Determination of three-dimensional corrective force in adolescent idiopathic scoliosis and biomechanical finite element analysis

Tianmin Guan
Dalian Jiaotong University
Yu-fang Zhang (zyfdjtu@163.com)
Dalian Jiaotong University

Research

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Abstract

Background: Adolescent idiopathic scoliosis is a complex three-dimensional deformity of spine and soft tissues. It is usually treated with spinal brace if Cobb angle is less than 40°. To date, displacement and rotation of human vertebrae and ribs in three-dimensional space have not been fully considered in treatment of scoliosis. Three dimensional corrective forces in treatment of scoliosis are still unclear and very less attention has paid on it.

Methods: An objective function of corrective force in three-dimensional space is defined. Computed tomography images were used to reconstruct three dimensional model of scoliotic trunk. Computer aided engineering software Abaqus was used to establish finite element model of deformed spine and its biomechanical characteristics were analyzed. By adjusting magnitude and position of corrective forces, objective function is minimized to achieve best orthopedic effect. The proposed corrective conditions were divided into three groups: 1. thoracic deformity; 2. lumbar deformity; 3. both thoracic and lumbar deformities were considered.

Results: In all three cases, the objective function was reduced by 58%, 52%, and 63%, respectively. The best correction force point is located on convex side of maximum displacement of vertebral body.

Conclusions: Using minimum objective function method, spinal deformity in three-dimensional space can be sufficiently reduced. This study provides scientific basis for design of a new corrective brace for treatment of scoliosis.

Background

Adolescent idiopathic scoliosis (AIS) is a three-dimensional deformity of spine and ribs in adolescence, which results in trunk deformation. At present, brace is frequently used in treatment of AIS with Cobb angle 20° to 40°. This treatment modality is advocated as an effective method for treatment of scoliosis [1–3]. Traditionally designed models and fabrication methods of thoraco-lumbo-sacral orthosis (TLSO) are filled with plaster. Plaster is added or removed according to clinical experience. Finally, brace is thermo plastically formed on trimmed plaster cast. It’s shown that the correction effects of brace made by this method is often different from predicted effects and patients may have different degrees of flat back deformity in sagittal plane [4, 5]. In order to quantify spinal deformity condition with and without braces, it is necessary to provide a set of effective parameters to analyze patient specific treatment effect of patients.

The finite element model (FEM) has been applied in biomechanical study of scoliosis treatment [10, 11]. In clinical series of 5 patients treated with Milwaukee braces, Andriacchi et al [12] has found that braces had better correction effects in patients with coronal deformity, but it increased incidence of flat back in sagittal plane. Goldberg et al [2] studied the treatment effects of Boston braces, and found that Boston braces not only improve coronal plane deformity but also minimizes rate of flat back syndrome. A recent study compared the correction effects of computer-aided design and manufactured braces with that of traditionally designed braces [11]. Moreover, braces obtained by computer aided design and manufacture technology are lighter and have better correction effects. Although many scholars have used FEM simulation to design personalized braces, but they mainly concentrated on improvements in coronal plane. Sagittal and horizontal planes are not fully considered. So the ideal corrective effects of braces are lacking in recent literature. In this study, deviation distance and rotation angle of scoliosis and normal spine in three directions (that is in coronal, sagittal and horizontal planes) are set as parameters to construct objective function. The biomechanics was simulated using FEM and analyzed the effects of brace correction. This method provides theoretical basis for designing a better correction brace.

Results

The defined aim index was as: a. Cobb angle of thoracic and lumbar lateral bending in coronal plane; b. sagittal rib kyphosis and lumbar lordosis; c. values of variables in objective function were the lateral bending distances on each plane.

By optimizing iteration calculation, optimal correction forces can be obtained after convergence condition is reached. The degrees of coronal correction of thoracic and lumbar vertebrae were 58% and 52% respectively. The simulation results were compared with the orthopedic condition of the patient wearing the brace at 6 months as show in Table 2. Relative to clinical data simulation to achieve the effect of correcting the flat back phenomenon, and is closer to the normal spine physiological curve. We get the best correction forces were: F_1 of 58N, located at seventh rib on right side of model; F_2 of 36N, located in front of left fifth rib; F_3 of 41N on left side of third lumbar vertebrae and F_4 of 23N on posterior part of second lumbar spine.

This study shows that correction results in coronal plane are similar to those achieved by clinical wearing braces. However, it can reduce adverse effects of traditional braces in sagittal plane and horizontal plane, such as flat back in sagittal plane and rotation in horizontal plane. Although this simulation is a correction effect produced at moment of application of force, it has a certain correlation with long-term treatment.

The displacement of the spine under three sets of orthopedic conditions is shown in Fig. 3. The simulation results showed that maximum displacement of first group was at T8 (the maximally deformed vertebral body) and the displacement was 74.1 mm. Thoracic lateral bending was corrected to a certain extent and rotation of horizontal ribs was minimized due to application of correction force F_2 on ribs at the same time. In second group, F_3 and F_4 correction forces were applied to L3 and posterior lumbar spine respectively. Lumbar scoliosis was corrected and lower back physiological curve was improved. Because in second group only lumbar correction forces were applied, the side bending of chest was affected and improvement effect was poor. In third group, when thoracolumbar and lumbar correction forces were applied simultaneously, overall scoliosis was corrected and spinal curve was close to that of normal spine curve. Scoliosis was improved in 3D space and increased in vertical direction to a certain extent. It showed that objective function has significance in correction of scoliosis in 3D space.

Due to constraints on upper and lower boundaries of model, the maximum stress was at upper and lower ends of trunk. The maximum stress value in three groups was 354 Kpa, which is lower than maximum stress threshold that human body can bear [6], as show in Fig. 4. Practically, human body can be adjusted automatically because there is no constraint of upper and lower boundaries, and stress is smaller than simulation results.

Discussion

In this study, a new objective function is proposed, which considers correction of spine and ribs in 3D space more comprehensively using the braces. This provides a theoretical basis to design much better braces with higher efficiency. According to our findings, application of different correction forces in 3D space can reduce rate of flat back phenomenon.

When defining the objective function, seven deformations were defined to consider displacements and rotation in 3D space. The weightings definition of a function was related to value of variable, and weight changed when variable changed. In order to avoid large errors in simulation process, convergence conditions were set up in this study. However, residual deformities or overcorrection were allowed in course of correction. In actual clinical correction process, best correction effects of spine which are consistent with normal spine cannot be produced permanently. This indirectly proves rationality of convergence conditions set in this work. In treatment of scoliosis patients, physicians can measure 7 variables according to actual situation of patients. According to individual condition of patients, weight of objective function is individually set to get the best correction effects.

In simulation of spine model, as 3D correction forces increased, Cobb angle of thoracolumbar segment gradually decreased and rotation angle of vertebral body decreased too. Combined correction effects of combined forces were better. After applying correction forces, stress at intervertebral discs in deformed area changed significantly. When correction force was 20 N, stresses on convex side and concave side of disc between T7 and T8 were 120 KPa and 93 KPa respectively. When correction force was increased to 40 N, stress on convex side reached to 153 KPa and stress on concave side was 84 KPa. Stress on convex side and back side of vertebral body was larger than that was on concave side, and stress level of cortical bone was higher than that of cancellous bone. The maximum stress value appeared in cortical bones of thoracic and lumbar segments. Discs with greatest stress were located on the most convex side of deformed spine, and maximum stress value was between T7-T8. According to Wolff's law [7], the change of stress on intervertebral discs is beneficial to growth of concave vertebral body and intervertebral disc, which is beneficial to provide endogenous stability for spinal correction. 3D correction forces compensated the deficiency in 2D plane correction, and did not produce side effects such as flat back. The overall correction effects were better than that was reported in previous work [8]. Changes in coronal and sagittal spine curves are shown in Fig. 5.

This study results are similar to results by the Cheneau brace. Results generally are same as current brace theories using three point forces per curve. However, the simulations have shown that the correction can be achieved by reducing some adverse effects of the current bracing systems, as in the sagittal plane (flat back problem) and the limited rotation in the transverse plane. Chase A et al [9] showed that short-term correction has the same characteristics as correction resulting from long-term treatment.

Some factors associated with the methods may influence the overall results of this study, such as the boundary conditions applied to the model. In all biomechanical modeling of the human trunk, selecting the boundary conditions is complicated because there is no part of the spine that is completely fixed in space. These conditions should be plausible and carefully chosen as they produce reaction forces at the boundary vertebrae (T1 and L5), which are not necessarily present in a real brace.

These weightings can be changed by doctor as desired by the physician aiming to focus on a specific correction for each patient s deformities. All weightings in the objective function and the range allowed for the descriptors should be individualized to each patient in order to choose where to put the emphasis of the correction.

Conclusions

This study demonstrates an optimization approach to finding effective corrective forces in the three-dimensional treatment of idiopathic scoliosis on the basis of the individual spinal and rib cage geometry of patients. It could now be performed to analyze different three-dimensional descriptors and their effect on the clinical indices, especially the ones that were not improved (ie, axial rotation and planes of maximum deformity). In the long term, this study could help physicians and orthotists to optimize the placement of correction in braces to produce better correction and improve bracing in idiopathic scoliosis.

Methods

Patient specific 3D finite element model was constructed from CT images data of AIS. The correction effects of different applied forces were analyzed by using biomechanical principles.

Personalization of the finite element model

The patient was an 11 year old with AIS, having typical right thoracic and left lumbar scoliotic curvatures, with thoracic Cobb angle of 36° and lumbar Cobb angle of 24°. The maximum lateral bending displacement was in T9 and L3 vertebrae. In coronal plane, displacements were 63mm and 32mm whereas in sagittal plane these were 46mm and 35mm respectively. The maximum rotation angle of ribs was 13° and kyphotic displacement of ribs was 23 mm. A 3D model of AIS was established by using medical image processing software (Mimics 19.0; Materialise, Leuven) from DICOM CT images. Spine model was extracted according to bone threshold and then region growing tool was used to segment the different regions. According to the knowledge of human anatomy, a three-dimensional model of AIS including the thoracolumbar vertebrae, intervertebral disc, ribs and pelvis was established. By comparing the finite element model with the patient's X-ray film, the distance between the length of the vertebral mass and the tibia and the distance between the X-ray films are less than 1 mm. The angle between the midline of each vertebra and the tibia and the angle between the X-ray films are less than 0.5 degrees. This proves that the model is basically consistent with the patient's scoliosis features. Point cloud data was obtained by scanning patient's body surface with a 3D scanner. In Geomagic studio we denoised and smoothened to reconstruct a 3D model of body contour. Patient's body surface was assembled with its skeletal model to obtain 3D geometric model of trunk. The software (Hypermesh13.0; Altair; America) was used to mesh 3D model.

A personalized model contains 152301 nodes and 302357 tetrahedral elements representing the thoracic and lumbar vertebrae, intervertebral disks, ribs, sternum, intercostal ligaments, anterior longitudinal ligament, posterior longitudinal ligament, ligamentum flavum, intertransverse ligaments, interspinal ligament, supraspinal ligament, costovertebral and costotransverse joints, and zygapophyseal articulations. Muscle tissues (trapezius muscle, abdominal muscles, rectus muscle and so on) were simulated by spring elements. Mechanical properties of anatomical structures were taken from published data (see Table 1) [10, 13-16]. Finally 3D model was processed and simulated in CAE (computer assisted engineering) software (Abaqus 6.14-2; Dassault; France) to simulate.

Objective function

Due to symmetry of human body, there is no hump in coronal plane of the normal spine and ribs are symmetrically in horizontal plane of midline of spine [17]. In order to obtain curve of normal spine in sagittal plane, 12 healthy adolescents with no medical history of spinal disorders were selected, with an average age of 12.0 years consisting of 6 males and 6 females. The results in form of normal curve of spine were obtained.

Defining parameters

In coordinate system, coronal axis was X axis, sagittal axis Y axis and vertical axis as Z axis. Curvature curve was defined by connecting the center of each vertebral body in 3D model. Similarly, normal spinal curve was defined. In this work, objective function was consisted of seven variables; displacement distance between scolitic and normal spinal curves in thoracic and lumbar regions in coronal plane as X_1, X_2, and in sagittal plane as Y_1, Y_2 respectively. Distance between left and right ribs kyphosis in horizontal plane was defined as C. The rotation angle of first scoliosis rib was θ_{z0}, and initial rotation angle of vertebral body was θ_{y0}. These variables are represented in the 3D view interface as shown in Fig. 1.

Since geometric variables may be positive or negative, the square of displacement was chosen to be square and square root to ensure that the numerical value is positive. The objective function is expressed as follows:

See Formula 1 in the Supplementary Files.

Where α, β, γ and δ, ε are the weightings of distance and rotation in coronal, sagittal and horizontal planes respectively.

The interval ranges from 0 to 1, N is total number of vertebrae, and $d\theta$ is increment of relative initial rotation value of each vertebra. In order to fully consider the lateral bending of vertebrae and ribs, rotation angles of each rib and vertebrae were counted. The optimal solution of objective function was to reach the minimum value that is 0 in equation.

The optimal scoliosis correction curve was considered when body was in upright position whereas lateral displacement, tilt angle and axial rotation of spine in 3D space were all 0. In actual clinical scoliosis correction process, due to complexity of muscles, skeleton and other structures, the correction often cannot achieve ideal state. Therefore, in treatment of scoliosis, goal is to minimize the objective function. It can be done to minimize each variable in space as much as possible. In patients with bilateral bends, displacement of coronal plane is statistically larger than that of sagittal plane [11]. Clinically, physicians usually consider coronal deviation. Instead of vertebral rotation, ribs rotation angle is smaller. In order to distribute the weight of each variable reasonably, which is to define the proportion of each deformation amount to the total deformation amount. Larger the migration distance, larger the weight.

Loading condition

Three point force principle is applied to scoliosis model to achieve correction effects in 3D space. Applied force was divided into three groups: a. F_1 and F_2 were applied to thoracic vertebrae and ribs to examine correction of thoracic vertebrae in coronal axis and ribs kyphosis. b. Application of F_3 and F_4 in position of lumbar scoliosis, measuring lumbar vertebral correction; c. Four sets of correction forces, F_1, F_2, F_3 and F_4, were applied to position of lateral bend at same time. In coronal plane, F_1 and F_3 correction forces were applied to thoracic and lumbar vertebrae in coronal plane. In horizontal plane, deformity of vertebral rotation was improved by applying F_2 correction forces on kyphotic ribs. In sagittal plane, F_4 correction forces were applied to posterior lumbar spine to alleviate flat back and to maintain normal physiological curvature. The specific application of force is shown in Fig. 2. Proposed forces were transmitted to spine through thoracic and lumbar soft tissues.

Review of literature concerning non-surgical treatment of scoliosis showed that initial maximum displacement of thoracic and lumbar vertebrae is within 100 mm and rotation angle of ribs is within 40° [18]. In order to make this study more practical, the maximum displacement of vertebrae and maximum rotation angle of ribs were set at 200 mm and 50° respectively.

Finite element analysis

When body is under high pressure, it will not only produce discomfort but also cause pressure ulcers. In order to ensure that local pressure is within tolerance range, it is necessary to select a reasonable stress area. Visser D et al [19] used visual analogue scales and pressure sensors to measure pressure generated by braces. Other researchers measured pressure on trunk of in-braces patients in range of 10 cm^2 and withstanding 10 Kpa pressure while contact point force was kept within 10 N, [20-22]. Cobetto et al [23] collected the pressure between brace and patient's body surface and gave a withstanding threshold of 35

Kpa. The correction forces were guaranteed in range of human pain threshold. The applied range of force was 0 ~ 100N, and the applied area was 64 ~ 225 cm^2 [23,24].

Convergence conditions

Internal and external geometries were linked together. Using an iterative nonlinear resolution method, applied forces on selected vertebrae lead to correction of external trunk model [24,25]. The simulation boundary conditions included a fixed pelvis in rotation-translation. T1 was limited to transverse plan movements. According to scoliosis curve of patient, the values of variables in objective function were calculated, and initial variables were taken as original values. By adjusting magnitude of correction forces, the deformation of objective function was obtained. As the initial value of next iteration, results of the previous step were iterated until convergence conditions were satisfied. Convergence conditions used were as: 1. total number of iterations exceeding 200 steps; 2. variation of variables in two iterations was within 2mm; 3. difference between two iterations was less than 1 N, accounting for about 1% of the force range. When convergence condition was reached, the iteration was terminated.

Declarations

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Authors' contributions

(1) TG submitted final approval of the version (2) YZ drafted the article or revising it critically for important intellectual content

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Availability of data and materials

Not applicable.

Ethics approval and consent to participate

Bioetic Commission for this study is not necessary. Written informed consent was obtained from individual or guardian participants.

Consent for publication

Not applicable.

Competing interests

The authors declare that they have no competing interests.

Author details

[1]School of Mechanical Engineering, Dalian Jiaotong University, No. 794 the Yellow River Road, Shahekou District, Dalian 116028, China

Abbreviations

CT: computed tomography; AIS: adolescent idiopathic scoliosis; 3D: three dimensional; FEM: finite element method; DICOM: digital imaging and communication in medicine; TLSO: thoraco-lumbo-sacral orthosis

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Tables

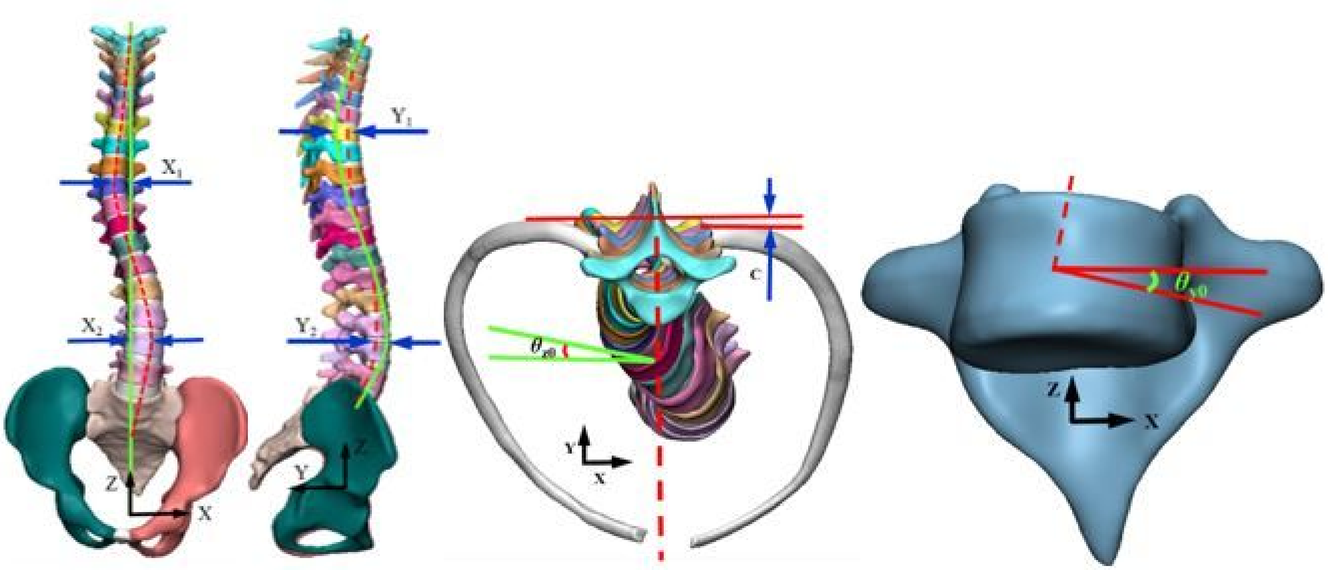
Table1 Material properties of each part

Components	Young’s modulus E (Mpa)	Poisson’s ratio ν	Cross sectional area(mm^2)	Stiffness	Density ρ kg/m^3
Cortical bone	12000	0.3			
Cancellous bone	100	0.2			
Posterior structure	3500	0.25			
Nucleus	1	0.49			
Annulus	4.2	0.45			
Rib	5000	0.1			
Endplate	24	0.4			
Pelvis	5000	0.2			
Sacrum	5000	0.2			
Costal cartilage	480	0.1			2000
Skin	31.5	0.42			1000
Muscle	1.33	0.14			1100
anterior longitudinal ligament			22.4	8.74	
posterior longitudinal ligament			7.0	5.83	
ligamentum flavum			14.1	15.38	
intertransverse ligaments			0.6	0.19	
interspinal ligament			14.1	10.85	
supraspinal ligament			10.5	2.39	

Table 2 Index parameter

Parameter（mm）	Initial value	Clinical data	Simulation results	Parameter（mm）	Initial value	Clinical data	Simulation results
Cobb (°)				Rib kyphosis	31	25	45
Thoracic	36	21	14	lumbar lordosis	38	31	31
Lumbar	24	16	10	X_1	63	42	15
Sagittal plane (°)				X_2	-32	-21	-10
Thoracic	34	27	36	Y_1	-34	-28	-8
Lumbar	35	31	38	Y_2	26	13	4
				C	23	14	6
				θ_{z0}(°)	13	15	5
				θ_{y0}(°)	14	8	4

Figures



a. Coronal plan b. Sagittal plan c. Transverse plan d. Vertebral rotation angle

Figure 1

Representation of variables in a three-dimensional view

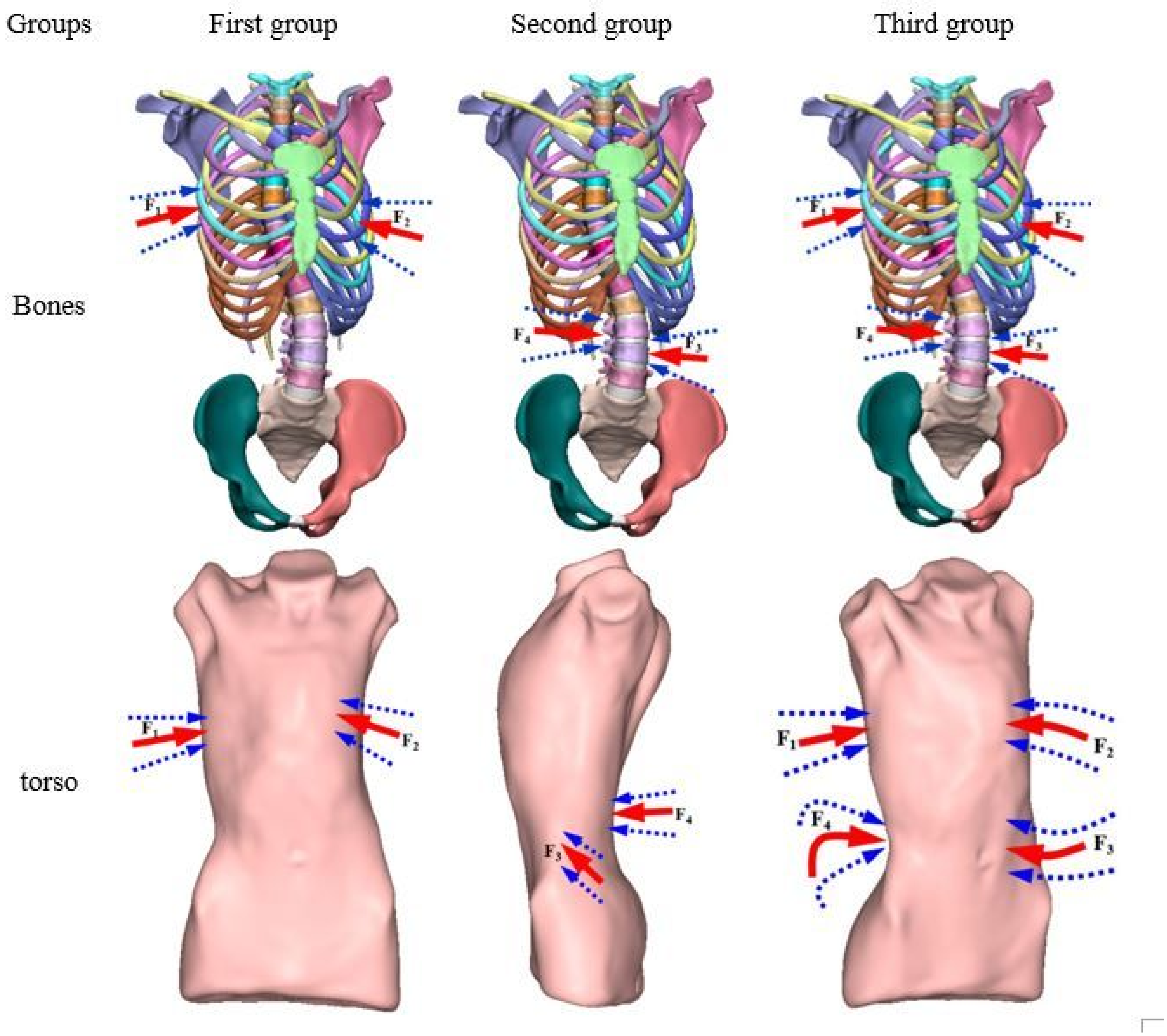


Figure 2

Force position

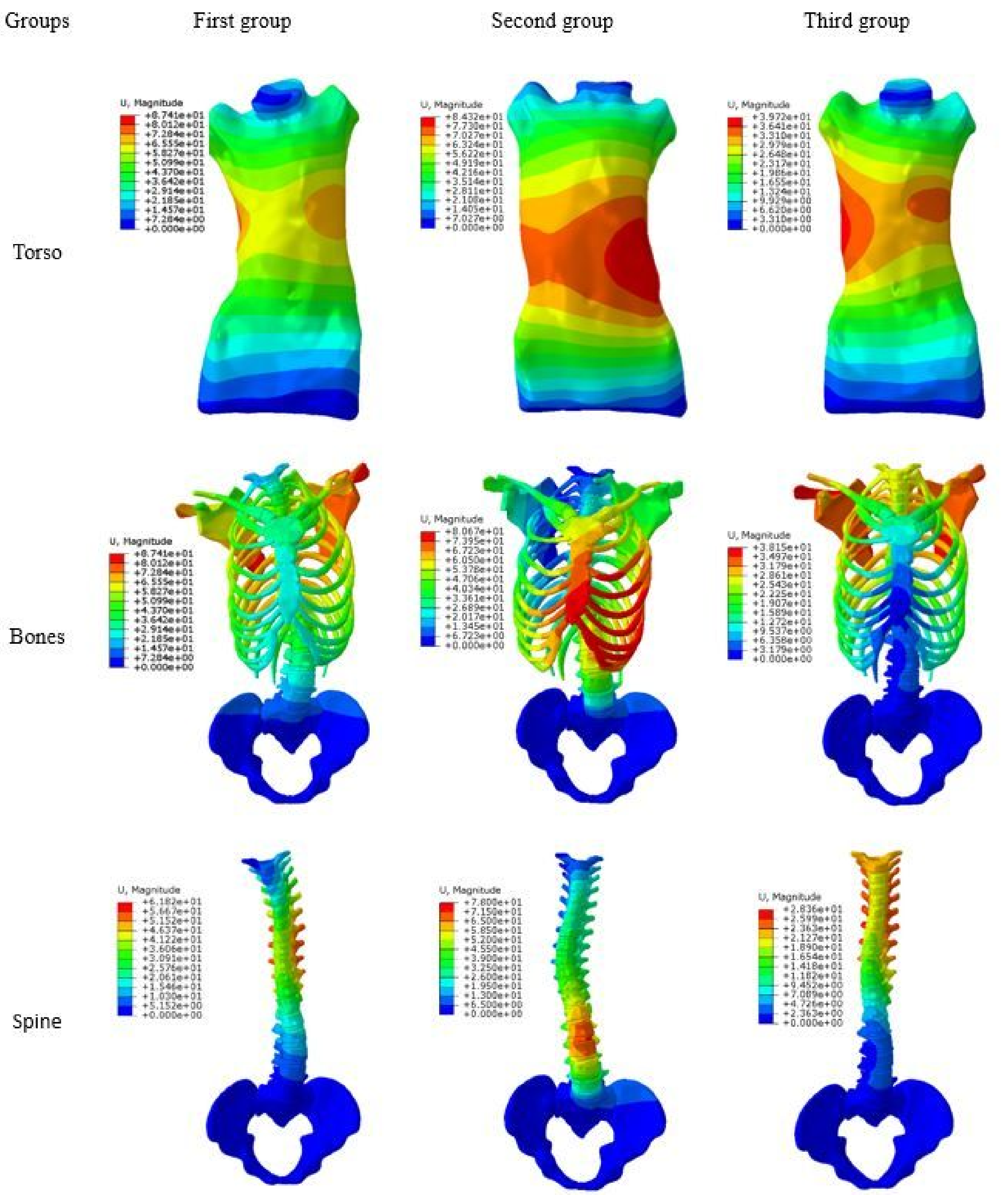


Figure 3

Displacement variables

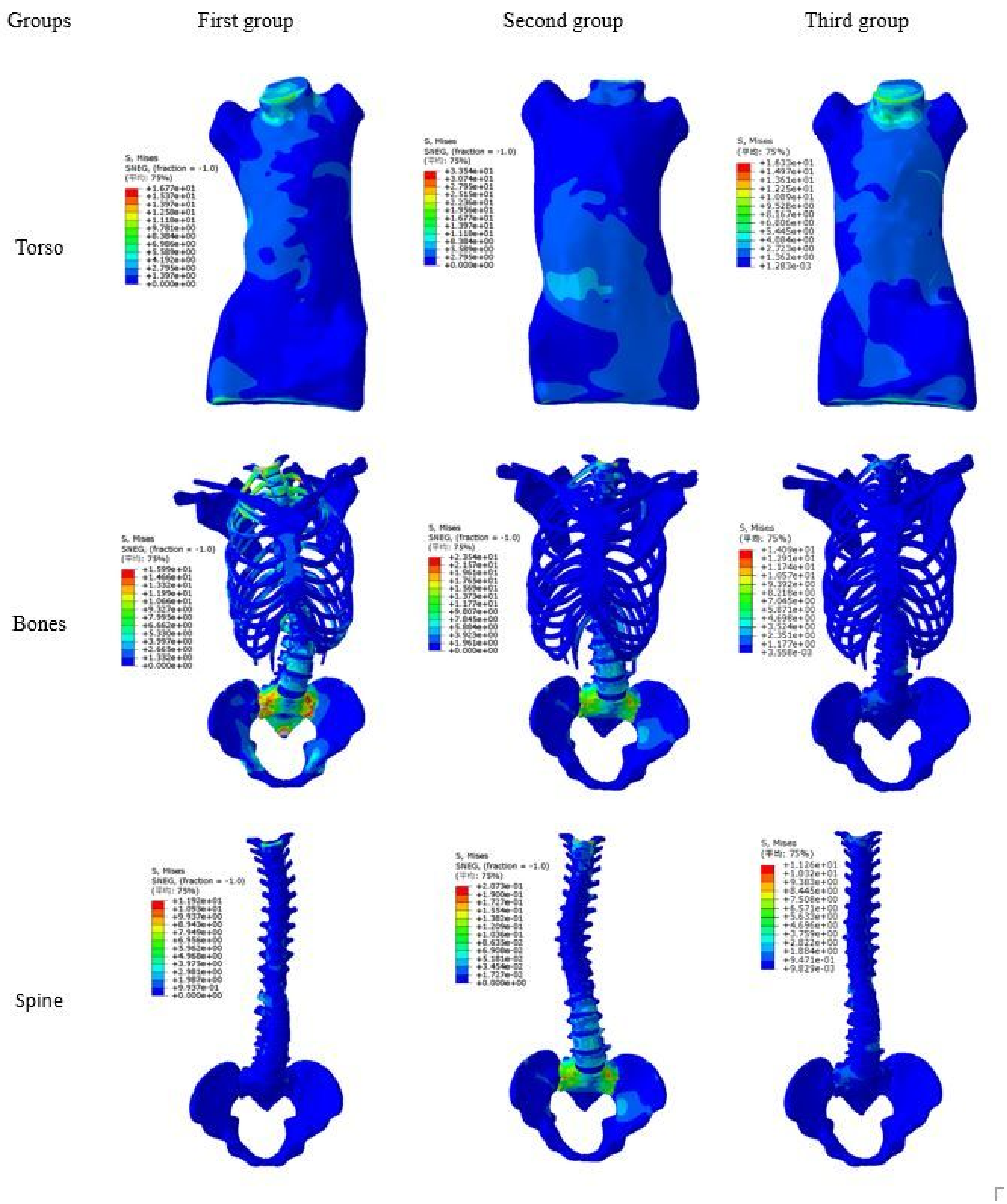


Figure 4

Stress Diagram

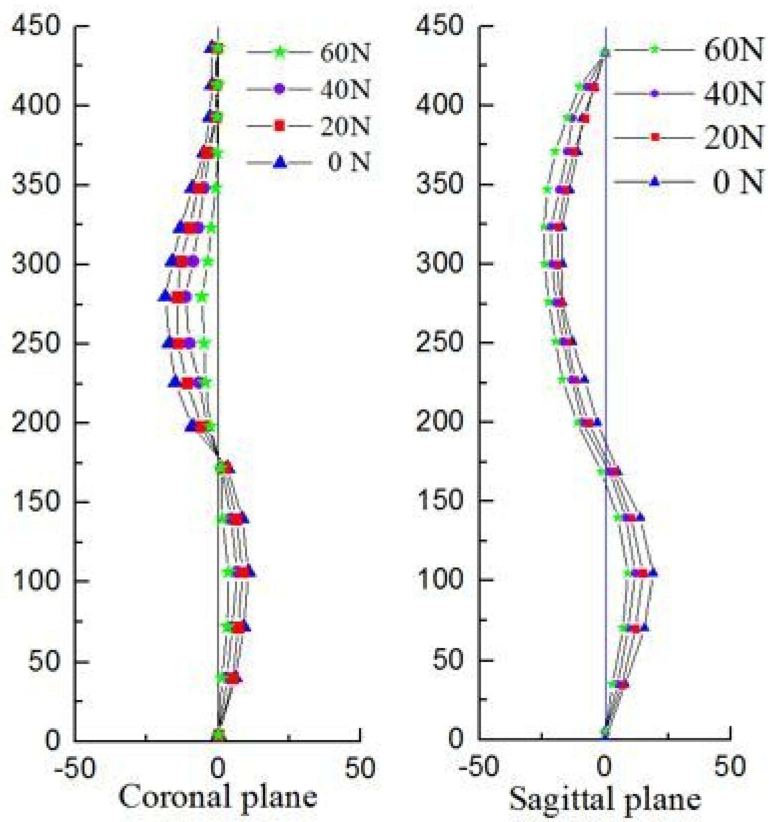


Figure 5

Changes in the curvature of the spine with different correction forces

Supplementary Files

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- Formula1.JPG