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Biomechanical analysis of iliac screw fixation in spinal deformity instrumentation

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ABSTRACT

Background: High rates of iliac screw fixation failures have been reported in spinopelvic instrumentations. The objective was to assess the iliac screw loads as functions of instrumentation variables.
Methods: Spinopelvic instrumentations of six neuromuscular scoliosis were simulated using patient-specific modeling techniques to evaluate the intra- and postoperative iliac screw loads as functions of instrumentation variables: the combined use of sacral screws, the uses of lateral offset connectors and cross-rod connectors, and the iliac screw insertion point and trajectory.
Findings: Sacral screws, lateral connectors and the iliac screw insertion point had significant effects on iliac screw axial forces (69–297 N) and toggle moments (0.8–2.9 N m) ($p < 0.05$). The addition of sacral screws made the iliac screw forces lower for some functional loads but higher for other functional loads, and resulted in an increase of intraoperative screw forces when attaching the rods onto these additional screws. When lateral offset connectors were used, the toggle moments were 16% and 25% higher, respectively for the left and right sides. Inserting iliac through the sacrum resulted in 17% lower toggle moment compared to insertion through the iliac crest. Cross-rod connectors had no significant effect on the intraoperative iliac screw forces. Postoperative functional loading had an important effect (additional 34% screw axial force and 18% toggle moment).
Interpretation: It is possible to reduce the iliac screw loads by adapting instrumentation variables and strategies. Reducing the loads could decrease the risk of failure associated with iliac screw fixations.

1. Introduction

Iliac screw fixation is used in spinopelvic instrumentation for deformity treatment as anchorage to ensure solid fusion and, subsequently, a balanced spine (Phillips et al., 2007). Pelvic fixation is also required when the patient has a pelvic obliquity greater than 15° (Modi et al., 2010). Many sacropelvic fixation techniques have been developed: the Galveston technique and the Dunn-McCarthy (S-Rod) technique, as well as transiliac screw, intrasacral rod, iliosacral fixation and iliac-screw-based techniques (Dayer et al., 2012; Miladi et al., 1997; Modi et al., 2010; Moshirfar et al., 2005b; Phillips et al., 2007; Schwend et al., 2003). It has been demonstrated that the use of iliac screws is safer and simpler compared to the standard Galveston technique (Phillips et al., 2007; Schwend et al., 2003).

Pelvic fixation presents surgical challenges because of, among other things, the complex local anatomy and large biomechanical loads (Moshirfar et al., 2005b). Iliac screws are located at the most distal end of instrumentation and, thus, are subject to large forces and moments generated intraoperatively by deformity-correcting maneuvers, and post-operatively by body weight, muscle contractions and functional loadings. Consequently, failures of iliac screws have been reported in spinopelvic instrumentations (Moshirfar et al., 2005a; Phillips et al., 2007; Tsuchiya et al., 2006). Phillips et al. reported that, of a group of 30 patients with neuromuscular scoliosis, 23.3% experienced complications directly related to iliac screws. Moreover, 13.3% of iliac screw-related complications were rod disengagement and 10% of these were screw loosening as revealed by radiolucencies (Phillips et al., 2007).

Pelvis fixation techniques using iliac screws involve inserting screws into the iliac crests and attaching them to spinal rods that are also anchored to vertebrae above (Dayer et al., 2012; Moshirfar et al., 2005b). There exist a variety of surgeon specific iliac screw fixation techniques. Iliac screws can be inserted solely into the iliac crest or through the iliac crest into the sacrum, therefore referred to as sacral alar iliac screws (Dayer et al., 2012). Lateral connectors linking the rods to the screws, cross-rod connectors joining the two rods, and sacral screws can also

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be used. The biomechanical effects of each of these instrumentation variables, and their combined effects on the screw loading remain poorly documented.

The objective of this study was to computationally analyze the forces and moments sustained by iliac screws as functions of instrumentation variables, body weight, and functional loadings.

2. Methods

2.1. Cases

To conduct this study six scoliosis instrumentation cases, under IRB approval, were selected. Patients with severe neuromuscular scoliosis who had instrumentation to the pelvis with iliac screws were included, while patients with prior surgery to the pelvis were excluded. The preoperative and postoperative clinical data of the patients are presented in Tables 1 and 2. Information required to define the simulations was extracted from the surgical documentation:

- Preoperative coronal and lateral radiographs with patient wearing a calibration belt, a rigid plate with four embedded radiopaque pellets of know coordinates (Aubin et al., 2011);
- Preoperative left and right side bending radiographs;
- Implant and rod geometric parameters and their mechanical properties (type and position of the screws, diameter and material of the rods);
- Instrumentation details;
- Postoperative coronal and lateral radiographs with the patient wearing a calibration belt (Aubin et al., 2011);
- Patient's weight.

2.2. Simulation of patient-specific spinopelvic instrumentation

2.2.1. Patient-specific biomechanical model of the spine and pelvic

The three dimensional (3D) spine and pelvis geometry of each of the six scoliosis patients was reconstructed using the calibrated preoperative coronal and lateral plain radiographs and reconstruction techniques (Cheriet et al., 2007). In brief, postero-anterior (PA) and lateral (LA) radiographs were taken with the patient wearing a calibration plate containing 4 radiopaque pellets. Fourteen anatomical landmarks (2 on the tips of each pedicle, and 4 on the periphery and one in the middle of each vertebral endplate) were manually identified on each vertebra and twenty-one on the pelvis (13 points on the sacrum (S1–S5 endplate centers, S1 articular processes, and sacral wings' extremities), 7 points on the iliac bones (iliac crest wing's superior, anterior, and posterior extremities), and 2 points on ipsilateral acetabulis) in the patient's biplanar radiographs. 3D coordinates were then computed using a self-calibration and optimization algorithm (Cheriet et al., 2007; Kadoury et al., 2007). A detailed geometry was then transformed and adjusted to match those landmarks using a free form deformation technique. The accuracy of this reconstruction technique is 3.3 mm on average (SD 3.8 mm) when considering all landmarks, but is much better for the pedicles (1.6 ± 1.1 mm) and the vertebral bodies (1.2 ± 0.8 mm) (Delorme et al., 2003). The reconstruction variations of a given scoliotic spine in terms of Cobb angles (≤0.6°), kyphosis and lordosis (≤6.7°) are below or within the error levels reported for equivalent 2D measurements used by clinicians (Delorme et al., 2003; Labelle et al., 1995). Screw tips and axes were reconstructed in 3D using the postoperative radiographs and the same technique.

To build the biomechanical numerical model of the spine, the vertebrae and pelvis were modeled as rigid bodies based on the assumption that the bone deformation was negligible compared to the intervertebral displacement during the instrumentation. In fact, the vertebral body is 100 times more rigid than the disc (Wang et al., 2012b). The intervertebral elements were represented with flexible connectors that used parametric curves to relate the intervertebral displacement and load. Intervertebral load–displacement data was acquired through mechanical tests on cadaveric spines (Gardner-Morse and Stokes, 2004; Panjabi et al., 1976). Then, these parametric curves were adjusted to the patient's specific spinal flexibility by adjusting the defining data such that side bending simulations reproduced the Cobb angles measured on the patient's side bending radiographs (Aubin et al., 2008; Petit et al., 2004). The

Table 1
Patient data and instrumentation details.

Case	Age	Sex	Weight	Diagnosis	Rods	
					Left	Right
1	9y 7 m	M	26.0 kg	Neuromuscular lumbar scoliosis, total involvement cerebral palsy	Stainless steel E = 197GPa, G = 75.7GPa Ø = 5.5 mm Length = 450 mm (L), 450 mm (R)	
2	11y	F	57.8 kg	Right thoracolumbar neuromuscular scoliosis, with microcephaly and cerebral palsy, seizure disorder	Cobalt chrome E = 240GPa, G = 100GPa Ø =5.5 mm Length = 495 mm (L), 495 mm (R)	
3	10y	F	35.5 kg	Spastic quadriplegia, neuromuscular scoliosis	Titanium E = 110GPa G = 37.6GPa Ø =5.5 mm Length = 376 mm	Cobalt chrome E = 240GPa G = 100GPa Ø =5.5 mm Length = 376 mm
4	10y 9 m	F	48.4 kg	Right scoliosis, T3 paraplegia from spinal cord injury	Cobalt chrome E = 240GPa, G = 100GPa Ø =5.5 mm Length = 372 mm (L), 375 mm (R)	
5	11y	F	36.1 kg	Spastic quadriplegia number, neuromuscular scoliosis	Cobalt chrome E = 240GPa G = 100GPa Ø =5.5 mm Length = 391 mm	Titanium E = 110GPa G = 37.6GPa Ø =6 mm Length = 394 mm
6	13y	M	41.9 kg	Static encephalopathy secondary to traumatic brain injury, neuromuscular scoliosis	Titanium E = 110GPa G = 37.6GPa Ø =5.5 mm Length = 640 mm	Cobalt chrome E = 240GPa G = 100GPa Ø =5.5 mm Length = 650 mm