Patient-specific mechanical properties of a flexible multi-body model of the scoliotic spine

Y. Petit[1,2] C.-É. Aubin[1,2] H. Labelle[2]

[1]École Polytechnique, Department of Mechanical Engineering, Montreal, Quebec, Canada
[2]Research Centre, Sainte-Justine Mother-Child University Hospital, Montreal, Quebec, Canada

Abstract—*The flexibility of the scoliotic spine is an important biomechanical parameter to take into account in the planning of surgical instrumentation. The objective of the paper was to develop a method to characterise* in vivo *the mechanical properties of the scoliotic spine using a flexible multi-body model. Vertebrae were represented as rigid bodies, and intervertebral elements were defined at every level using a spherical joint and three torsion springs. The initial mechanical properties of motion segments were defined from* in vitro *experimental data reported in the literature. They were adjusted using an optimisation algorithm to reduce the discrepancy between the simulated and the measured Ferguson angles in lateral bending of three spine segments (major or compensatory left thoracic, right thoracic and left lumbar scoliosis curves). The flexural rigidity of the spine segments was defined in three categories (flexible, nominal, rigid) according to the estimated mechanical factors (α). This approach was applied with ten scoliotic patients undergoing spinal correction. Personalisation of the model resulted in an increase of the initial flexural rigidity for seven of the ten lumbar segments ($1.38 \leq \alpha \leq 10.0$) and four of the ten right thoracic segments ($1.74 \leq \alpha \leq 5.18$). The adjustment of the mechanical parameters based on the lateral bending tests improved the model's ability to predict the spine shape change described by the Ferguson angles by up to 50%. The largest differences after personalisation were for the left lumbar segments in left bending ($4^\circ \pm 3^\circ$). The* in vivo *identification of the mechanical properties of the scoliotic spine will improve the ability of biomechanical models adequately to predict the surgical correction, which should help clinicians in the planning of surgical instrumentation manoeuvres.*

Keywords—*Spine biomechanics, Multi-body model, Bending test, Scoliosis, Spine instrumentation surgery*

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1 Introduction

THE FLEXIBILITY of the scoliotic spine is an important biomechanical parameter for the planning of surgical instrumentation to assess the reducibility of the curves as well as the levels to instrument. It is evaluated by means of flexibility tests such as side bending (ARONSSON *et al.*, 1996; KLEPPS *et al.*, 2001; POLLY and STURM, 1998), traction (MATSUMOTO *et al.*, 1997; POLLY and STURM, 1998) and fulcrum bending (CHEUNG and LUK, 1997; KLEPPS *et al.*, 2001). Even if the maximum voluntary supine side bending test is considered as the gold standard (KLEPPS *et al.*, 2001), there is currently no consensus on the optimum flexibility test. Moreover, these tests measure the mobility rather than the 'flexibility' of the spine, as the forces involved are not known (SEVASTIK and STOKES, 2000).

Biomechanical models were developed (AUBIN *et al.*, 2003; GARDNER-MORSE and STOKES, 1994; GHISTA *et al.*, 1988; LEBORGNE *et al.*, 1999; POULIN *et al.*, 1998; VANDERBY *et al.*, 1986) to simulate scoliotic spine instrumentation manoeuvres and to estimate reaction forces at intervertebral levels. Although finite element models may prove essential for the study of anatomical stress levels during surgery, flexible multi-body models are more efficient in predicting changes in the shape of the spine resulting from different surgical instrumentation strategies and have the potential to assist in the pre-operative planning (AUBIN *et al.*, 2003). In general, these models adequately predict the correction of the instrumented spine in the frontal plane, but are less accurate for adjacent segments and the other planes in space (STOKES *et al.*, 1999). Plausible total reaction forces between the rods and the implants were predicted by the models, but the maximum reactions were close to pedicle screw pull-out forces (AUBIN *et al.*, 2003). It is suspected that the mechanical properties of motion segments generally obtained from cadaver spines, without

Correspondence should be addressed to Professor Carl-Éric Aubin; email: carl-eric.aubin@polymtl.ca

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consideration of *in vivo* loading conditions, inadequately represent the spine behaviour of scoliotic patients during surgical instrumentation (AUBIN *et al.*, 2003; GARDNER-MORSE and STOKES, 1994; STOKES *et al.*, 1999).

Methods have been proposed to personalise the mechanical properties of finite element models of the spine (GHISTA *et al.*, 1988; LEBORGNE *et al.*, 1999; VANDERBY *et al.*, 1986). For instance, VANDERBY *et al.* (1986) developed an optimisation method based on the curve reduction after the application of distraction forces during surgery. GHISTA *et al.* (1988) proposed an optimisation method based on pre-operative traction radiographs, but only represented the spine using a 2D finite element model. LEBORGNE *et al.* (1999) have developed an heuristic approach to introduce personalised mechanical properties into a 3D finite element model based on lateral bending radiographs. However, this approach is subjective and dependent on the adjustment strategy adopted from the knowledge of important mechanical parameters and *a priori* thoughts of where structural stiffening appears in scoliosis.

The purpose of this paper was to develop a biomechanical model of the scoliotic spine, incorporating patient-specific geometric and mechanical properties.

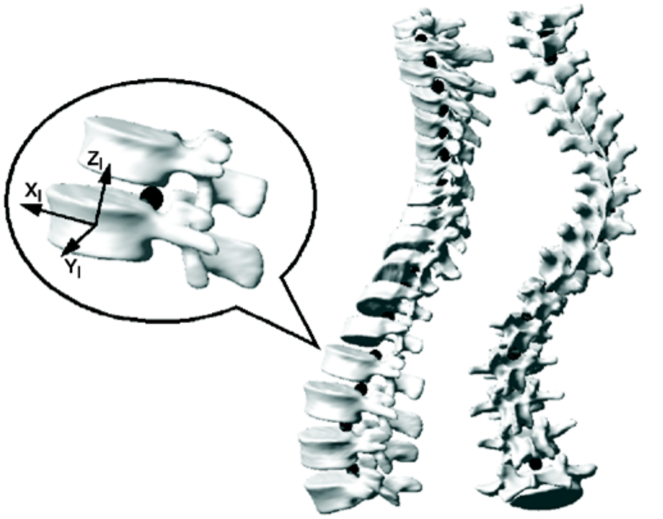

Fig. 1 *Geometric representation of multi-body model of spine and of location of intervertebral joints. Local co-ordinate system of functional unit is located at centre of vertebral body of inferior vertebra; X_1, Y_1 and Z_1 local axes define anterior, left and cephalad directions, respectively*

2 Material and methods

The flexible multi-body model of the scoliotic spine and the method of personalisation of intervertebral mechanical properties were developed and implemented using ADAMS 11.0 flexible mechanism simulation software*. The geometrical and mechanical properties of the model's functional units were defined in a local co-ordinate system (Fig. 1) having its origin at the centre of the vertebral body of the inferior vertebra, where the X_1, Y_1 and Z_1 axes define the anterior, left and cephalad directions, respectively.

2.1 *Flexible multi-body model of the spine*

Postero-anterior (PA) and lateral (LAT) radiographs of the spine were taken one day before surgery in a cohort of ten scoliotic patients, presenting right thoracic (RT, $n = 5$) or right thoracic and left lumbar (RT-LL, $n = 5$) curves. The initial 3D geometry of the model was personalised to the patients from the identification of six anatomical landmarks for each vertebra on the radiographs using the 3D reconstruction method of CHERIET *et al.* (1999).

A comprehensive geometric representation of each vertebra was obtained (Fig. 1) from the deformation of an atlas of detailed vertebrae to fit the reconstructed landmarks (AUBIN *et al.*, 1995). Vertebrae were represented using rigid bodies. The intervertebral elements of every functional unit were defined using a spherical joint and three torsion springs. The spherical joints were located at the posterior extremity of the superior endplate of the functional unit's lowest vertebra (Fig. 1) based on a study made of the kinematics of the scoliotic spine surgical instrumentation in a cohort of 82 patients (PETIT *et al.*, 2003). These joints allow 3 degrees of freedom (DOFs) in rotation and constrain all relative translations between the vertebrae. The torsion springs represent the principal flexible behaviour of motion segments in rotation (flexural rigidity) as a linearisation of load-displacement curves reported in the literature (OXLAND *et al.*, 1992; PANJABI *et al.*, 1976*a*; *b*; 1994), as well as the coupling behaviour between transverse and frontal plane rotations, as

$$\begin{Bmatrix} M_x \\ M_y \\ M_z \end{Bmatrix} = \begin{bmatrix} \alpha_i K_{xx} & 0 & K_{zx} \\ 0 & \beta_i K_{yy} & 0 \\ K_{xz} & 0 & \delta_i K_{zz} \end{bmatrix} \begin{Bmatrix} R_x \\ R_y \\ R_z \end{Bmatrix} \qquad (1)$$

where each of the torsion springs produces a moment (M_x, M_y or M_z) from the rotations (R_x, R_y, R_z) and the stiffness coefficients (K_{xx}, K_{yy}, K_{zz}, K_{xz}, K_{zx}). Mechanical modulation parameters (α_i, β_i, δ_i) were also affected by the principal stiffness coefficients (K_{xx}, K_{yy}, K_{zz}) and initially set to 1 (100%) for all functional units of each scoliotic segment *i* defined in the following Section. The initial mechanical properties of the model's functional units are presented in Table 1.

2.2 *Model's mechanical parameters personalisation*

The frontal plane mechanical modulation parameters α_i were personalised to specific patients using an optimisation algorithm to improve the model's behaviour for movements presenting a correction of the scoliotic curves. Two antero-posterior radiographs were taken pre-operatively while the patient performed maximum voluntary bending movements to the left and right sides in the supine position.

Table 1 Initial mechanical properties of multi-body model of scoliotic spine

Functional units	Stiffness coefficients, Nm rad^{-1}				
	K_{xx}	K_{yy}	K_{zz}	K_{xz}	K_{zx}
T1-T2	210	90	164	−50.9	−50.9
T2-T3	135	115	91	−10.1	−10.1
T3-T4	191	185	198	−6.2	−6.2
T4-T5	187	241	185	−3.5	−3.5
T5-T6	158	141	121	−9.5	−9.5
T6-T7	122	152	111	−5.7	−5.7
T7-T8	239	163	107	−7.0	−7.0
T8-T9	143	158	161	−9.2	−9.2
T9-T10	122	153	132	−11.5	−11.5
T10-T11	197	161	195	−29.0	−29.0
T11-T12	110	148	214	9.5	25.2
T12-L1	99	110	320	−2.4	34.5
L1-L2	114	102	389	−2.0	−123.6
L2-L3	95	85	327	−43.9	−104.8
L3-L4	97	91	292	−64.6	−37.9
L4-L5	64	71	324	−131.8	−131.8

*MSC Software Corporation, Santa Anna, USA

Three spine segments were defined corresponding to the scoliotic curves (major and compensatory) and were affected by a mechanical modulation parameter: α_1 for the LT segment and α_2 and α_3 for the LL and RT segments. The Ferguson angle *Fb*, which is the angle between the lines drawn through the mid-points of the end vertebrae and the apical vertebra, was measured for every spine segment (Fig. 2). It has been reported previously (STOKES *et al.*, 1993) that the Ferguson angle is adequate to measure curve magnitude with an inter-observer variability of 1.8° (standard deviation). The amplitude of the lateral bending movement was also measured on both radiographs from the angle between the line drawn through the mid points of T1 and L5 and the normal to the line drawn through the superior tips of the left and right iliac crests. Simulation of the lateral bending (left and right) was then defined by fixing all DOFs of the lowest 3D reconstructed vertebra (L5) and by imposing a lateral displacement to the superior vertebra (T1) until the measured bending amplitude of the spine was reproduced.

During the simulation, the simulated Ferguson angles *Fs* of every segment were calculated. A cost function ω was defined as the sum of the squared differences between simulated and experimentally measured Ferguson angles for the three spine segments

$$\omega = \sum_{i=1}^{3} (Fs_i - Fb_i)^2 \tag{2}$$

A Fletcher–Reeves conjugate gradient optimisation algorithm implemented in the simulation software was then used to modify the design variables α_i until the cost function ω was minimised. A first minimisation was performed for the left bending simulation to determine the optimum values of the modulation parameters corresponding to the LL and LT curves (α_1 and α_3). Starting with the results of the first optimisation, a second minimisation was performed for the right bending to find out the optimum α_2 (RT curve). The adjusted parameter set (α_1, α_2 and α_3) was considered to be optimum for a movement reducing the curve of all segments, which is expected to result from scoliotic spine surgical instrumentation.

2.3 *Evaluation of the spine model sensitivity*

The influence of the variability of the intervertebral articulation location previously documented (PETIT *et al.*, 2003) was verified. This was done by incorporating a randomly distributed perturbation with a standard deviation of 10 mm into the initial location defined in the model, while performing 30 successive optimisation simulations to personalise the model's mechanical parameters. It was tested on one subject of this study presenting a single RT curve (Ferguson angle of 38° standing). The simulated Ferguson angles of the major curve and of the two compensatory curves were measured in left and right bending before and after personalisation. The mechanical modulation parameters resulting from the personalisation were also recorded.

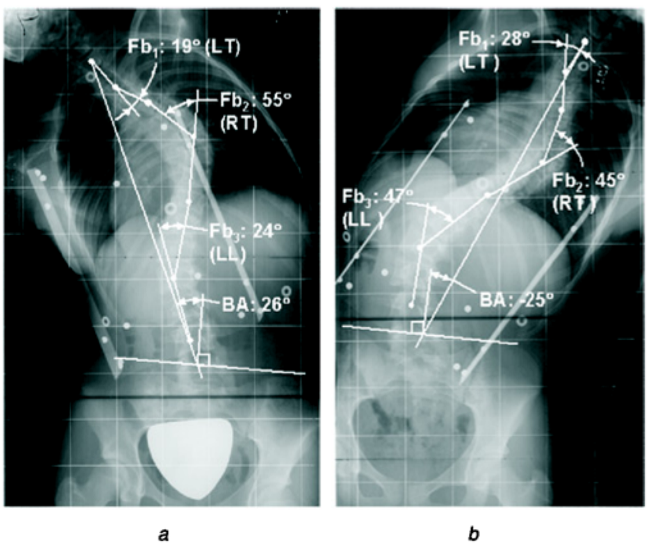

Fig. 2 *Ferguson angles of spine segments and amplitude of lateral bending movement measured on (a) left and (b) right bending radiographs*

3 Results

3.1 *Patient-specific mechanical parameters of the scoliotic spine*

On average, the Ferguson angles decreased by 24% for the LT segments and by 44% for the LL segments in left bending, and they decreased by 21% for the RT segments in right bending after optimisation of the mechanical properties (Table 2). The amplitude of the lateral bending movement measured on the radiographs presents a large variability, with standard deviations of 7° and 6° for the left and right sides, respectively. The personalisation of the mechanical parameters allowed an average improvement of the cost functions of 9% for the left bending and of 50% for the right bending (Table 2). Before the adjustment of mechanical properties, the differences between the measured and simulated Ferguson angles were $3° \pm 2°$ (average $\pm$ standard deviation) and $3° \pm 3°$ for the LT and LL segments in left bending, and they were $4° \pm 3°$ for the RT segments in right bending. After the personalisation, these differences were $1° \pm 2°$ and $4° \pm 3°$ for LT and LL segments in left bending and $2° \pm 1°$ for RT segments in right bending.

The flexural rigidity of spine segments was divided into three categories according to the estimated mechanical modulation parameters α: flexible ($\alpha < 0.8$), nominal ($0.8 \leq \alpha \leq 1.2$) and rigid ($\alpha > 1.2$). LL segments were rigid in seven cases ($1.38 \leq \alpha \leq 10.0$) and flexible in only one case ($\alpha = 0.23$). The rigid RT segments ($n = 4$, $1.74 \leq \alpha \leq 5.18$) were all associated with a rigid LL segment. All other RT segments were flexible ($0.15 \leq \alpha \leq 0.45$). LT segments were flexible in seven cases ($0.15 \leq \alpha \leq 0.74$), and five of these were associated with a flexible RT segment. Only one LT segment was rigid ($\alpha = 7.6$), and only two LT segments and two LL segments were nominal.

The adjustment of the mechanical properties produced an overall increase in intervertebral reaction moments in the frontal plane (Table 3). However, reactions at the apex of the flexible RT segments decreased by 46% and 24% in left and right bending, respectively, whereas it increased by a factor of 2.3 and 2.2 for rigid RT segments. Similarly, intervertebral reactions increased by 129% and 125% in left and right bending at the apex of the rigid LL segments, whereas they decreased by 74% and 31% for nominal and flexible segments.

3.2 *Validation results*

Variabilities of 0.13° (standard deviation) and 0.33° were found for the computed Ferguson angle before and after personalisation, when a random perturbation up to 10 mm was generated on the location of the spherical joints. The maximum deviation from the average was obtained after personalisation on the left thoracic curve during the left bending simulations (1.1°), whereas the right thoracic and left lumbar curves were less influenced by the perturbations (maximum 0.53° and 0.28°, respectively, during the left bending simulations). The variability of the mechanical modulation parameters was 0.02, 0.03 and 0.01, respectively, for α_1, α_2 and α_3, and the maximum deviation from the average was obtained for the right thoracic segment (α_2: 0.05).

Table 2 Average values (and standard deviation) of Ferguson angles and lateral bending amplitude measured on radiographs and predicted from bending simulations with and without personalisation of mechanical properties

	Radiographs			Bending simulations			
		bending		left		right	
	standing	left	right	initial	personalised	initial	personalised
Ferguson angles, °							
left thoracic	21 (12)	16 (9)	26 (11)	19 (9)	17 (10)	20 (9)	22 (9)
right thoracic	43 (9)	48 (6)	34 (8)	44 (6)	47 (9)	38 (8)	35 (8)
left lumbar	29 (8)	16 (9)	32 (10)	17 (6)	20 (7)	34 (9)	33 (10)
Bending amplitude, °	3 (3)	18 (7)	16 (6)				
Cost function ω				64 (54)	58 (43)	104 (92)	52 (37)

Table 3 Average (and standard deviation) intervertebral moments (Nm) in frontal plane at selected intervertebral levels

	Left bending simulations		Right bending simulations	
	initial	optimised	initial	optimised
LT apex	0.92 (0.53)	1.47 (1.49)	0.79 (0.44)	1.39 (1.3)
T limit	1.66 (0.57)	2.64 (2.22)	1.79 (0.71)	3.19 (2.58)
RT apex	2.67 (0.79)	4.46 (3.85)	2.91 (0.99)	5.42 (4.59)
TL limit	4.37 (1.31)	7.39 (5.98)	3.88 (1.24)	7.36 (6.24)
LL apex	6.24 (2.24)	10.55 (8.63)	4.95 (1.62)	9.53 (8.1)

4 Discussion

The adjustment of the mechanical parameters based on the lateral bending tests clearly indicates an improvement of up to 50% of the model's ability to predict the Ferguson angles in lateral bending. However, this adjustment was limited to the frontal plane, as only 2D information was available from the bending radiographs, and the change in spine shape resulting from side-bending is expected to occur mostly in that plane. As coupled displacements between axial and frontal rotations were defined in the model, it also affected the behaviour in the transverse plane. The simulation of side bending by only constraining a lateral displacement is also a simplification of the reality, because it neglects the effect of gravity on the change in posture, the interaction between the table and the trunk and the muscle forces distributed along the spine. To address combined effects, an additional test was performed on one patient. The segmental optimisation of the model's mechanical parameters was performed in the frontal and the sagittal planes (α_i and β_i, (1)) using the same technique. The Ferguson angles of the three spine segments in lateral bending were measured in the frontal and the sagittal planes from a 3D model obtained by 3D reconstruction (CHERIET *et al.*, 1999) using an additional radiograph. The mechanical behaviour of the model was improved by 65% in the sagittal plane and 69% in the frontal plane. However, the use of lateral radiographs during the bending test cannot be done routinely in clinical practice.

The adjustment of the model's mechanical parameters was the same for all vertebrae of a given segment. PERDRIOLLE (1979) found that the functional units at the apex (apex ± 1 vertebra) account for 63% of the scoliotic curve reduction with the lateral bending test, whereas the overlying and underlying functional units of the curve account for 10% and 27%, respectively. This suggests that the flexural rigidity of scoliotic segments may not be homogeneous and that shorter segments should have been considered. However, PERDRIOLLE (1979) measured curve reduction using the Cobb angle method, which relies on the angulation of the endplate of vertebrae and presents a variability (1.3°) similar to that of the Ferguson method (STOKES *et al.*, 1993). Thus Ferguson and Cobb angles are actual limits of the proposed personalisation method, and using shorter spine segments will only be possible if a more sophisticated measurement method is considered.

The large variability of the lateral bending amplitude suggests a poor reproducibility of this test. For instance, the lateral bending amplitudes measured to the left (19°) and right (7°) sides for one of the subjects in this study were not important compared with the initial balance of the spine (6°). Consequentially, the cost function was much smaller in the left bending (2.8) than in the right (145.6) bending before the optimisation. The adjustment of mechanical parameters resulted in slight changes in the cost function values for the left (+6.7) and right (−27.3) bending, respectively. This suggests that the algorithm could converge to a sub-optimum solution of mechanical parameters when the lateral bending amplitude is not sufficient. This observation draws our attention to the controversy about the efficiency of the lateral bending test to predict the surgical correction, compared with other flexibility tests (POLLY and STURM, 1998; KLEPPS *et al.*, 2001). The application of the proposed personalisation method to the traction and fulcrum bending tests can easily be carried out and is considered a possible improvement.

A classification of spine segment flexibility was proposed, based on computed mechanical modulation parameters. This allowed discrimination between flexible and rigid scoliotic curves, compared with published data from cadaveric spines. Such a classification was not possible using only the reducibility of spine curves measured after the lateral bending tests. The small regression coefficients between the measured reducibility and the adjusted mechanical properties ($R^2 \leq 0.2$) suggest that the reducibility, as expressed only by the lateral bending test, is not sufficient to predict the flexural rigidity of scoliotic segments. This is in agreement with recent studies (KLEPPS *et al.*, 2001) showing that none of the current clinical tests allows the actual correction observed post-operatively to be fully predicted. Consequently, the proposed mechanical property adjustment method allows better approximation of the segmental flexibility when intervertebral reaction forces are being estimated. This classification should be extended to a larger number of scoliotic patients in each category to confirm its clinical relevance.

The personalisation of the model for the ten patients yielded an increase in the initial stiffness from cadaver specimens in 70 % of the lumbar segments and 40% of the thoracic segments, with a large inter-individual variability. This finding is in agreement with clinical experience (MOE *et al.*, 1978), but is in contrast with the results of VANDERBY *et al.* (1986), who reported no significant changes in segmental flexural rigidity in a preliminary study on only one scoliotic patient. The inter-individual variability of the adjusted mechanical properties also underlines the importance of considering the flexibility of the spine in the planning of surgical instrumentation manoeuvres. The general increase in the initial motion segment stiffness could also be considered to take into account the *in vivo* loading in the spine provided by gravity, active muscles, rib and other structures that were not present in the cadaver data. In fact, axial compressive preloads used in experimental studies to emulate physiological loading conditions have been shown to increase the motion segment stiffness and linearity (ADAMS, 1995; CRIPTON *et al.*, 2000; PATWARDHAN *et al.*, 1999). Thus the proposed method allows the global effect of these structures to be considered, as well as the structural stiffening associated with scoliosis deformities, and may lead to an increased ability to predict how well surgical instrumentation will benefit a specific patient.

5 Conclusions

A new method was presented for the identification of patient-specific mechanical properties of the scoliotic spine using a flexible multi-body model and an optimisation algorithm based on the lateral bending test. This method was used to estimate *in vivo* the segmental mechanical properties of the spine of ten scoliotic subjects and allowed flexible and rigid scoliotic curves to be discriminated. The inter-individual variability of the scoliotic spine flexural rigidity is important and should be considered in biomechanical simulation models. The *in vivo* identification of the mechanical properties of the scoliotic spine will improve the ability of biomechanical models adequately to predict surgical correction as a function of the instrumentation strategy. The exploitation of such a biomechanical model should help clinicians in the planning of surgical instrumentation manoeuvres.

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Authors' biographies

C.-É. AUBIN received his B.Eng and PhD degrees in mechanical engineering from the Ecole Polytechnique, Montreal, Canada. He is an Associate Professor at Ecole Polytechnique in the Department of Mechanical Engineering and at the Biomedical Engineering Institute, and a Researcher at the Research Center of Sainte-Justine University Hospital, Montreal. He is also Director of the Biomechanical Modeling & Computer Assisted Surgery Lab. Since 2001, he has held the Canada Research Chair on CAD Innovations in Orthopedic Engineering. His research interests include orthopedic biomechanics, computer modelling of the spine and thorax, pathomechanism of musculoskeletal deformities, computer assisted surgery as well as experimental evaluation and biomechanical simulations of orthotics and surgical corrections.

Y. PETIT received the B.Eng and M.A.Sc degrees in mechanical engineering and a PhD in biomedical engineering from École Polytechnique, Montreal, in 1992, 1995, and 2002, respectively. He held 'Research Assistant' and 'Director of Operations' positions at the 3-D Scoliosis Laboratory at Sainte-Justine Hospital between 1995 and 1999. Since January 2003, he has been a Postdoctoral Fellow at Queens University, Kingston, ON, Canada, in the Department of Mechanical Engineering. His research interests include orthopedic biomechanics, 3-D geometrical and biomechanical modelling of the musculoskeletal system, medical image analysis, and computer-assisted surgery.

H. LABELLE received the M.D. degree and completed residency in orthopaedics at University of Montreal, in 1981. He completed a two-year fellowship in paediatric orthopaedics at Sainte-Justine Hospital, Montreal, Rancho Los Amigos Hospital, CA and the A.I. Du Pont Institute, DE. Since 1982, he has held a post in the Department of Surgery, University of Montreal and at Ste-Justine Hospital. He is currently Professor of Surgery and Titular of the Movement Sciences Research Chair of Sainte-Justine Hospital and University of Montreal. His clinical work is focused on the evaluation and treatment of scoliotic deformities in children. He is Head of the Musculoskeletal Research Group at Sainte-Justine Hospital Research Center and Director of the 3-D Scoliosis Laboratory. His research interests are concentrated on the 3-D evaluation and treatment of spinal deformities, with a particular emphasis on computer-assisted surgery, 3-D design and evaluation of braces for the treatment of idiopathic scoliosis, and 3-D evaluation and simulation of surgery for scoliotic deformities.