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ORIGINAL ARTICLE

Patient-specific spinal stiffness in AIS: a preoperative and noninvasive method

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Abstract
Introduction The clinical tests currently used to assess spinal biomechanics preoperatively are unable to assess true mechanical spinal stiffness. They rely on spinal displacement without considering the force required to deform a patient's spine. We propose a preoperative method for noninvasively quantifying the three-dimensional patient-specific stiffness of the spines of adolescent idiopathic scoliosis patients.
Methods The technique combines a novel clinical test with numerical optimization of a finite element model of the patient's spine.
Results A pilot study conducted on five patients showed that the model was able to provide accurate 3D reconstruction of the spine's midline and predict the spine's stiffness for each patient in flexion, bending, and rotation. Statistically significant variation of spinal stiffness was observed between the patients.
Conclusion This result confirms that spinal biomechanics is patient-specific, which should be taken into consideration to individualize surgical treatment.

Keywords Spine · Stiffness · Adolescent · Scoliosis · Noninvasive

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Introduction

Spinal fusion is the treatment of choice when the thoracic curve of an adolescent idiopathic scoliosis (AIS) patient is expected to reach at least 50° by skeletal maturity [1]. Accurate planning of surgery requires a good understanding of both patient-specific spinal morphology and stiffness. Spinal morphology is usually obtained from medical imagery such as X-ray radiographs, but the surgeon has only limited information on the mechanical behavior of the patient's spine. Methods such as side bending or fulcrum bending tests [2], push-prone [3], traction under general anesthesia [4] and push-traction films [5] have been proposed to quantify the stiffness of the patient's spine prior to surgery. Unfortunately, these tests solely rely on the displacement of the spine without taking force information into account. Only reducibility (displacement of the spine) can be evaluated by these methods, not mechanical stiffness.

To overcome this issue, alternative clinical tests have been proposed to simultaneously measure both spinal displacement and the forces causing it. Ghista et al. [6]

Table 1 Patient demographics and Cobb angles before and after application of a traction load corresponding to 30 % of patient body weight

Subjects	Age (years)	Weight (kg)	Apex location	Apex rotation "standing"	Cobb angle "standing"	Cobb angle "suspension"	3D reconstruction
1	15	59.3	L1	43°	67°	39°	L4–C5 (19)
2	16	56.3	T9	38°	71°	63°	L4–T1 (16)
3	13	53.6	T8	28°	46°	34°	L4–T2 (15)
4	18	48.1	T8	31°	50°	46°	L4–T3 (14)
5	15	57.0	L1	34°	60°	50°	L4–C4 (18)

The number of vertebrae reconstructed in 3D varied between 14 and 19, but always started at L4

developed a traction system in which force was directly applied to the patient's head and thus to his spine. However, the traction seems not to have been aligned with the spine's axis (the line crossing the bottommost and topmost vertebra) and cannot be considered as a pure vertical load. In 2009, Lamarre et al. [7] proposed a similar traction mechanism in which force was applied under the patients' armpits. But it is not clear how much of the traction load under the armpit is transferred to the spine. In addition, both studies evaluated spinal displacement only in the coronal plane, which is questionable due to the three-dimensional (3D) nature of the scoliosis deformity. More recently, a spinal suspension test (SST) developed in our institute quantifies spinal stiffness in three dimensions [8]. Although biplanar X-ray images of the patients were acquired, the study reported only the coronal stiffness of the curve.

Numerical approaches have also been proposed to determine spinal stiffness preoperatively [6, 9]. These models rely on a drastic simplification of the complex spinal anatomy to a series of connected, two-dimensional (2D) beam elements [6], or use a 3D rigid body connected by torsional springs [9]. Inverse finite element modeling was used to optimize the mechanical properties of the model to match experimental radiographic data. However, these optimization approaches rely either on two-dimensional numerical models or on bidimensional side bending data, which do not include force measurement. Therefore, none of the current methods is able to provide the clinician with three-dimensional stiffness measurement preoperatively.

Therefore, the aim of this study was to evaluate the feasibility of quantifying 3D spinal stiffness of AIS patients. To achieve this, we combined the recent spinal suspension test of Büchler et al. [8] with a patient-specific finite element model of the spine to identify the stiffness parameters in flexion/extension, lateral bending, and axial rotation.

Materials and methods

Five patients suffering from AIS and scheduled for surgery, who also were recruited for the previous study reported in [8], were involved in this study. The inclusion criterion for the adolescents (15.4 ± 1.8 years) was moderate to severe scoliosis (Cobb angle 58.8° ± 10.7°, Table 1). The study was carried out in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki for research involving human subjects and was approved by the local ethics committee.

The spinal suspension test was developed to apply an axial traction force on the patient's spine [8]. This system consists of a frame structure supporting a suspension platform, which can freely move horizontally to ensure a proper alignment of the force with the spine during the test. A motorized system on the platform is able to gradually apply a traction force of up to 30 % of the patient's body weight (Fig. 1). Frontal and lateral radiographic images of the patient's spine were acquired before and during the application of the traction force. A calibration unit was attached to the patient's waist and the three-dimensional spinal midline was reconstructed from these calibrated biplanar images [10].

The construction of the 3D finite element model of the patient's spine was based on the spinal midline. Two landmarks per vertebrae were used to define the position of the node of the finite element model, one for the center of each endplate. The lowermost and the topmost vertebra were used to align the spine with the vertical axis. The apex vertebrae‘s axial rotation was measured with Raimondi's table, which has been shown to be an accurate tool [11, 12]. Axial rotation of the other vertebrae was linearly interpolated as a function of their distance to the axial axis, with the apex vertebra being the most rotated. The field of view of the radiographic images differed from one patient to another; the number of vertebrae included in the 3D reconstruction therefore varied from 14 to 19. However, all reconstructions started at L4 and ended between T3 and C5 (Table 1). To evaluate the precision of the reconstruction process and its influence on the final stiffness parameters, totally independent reconstructions were done for each patient.

Spinal stiffness was identified using inverse finite element modeling. First, patient-specific finite element models

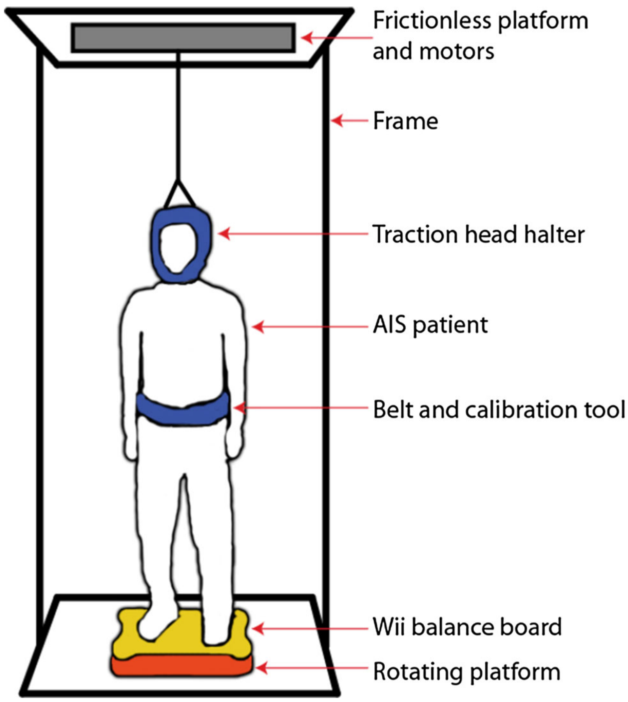

Fig. 1 Schematic representation of the Spinal Suspension Test. A traction force corresponding to 30 % of the patient body weight was applied to the patient's spine using a standard cervical traction head halter. A Nintendo Wii Balance Board was used to constantly monitor the traction force, and orthogonal radiographic images were acquired before and after application of the traction load to measure the three-dimensional displacement of the vertebrae under the applied force [8]

of the spine were constructed from the three-dimensional spine's midline in normal standing position. In the model, each vertebra was considered as a rigid body. The stiffness of the complete system was concentrated in the intervertebral disc's location (IVD), which was modeled with three linear torsional springs. Each spring was aligned with the anatomical axes and represents the spinal stiffness in lateral (K_x), axial (K_y), and frontal rotation (K_z) as previously defined by Panjabi et al. [13].

Boundary conditions of the model constrained the degrees of freedom of the lower vertebra (L4), while the most cranial vertebra was constrained to move only along the direction of the traction force. The suspension force corresponding to the experimental measurement was applied to the most cranial vertebra. The finite element problem was solved using the commercial package Abaqus (Abaqus 6.12, Dassault Systèmes, 2012).

An optimization process (Nealder–Mead algorithm) was used to determine the stiffness coefficients that best reproduce the deformation of the spine observed experimentally. The cost function E was calculated as,

$$E = \begin{cases} D + P & \text{if } P > 0 \\ D & \text{else} \end{cases}$$

where D is the mean distance between the positions of the IVDs calculated with the numerical model and measured on the patients:

$$D = \frac{1}{N} \times \sum_{i=1}^{N} \sqrt{(x_i^e - x_i^m)^2 + (y_i^e - y_i^m)^2 + (z_i^e - z_i^m)^2}$$

N is the number of IVDs, the superscript e indicates the experimental data, and m the numerical data. P is a regularization term to limit the possible deviation between the stiffness parameters along the different anatomical directions. Since the three stiffness components represent the mechanical behavior of the same soft tissues along different directions, the stiffness regularization term was included in the cost function to penalize any solution that presents high differences between the stiffness coefficients in different directions. The regularization was chosen so as to penalize stiffness parameters when the standard deviation of $\{K_x, K_y, K_z\}$ is higher than S_0:

$$P = \sqrt{\frac{1}{3} \times \sum_{i=1}^{3} (K_i - \overline{K})^2 - S_0}$$

Based on the intraoperative measurement of Reutlinger et al. [14], S_0 was chosen to be 1 Nm/°.

To evaluate the sensitivity of the optimization to the selection of the initial conditions, ten optimizations were performed for each reconstruction. The initial parameters were randomly chosen between 0.01 and 20.00 Nm/°. In addition, once the algorithm converged, the solution was used as the initial condition for a second optimization. After ten successive convergences, we regarded the optimization as converged on the optimal solution. The result of an optimization was considered as an outlier when the z-score of the cost function or of one of the stiffness coefficients exceeded 2.5.

Results

The precision of the 3D reconstruction was evaluated by comparison of ten independent reconstructions for each of the patients' spines. For all patients the reconstruction precision is good, that is, it is considerably smaller than the measured vertebral displacements under traction. Patient 1 presented the largest displacement, which was about 14 times higher than the reconstruction precision, while patient 4 presented the smallest displacement, which was on average 5 times higher than the reconstruction precision.

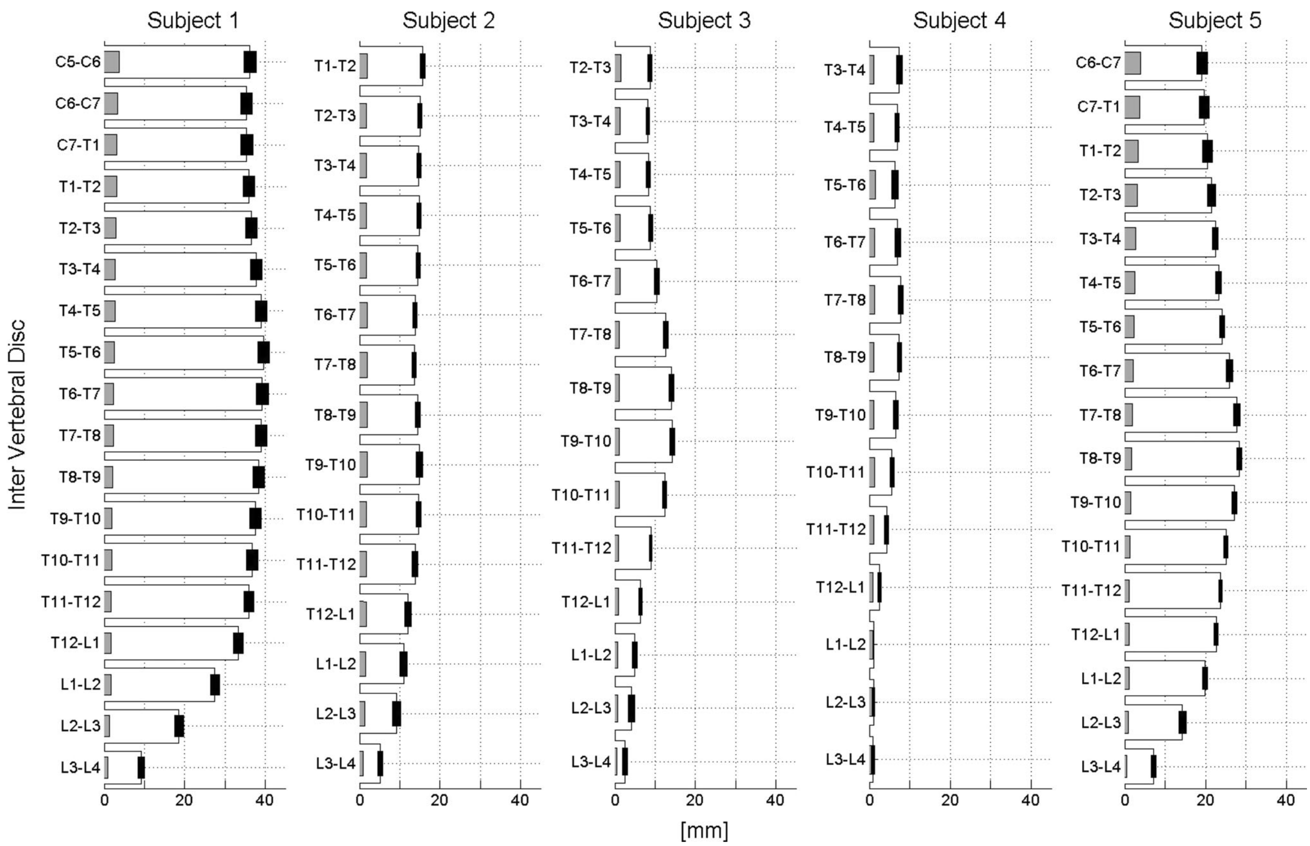

Fig. 2 The precision of the reconstruction (*gray*), the magnitude of the displacement of the IVDs from the normal to the suspension condition (*white*), and their 95 % confidence intervals (*black*). The displacements of the IVDs were on average nine times larger than the reconstruction precision

All other patients range in between these two extreme cases (Fig. 2).

To verify whether the optimization of a given reconstruction is likely to converge to the same result (global optimum), ten optimizations per reconstruction were run for each patient. Only 2.60 % of the optimizations were categorized as outliers and rejected from further analyses. The optimizations of a given reconstruction generally converged toward a single global optimum with high reproducibility (std = 0.1 Nm/°). The only exceptions were the stiffness coefficients along the axial direction of patients 1 and 5, which showed higher variation (std = 0.3 Nm/°). Therefore, the optimization procedure was robust and the standard deviation observed on the stiffness coefficients was attributable largely to the reconstruction procedure (Fig. 4).

For all patients, the quality of the fit between the model's output and the experimental suspension measurements was generally between 1 and 5 mm (Fig. 3). After optimization, the regularization term of the cost function was always equal to zero. Therefore, the regularization term is important to guide the optimization process, but has no effect on the final results.

Stiffness coefficients of a given reconstruction were computed as the mean of the ten optimizations. This process was repeated for each of the ten reconstructions. The average and standard derivation of these ten reconstructions were chosen as representatives of the spinal stiffness for each patient. The stiffness coefficients are statistically different (*t* test, $p < 0.05$) for each patient, except for K_x of patients 2 and 4, which emphasizes the patient-specific nature of spinal stiffness. The stiffness coefficients range from 0.2 to 3.2 Nm/° (Fig. 4).

Discussion

This method assesses the mechanical behavior of the spine preoperatively by combining a clinical test with an inverse finite element model. The pilot study on five patients showed that the model was able to provide accurate 3D reconstruction of the spine and predict spinal stiffness for each patient. To the best of our knowledge, this work is the first to quantify the spinal biomechanics in 3D based on data acquired preoperatively.

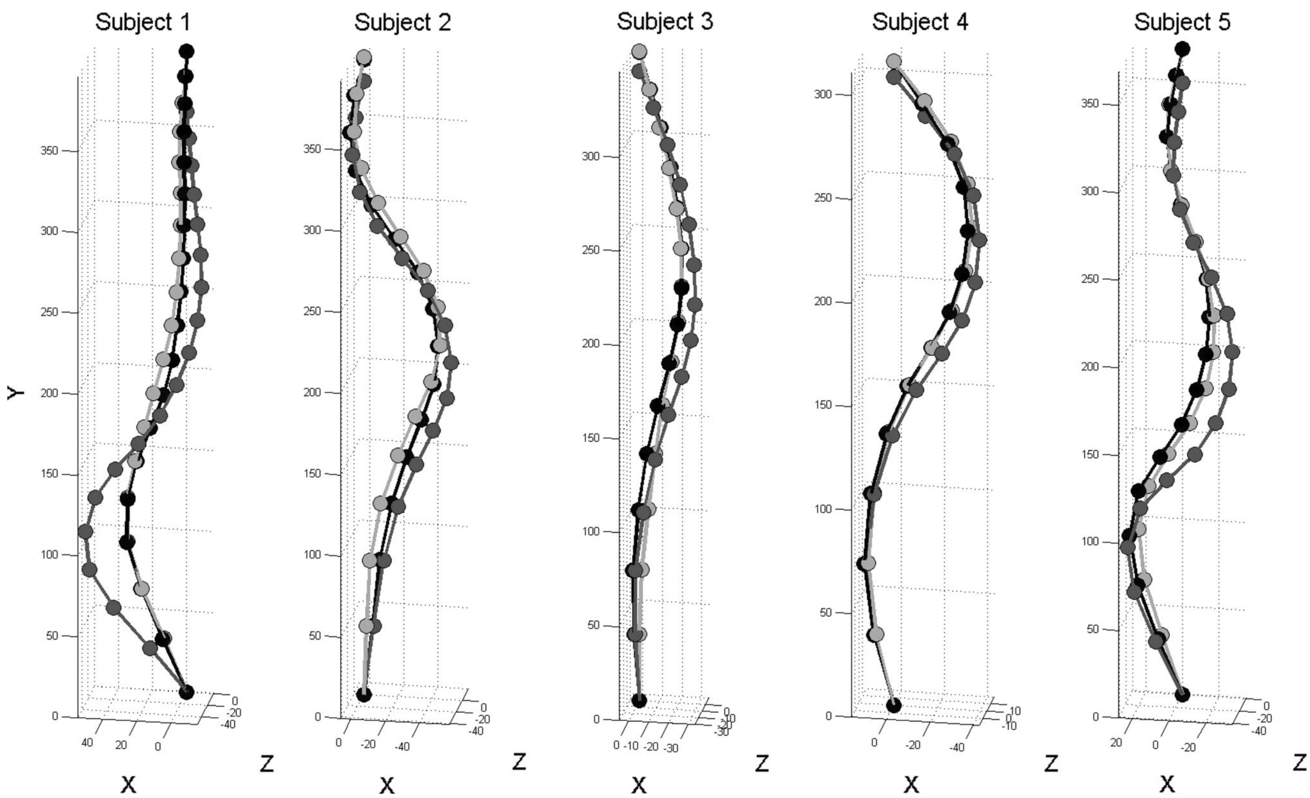

Fig. 3 Three-dimensional shape of the spinal midline in the experimental normal standing condition (*dark gray*), experimental suspension condition (*light gray*), and suspension condition calculated with the optimal set of mechanical parameters (*black*) for each subject. *Dots* represent the position of the IVDs

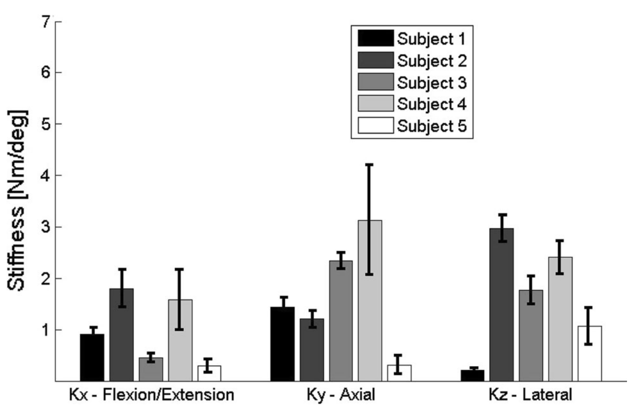

Fig. 4 Mean and standard deviation stiffness coefficients for each patient

Results indicate a large variation in spinal stiffness between patients. The stiffness of the most flexible spine was about four times lower than that of the stiffest spine. Analyses also show that the displacement of the spine induced by the suspension test is always larger than the precision of the reconstruction process. Therefore, the measured position of the IVDs was not significantly influenced by the intraobserver variability of the reconstruction process. This observation is confirmed by the robust convergence of the optimization to the optimal stiffness coefficients regardless of the error linked with the reconstruction process. These results indicate that individualized parameters can be quantified preoperatively and could be used to personalize the treatment strategy.

The mechanical properties observed in this study can be compared with the existing in vivo data (Fig. 5). In a suspension experiment, Lamarre et al. [7] quantified the stiffness of the apical vertebra based on anterio-posterior radiographs. Their results showed an axial stiffness larger than that reported in our study while the bending stiffness was lower. Contrary to our results, the standard deviation in the axial direction is much higher than that for the bending direction. The high differences between the axial and lateral stiffness might be explained by the fact that Lamarre's study considered only the apical vertebra, as compared to the whole spine in the present study. Since a single bidimensional radiograph was used for the calculations, no flexion stiffness was provided.

Petit et al. [9] calculated the stiffness of each of the motion segments from T1 to L5 and observed higher stiffness than we did. This difference can be explained by the use of side bending data without direct force measurement.

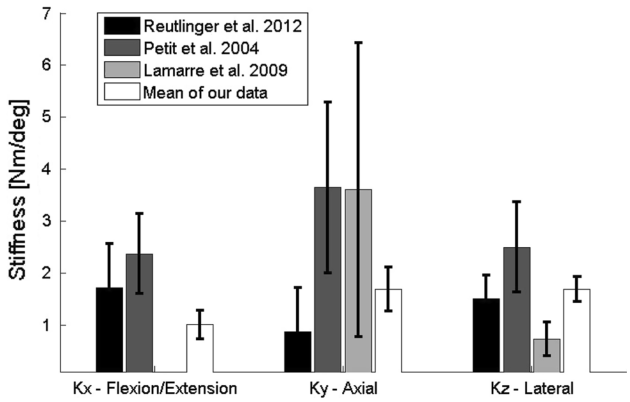

Fig. 5 Mean and standard deviation of the stiffness coefficients reported in previous studies. The *black bars* correspond to the intraoperative data collected by [14], *dark gray* corresponds to preoperative measurements made by Petit et al. [9], *light gray* represents the data acquired by Lamarre et al. [7] using a suspension technique. The *white bars* represent the mean results of our study

The only direct measurement of spinal stiffness in AIS patients was performed by Reutlinger et al. [14]. Even though only two patients were studied, the stiffness measured in lateral bending and flexion/extension is very similar to the data we obtained. Interestingly, they observed axial stiffness lower than the values obtained for the preoperative methods presented here. A possible explanation for the difference may be the very small axial rotation induced by the measurement method, which limits the accuracy of the stiffness measurement.

The stiffness coefficients we observed fall within one standard deviation and compare favorably with the intraoperative measurements of previous studies. However, comparison is hindered by the use of very different clinical tests as well as the different amount of information used to estimate the stiffness parameters. In addition, the mechanical properties of the spine are patient-specific. For this reason, a complete validation would have required intraoperative measurement of the patients who underwent our suspension test, which was beyond the scope of this study.

The lack of precise information about the axial rotation of the spine during suspension does not significantly affect our results since axial rotation is not taken into account during optimization. Another simplification of the numerical model is the uniform stiffness used for all segments along the spine. The stiffness of the spine very likely differs from one level to another. Optimizing IVD-specific stiffness would allow better understanding of how stiffness changes along the spine's curvature. However, the ability of the model to accurately represent the three-dimensional shape of the spine under load indicates that the approximation is not critical to the prediction of spinal deformation. It is important to realize that the stiffness parameters identified in this work correspond not only to the disc but also include all the connective tissues such as ligaments, muscles, or rib cage. As a result, the stiffness is not concentrated at single points in the discs, but distributed across several levels, which is an explanation why a single set of stiffness parameters is able to reproduce the spinal deformation. In addition, the regularization term only guides the optimization toward increasingly realistic results but does not contribute to the final results. Finally, due to the limited number of patients included in this preliminary study, it was also not possible to find a correlation between the mechanical stiffness of the spine and a morphologic classification system such as provided by Lenke.

This method was applied to five patients to quantify the 3D spine's stiffness. The approach can be the basis of a standardized test setup for the clinician to assess spinal stiffness in an accurate and reproducible manner. Since the shape and mechanical properties of the spine showed important variation across patients, this quantitative information is critical for the development of planning that considers patient-specific biomechanics. Such tools will become increasingly important due to the ever-increasing complexity of surgical instrumentation and procedures. Additionally, the method will also serve stiffness-adapted implants or surgical strategies and could be used to work out nonsurgical approaches, standards, and "safe zones" for daily clinical use.

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Conflict of interest None.

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