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Optimization method for 3D bracing correction of scoliosis using a finite element model

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Abstract Scoliosis is a complex three-dimensional deformity of the spine and rib cage frequently treated by brace. Although bracing produces significant correction in the frontal plane, it generally reduces the normal sagittal plane curvatures and has limited effect in the transverse plane. The goal of this study is to develop a new optimization approach using a finite element model of the spine and rib cage in order to find optimal correction patterns. The objective function to be minimized took account of coronal and sagittal offsets from a normal spine at the thoracic and lumbar apices as well as the rib hump. Two different optimization studies were performed using the finite element model, which was personalized to the geometry of 20 different scoliotic patients. The first study took into account only the thoracic deformity, while the second considered both the thoracic and lumbar deformities. The optimization produced an average of 56% and 51% reduction of the objective function respectively in the two studies. Optimal forces were mostly located on the convex side of the curve. This study demonstrates the feasibility of using an optimization approach with a finite element model of the trunk to analyze the biomechanics of bracing, and may be useful in the design of new and more effective braces.

Key words Scoliosis · Optimization · Biomechanical model · Brace · 3D correction

Introduction

Scoliosis is a complex three-dimensional (3D) deformation of the spine and rib cage that produces cosmetic asymmetries of the trunk, which represent the main complaints of patients [14]. Mild to moderate scoliosis generally is treated with an orthosis such as the Boston brace system to prevent curve progression and to reduce the deformity. Most studies agree that the Boston brace system produces significant corrections of Cobb angles in the frontal plane [12]. However, limited effects on the 3D aspect of the deformity have been reported [2, 10, 15], and results are generally different than expected from the forces applied to the trunk [6]. For instance, looking at the 3D aspects of scoliosis, the Boston brace system tends to decrease normal sagittal curvatures, has limited effect on the plane of maximum deformity, and has no significant derotational effect in the transverse plane [10]. These 3D effects achieved by the Boston brace system may be attributable to inappropriate application of forces to the torso, i.e., to incorrect personalization of the brace.

In this context, a different treatment approach has been proposed [6, 9], consisting of the application of forces to the anterior rib hump and on the convex side at the thoracic apical level, while preventing posterior displacement of the rib hump. This approach was simulated on a biomechanical model, and although better correction was obtained than with the Boston brace system, it was found that a systematic (rationalized) method was needed to personalize the treatment for each patient [5].

An optimization procedure was developed by Wynarsky and Schultz [17] to find the optimal active muscle and passive brace forces required to correct a right mid-thoracic scoliosis. The objective function took into account axial and frontal vertebral rotations, and offsets from a normal spine, although the scoliotic configuration had sagittal curvatures different from those normally observed in a scoliotic population. Furthermore, their objective function did not include the rib hump, which for the patients is in fact the most important cosmetic deformity [14].

This paper introduces an optimization approach using a personalized finite element model representing the current geometry of each patient. This innovative approach was developed to study the best loading patterns required to correct both the spine and the rib cage scoliotic deformities in 3D.

Methods

Biomechanical finite element model

A personalized model of the trunk previously used for the simulation of Boston brace treatment was utilized [4]. It contains 1411 nodes and 3011 elements representing the thoracic and lumbar vertebrae, intervertebral disks, ribs, sternum, intervertebral and intercostal ligaments, costovertebral and costotransverse joints, and zygapophyseal articulations [3]. Material properties were taken from published experimental data [8, 11]. Muscles were not considered in the model as they were not proven to have an active contribution in braces [16]. L5 was blocked to simulate the brace's restraint on the pelvis, and translation of T1 was constrained in the transverse plane to keep the trunk in a balanced posture (righting reflex) and to represent the counterforces generated by the upper section of the brace.

Personalization of the model

The initial geometry of the model was obtained using a multiview radiographic reconstruction technique [3, 7]. Twenty different models representing adolescents with idiopathic scoliosis (17 girls, 3 boys) were generated. The patients had typical right thoracic–left lumbar scoliotic curvatures justifying brace treatment. They were 12.8 ± 2.4 years old, with mean thoracic Cobb angles of 32.0 ± 5.9° and mean lumbar Cobb angles of 34.2 ± 7.3° (Table 1). The apical level of their thoracic and lumbar curves were respectively between T8 and T10 and between L1 and L3.

Optimal spine and rib cage

Optimal spine and rib cage, corresponding to those of a normal subject, were defined as follows. In the frontal plane, the symmetry of the spine was considered with respect to the spinal axis (defined by Stokes et al. [13]). For the sagittal plane, a normal profile was defined using 11 healthy subjects (24.5 ± 3.0 years old; 5 male, 6 female). In this plane, the distance of each vertebral body from the spinal axis was computed and normalized with respect to the spinal length. As for the rib cage, a back without rib hump was considered as optimal.

Objective function

The objective function included five terms representing 3D descriptors of the trunk deformity:

- The distances in the sagittal plane between the normal spine and the kyphosis and lordosis apices (X_K and X_L; Fig 1 a)
- The distances in the coronal plane between the vertebral body and the spinal axis at the thoracic and lumbar apices (Y_T and Y_L; Fig. 1 b)
- The rib hump (G) calculated as the distance between the most posterior points on the left and right ribs at the level of the greatest deformity (Fig. 1 c)

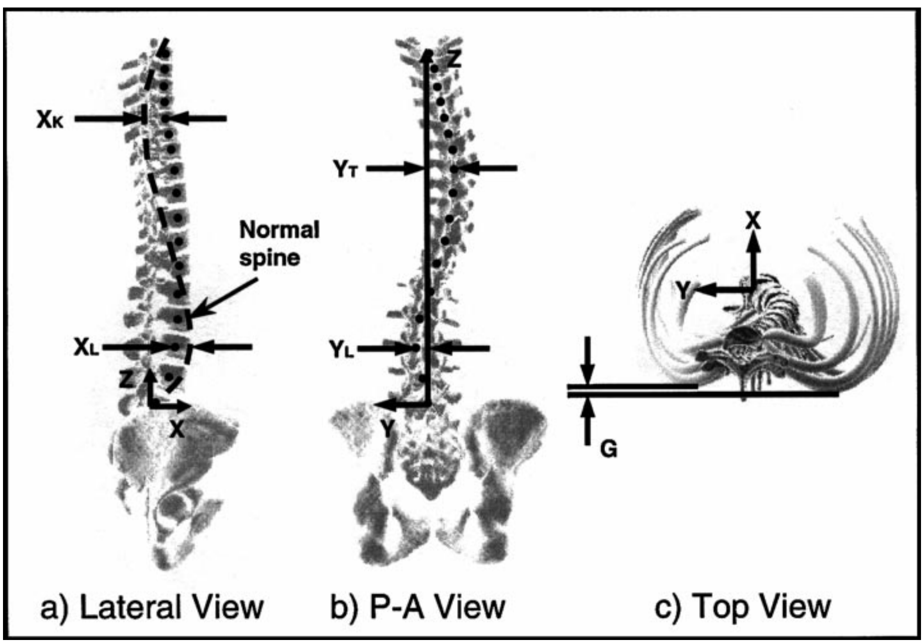

Fig. 1 The 3D descriptors, lateral, posteroanterior, and top views

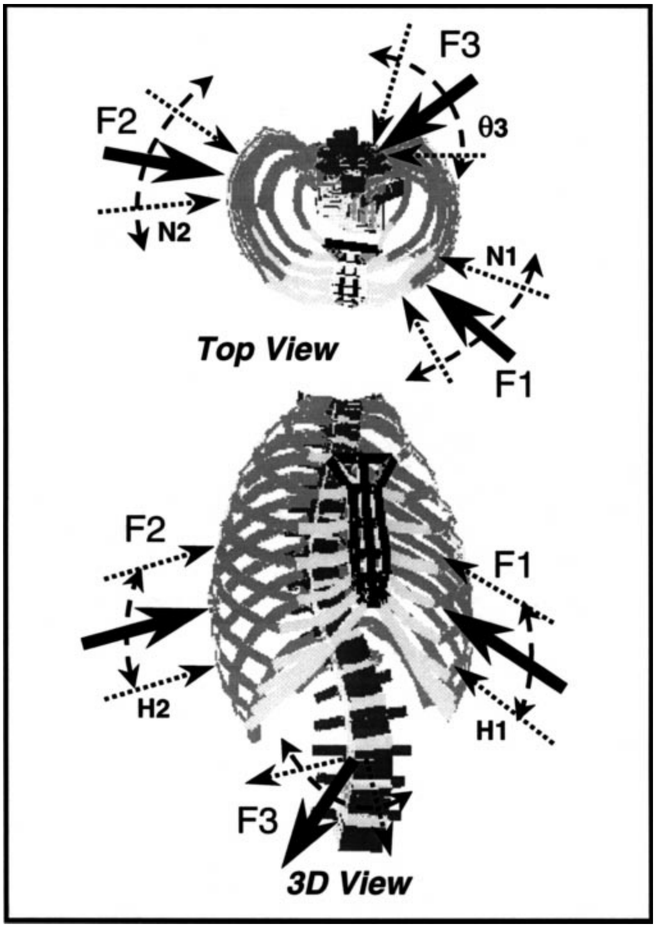

Fig. 2 Design variables used for the optimization studies. F3 and θ3 are only used for the second study

The objective function φ is calculated as the weighted sum of the square of these descriptors:

$$\varphi = W_1 \cdot (G)^2 + W_2 \cdot [(Y_T)^2 + (Y_L)^2] + W_3 \cdot [(X_K)^2 + (X_L)^2]$$

where W_1, W_2, and W_3 are the weightings assigned for the correction of descriptors in the transverse, coronal and sagittal planes respectively. These weightings can be changed [from 0 to 1 (100%)] proportionally to the importance given for each descriptor. In this study, W_1 and W_2 were set equal to unity (100%). They were chosen to focus equally on the correction of the frontal curvatures and the rib deformity. Most of the patients used in this study had a flat back, as is observed in many scoliotic subjects. A significant but less important weighting was assigned to sagittal indices, with W_3 set to 0.5. The objective function reaches a minimum at zero, which corresponds to the optimal situation defined before, corresponding to a normal subject.

Optimization variables

Two generic forces were applied on the thorax and a third one on the lumbar spine (Fig 2). The variables F_1, F_2, and F_3 represent the magnitude of each force, which could vary from 0 to 100 N, corresponding to the range of forces measured inside braces [2]. Forces F_1 and F_2 were always applied perpendicularly to the trunk. H_1 and H_2 specify the anatomic levels of forces F_1 and F_2, on either the fifth, seventh, or ninth rib. N_1 and N_2 relate to the nodes on the rib where forces F_1 and F_2 were applied. Finally, θ3 represents the angle in the transverse plane of F_3 with respect to the x direction, counterclockwise.

Convergence conditions

Conditions to stop the optimization process were defined as follows:

1. A total number of iterations exceeding 200.
2. A variation smaller than 1 N of any force between the last two iterations, or the last iteration and the best one. This variation corresponds to a 1% change in the allowed range. The best iteration is the one with the smallest objective function.
3. A variation smaller than 0.1 mm² of the objective function between the last two iterations or the last iteration and the best, which is considered a very small change.

Optimization procedure

Two different optimization studies were conducted using the Ansys V5.3 advanced zero order optimization algorithm (Ansys Inc., USA). The first one took into account only the thoracic deformity by applying two forces to the ribs (F_1 and F_2). The second study additionally used a third force (F_3), applied to the apical lumbar vertebra.

The first step of each optimization study consisted of performing an initial set of iterations using arbitrarily specified variables. The algorithm required eight iterations before starting the optimization process for the first study, and ten iterations for the second one. For the second step the software calculated values for each variable and performed a new iteration. After each iteration, the descriptors (G, X_K, X_L, Y_T, and Y_L) were calculated and the objective function was computed. If the descriptors were within a range of ± 15 mm, the iteration was then considered as a possible solution, and a convergence check was done. The program repeatedly calculated possible values for each variable and performed new iterations until convergence.

The resulting geometry of the optimization was compared to the initial and in brace configurations using the following clinical indices: frontal and sagittal Cobb angles, angle of the plane of maximum deformity measured with respect to the sagittal plane, axial rotation at the thoracic apical level, and maximum rib hump [10]. A paired Student's t-test was used to compare significant changes with respect to the initial configuration.

Results

Convergence was obtained for 60% of the patients in the first study and for 35% in the second one using the specified criteria. It took an average of 12 h and 125 iterations to converge using a SGI R-8000 workstation. The best iteration for runs that stopped because they reached 200 iterations reduced the objective function by an average of 49%. They were also included in the results because they were converging slowly and would eventually reach an optimum.

For all the patients, the objective functions were reduced by an average of 56% and 51% in the first and the second studies respectively. The descriptors G, X_K, X_L, Y_T, and Y_L were respectively reduced by an average of 43%, 51%, 3%, 10%, and 17% in the first study (two thoracic

Table 1 Clinical indices (mean ± SD) evaluated on the trunk geometry of the 20 patients

Index	Initial geometry	Patient in Boston brace	Best results for the first study[a]	Best results for the second study[b]
Cobb angles in the frontal plane				
Thoracic	32° ± 6°	30° ± 6°*	27° ± 6°*	27° ± 7°*
Lumbar	34° ± 7°	31° ± 7° *	34° ± 7°	33° ± 6°*
Angles in the sagittal plane				
Thoracic kyphosis	33° ± 12°	28° ± 9°*	34° ± 10°	37° ± 11°*
Lumbar lordosis	38° ± 12°	30° ± 13°*	36° ± 12°*	32° ± 11°*
Angles of the plane of maximum deformity with respect to the sagittal plane				
Thoracic	38° ± 20°	42° ± 23°	33° ± 20°*	36° ± 19°
Lumbar	39° ± 19°	48° ± 25°*	38° ± 21°	45° ± 23°*
Angles in the transverse plane				
Axial rotation at thoracic apex	–6° ± 6°	–5° ± 6°	–5° ± 8°	–6° ± 8°
Maximum rib hump angle	–7° ± 4°	–6° ± 6°	–5° ± 4°*	–5° ± 5°*

* Significant change from the initial configuration, paired Student's *t*-test, $P < 0.05$
[a] Two thoracic forces
[b] Two thoracic and one lumbar force

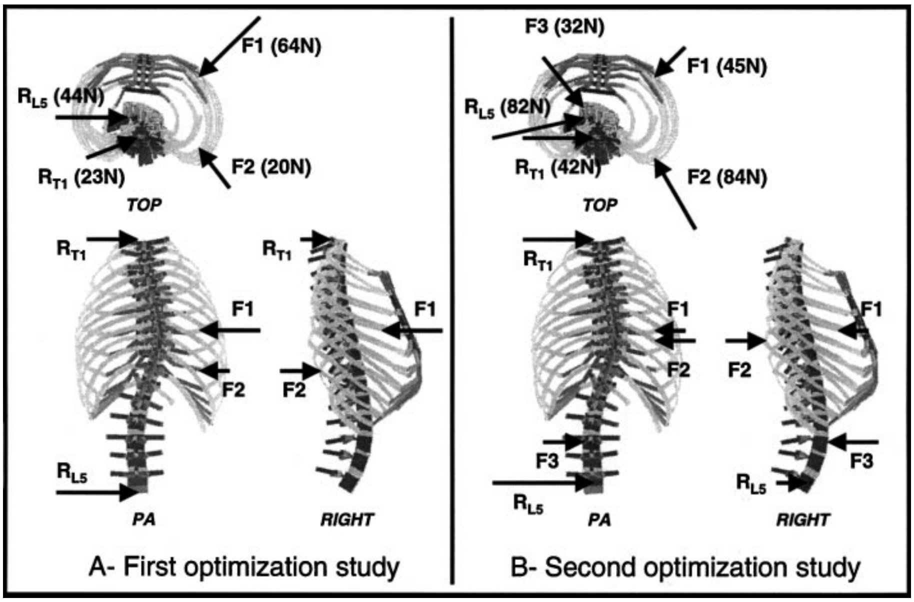

Fig. 3 The optimal forces (F_1, F_2, F_3) and the reaction forces (R_{T1}, R_{L5}) for the two optimization studies on a typical patient. This patient had resulting forces close to the average values reported

forces), and by 30%, 48%, 0.1%, 40%, and 45% in the second study (three thoracic and lumbar forces). The clinical indices computed were mostly improved on the resulting geometry, except for the axial rotation at the thoracic apex and the orientation of both planes of maximum deformity. To show the average trend, Table 1 lists the means and standard deviations of clinical indices of the 20 patients before their treatment, with their Boston brace, and the best iteration of the two optimization studies. As can seen by the rather large standard deviation, the clinical indices vary even though the patients had a similar right thoracic–left lumbar scoliosis.

The best forces for the first optimization study were as follows: F1 was mostly located on the anterior convex side of the fifth rib with an average amplitude of 63 N. F_2 was located posterolaterally on the convex side of the ninth rib with an average amplitude of 34 N. The reaction forces at T1 due to the boundary conditions had average magnitudes of 0 N and –19 N in the *x* and *y* directions and 4 N, –33 N and 0 N in the *x*, *y*, and *z* directions at L5.

For the second optimization study, F_1 was mostly located posterolaterally on the convex side of the fifth rib, with an amplitude of 64 N. F_2 was located on the anterior convex side of the seventh rib, with an amplitude of 34 N. The lumbar force (F_3) was oriented at about 190° with respect to the *x* axis, with an amplitude of 57 N. The reaction forces had amplitudes of 8 N and –16 N in the *x* and *y* directions at T1, and 31 N, –24 N and 3 N in the *x*, *y*, and *z* directions at L5. Typical results for a given patient are illustrated in Fig. 3.

Discussion

The results found in this study are similar to results achieved by the Boston brace system if only the frontal plane deformity is considered. 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. Even if the correction is small (the simulation gives the effect instantaneously), Andriacchi et al. [1] showed that short-term correction has the same characteristics as correction resulting from long-term treatment.

Results generally are in agreement with current bracing theories using three point forces per curve. The main difference in this study is the lumbar force, which is mainly oriented posteriorly. Wynarsky and Schultz [17] also reported optimal corrective forces located on the convex side of the curve, but mostly oriented anteriorly. This difference is probably due to their initial patient configuration, which did not have the typical scoliotic flat sagittal curvatures. Moreover, the rib hump deformity was not taken into account in their study.

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. For instance, the reaction force at T1 may represent the combined effect of the counterweight needed to oppose the thoracic forces (F_1 and F_2) and the possible forces required to keep the trunk balanced. This study also questions the efficacy of using only three point forces applied on the torso to reduce the multiple degrees of freedom components of the scoliotic deformities. Other factors intrinsic to the finite element model, such as the personalization of the mechanical properties, may limit the current results in regard to the direct application of these results to real brace treatment.

The arbitrary choice of the weightings of the objective function was quite appropriate because the reductions of the descriptors were of similar magnitude. However, these weightings can be changed as desired by the physician aiming to focus on a specific correction for each patient's deformities. The range allowed for the five descriptors was arbitrarily set at ± 15 mm because this was slightly lower than the average value of all descriptors and seemed acceptable for a residual deformity. If the range is set to be smaller, fewer simulations will converge, but the ones that do will obtain greater correction. By contrast, allowing a larger range will produce poorer correction. Both the weightings of terms 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.

This study demonstrates the feasibility of an optimization approach to finding effective corrective forces in the 3D treatment of idiopathic scoliosis on the basis of the individual spinal and rib cage geometry of patients. Other studies could now be performed to analyze different 3D descriptors and their effect on the clinical indices, especially the ones that were not improved (i.e., lumbar Cobb angle, axial rotation, and planes of maximum deformity). In the long term, this approach could help physicians and orthotists to optimize the placement of pads and cushions in braces to produce better correction and improve bracing in idiopathic scoliosis.

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