Clinical Biomechanics 39 (2016) 115–121

Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

How does differential rod contouring contribute to 3-dimensional correction and affect the bone-screw forces in adolescent idiopathic scoliosis instrumentation?

Xiaoyu Wang, PhD [a,b], Laure Boyer, M.Eng [a,b], Franck Le Naveaux, B.Eng [a,b], Richard M. Schwend, MD [c], Carl-Eric Aubin, PhD, Eng [a,b,*]

[a] Polytechnique Montréal, Department of Mechanical Engineering, P.O. Box 6079, Downtown Station, Montreal (Quebec), H3C 3A7, Canada
[b] Sainte-Justine University Hospital Center, 3175, Cote Sainte-Catherine Road, Montreal (Quebec), H3T 1C5, Canada
[c] Children's Mercy Hospital, 2401, Gillham Rd, Kansas City, (Missouri) 64108, USA

ARTICLE INFO

Article history:
Received 13 April 2016
Accepted 4 October 2016

Keywords:
Biomechanical modeling
Pedicle screw
Scoliosis
Spine instrumentation
Differential rod contouring
Vertebral derotation

ABSTRACT

Background: Differential rod contouring is used to achieve 3-dimensional correction in adolescent idiopathic scoliosis instrumentations. How vertebral rotation correction is correlated with the amount of differential rod contouring is still unknown; too aggressive differential rod contouring may increase the risk of bone-screw connection failure. The objective was to assess the 3-dimensional correction and bone-screw forces using various configurations of differential rod contouring.

Methods: Computerized patient-specific biomechanical models of 10 AIS cases were used to simulate AIS instrumentations using various configurations of differential rod contouring. The tested concave/convex rod configurations were 5.5/5.5 and 6.0/5.5 mm diameter Cobalt-chrome rods with contouring angles of 35°/15°, 55°/15°, 75°/15°, and 85°/15°, respectively. 3-dimensional corrections and bone-screw forces were computed and analyzed.

Findings: Increasing the difference between the concave and convex rod contouring angles from 25° to 60°, the apical vertebral rotation correction increased from 35% (SD 17%) to 68% (SD 24%), the coronal plane correction changed from 76% (SD 10%) to 72% (SD 12%), the thoracic kyphosis creation from 27% (SD 60%) to 144% (SD 132%), and screw pullout forces from 94 N (SD 68 N) to 252 N (SD 159 N). Increasing the concave rod diameter to 6 mm resulted in increased transverse and coronal plane corrections, higher thoracic kyphosis, and screw pullout forces.

Interpretations: Increasing the concave rod contouring angle and diameter with respect to the convex rod improved the transverse plane correction but with significant increase of screw pullout forces and thoracic kyphosis. Rod contouring should be planned by also taking into account the 3-dimensional nature and stiffness of the curves and combined with osteotomy procedures, which remains to be studied.

1. Introduction

Pedicle screw fixation uses the strongest part of the vertebra to provide solid anchoring of the spinal rods and enables corrective forces/torques to be applied in three dimensions. It has been observed that transverse corrective forces tended to further rotate the spine; the rib hump was worsened in some cases after segmental instrumentation (Ogilvie and Millar, 1983; Sullivan and Conner, 1982). The transverse plane correction has been the focus of studies on modern segmental spinal instrumentation. In pedicle-screw-based spinal instrumentation, transverse plane correction can be achieved with apical vertebral derotation (Cheng et al., 2010) and/or segmental vertebral derotation (Hwang et al., 2012; Parent et al., 2008). Another important technique to improve transverse plane correction is differential rod contouring (Newton, 2013; Shah, 2011). The underlying principle of differential rod contouring is that the over-bent concave rod pulls the concave side of the apical spine dorsally out of the chest and the under-bent convex rod pushes the convex side of the apical spine to generate a vertebral derotation torque to reduce the vertebral rotation in the transverse plane (Newton, 2013; Shah, 2011). Differential rod contouring was thought to be an effective way to improve transverse plane correction and its effectiveness has been observed clinically, however, biomechanical assessment has yet to be performed to acquire essential knowledge on the appropriate differential contouring angle and the associated bone-screw forces. The objective of this study was to biomechanically assess corrections in the three anatomical planes

* Corresponding author.
E-mail address: carl-eric.aubin@polymtl.ca (C.-E. Aubin).

http://dx.doi.org/10.1016/j.clinbiomech.2016.10.002
0268-0033/

Table 1
Patient information and clinical indices (PT: proximal thoracic; MT: main thoracic; TL/L: thoracolumbar/lumbar; UIV: upper instrumented vertebra; LIV: lower instrumented vertebra).

Case #		1	2	3	4	5	6	7	8	9	10
Sex		Female	Female	Female	Male	Female	Female	Female	Female	Female	Female
Age		14	17	14	15	16	15	16	15	19	14
Height (cm)		154	168	170	172	165	170	162	159	162	148
Weight (kg)		52	56	59	55	53	48	56	59	47	39
Lenke classification		1 A	4 A	3B	3C	1 A	1 A	1 A	3C	3B	2 A
MT curve superior end vertebra		T6	T5	T5	T7	T5	T4	T6	T6	T5	T5
Apical vertebra		T9	T8	T8	T10	T9	T8	T11	T9	T8	T9
Inferior end vertebra		T12	T11	T12	L1	L1	T12	L2	T12	T11	T11
UIV		T4	T3	T3	T4	T2	T3	T4	T3	T4	T4
LIV		L2	T12	L1	L2	L1	L1	L3	L1	T12	L1
PT Cobb (°)	Preoperative	32	52	31	34	28	9	31	40	39	42
	Left bending	22	17	24	6	20	5	10	15	12	28
	Right bending	35	54	35	37	32	20	41	41	42	46
MT Cobb (°)	Preop.	50	58	62	59	40	46	50	67	59	52
	Left bending	59	64	65	62	73	55	60	63	63	60
	Right bending	29	38	50	27	10	30	25	35	29	30
MT apical vertebral rotation (°)	Preoperative	16	19	19	20	19	18	18	22	18	19
TL/L Cobb (°)	Preoperative	37	30	42	48	35	39	37	40	39	32
	Left bending	2	9	5	30	10	10	5	9	20	2
	Right bending	49	40	65	50	42	45	50	70	42	49
TL/L apical vertebral rotation (°)	Preoperative	5	7	3	11	6	11	3	9	7	3
T4–T12 kyphosis (°)	Preoperative	8	32	27	17	24	23	24	8	37	15
L1–L5 lordosis (°)	Preoperative	42	45	33	15	47	40	37	30	27	32

and bone-screw forces associated with various configurations of differential rod contouring.

2. Methods

To quantify the effects of differential rod contouring on 3D corrections and bone-screw forces, numerical simulations were performed using patient-specific models of 10 adolescent idiopathic scoliosis (AIS) patients. With the institutional review board approval, the cases were randomly selected from AIS patients having undergone instrumented spinal fusion at our university hospital center during the last 6 years with preoperative apical vertebral rotation angle $>15°$ and a variety of coronal plane deformities and thoracic kyphosis. The collected clinical data of the 10 cases are provided in Table 1.

The independent variables were the concave rod diameter and its sagittal plane contouring angle with respect to that of the convex rod. The dependent variables were the apical vertebral rotation, Cobb angles in the coronal and sagittal planes, and bone-screw forces. The instrumented fusion levels in the actual surgeries were initially used in the modeling and simulations for verification purposes. The instrumentation levels were then adapted and the constructs were changed to uniaxial pedicle screws and Cobalt-chrome rods in all simulations in order to isolate the effect of the differential rod contouring. Modeling and simulation details are presented in the following sub-sections.

2.1. Patient-specific biomechanical spine model

3D spine geometries of the selected cases were built using calibrated preoperative coronal and lateral radiographs and 3D multi-view reconstruction techniques (Cheriet et al., 2007). The process began with the identification of key anatomical landmarks of each vertebra, typically, the pedicles, vertebral endplate middle and corner points, and transverse and spinous process extremities. 2D coordinates of these landmarks allowed the determination of their 3D coordinates in space, which was done using a self-calibration and optimization algorithm (Cheriet et al., 2007; Delorme et al., 2003). The reconstruction process was completed by registering detailed vertebral models using the 3D coordinates of the key landmarks and a free form deformation technique (Cheriet et al., 2007; Delorme et al., 2003). Average accuracies for pedicles and vertebral bodies were 1.6 mm (SD 1.1 mm) and 1.2 mm (SD 0.8 mm), respectively (Delorme et al., 2003).

Vertebrae from T1 through L5 and the pelvis were modeled as rigid parts which were connected through multiple flexible elements respectively representing the intervertebral discs, anterior longitudinal ligaments (ALL), posterior longitudinal ligaments (PLL), ligamenta flava (LF), intertransverse ligaments (ITL), facet joint capsules (FC), and interspinous ligaments (ISL) combined with supraspinous ligaments (SSL). Six translational springs were used to respectively represent 1) ALL, 2) PLL, 3) LF, 4) left ITL, 5) right ITL, and 6) the combined effect of ISL and SSL. The biomechanical behavior of the facet joints is more

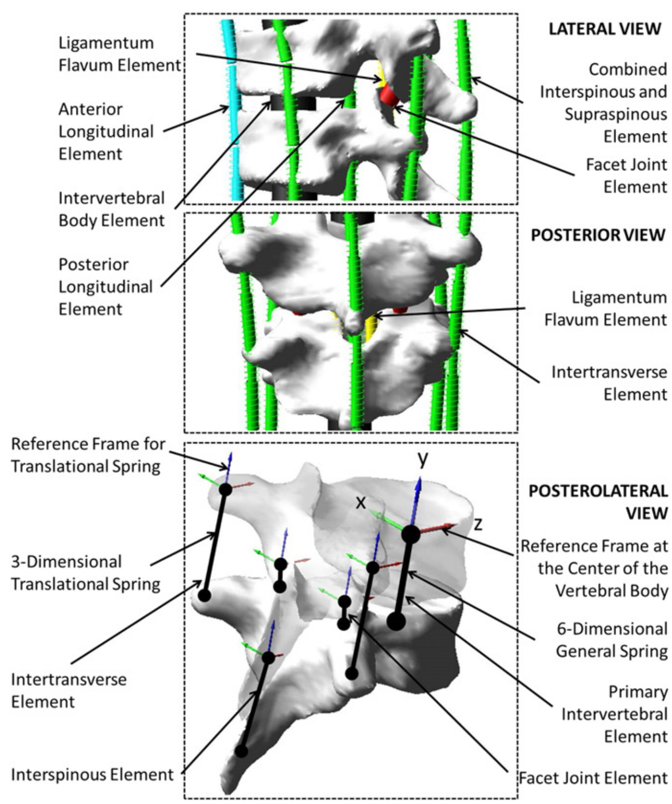

Fig. 1. Multibody modeling elements of a functional spinal unit and typical modeling element reference frame definition.