OPTIMIZATION OF SKELETAL CONFIGURATION: STUDIES OF SCOLIOSIS CORRECTION BIOMECHANICS

GEORGE T. WYNARSKY and ALBERT B. SCHULTZ*

Department of Mechanical Engineering and Applied Mechanics, University of Michigan, Ann Arbor, MI 48109-2125, U.S.A.

Abstract—A scheme for optimizing configurations in models of skeletal structures is presented. Use of the scheme is illustrated through determination of biomechanically optimal correction of a right-thoracic scoliosis by passive brace and active muscle forces. The locations and magnitudes of the passive brace forces, and the trunk muscle groups and their corresponding contraction intensity magnitudes that would optimally correct the geometric deformities of the spine were determined. The results suggest that, from a biomechanical viewpoint, both brace and muscle forces are capable of substantial correction of a model thoracic scoliosis. However, comparison of model results with long-term clinical results suggests that even under optimal conditions it is unlikely that scoliosis can be fully corrected by passive brace forces or active muscle contractions.

INTRODUCTION

Bracing remains the most common method of non-operative treatment of idiopathic scoliosis (Benson, 1987; Lonstein and Winter, 1988; Kehl and Morrissy, 1988; Keller, 1989; Gunnoe, 1990). Two current views as to how bracing corrects scoliosis are that (1) correction is produced by passive response to the forces applied by the brace; or (2) brace-wearing engenders active contractions of trunk muscles, and those muscle forces correct the scoliosis. Moreover, some cases of scoliosis are treated conservatively by electrostimulation of muscles on the convex side of the scoliotic curve (Bobechko *et al.*, 1979; Axelgaard and Brown, 1983; Axelgaard *et al.*, 1983; Brown *et al.*, 1984). We undertook studies to examine what passive brace and active muscle contraction forces would be required to produce biomechanically optimal correction of a spine with a scoliosis curve. These studies examined (1) to what extent a scoliosis can be corrected when corrective forces are optimally selected; and (2) what sets of passive brace or active muscle forces produce those optimal corrections.

A number of structural models suitable for studies of skeletal configurations exist, generally consisting of a collection of rigid bodies to represent the bony elements of the skeleton, interconnected by a series of deformable elements to represent the soft tissues. Forces and moments are applied to the models, and their resulting displacements and rotations are computed. Such models have been used to study the biomechanics of correction of scoliosis, for example by Harrington instrumentation (Schultz and Hirsch, 1973a), by Milwaukee brace (Andriacchi *et al.*, 1976), and by muscle stimulation (Schultz *et al.*, 1981). Beam-column structural models have been used to analyse a

functional scoliosis (Lindbeck, 1985) and curve progression and orthotic stabilization in idiopathic scoliosis (Patwardhan *et al.*, 1986). Ghista *et al.* (1988) used a two-dimensional finite element model of the spine to determine the axial distraction and lateral traction forces required for optimal correction of scoliosis by spinal instrumentation. These models have not so far been used to examine optimal correction by bracing.

The primary purpose of this paper is to describe a computational technique to optimize configurations of rigid-body deformable-element models of the skeleton. The technique has general applicability, but its use will be illustrated here through finding sets of brace and trunk muscle forces that produce biomechanically optimal, non-surgical correction of a scoliosis.

METHODS

The effects of passive brace forces and active muscle contractions were simulated in an existing computer model of the spine and rib cage (Belytschko *et al.*, 1973; Schultz *et al.*, 1973b; Andriacchi *et al.*, 1974; Schultz *et al.*, 1974a, b; Takashima *et al.*, 1979). This model incorporated 39 rigid bodies to represent the vertebrae of the thoracic and lumbar spine, the sacrum, rib pairs 1–10, and the sternum. These 39 rigid bodies were interconnected by 236 spring and 59 beam deformable elements to represent the soft tissues of the motion segments, the costal cartilages and other ligamentous tissues of the trunk. Each spring element was characterized by one axial stiffness and each beam element by six stiffnesses; one each for axial deformation, torsional rotation, lateral bending, lateral shear, antero-posterior bending and anteroposterior shear. Skeletal geometries and soft-tissue property data were those from the background papers cited. Trunk muscles capable of applying significant forces to the spine or rib cage were represented by 78 model muscle slips

(Table 1, column 1), using the techniques of Taka-shima *et al.* (1979) with minor modifications to their data (revised locations of muscle origins, insertions, and lines-of-action). In all computations, the sacrum was fully fixed while the vertebrae, ribs, and sternum were free to displace and rotate within prescribed bounds. Since the initial scoliotic spine configuration was taken as an equilibrium configuration, the model spine was not axially loaded. The balancing of body weight by muscle and connective tissue actions would superpose on the results to be described, since the force–deformation behavior of the model used was assumed to be linear.

Table 1. Left column: muscle groups included in model with corresponding cross-sectional areas. Right column: muscle slips and corresponding contraction intensity magnitudes selected by the optimization program (objective function *A*) for correction of the mid-thoracic scoliosis

Muscle name	Area (cm^2)	Contraction intensity (N cm^{-2})
Midline erector L	13.2	
Midline erector H	5.6	
Lt. thoracic erector H	6.9	10.0
Lt. thoracic erector M	6.0	
Lt. lumbar–thoracic erector	15.0	
Rt. thoracic erector H	6.9	
Rt. thoracic erector M	6.0	
Rt. lumbar–thoracic erector	15.0	10.0
Lt. rectus abdominis	3.9	2.1
Rt. rectus abdominis	3.9	
Lt. external abdominal oblique H	2.6	8.2
Lt. external abdominal oblique M	3.2	10.0
Lt. external abdominal oblique L	3.8	
Rt. external abdominal oblique H	2.6	10.0
Rt. external abdominal oblique M	3.2	
Rt. external abdominal oblique L	3.8	10.0
Lt. internal abdominal oblique H	3.9	4.3
Lt. internal abdominal oblique M	3.2	10.0
Lt. internal abdominal oblique L	3.8	
Rt. internal abdominal oblique H	3.9	4.2
Rt. internal abdominal oblique M	3.2	
Rt. internal abdominal oblique L	3.8	
Lt. pectoral major H	7.4	
Lt. pectoral major L	6.5	10.0
Lt. pectoral minor	1.6	10.0
Rt. pectoral major H	7.4	
Rt. pectoral major L	6.5	
Rt. pectoral minor	1.6	
Lt. psoas major	11.6	
Rt. psoas major	11.6	
Lt. trapezius H	2.4	10.0
Lt. trapezius M	2.4	10.0
Lt. trapezius L	1.9	9.1
Rt. trapezius H	2.4	
Rt. trapezius M	2.4	
Rt. trapezius L	1.9	

Table 1. (*Contd.*)

Muscle name	Area (cm^2)	Contraction intensity (N cm^{-2})
Lt. latissimus H	1.5	10.0
Lt. latissimus M	2.7	
Lt. latissimus L	3.5	
Rt. latissimus H	1.5	
Rt. latissimus M	2.7	
Rt. latissimus L	3.5	10.0
Lt. serratus H	3.7	0.3
Lt. serratus M	7.9	9.8
Lt. serratus L	4.9	10.0
Rt. serratus H	3.7	
Rt. serratus M	7.9	10.0
Rt. serratus L	4.9	10.0
Lt. intercostals H	8.6	10.0
Lt. intercostals M	8.4	
Lt. intercostals L	6.3	
Rt. intercostals H	8.6	
Rt. intercostals M	8.4	10.0
Rt. intercostals L	6.3	10.0
Lt. transverse abdominal H	3.4	10.0
Lt. transverse abdominal M	3.4	3.0
Lt. transverse abdominal L	3.4	0.6
Rt. transverse abdominal H	3.4	
Rt. transverse abdominal M	3.4	10.0
Rt. transverse abdominal L	3.4	10.0
Lt. quadratus lumborum	2.6	10.0
Rt. quadratus lumborum	2.6	

Lt. = left; Rt. = right; H = high; M = mid, L = low.

The basic structural computational technique employed was the direct-stiffness method. Model behavior is governed by the set of simultaneous linear algebraic equations

$$[K]\{D\} = \{F\} \qquad (1)$$

where $[K]$ is the assembled global stiffness matrix that represents the geometric and material properties of the model, $\{D\}$ is a column vector of the global three displacements and three rotations of each model rigid body, and $\{F\}$ is a column vector of the global three force and three moment components applied to each rigid body. The number of equations, N, is six times the number of rigid bodies included in the model, or 234 here.

In this study, the N global components of the applied loads $\{F\}$ consisted of those due to N_b brace forces and N_m muscle contraction intensities, so that equation (1) took the form

$$[K]\{D\} = [CB]\{F_b\} + [CM]\{F_m\} \qquad (2)$$

where $[CB]$ is a prescribed $N \times N_b$ matrix specifying the locations and directions of any applied brace forces, $\{F_b\}$ the vector of brace force magnitudes, $[CM]$ is a prescribed $N \times N_m$ matrix specifying muscle

lines-of-action and cross-sectional areas, and $\{F_m\}$ the vector of muscle contraction intensities. Since the brace forces $\{F_b\}$, the muscle contraction intensities $\{F_m\}$, and the translations and rotations of the rigid bodies $\{D\}$ were all unknown, equation (2) was highly indeterminate. Solutions were therefore obtained by optimization techniques.

The optimization problem took the general form

$$\text{minimize an objective function } Q \qquad (3a)$$

subject to

$$[K \mid -CB \mid -CM] \left\{ \begin{array}{c} D \\ \hline F_b \\ \hline F_m \end{array} \right\} = \{0\} \qquad (3b)$$

and lower and upper bounds on the force and displacement variables

$$l_i \leqslant x_i \leqslant u_i \qquad (3c)$$

where equation (3b) is a rearrangement of equation (2).

Brace force vector $\{F_b\}$

The forces exerted by a Boston brace were represented by a network of $N_b = 196$ potential point loads, evenly distributed over the patient–orthosis contact surface. Point loads were located at the centers of $10.0\ \text{cm}^2$ contact areas. Since braces seldom exert significant shear forces, each point load was assumed to act normal to the body surface. Brace forces were assumed to be transmitted directly to the rib cage in the thoracic region and directly to the vertebrae in the lumbar region. Force transmission through soft tissue (lumbar region) was also considered by assuming that a fraction of the external force was transmitted to the vertebrae. Since brace force locations and directions were prescribed, only the 196 brace force magnitudes were unknown.

Muscle contraction intensity $\{F_m\}$

Following Takashima et al. (1979), each muscle slip was modelled using a series of secondary nodes run from origin to insertion. Secondary nodes not attached directly to a model rigid body were assumed to be rigidly linked to the vertebra closest to their superior–inferior level. Muscle slip contraction was represented by the application of equal and opposite forces to each node pair along the line of action. With muscle lines-of-action and cross-sectional areas prescribed, only the $N_m = 78$ contraction intensity magnitudes were unknown.

Objective functions Q

The objective functions to be minimized were constructed using the geometric parameters that describe the spine deformity. The results of initial computational studies (Wynarsky, 1988) led to the exclusion of force parameters in the objective functions. Although each vertebral body had six degrees of freedom (three displacements and three rotations), only four of these
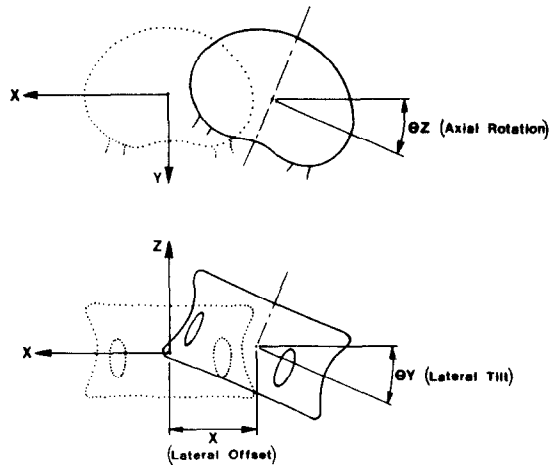

Fig. 1. The geometric components of scoliosis. Superior and posterior views of a normal (dotted) and displaced vertebra.

were included in the objective function. The geometric parameters were the lateral and anteroposterior global coordinates of the primary nodes of a model vertebra, represented by X and Y, and the lateral and axial global orientations of that body, represented by θ_y and θ_z. Vertebrae included in a scoliosis curve would have a lateral offset X, a lateral tilt θ_y (related to the Cobb angle of the curve), and an axial rotation θ_z (Fig. 1), while those in a structurally normal spine would have small or zero values for these variables. The Y terms were included in the objective function to prevent displacements in the sagittal plane rather than displacements to correct the scoliosis. Geometric parameters for the rigid bodies other than the vertebrae were not included in the objective functions.

The vector of displacements in equation (3b) does not involve three out of the four geometric parameters directly. Rather, it involves small changes in these parameters, where the parameters, their initial values, and the changes that occur in them are given by

$$X = X_0 + dX \qquad (4a)$$

$$\theta_y = \theta_{y0} + d\theta_y \qquad (4b)$$

$$\theta_z = \theta_{z0} + d\theta_z. \qquad (4c)$$

Since the Y coordinate (sagittal plane) differences between a normal model spine and a thoracic scoliosis model spine were small, the anteroposterior displacements were dealt with directly in the form of dY. Thus, dX, dY, $d\theta_y$ and $d\theta_z$ were the elements of the displacement vector for each vertebra in the model. Those displacements were to take on values such that the final vertebral configuration in terms of parameters X, Y (or dY), θ_y, and θ_z was optimized. The initial values X_0, θ_{y0}, and θ_{z0} were prescribed based on the original spine configuration for the specific configuration of scoliosis studied.

Because the geometric variables X, Y, θ_y, and θ_z could take on either positive or negative values, the sum of the squares of these variables was minimized.

This made the objective function a convex function subject to linear inequality constraints, which guaranteed that a global optimum would be found (Murtagh and Saunders, 1983).

To eliminate dimensional dependence, the sums of the squares of the X, $d\theta_y$ and $d\theta_z$ were divided by the sum of the squares of the initial vertebral body offsets or rotations [equation (5)]. Values of dY (A–P displacements) remained small, so that normalization of these terms was not necessary. The sums were also weighted, as will subsequently be detailed.

In summary, the objective functions minimized the weighted sums of the squares of the major geometric components of the scoliosis deformities:

$$Q = C_1 \frac{\sum_{i=1}^{N_v} (X_0 + dX)_i^2}{\sum_{i=1}^{N_v} (X_0)_i^2} + C_2 \sum_{i=1}^{N_v} dY_i^2$$

$$+ C_3 \frac{\sum_{i=1}^{N_v} (\theta_{y_0} + d\theta_y)_i^2}{\sum_{i=1}^{N_v} (\theta_{y_0})_i^2} + C_4 \frac{\sum_{i=1}^{N_v} (\theta_{z_0} + d\theta_z)_i^2}{\sum_{i=1}^{N_v} (\theta_{z_0})_i^2} \quad (5)$$

where X is the lateral coordinates, dY the anteroposterior displacements, θ_y the lateral tilts, θ_z the axial rotations, N_v the total number of vertebral bodies, and C_1, C_2, C_3, and C_4 are weighting coefficients.

Selection of weighting coefficients

Ideal correction of a scoliosis curve in an upright spine would bring all lateral offsets, lateral tilts, and axial rotations to zero and keep anteroposterior displacements reasonably small. Within the constraints used in these studies, ideal correction cannot be achieved. When correction cannot be ideal, what precisely constitutes optimal correction? Lateral tilts, lateral offsets and axial rotations clearly should be made as small as possible. But, is it more important to correct lateral tilts than axial rotations? So, to at least a small degree, the choice of objective function weightings is a matter of clinical judgement. Once those weightings are chosen, the computational techniques used here insure that the solutions obtained will be globally optimal.

To choose sets of weighting coefficients that would produce clinically pleasing spine final configurations, analyses were conducted in which each weighting coefficient was systematically varied over values from 1 to 100 (Wynarsky, 1988). The resulting configurations were visually judged. The most pleasing overall corrections were attained by weighting X modestly, Y minimally, and θ_y heavily. Changes in θ_z weighting had little visual impact, so a wider range of its values was ultimately explored. Based on the results, three sets of weightings [C_1, C_2, C_3 and C_4 in equation (5)] were examined in detail:

$$Q_A = 10 \ f(X) + 1 \ f(Y) + 50 \ f(\theta_y) + 2 \ f(\theta_z) \quad (6a)$$

$$Q_B = 10 \ f(X) + 1 \ f(Y) + 50 \ f(\theta_y) + 10 \ f(\theta_z) \quad (6b)$$

$$Q_C = 20 \ f(X) + 1 \ f(Y) + 90 \ f(\theta_y) + 50 \ f(\theta_z) \quad (6c)$$

where $f(X)$, $f(Y)$, $f(\theta_y)$, and $f(\theta_z)$ represent the summation terms in equation (5).

Successively greater emphasis was placed on decreasing the final axial rotations in weightings A, B, and C, respectively.

Bounds on variables

The maximum initial lateral offset in any case of scoliosis studied was 4.0 cm, and the maximum initial vertebral rotation was 25°. Based upon these numbers, rigid body translations were arbitrarily limited to 10 cm and rigid body rotations to 90°. These bounds were not approached.

Based upon previously established skin pressure tolerances (Blount and Moe, 1973; Bunch, 1975; Bader and Gant, 1985) and brace–patient interface pressures measured by Chase et al. (1989) and Wynarsky (1988), brace point force magnitudes were kept within 10 N each, corresponding to pressures of 10 kPa over the 10 cm² areas. To account for non-maximal brace pressures, brace point force bounds of 2.5, 5.0, 7.5, and 10.0 N were explored.

Maximum voluntary muscle contraction intensities have been estimated to range from 400 to 1000 kPa (Ikai and Fukunaga, 1968) in healthy adult males and to be approximately 200 kPa in healthy adolescent females (Portillo et al., 1982). Accordingly, contraction intensities were kept within 100 kPa. Muscle intensity bounds of 25, 50, 75, and 100 kPa were explored.

Optimization routine

Optimizations were carried out using the modular in-core non-linear optimization system (MINOS) of Murtagh and Saunders (1978, 1983). This is a large-scale optimization system for the solution of sparse linear and non-linear programs. For a linearly constrained non-linear (objective function) problem, MINOS uses a reduced-gradient algorithm in conjunction with a quasi-Newton algorithm based on the primal simplex method. To examine the effects on correction of only passive brace forces, the routine found the brace–force locations and corresponding magnitudes which yielded optimal correction in light of the selected weighting coefficients and the bounds on the variables. To examine the effects on correction of only active muscle forces, the routine found the muscles and their corresponding contraction intensities which yielded optimal correction. Combinations of both passive brace and active muscle forces within the prescribed bounds were also explored.

Initial scoliosis morphology

The computational method described here could be applied to any initial spine configuration. The initial morphology of the scoliosis curve for which results will be reported here was that of a typical right midthoracic curve (MT spine). This type of curve has the highest incidence (Moe and Kettleson, 1970). The initial curve configuration (Fig. 2) had its apex at the
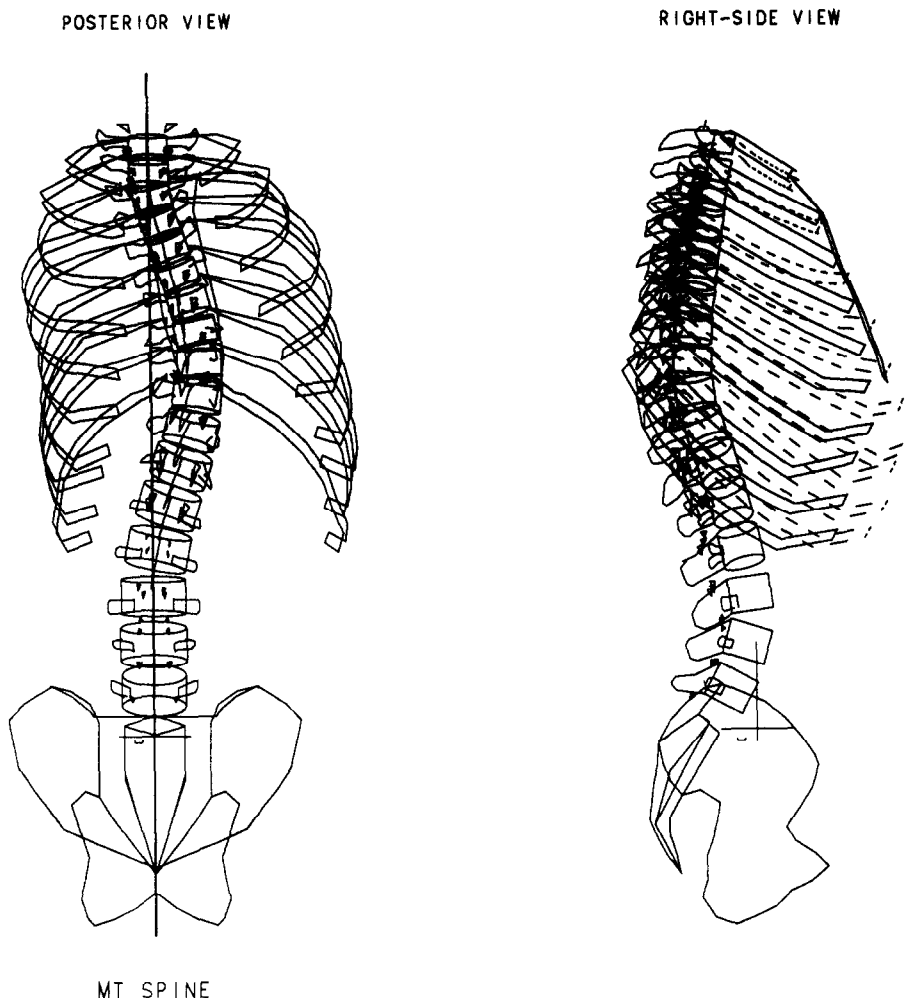
POSTERIOR VIEW

RIGHT-SIDE VIEW

MT SPINE

Fig. 2. Mid-thoracic scoliosis (MT Spine). Posterior and right-side view of initial configuration. Curve data: Cobb angle = 42°; lateral offset of T9 = −4.0 cm; axial rotation of T9 = −20°.

T9 level, a Cobb angle of 42°, an apical lateral offset of 4.0 cm to the right, and an apical axial rotation of 20° clockwise in a superior view.

Analysis

The spine configuration changes resulting from the applied brace and muscle forces were evaluated in terms of the final Cobb angle and lateral offset of the apical vertebrae of the original curve and those for any secondary curves which developed. The final axial rotations of the apical vertebra of the original curve and the two adjacent vertebrae were also noted. Moreover, the recruitment patterns of brace and muscle forces and their corresponding magnitudes were noted.

RESULTS

Correction by brace forces

Application of passive brace forces had a substantial corrective effect on the Cobb angle and lateral alignment (Table 2, Fig. 3), even at non-maximal upper bounds. For example, with the 5.0 N bound, the

initial Cobb angle was reduced to 27° and the lateral offset of the apical vertebra was reduced to −1.2 cm (objective A). However, under all circumstances examined, brace forces caused small left lumbar and left high thoracic secondary curves, and produced only modest spine de-rotations. The Cobb angle and apical lateral offset were significantly reduced independent of the objective function (A, B, or C), but the axial de-rotation was dependent on the objective function; progressively greater axial de-rotation was achieved by using objective functions Q_B and Q_C.

The locations of brace forces to optimally correct the model thoracic scoliosis are shown in Fig. 4. The optimal distribution of brace forces was: (1) primary forces on the convex side of the curve on ribs 7–10; (2) counter forces on the concave side of the curve on ribs 5 and 6; and (3) counterforces on the concave side of the curve at the L2–L4 vertebral levels.

Correction by muscle forces

Use of active muscle forces also had a substantial corrective effect on the Cobb angle and lateral alignment (Table 3, Fig. 5), again, even at non-maximal

Table 2. Final spine configurations resulting from correction by brace forces for increasing upper force bounds and objective functions A, B, C. Initial values are denoted by parentheses. Lum = lumbar, HTh = high-thoracic

Force (N)	Cobb angles (degrees)			Lateral offsets (cm)			Axial rotations (degrees)		
	Lum.	Primary	HTh	L2	T9	T2	T8	T9	T10
	$(0)^*$	$(42)^*$	$(0)^*$	$(0.0)^*$	$(-4.0)^*$	$(0.0)^*$	$(17)^*$	$(20)^*$	$(15)^*$
Objective A									
2.5	22	34	22	1.1	−2.1	1.0	14	18	13
5.0	22	27	24	1.3	−1.2	1.1	12	16	12
7.5	20	23	24	1.1	−0.7	1.2	10	15	12
10.0	19	21	24	0.9	−0.6	1.2	10	15	12
Objective B									
2.5	22	34	22	1.1	−2.2	1.1	11	15	11
5.0	22	28	24	1.5	−1.2	1.2	9	13	10
7.5	21	24	24	1.3	−0.8	1.2	8	13	10
10.0	20	22	23	1.2	−0.7	1.2	8	13	10
Objective C									
2.5	22	35	22	1.0	−2.2	1.2	10	14	10
5.0	22	28	23	1.5	−1.3	1.1	8	13	9
7.5	21	24	24	1.4	−0.9	1.1	7	12	9
10.0	21	23	24	1.3	−0.8	1.1	7	12	9

*Initial spine configuration.
Objective $A = 10f(X) + 1f(Y) + 50f(\theta_Y) + 2f(\theta_Z)$.
Objective $B = 10f(X) + 1f(Y) + 50f(\theta_Y) + 10f(\theta_Z)$.
Objective $C = 20f(X) + 1f(Y) + 90f(\theta_Y) + 50f(\theta_Z)$.

upper bounds. For example, with the 25 kPa bound, the initial Cobb angle was reduced to 24° and the lateral offset of the apical vertebra to − 1.0 cm (objective A). Moderate de-rotations were produced, but use of muscle forces still created small left lumbar and left high thoracic secondary curves. The Cobb angle and

apical lateral offset again were significantly reduced independent of the objective function, and again, the axial de-rotation was dependent on the objective function weighting used.

The muscle groups and corresponding contraction intensities selected for optimal correction of the model
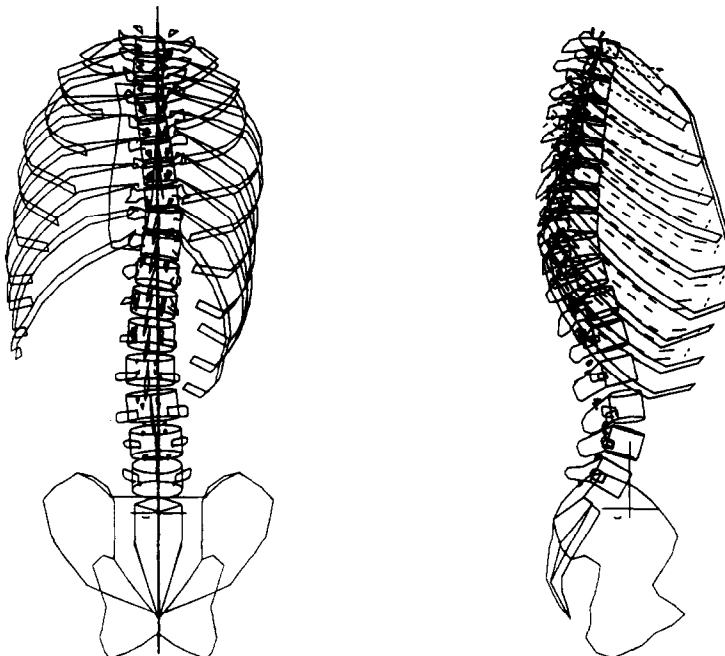

Fig. 3. Correction of model thoracic scoliosis by brace forces with an upper force bound of 10.0 N and using objective function A. See Table 2 for quantitative results.

LEFT SIDE   POSTERIOR   RIGHT SIDE
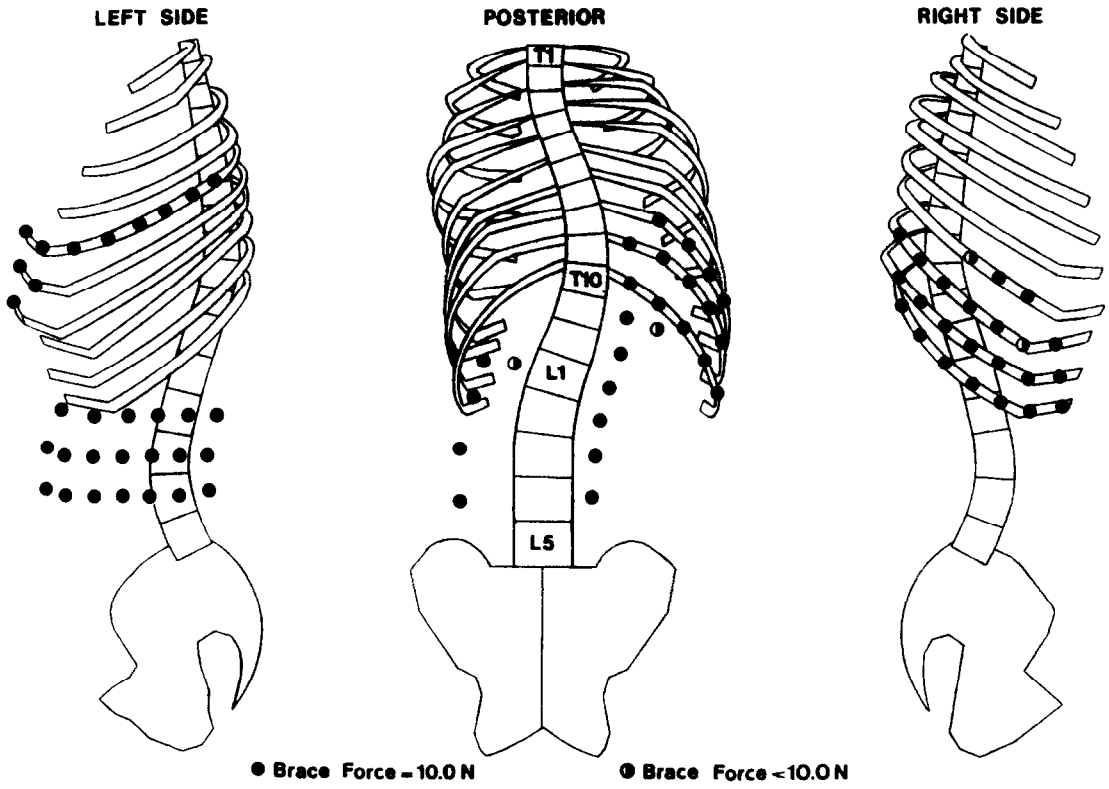
● Brace Force = 10.0 N   ◐ Brace Force < 10.0 N

Fig. 4. Optimal distribution of brace forces for correction of the thoracic scoliosis. Each dot represents a 10.0 N point force exerted by the brace (objective function A).
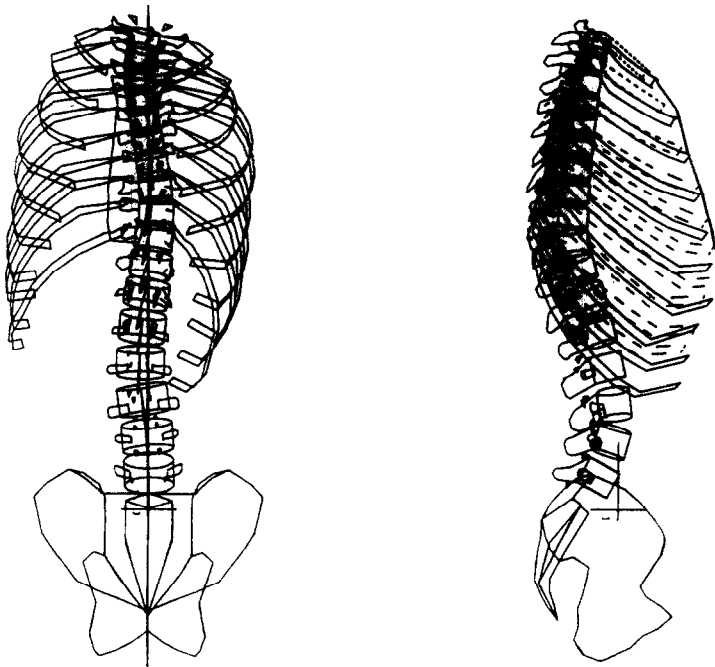
Fig. 5. Correction of the model thoracic scoliosis by muscle forces with an upper contraction intensity bound of 100 kPa and using objective function A. See Table 3 for quantitative results.

Table 3. Final spine configurations resulting from correction by muscle forces for increasing upper bounds on contraction intensity and objective functions A, B, and C. Initial values denoted by parentheses. Lum = lumbar, HTh = high-thoracic

Intensity (N cm^{-2})	Cobb angles (degrees)			Lateral offsets (cm)			Axial rotations (degrees)		
	Lum.	Primary	HTh	L2	T9	T2	T8	T9	T10
	(0)*	(42)*	(0)*	(0.0)*	(−4.0)*	(0.0)*	(17)*	(20)*	(15)*
Objective A									
2.5	22	32	21	1.0	−2.1	0.7	12	16	12
5.0	22	24	19	1.4	−1.0	0.9	11	15	11
7.5	19	21	18	1.1	−0.7	1.0	8	14	11
10.0	16	18	16	0.8	−0.5	1.0	7	13	11
Objective B									
2.5	22	32	21	1.1	−2.2	0.7	10	14	9
5.0	22	24	21	1.5	−1.1	0.9	8	12	8
7.5	20	21	19	1.3	−0.7	0.9	7	12	9
10.0	18	19	17	1.2	−0.7	0.8	7	12	9
Objective C									
2.5	24	32	22	1.2	−2.2	0.8	8	12	7
5.0	22	24	21	1.5	−1.1	0.9	7	11	7
7.5	21	22	20	1.5	−0.9	0.9	6	11	7
10.0	20	21	19	1.3	−0.8	0.8	6	10	6

*Initial spine configuration
Objective $A = 10f(X) + 1f(Y) + 50f(\theta_Y) + 2f(\theta_Z)$.
Objective $B = 10f(X) + 1f(Y) + 50f(\theta_Y) + 10f(\theta_Z)$.
Objective $C = 20f(X) + 1f(Y) + 90f(\theta_Y) + 50f(\theta_Z)$.

scoliosis are listed in column 2 of Table 1 and are shown graphically in Fig. 6. Typical responses employed the convex side erector spinae, lower latissimus, intercostals, external abdominal obliques, rectus abdominis, and transverse abdominal muscles. On the concave side, the upper trunk model muscles used were serratus, trapezius, pectorals, superior intercostals, superior thoracic erectors, and superior latissimus; and the lower trunk model muscles used were the external and internal abdominal obliques, transverse abdominals, and quadratus lumborum.

In summary, both brace and muscle forces were capable of substantial correction of a model thoracic scoliosis. Within the constraints imposed, muscle forces were somewhat more effective than brace forces in correcting all aspects of a thoracic scoliosis. The final spine configurations resulting from correction by brace and muscle forces are compared in Fig. 7. Since the focus of the optimization program was on correction of the spine, the rib cage has been omitted in these comparisons.

DISCUSSION

The results show the extent to which, under the optimal conditions outlined, brace and muscle forces are capable of correcting a model thoracic scoliosis. It is unknown if the muscle groups selected by the optimization program can contract simultaneously *in vivo* in the manner specified or if they can sustain their force over a period of time. The active correction theory of bracing assumes that the brace functions by

stimulating the contraction of trunk muscles to correct a scoliosis. There is no substantial evidence to support this theory (Wynarsky, 1988). Recently, it was shown that there were no differences in trunk muscle myoelectric activity in patients when wearing and when not wearing a Boston brace (Wynarsky and Schultz, 1990). Thus, although the biomechanical analysis here shows that muscles have a capacity for substantial curve correction, at present there is no evidence that correction by active muscle contraction can be obtained other than through electrostimulation. However, scoliosis treatment by electrostimulation remains controversial, with some investigators reporting good short-term results (McCollough *et al.*, 1980; Axelgaard and Brown, 1983; Axelgaard *et al.*, 1983; Brown *et al.*, 1984; Fisher *et al.* 1987; Katznelson and Nerubay, 1987), while others report poor results or question its effectiveness (Sullivan *et al.*, 1986; Benson, 1987; O'Donnell *et al.*, 1988).

The model curve corrections resulting from brace forces represent instantaneous elastic responses, and so correspond clinically to curve reductions upon immediate application of a brace. Watts *et al.* (1977) reported a mean instantaneous Boston brace reduction of initial 25° curves to 10° (60%) in 44 patients with a thoracolumbar scoliosis. In a study of 14 patients treated for adolescent idiopathic scoliosis by the Boston brace, Chase *et al.* (1989) reported a mean initial correction of 37 + 21% for thoracic curves and 36 + 22% for lumbar curves. Most clinical reports document the maximum Cobb angle reduction achieved during the treatment period (Emans *et al.*,
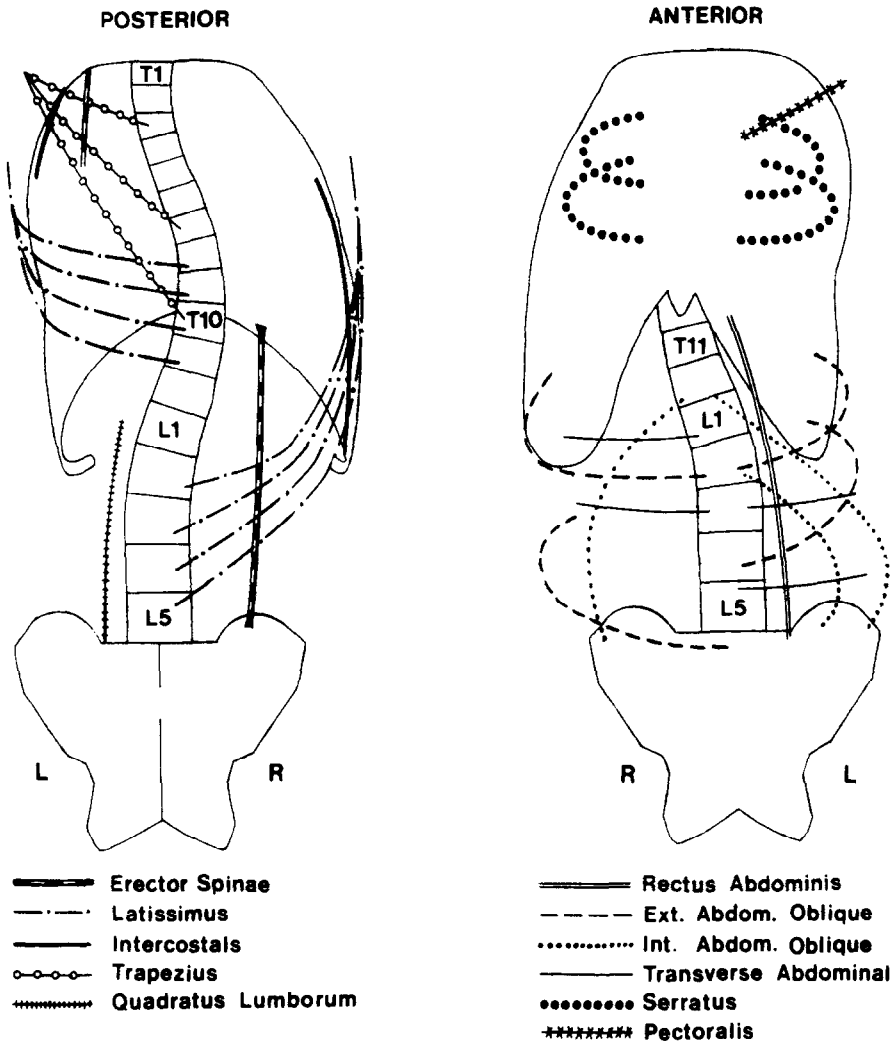
POSTERIOR   **ANTERIOR**

▬▬► Erector Spinae	══════ Rectus Abdominis
─·─·─· Latissimus	──── ── Ext. Abdom. Oblique
──────── Intercostals	············ Int. Abdom. Oblique
o-o-o-o- Trapezius	──────── Transverse Abdominal
++++++++ Quadratus Lumborum	●●●●●●●● Serratus
	++++++++ Pectoralis

Fig. 6. Muscle groups selected by the optimization technique to achieve optimal correction of the thoracic scoliosis. Specific muscle slips and their corresponding contraction intensity magnitudes are listed in Table 1, column 3.

1986; Gardner et al., 1986; Jonasson-Rajala et al., 1984; Willner, 1984; Laurnen et al., 1983; Uden and Willner, 1982). These show thoracic curve reductions, on average, of approximately 44%. In the present study, the initial model Cobb angle, which was considerably larger than those studied by Watts et al., was reduced 50% by optimal brace forces and 57% by optimal muscle contractions.

Secondary lateral curves in the high-thoracic and lumbar regions were produced by the optimal correction schemes. This finding is also consistent with clinical observations. Jonasson-Rajala et al. (1984) reported that the Boston brace treatment may cause secondary curves of approximately 15–19°. In the present study, the secondary curve magnitudes were only slightly larger than those reported with correction by brace forces (Table 2) and about the same magnitude with correction by muscle forces (Table 3).

Vertebral body lateral offsets and lateral tilts were closely related, but neither was substantially related to the axial rotations of the spine. In general, solutions which emphasized correction of the Cobb angle and lateral alignment did not yield substantial axial derotations of the vertebrae, and vice versa. This implies that correction of the rib cage deformity (associated with vertebral rotations) is not necessarily coupled with the Cobb angle. Weisz et al. (1989) showed that bracing may improve the cosmetic appearance of a scoliotic back, although the underlying curve remains unchanged. In 32 patients treated with the Boston brace, back surface shape improved in 41% of the patients, while roentgenographic improvement occurred in only 9%.

Few clinical investigations of bracing report axial de-rotations. Aaro et al. (1981) found a mean apical de-rotation of 38% in 33 patients with thoracic idiopathic scoliosis. The present study found a reduction of 40% when brace forces and objective weighting C were used. Saraste and Ostman (1986) found a mean de-rotation of 14° (0–38° range) in 21 patients treated
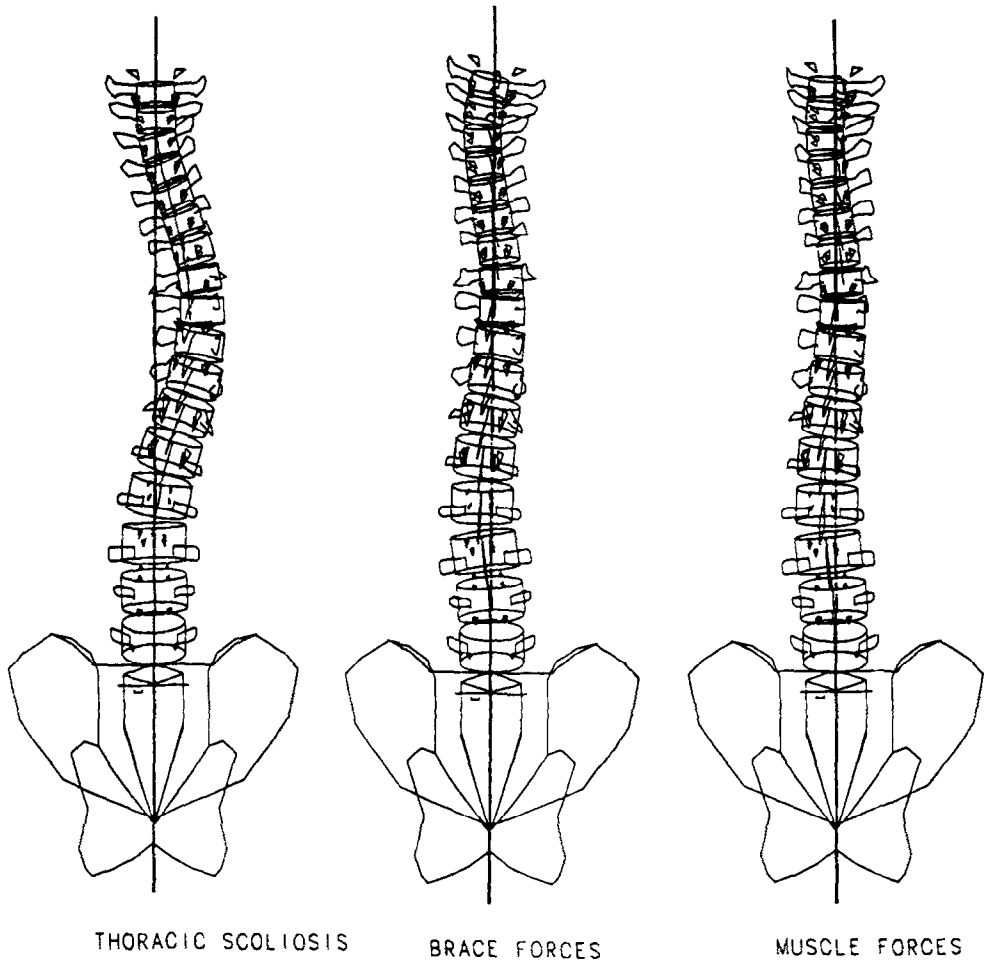
Fig. 7. Comparison of thoracic curve correction by brace and muscle forces. Rib cage omitted for clarity.

with the Boston brace, but they did not report initial axial rotations.

The structural model assumed, for computational tractability, both geometric and material linearity. Wynarsky (1988) explains why neither assumption was likely to qualitatively bring into question any study findings. Model spine configuration changes were usually small and only the deformable elements connecting the ribs to the sternum underwent large deformations. Model geometric changes did not relate linearly to applied forces because of the inequality constraints.

There is substantial clinical controversy regarding the use of bracing for the treatment of idiopathic scoliosis. The curve reductions achieved under optimal biomechanical conditions were only slightly larger than the largest reductions ('best-in-brace') achieved clinically by brace treatment. Thus, the outcomes resulting from optimal brace treatment could not be expected to be much different from those obtained with present bracing protocols. Clinically, long-term brace treatment results show that, on aver-

age, the final post-treatment Cobb angles are nearly the same as the pre-bracing Cobb angles (Mellencamp and Blount, 1986; Emans et al., 1986; Bassett et al., 1986; Peltonen et al., 1988; Edgar and Mehta, 1988; Andrews and MacEwen, 1989; Montgomery et al., 1990; Piazza and Bassett, 1990). Thus, even optimal bracing should not be expected to produce either permanent or complete curve correction.

In summary, a computational scheme for optimizing responses of skeletal configuration models has been presented. Use of the scheme was illustrated by studies of non-surgical correction of a model thoracic scoliosis. The outcomes of the model study were in good agreement qualitatively with clinical outcomes, but quantitatively tended to be somewhat better. Even under optimal conditions it is unlikely that scoliosis can be fully corrected by passive brace forces or active trunk muscle contractions.

This optimization program has been used to study thoracolumbar double curve, single lumbar curve, and thoracic double curve spine morphologies (Wynarsky, 1988). This scheme could also be used to determine

forces necessary to correct the rib cage deformity as well as the spinal deformity, and other spine deformities such as kyphosis, lordosis, and lordoscoliosis.

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