cobb angles under the same traction loads between patient A and model A with disc E = 5MPa minimally demonstrated that 5MPa was the optimal disc elastic modulus of model A (Table 4). The fit curves between Cobb angles and longitudinal traction loads of patient A and model A with disc E = 5MPa were the closest (Fig. 6).

Table 3
Cobb angles of patient A and model A with different elastic moduli (E) under different loads

Traction load (N)	Patient A Cobb Angle (°)	Model A Cobb Angle (°)				
		E = 3MPa	E = 4MPa	E = 5MPa	E = 6MPa	E = 7MPa
0	50	51.06	51.06	51.06	51.06	51.06
20	44	37.80	38.53	43.76	44.29	44.82
40	40	35.90	36.67	39.93	40.07	41.61
60	35	32.14	32.90	34.32	37.21	38.51
80	33	30.70	31.32	33.82	35.16	35.29
100	32	29.45	30.96	31.01	32.75	33.49
120	30	27.39	29.35	29.75	31.30	31.84
140	28	25.86	27.89	28.55	29.07	31.00
160	27	24.27	25.75	27.51	27.96	28.21

Table 4
Fit equations between Cobb angle and load of model A with different elastic modulus(E) and patient A

Subject	E(Mpa)	Fit equation	R ²	SSD
model A	3	Cobb angle = 0.001Load ² -0.3033Load + 47.4902	0.99	14.91
	4	Cobb angle = 0.001Load ² -0.2873Load + 47.6005	0.97	3.12
	5	Cobb angle = 0.001Load ² -0.2928Load + 50.0767	0.95	0.52
	6	Cobb angle = 0.0007Load ² -0.2471Load + 49.8643	0.99	1.84
	7	Cobb angle = 0.0006Load ² -0.2320Load + 49.8643	0.99	2.93
patient A	-	Cobb angle = 0.0008Load ² -0.2661Load + 50.1616	0.99	-
<i>Note:</i> R ² : coefficient of determination; SSD: sum of squared difference				

Discussion

Studies have shown that assessment of the spinal flexibility of AIS patients is of great significance in determining fusion segments, selecting surgical approaches, and predicting postoperative orthopedic effects[24]. For AIS patients with relatively better flexibility, orthopedic surgery can be completed by reducing the screw density or shortening the fixed fusion segment. In comparison, patients with comparatively poorer flexibility need to increase the screw density or extend the fixed fusion segment to

achieve the designated orthopedic purposes[24–26]. In addition, the role of spinal flexibility assessment in conservative treatment is also important[27]. Scoliosis flexibility evaluation is the overall deformability of the spine under active or passive external force, and it is a quantification of the degree of spinal stiffness. At present, the commonly used methods for clinical evaluation of scoliosis flexibility include the supine bending method, fulcrum bending method, traction method, push prone method, and suspension-traction method[28, 29]. Berger et al.[30] used flexibility index for systematic review of the predictive effect of the above five methods on the surgical orthopedic outcome, and concluded that the supine bending method was the most commonly used method, and the fulcrum bending method was the most accurate method. However, both assessment methods are conducted in a static and specific state[31], which hinders the overall assessment of spinal flexibility. In addition, the stiffness degree of AIS patients is often different under the same traction loads, which emphasizes the necessity of overall assessment of the patients' flexibility.

Although the finite element model has been widely used in the field of spinal surgery, especially in the spinal biomechanical analysis of AIS patients[32, 33], for the first time it was used to simulate scoliosis correction surgery by comparing the effect of lateral and longitudinal forces on the correction of scoliosis[34]. Viviani et al.[35] optimized the surgical strategy by comparing the simulated surgical effect with the actual surgical effect. Ghista et al.[36] summarized the methods for biomechanical analysis of scoliosis orthopedic surgery. With the advancement in modeling technology, scholars have also performed finite element simulations of scoliosis orthopedic surgery from different perspectives[37, 38] and now the finite element model can also simulate internal fixation devices, surgical strategy, vertebral body growth rate and other elements[8–10, 39]. Further efforts have also been made to improve the simulation degree of the finite element model, including performance optimization and structure optimization, especially in terms of biomechanical characteristics. Lafage et al.[40] established a simulated finite element model of scoliosis by referring to the supine bending method to optimize the mechanical parameters of soft tissues. Little et al.[16] optimized the finite element model by using the fulcrum bending method, and after a detailed report of the specific simulation process and the adjustment of the relevant soft tissue stiffness index, they believed that the stiffness of the intervertebral disc fiber had a relatively strong impact on the flexibility of the spine. In a subsequent study, the authors also found that spinal flexibility in the fulcrum bending test was not governed by any single soft tissue structure acting in isolation. More detailed biomechanical characterisation of the fulcrum bending test is required to provide better data to determine the properties of patient-specific soft tissues[41]. Kamal et al. [39] achieved optimization by adding muscle simulation to the finite element model. In addition, other scholars were also committed to improving the simulation accuracy of the model[42]. The individualized finite element model based on hexahedral meshing can simulate the structural characteristics of different scoliosis spines with more precision and make the stress distribution more uniform[43]. However, the construction of a high-precision individualized model requires manual matching with the details of scoliosis features, which can be extremely time-consuming. As a result, Hadagali et al.[17] proposed a block template method to construct a scoliosis thoracic spine model to improve the efficiency of modeling.

We believe that spinal flexibility in AIS patients is an important manifestation of biomechanical characteristics, and optimization of the finite element model needs to consider spinal flexibility. However, to the best of our knowledge, most models currently used to optimize the flexibility statically and in a specific state, and only the reduction degree in the range of flexion, extension, lateral flexion and rotation of the spine is consistent with AIS patients[18, 44]. In this study, we utilized the dynamic flexibility of AIS patients to optimize the finite element model. Although the dynamic flexibility in vivo is only under a longitudinal traction load, it can still be viewed as an exploration of a novel method of model optimization.

The present study has some limitations. Firstly, only a one-case model was optimized, the type (Lenke 1) of scoliosis was simple, and the representativeness of the main curve Cobb angle was questionable. Secondly, the flexibility of the spine in patients with scoliosis is affected by many factors, such as soft-tissue properties, but the model of the present study only optimized the elastic modulus of the intervertebral disc which was modelled as isotropic with the same mechanical properties, while they may change depending on the spinal level. Meanwhile, the intervertebral disc was not distinguished between the nucleus pulposus and the annulus fibrosus. Instead, the annulus fibrosus was used as the main assignment subject. However, this does not affect the purpose that provide a new perspective and approach for model optimization of this study. Thirdly, the model was too simplified neglecting specific anatomical structure (i.e. nucleus pulposus, facet joints) and simplifying the boundary conditions. Although the purpose was to present a novel optimization method, more sophisticated model and realistic boundary conditions would undoubtedly increase the applicability. Finally, the dynamic flexibility was only under longitudinal traction load and may not represent the overall flexibility of one specific patient. And also, it is not clear to which degree the disc properties are related to the traction forces. More comprehensive dynamic flexibility evaluation methods would be worth studying in the future.

Conclusion

The use of dynamic flexibility in vivo to optimize the finite element model can provide a new perspective and approach for model optimization in that it can reproduce the biomechanical characteristics in vivo of AIS patients.

Abbreviations

AIS: adolescent idiopathic scoliosis; BMI: body mass index; ICC: intraclass correlation coefficients; SSD: sum of squared differences

Declarations

Acknowledgments

Not applicable.

Authors' contributions

CWY and ML were corresponding authors and contributed to the study conception, design and revised the draft critically. JYB, HY and DSZ performed the trial, acquisition of data, analysis and interpretation of data, SHN drafted the manuscript, DSZ and SYL established and tested the finite element model, JHW drew the tables and figures. SBN, JYB, HY and DSZ were co-first authors, all authors have read and approved the final manuscript.

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Availability of data and materials

The datasets used or analyzed during the study are available from the corresponding author on reasonable request.

Ethics approval and consent to participate

The study was approved by the Ethics Committee of Changhai Hospital, Naval Medical University, (the committee's reference number is not applicable), and all patients provided written informed consent to participate before enrollment.

All procedures performed in studies involving human participants, human material and data were in accordance with the 1964 Declaration of Helsinki and its later amendments or comparable ethical standards.

Consent for publication

A statement of consent to publish from the patient is not applicable.

Competing interests

The authors declared no competing non-financial/financial interests.

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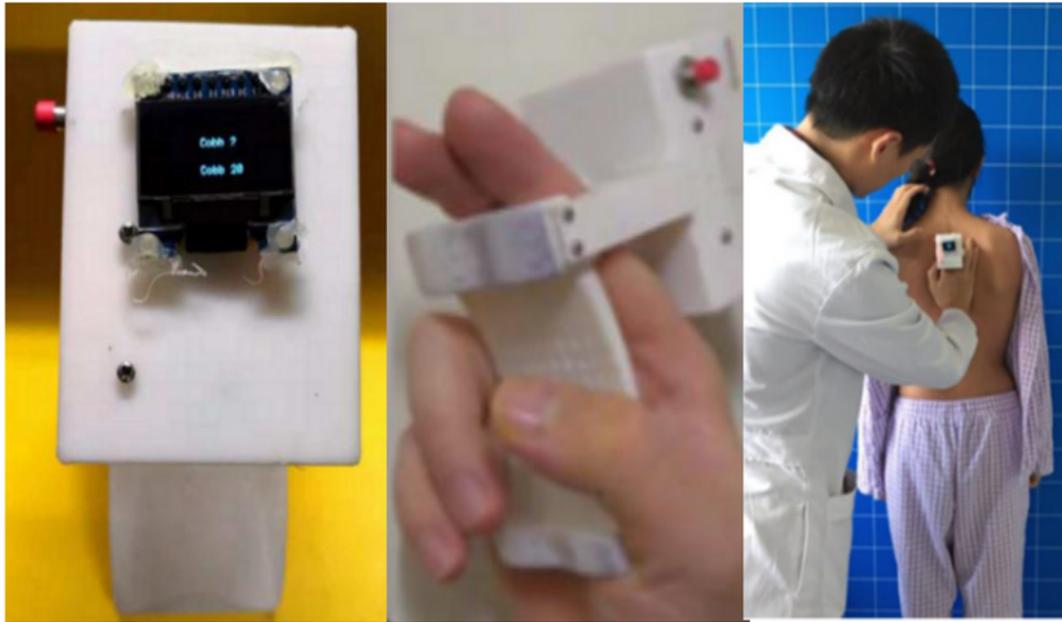
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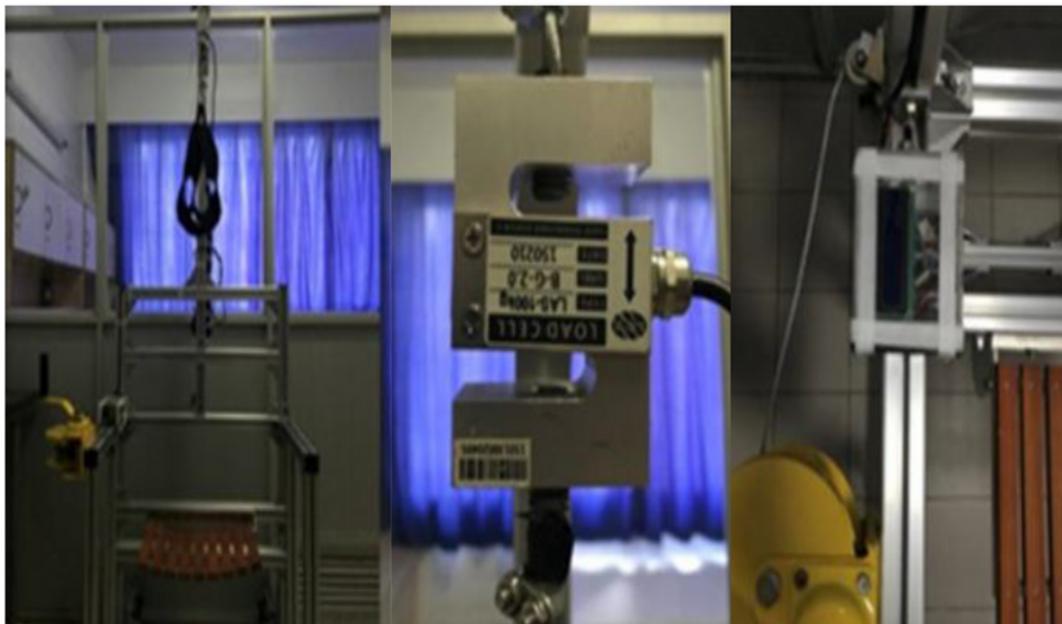
Figures



a

b

c



d

e

f

Figure 1

The spine morphometer and scoliosis longitudinal traction device: (a) the spine morphometer; (b) The spine morphometer is fixed on the back of the second and middle fingers of the measuring personnel; (c) the spinous processes of AIS patients are palpated through the sliding of the two fingers, and the spine morphometer records the sliding track the fingers; Cobb angle was obtained by connecting the surface

projection position of the spinous processes; (d) traction frame ; (e) mechanical sensor; (f) traction indicator.

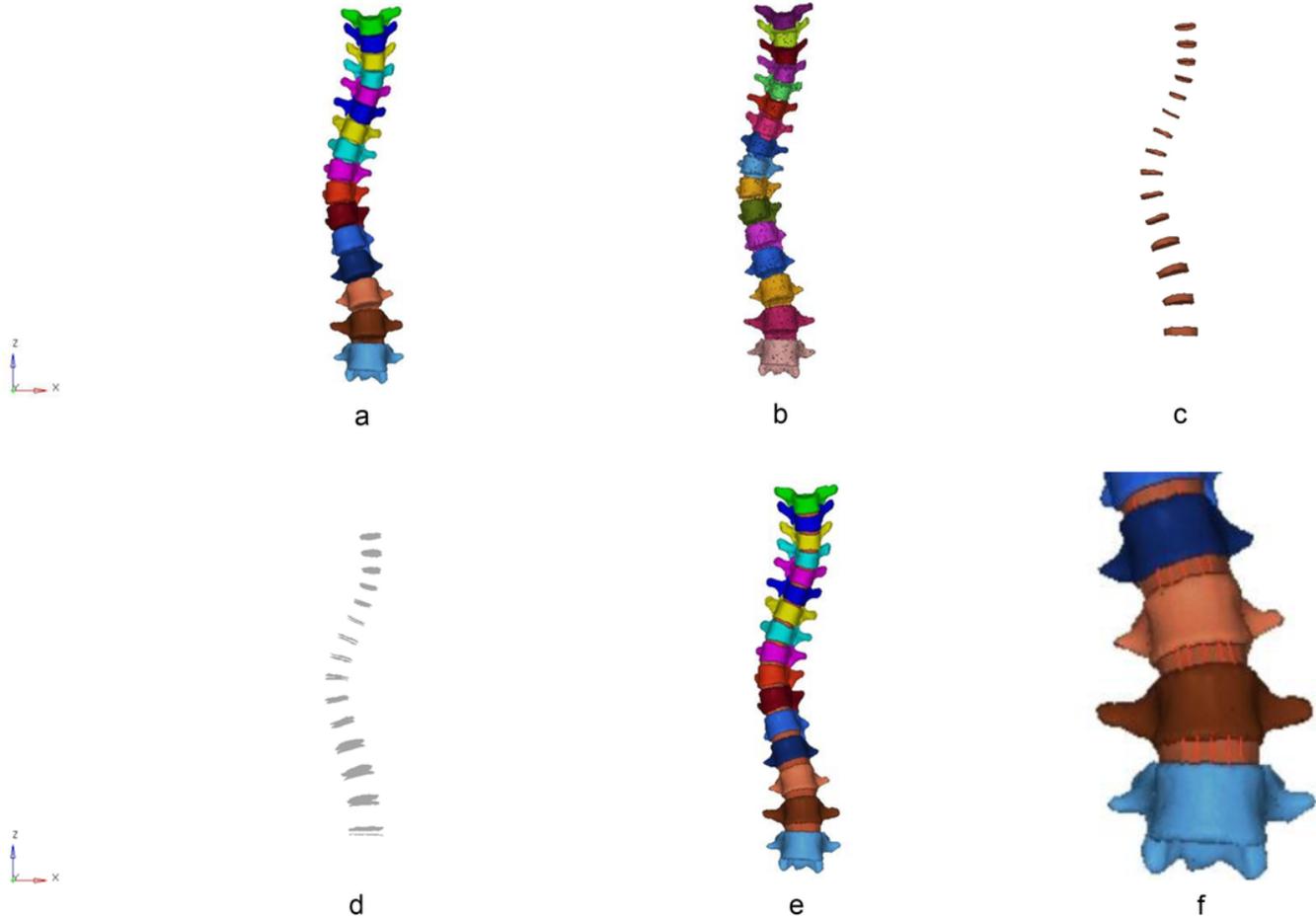


Figure 2

The finite element model A: (a) established cortical bone by the Hypermesh 14.0; (b) established cancellous bone by the Hypermesh 14.0; (c) established intervertebral disc; (d) Defining Tie contact connects the grid of vertebral segments and intervertebral discs; (e) The intervertebral segments were connected by the T3D2 unit to simulate ligaments; (f) Five anterior longitudinal ligaments and posterior longitudinal ligaments were set,respectively.

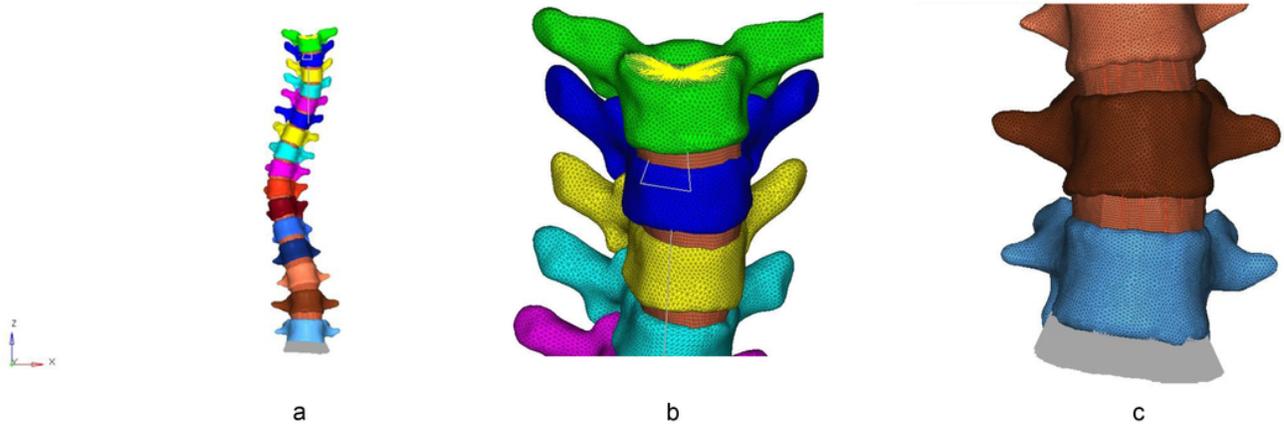


Figure 3

(a) Model A, (b) coupling constraint of T1 top surface, (c) constrained L4 bottom surface.

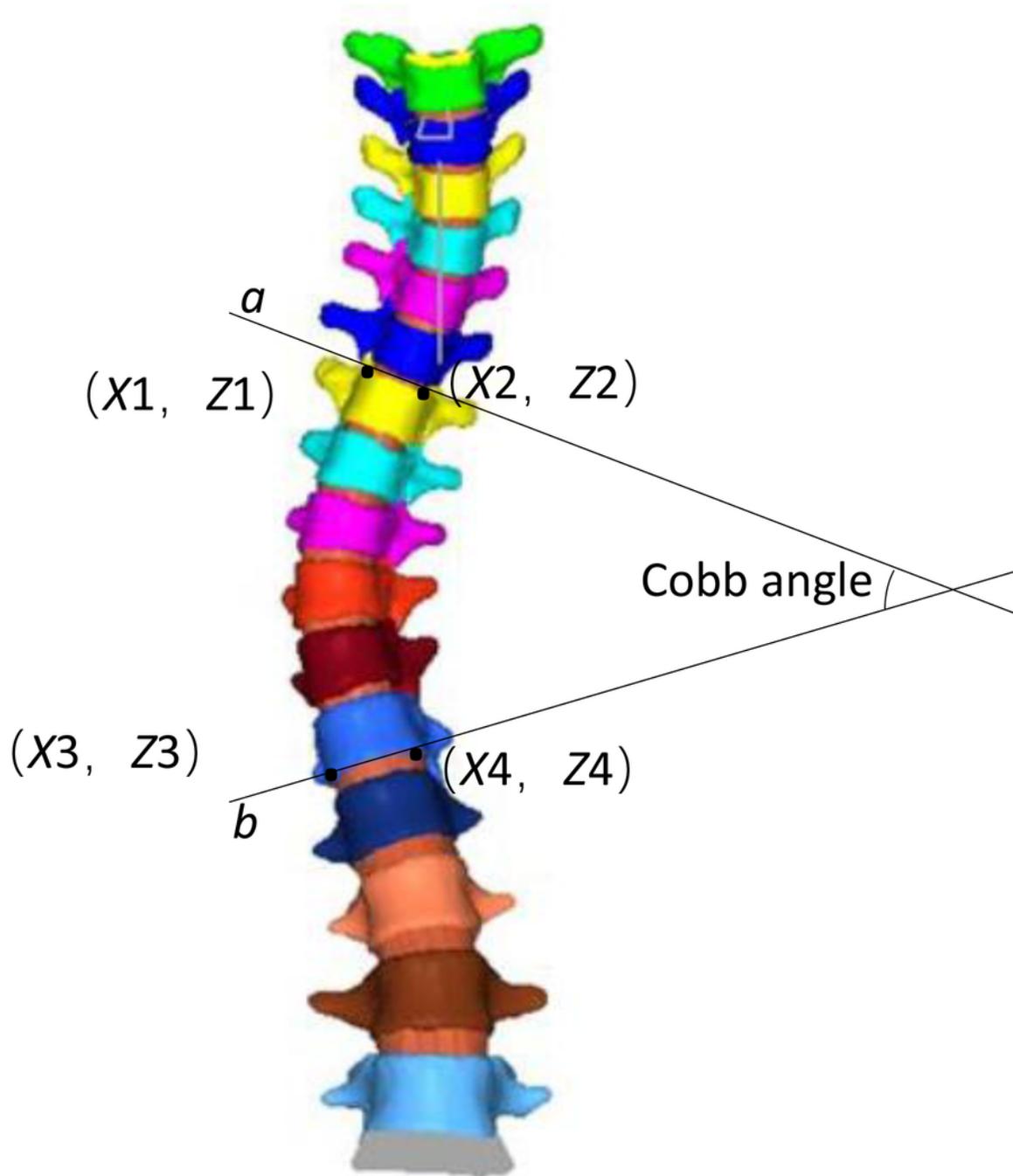


Figure 4

Calculation method of Cobb angle in the finite element model. Two points of the superior endplate of the upper end vertebrae were selected and their coordinates were recorded as $(X1, Z1)$, $(X2, Z2)$, which were connected to obtain a straight line a, with the slope being k_1 ; then the two final points of the inferior endplate of the lower end vertebrae were selected and their coordinates were recorded as $(X3, Z3)$, $(X4,$

Z4), which were connected to get a straight line b, with the slope being k_2 . The angle between the two straight lines was the Cobb angle.

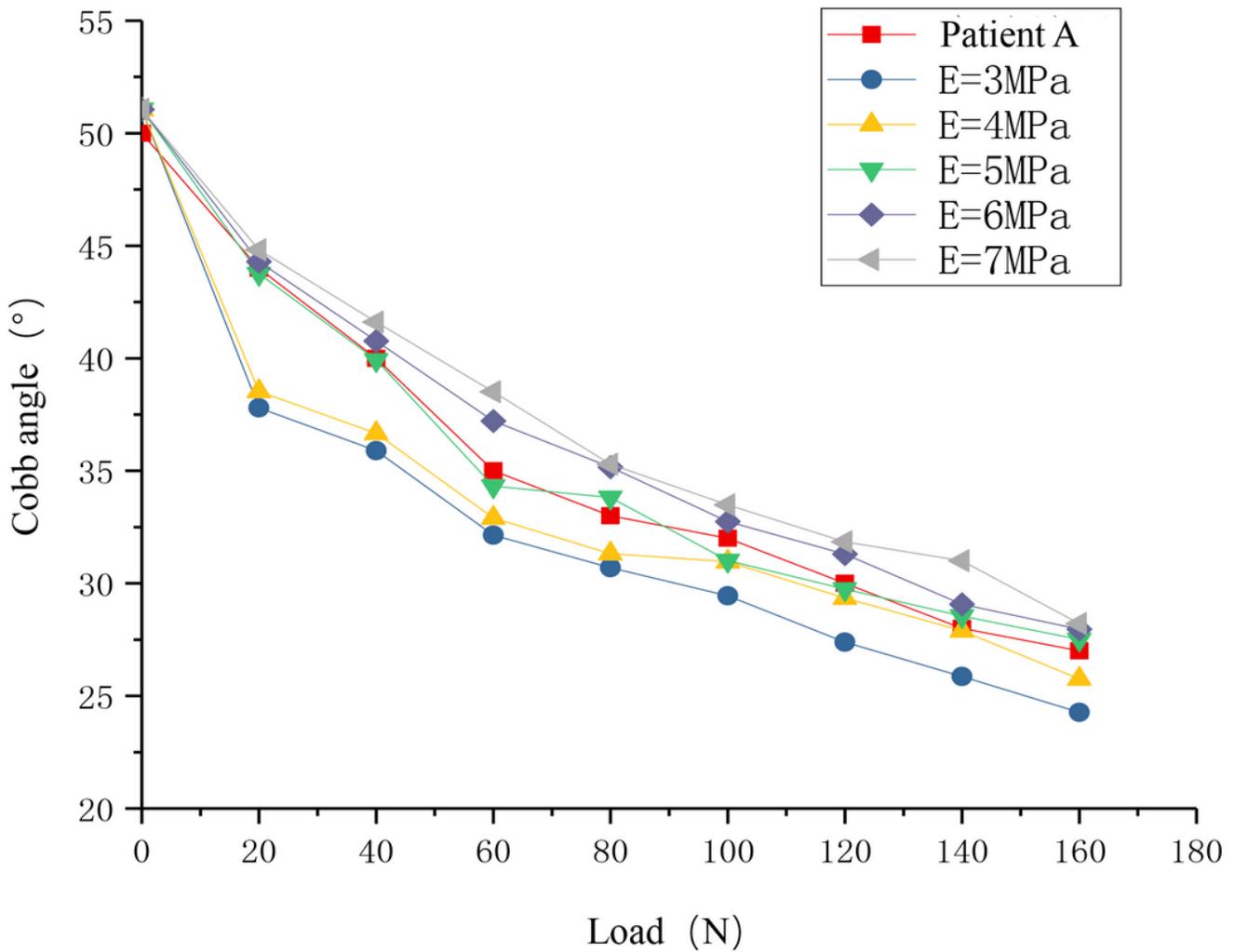


Figure 5

Dynamic flexibility curve of patient A and dynamic flexibility curves of model A with different E(elasticity modulus).

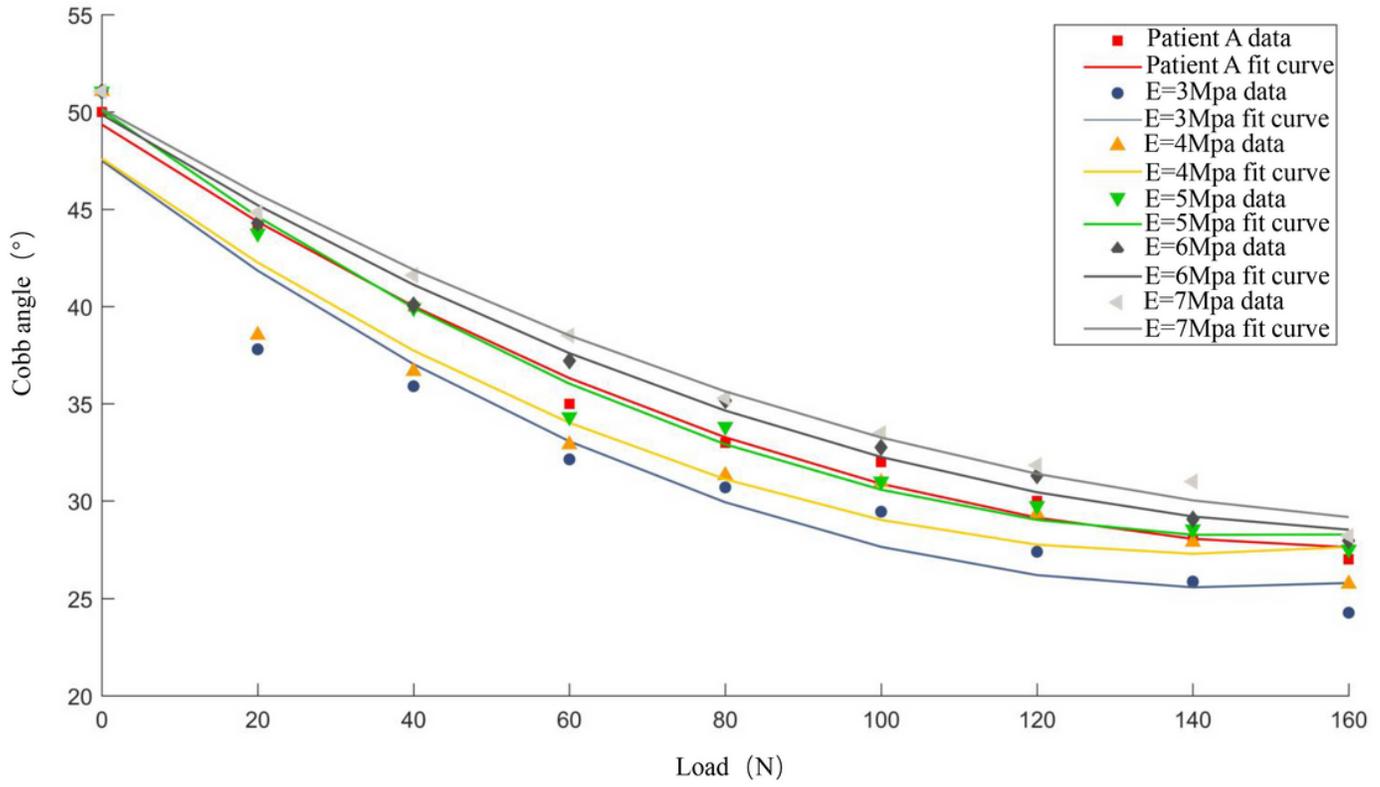


Figure 6

Fit curves of dynamic flexibility curve of patient A and dynamic flexibility curves of Model A with different E(elasticity modulus).