Biomechanical simulations of scoliotic spine correction due to prone position and anaesthesia prior to surgical instrumentation

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Abstract

Background. The positioning of patients during scoliosis surgery has been shown to affect the scoliosis curve, yet positioning has not been exploited to help improve surgical outcome from a biomechanics point of view. Biomechanical models have been used to study other aspects of scoliosis. The goal of this study is to simulate the specific influence of the prone operative position and anaesthesia using a finite element model with patient personalized material properties.

Methods. A finite element model of the spine, ribcage and pelvis was created from the 3D standing geometry of two patients. To this model various positions were simulated. Initially the left and right supine pre-operative bending were simulated. Using a Box–Benkin experimental design the material properties of the intervertebral disks were personalized so that the bending simulations best matched the bending X-rays. The prone position was then simulated by applying the appropriate boundary conditions and gravity loads and the 3D geometry was compared to the X-rays taken intra-operatively. Finally an anaesthesia factor was added to the model to relax all the soft tissues.

Findings. The behaviour of the model improved for all three positions once the material properties were personalized. By incorporating an anaesthesia factor the results of the prone intra-operative simulation better matched the prone intra-operative X-ray. However, the anaesthesia factor was different for both patients. For the prone position simulation with anaesthesia patient 1 corrected from 62° to 47° and 43° to 31°. Patient 2 corrected from 70° to 55° and 40° to 32° for the thoracic and lumbar curves respectively.

Interpretation. Positioning of the patient, as well as anaesthesia, provide significant correction of the spinal deformity even before surgical instrumentation is fixed to the vertebra. The biomechanical effect of positioning should be taken into consideration by surgeons and possibly modify the support cushions accordingly to maximise 3D curve correction. The positioning is an important step that should not be overlooked by when simulating surgical correction and biomechanical models could be used to help determine optimal cushion placement.

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1. Introduction

Scoliosis is a three-dimensional (3D) deformity of the spine and trunk affecting between 1.5% and 3% of the population (Longstein, 1994). Scoliosis often occurs secondary to neuromuscular diseases but the most
common type is adolescent idiopathic scoliosis and the cause is unknown. Although scoliosis is recognized as a 3D deformity, the gold standard for quantifying the curve is the Cobb angle of the spine measured on a two-dimensional radiograph (Cobb, 1948). For severe cases of scoliosis, where the Cobb angle is greater than 45°, surgical correction is often required to promote vertebral fusion and straighten and stabilise the curve. During a typical surgical procedure, the patient is anesthetized, placed prone on the operating table, an incision is made down the centre of their back, and instrumentation is fixed to the vertebrae straightening the curve. An average total correction in the Cobb angle of about 57% is obtained after surgery but over half of that total correction (37%) is due to the positioning and anaesthesia and the remaining correction is due to the instrumentation (Delorme et al., 2000).

The positioning of patients with scoliosis is a critical step in the surgical procedure. The fathers of the modern prone spinal frame are Relton and Hall (1967), whom emphasize that the abdomen must remain free and pendulous during the surgery to minimize blood loss. Callahan and Brown (1981) described various positioning techniques for spinal surgery. They identified the three most important factors attributing to optimal position as stability of the spine, exposure required and physiological limitations. They recommended the Relton–Hall frame. Tables similar to the Relton–Hall frame are sometimes referred to as four post, chest roll, and the Jackson table (OSI, Union City, CA, USA).

The positioning of patients undergoing posterior spine surgery has been shown to have an effect on the sagittal alignment of the spine (Marsicano et al., 1998; Stephens et al., 1996; Tribus et al., 1999). It is important that normal standing lumbar lordosis is maintained after surgery. In general, a lordosis, similar to that found in standing, is maintained when the hips are flexed less than 30° (Peterson et al., 1995).

A few studies have shown that the spinal deformity can decrease prior to the insertion of posterior spinal instrumentation (Behairy et al., 2000; Delorme et al., 2000; Labelle et al., 1995). This improvement can be attributed to the prone positioning of the patients under anesthesia and surgical exposure. Other studies have shown that the trunk deformity of patients with adolescent idiopathic scoliosis can be altered while the patients are lying on the surgical table (Duke et al., 2002; Mac-Thiong et al., 2000). While the positioning of the patient is recognized as an important step in the surgery, it has not been exploited to help improve surgical correction from a biomechanics point of view.

Biomechanical models have been used to aid in the study of scoliosis biomechanics. In particular, models have been used to simulate the surgical instrumentation (Aubin et al., 2003; Gardner-Morse and Stokes, 1994; Ghista et al., 1988). Supine bending test X-rays aid in the assessment of patient flexibility, the selection of fusion levels, and are also useful in the prediction of surgical outcome. Recent studies have looked at personalizing the material properties based on the results of the bending test X-rays (Lafage et al., 2004; Petit et al., 2004). In all of these models, the effect of gravity on the positioning of the patient during surgery was not considered as an independent step. Stokes et al. (1999) noted that before biomechanical simulations can become a reliable tool to assist with pre-operative planning, the intra-op changes due to positioning and anaesthesia are some of the issues that must be addressed. A preliminary study showed that simulating the absence of gravity as a traction type load in the cranial direction is a first step in simulating scoliotic patients positioned on the operating table (Duke et al., 2004). This model was limited in that it did not simulate the anaesthesia.

The purpose of this study is to develop a biomechanical model that can simulate and analyse the specific influence of the prone operative position and anaesthesia on the correction observed prior to posterior scoliosis surgical instrumentation.

## 2. Methods

### 2.1. Patient description

Two adolescent idiopathic scoliosis (AIS) patients with different curve types were selected: one had a double, right thoracic, left lumbar, King II curve and the other had a single right thoracic curve, King III (King

<table>
<thead>
<tr>
<th>Clinical information</th>
<th>Cobb angles (°)</th>
<th>Standing</th>
<th>Left bending</th>
<th>Right bending</th>
<th>Prone intra-op</th>
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<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Age</td>
<td>18</td>
<td>Proximal thoracic</td>
<td>45</td>
<td>22</td>
<td>42</td>
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<tr>
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<td>Main thoracic</td>
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<td>11</td>
<td>42</td>
</tr>
<tr>
<td><strong>Patient 2</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Age</td>
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<td>Proximal thoracic</td>
<td>39</td>
<td>11</td>
<td>33</td>
</tr>
<tr>
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<td>II</td>
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<td>70</td>
<td>73</td>
<td>53</td>
</tr>
<tr>
<td>Weight (kg)</td>
<td>36.8</td>
<td>Lumbar</td>
<td>40</td>
<td>16</td>
<td>33</td>
</tr>
</tbody>
</table>

* Data not available.
et al., 1983). A summary of the patient’s clinical information can be found in Table 1.

2.2. Model description

The 3D geometry of the patients was obtained from a bi-Di-planar reconstruction technique (Delorme et al., 2003). From this geometry a patient personalized finite element model of the spine was created (Aubin et al., 1995). In summary this model contains 1440 nodes and 2974 elements representing the osseo-ligamentous structures of the spine, ribcage, sternum and pelvis. The vertebrae, intervertebral disks, ribs, sternum, pelvis and cartilage are represented by 3D elastic beams. The costo-vertebral, costo-transverse and zygapophyseal joints are modelled in greater detail with shell, multilinear and point-to-surface contact elements. The vertebral and intercostal elements are represented by 3D elastic spring elements. The initial material properties were obtained from literature and experimental trials from cadaver specimens but these were not yet personalized to the patient (Descrimes et al., 1995).

The reference plane used in the model is that which is defined as the global (body) coordinate system by Stokes (1994). The origin is defined at S1, the z-axis is the gravity line when standing, the y-axis is a line parallel to the plane containing the anterior superior iliac spines pointing in the left direction and the x-axis is orthogonal to these two generally in the anterior direction. This reference is held fixed during the simulations maintaining the direction of the z-axis in the cranial direction.

To this initial model various simulations were performed as described in detail in the following sections. The main idea is to simulate the prone position using a model with patient personalized properties. To personalize the material properties, supine pre-operative bending simulations are performed with different material properties and the results are compared and optimized to the supine pre-operative bending X-rays. Once the patient personalized material properties are found, the patient is modelled in the prone surgical position and the results of this simulation are compared to the X-rays routinely taken during surgery to verify screw placement. The discrepancy between the prone simulation and the intra-op X-ray can be attributed to anaesthesia or limitations in the model. Finally, all the soft tissue material properties in the model are multiplied by an anaesthesia factor ($\alpha$) until the model behaviour best matches the intra-operative X-ray.

2.3. Personalization of the mechanical properties

The personalization of the mechanical properties is done using the flexible tests routinely done prior to the surgery using side bending films. The patient lies on their back (supine) and they are asked to bend as far as possible to the left and right sides; the correction of the scoliotic spine during this test is used to characterise the spine flexibility.

As an initial step in the pre-operative bending simulation the supine position was simulated. The supine position of the patient was simulated by constraining the nodes of the pelvis corresponding to the posterior iliac spine in all directions. Then, to simulate contact of the patients back with the table the most posterior points of three ribs on each side were deflected in the x direction until they were level with the pelvis and then constrained from moving in the x direction.

To simulate the effect of gravity on the model, forces were then applied to the spine. The mass for each trunk slice at every vertebral level were taken from Liu et al. (1971) and averaged for the entire trunk as well as by thoracic and lumbar regions for simplification. First, to simulate gravity acting in the negative x direction a force equivalent to 2% of the patient’s body mass was applied to the centre of each vertebra in the negative x direction. Since the model is made from standing X-rays, it is necessary to “remove” the gravity that is naturally compressing the spine when standing which, is absent from the longitudinal axis when lying prone. To do this a “traction type” load was applied in the positive z direction equivalent to 1.6% and 2.8 % body mass for each thoracic and lumbar vertebra respectively. In addition to the mass of each trunk slice, the mass of the head and arms (18.6%) were added to T1 in the negative x direction (Winter, 1990). Finally, since the centre of mass of each trunk slice is not located at the centre of the vertebra, corresponding moments are applied. For a normal patient these would include only moments in the negative y direction but, for a scoliotic patient, moments around the x-axis equivalent to the mass of the slice multiplied by deviation of the vertebral centre from the central sacral line (z-axis) are also included.

To simulate the supine side bending all the boundary conditions and forces as described above are applied on the patient. In addition, T1 is displaced in the y direction equivalent to the displacement measured on the bending X-rays.

Simulation of this bending with the initial material properties resulted in discrepancies of the spine curves (as measured by the Cobb angles) observed from the simulation and those observed on the patients X-rays. In order to personalize the material properties of the model a Box–Benkin experimental design was created with four manipulated variables at three modalities for a total of 27 runs. The manipulated variable included a multiplication factor for the material properties of the intervertebral disks at four different sections of the spine: proximal thoracic, main thoracic, thoracolumbar and lumbar. Using a method similar to Petit et al. (2004) the three modalities are 0.2 (flexible), 1 (neutral) and 8 (stiff). The results of the 27 runs are tabulated in
Statistica (StatSoft Inc. Tulsa, USA) where equations to represent the various Cobb angles are found. The Cobb angles that were optimised were the proximal thoracic (PT), the main thoracic (MT) and the lumbar (L). The model is then optimised to best represent the bending X-rays and the multiplication factor for the material properties for the four different sections of the spine are found.

In order to optimize the Cobb angles during bending simulations the difference between the simulated bending geometry and the actual 3D geometry was minimised. The difference was calculated from Eq. (1) as the weighting of the sum of the squares of the differences from the actual Cobb angles measured from the left (l) and right (r) bending X-rays and the Cobb angles measured from the simulated patient.

\[
\text{Difference} = 0.1^2(\text{PT}_\text{rActual-Simulated})^2 + 0.3^2(\text{MT}_\text{rActual-Simulated})^2 + 0.1^2(\text{L}_\text{rActual-Simulated})^2 + 0.1^2(\text{PT}_\text{lActual-Simulated})^2 + 0.3^2(\text{L}_\text{lActual-Simulated})^2
\]

where \(\text{PT}_\text{r}\) is equal to the proximal thoracic Cobb angle during right bending. Greater weighting was given to the main thoracic curve (MT) during right bending, as well as to the lumbar curve (L) during left bending, as those are more significant clinically. The multiplication factors for the four different sections of the spine were selected based on which combination gave the minimum difference. Once the best combination was found the supine, left bending and right bending were again simulated but with the new patient personalized material properties.

2.4. Simulation of the intra-operative prone position

The prone position of the patient in the operating room was simulated by constraining the nodes of the pelvic corresponding to the most predominant nodes on the left and right sides of the anterior iliac spine in all directions as shown in Fig. 1. To prevent rotation, a second node on the iliac spine was constrained in the \(x\) direction. Then, to simulate contact of the patient’s ribs with the cushions, the most anterior points of three ribs on each side were constrained from moving in the \(x\) direction.

Gravity was added to the model as described in the supine position except that a force equivalent to 2% of the patient’s body mass was applied to the centre of each vertebra in the positive \(x\) direction, as opposed to the negative \(x\) direction. The “traction type” loads and moments were applied as described above.

The results of these simulations were then compared with the 3D geometry reconstructed from the routine intra-op X-rays taken after screw placement before rod insertion. Since these X-rays are taken on a standard \(36 \times 43\) cm (14 × 17 in.) film the entire spine and pelvis is not visible. Also, the lordosis and kyphosis cannot be measured intra-operatively. To compare the simulations to the 3D reconstructed intra-op geometry in the same coordinate system the most inferior vertebra in the intra-op spinal geometry was translated to its partner vertebra in the simulation geometry. A triangular plane was then created between the most inferior, most superior and centre vertebra. A transformation matrix was then applied to the 3D geometry reconstructed from intra-op X-rays so that the triangular planes of the intra-op geometry and the simulation geometry were co-planar.

The prone position was first simulated with the model’s original material properties to provide a basis for comparison. The prone position was then simulated with the personalized material properties obtained from the bending tests. Finally, a relaxation factor (\(x\)) to represent the anaesthesia was applied to the model as described below.

As it is known, the anaesthesia involves the relaxation of muscles and the reduction of the spine stiffness. We introduced an anaesthesia factor to modify the Young’s modulus of the spine functional units. The Young’s moduli of all the soft tissues in the model were first multiplied by relaxation factors varying from 0.2 to 1.2 at 0.2 increments. Iterative simulations were performed to minimise the difference in the vertebral centroid positions between the 3D intra-op geometry and the 3D
simulation geometry. A local minimum was found, and further iterations were performed around that local minimum.

3. Results

3.1. Personalization of the mechanical properties

The results of the supine simulation are presented in Table 2. The initial simulation showed that, for the first patient, the thoracic spine corrected from 62° to 50° (20%) and the lumbar spine corrected from 43° to 35° (18%). For the second patient the thoracic spine corrected from 70° to 61° (14%) and the lumbar spine corrected from 40° to 37° (7%). In both patients there was also a decrease in kyphosis and lordosis in the sagittal plane. For the first patient the kyphosis decreased from 41° to 32° and the lordosis decreased from 37° to 24°. For the second patient the decrease was 41–33° and 18–13° for the kyphosis and lordosis respectively.

After the personalized material properties are obtained the simulations of the supine position are repeated with the new properties. As seen in Table 2 there is slight improvement in the amount of thoracic correction for both patients. For the first patient, the thoracic spine corrected from 62° to 47° (25%) and the lumbar spine corrected from 43° to 33° (24%). For the second patient there was less of a difference as the thoracic spine corrected from 70° to 59° (15%) and the lumbar spine corrected from 40° to 38° (6%).

The results of the bending simulations can be found in Table 2. As expected, personalization of the material properties improved the behaviour of the bending simulations for each patient. There was an overall reduction of the weighted sum of the squared of the differences.

For the first patient the thoracic spine corrected from 62° to 35° (44%) on right bending. The lumbar spine corrected from 43° to 11° (74%) on left bending. Initial simulations were able to produce corrections to 49° and 30° for right and left bending respectively. After personalization of the properties corrections to 44° thoracic and 23° lumbar, were obtained. The total difference, calculated from Eq. (1), was reduced from 263 mm² to 170 mm².

For the second patient the thoracic Cobb corrected from 70° to 53° (24%) on right bending. The lumbar Cobb corrected from 40° to 16° (60%) on left bending. Initial simulations were able to produce corrections to 59° and 17° for right and left bending respectively. After personalization of the disk properties corrections to 57° and 19° were obtained. The difference, calculated from Eq. (1), was reduced from 113 mm² to 89 mm².

For both patients, a combination of material properties was not found that could completely satisfy both the thoracic and lumbar curves on left and right bending. The focus was therefore placed on minimising the error. For both patients, flexibility of the spine was decreased in the upper thoracic and thoracic regions while the stiffness of the lumbar spine was increased. For the first patient the flexibility decreased in the thoracolumbar region contrary to the second patient whose spine was made more rigid in this region. In summary, for the first patient the multiplications factors were 0.2, 0.2, 0.2 and 8 for the proximal thoracic, main thoracic, thoracolumbar, and lumbar sections respectively. For the second patient the factors were 0.2, 0.2, 2, 2 for the same respective regions.

3.2. Simulation of the intra-operative prone position

Fig. 2 represents the 3D reconstruction of the vertebral body line for the patients standing (X-ray), initial prone simulation, prone simulation with personalized material properties and anaesthesia factor, and the 3D reconstruction of the prone X-ray. The Cobb angles and difference in the vertebral centroid positions are in Table 3.

For the first patient the correction while lying prone in the operating room was 26% (62°–46°) and 37% (43°–27°) for the thoracic and lumbar curves respectively. The simulation with the original material properties was only able to provide 17% (62°–52°) and 22% (43°–33°) correction for the thoracic and lumbar regions. By using the optimised intervertebral disk material properties the behaviour of the model improved slightly. Finally with the anaesthesia factor, of 0.9 for this patient, the thoracic curve was corrected to 47° while the lumbar curve was 31°. Fig. 3 shows the results of the iterations used to determine the anaesthesia.

Table 2
Results of the supine and bending simulations

<table>
<thead>
<tr>
<th></th>
<th>Standing</th>
<th>Supine simulation</th>
<th>Left bending</th>
<th>Right bending</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X-ray</td>
<td>Sim 1</td>
<td>Sim opt</td>
<td>X-ray</td>
</tr>
<tr>
<td>Patient 1</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thoracic</td>
<td>62</td>
<td>50</td>
<td>47</td>
<td>59</td>
</tr>
<tr>
<td>Lumbar</td>
<td>43</td>
<td>35</td>
<td>33</td>
<td>11</td>
</tr>
<tr>
<td>Patient 2</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thoracic</td>
<td>70</td>
<td>61</td>
<td>59</td>
<td>73</td>
</tr>
<tr>
<td>Lumbar</td>
<td>40</td>
<td>37</td>
<td>38</td>
<td>16</td>
</tr>
</tbody>
</table>

Sim 1 is without the personalized material properties. Sim opt is with the optimal patient personalized properties.
factors for both patients. The total difference in the vertebral centroid positions was originally 167 mm and reduced to 97 mm and 96 mm for the personalized material properties and the anaesthesia simulations respectively.

For the second patient the actual curves corrected from 70° to 46° (34%) and from 40° to 21° (48%) for the thoracic and lumbar curves respectively. The initial simulation demonstrated weak correction to only 63° (10%) and 37° (11%). The optimisation of the material properties did not improve the behaviour of the model of this patient. For this patient the anaesthesia factor was 0.15. With this large reduction on the material properties of the soft tissues better correction was observed. With the anaesthesia factor the simulated correction in the prone position was 55° and 32° for the thoracic and lumbar curves respectively. The total difference in the vertebral centroid positions was originally 80 mm and reduced to 76 mm and 30 mm for the personalized material properties and the anaesthesia simulations respectively.

Fig. 2. Spine curves of the two patients obtained from the radiographic acquisitions standing and prone, and from the finite element model simulations. All data are in mm.
respectively. Although there is still some discrepancy in the Cobb angles, visual inspection on Fig. 2 confirms that the difference in the vertebral centroid positions is small and there is good agreement with the simulated prone position and the intra-operative X-ray geometry.

### 4. Discussion

The purpose of this study was to create a biomechanical model that can simulate the specific influence of the prone operative position and anaesthesia on the correction observed prior to posterior instrumentation.

First the supine position was simulated without and with the personalised material properties. For the first patient correction was around 25% for both curves however, for the second patient the thoracic curve corrects 15% and the lumbar only corrected 6%. The correction for patient 2 is lower than the correction in the supine position of 23% reported by Klepps et al. (2001). Comparing those studies to those of Behairy et al. (2000) and Delorme et al. (2000) who found between 33% and 37% correction in the operating room prior to instrumentation, there appears to be another 10–15% correction due to anaesthesia and opening of the patient. The model supports this showing between 6% and 14% increase in curve correction after the anaesthesia factor was added to the model. Polly and Sturm (1998) reported even more correction stating that there is 31% correction due to positioning (presumably under anesthesia), an additional 7–17% correction is due to the soft tissue resection and far greater correction is achieved if an anterior release discectomy is performed.

While looking at the results for the anaesthesia factor iterations shown in Fig. 3 it is interesting to note that there was insufficient correction if the model of the spine is too stiff. Contrary, if the spine is too supple there was a flattening in the sagittal plane and the spine became too elongated. This elongation is represented by an increased difference in the vertebral centroid positions for low anaesthesia factors.

For these two patients the anaesthesia factor was variable ranging from 0.9 to 0.15. The low value of the 2nd patient is perhaps due to the fact that the material properties of patient 2 were not fully optimised due to insufficient bending. The correction of their thoracic curve at bending was 24% while the correction intra-op was far greater at 35%. This is perhaps due to the fact that the personalized properties were then based on an insufficient bending and do not truly reflect the flexibility of the patient. Active patient participation is a known limitation on the bending test and perhaps tests such as the fulcrum bending should be used to get a better idea of the flexibility of the patient (Cheung and Luk, 1997).

There may also be an age factor that was not taken into account. Patient 1 is 18 years old while patient 2 is only 11. It has been shown that flexibility decreases

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**Table 3**

<table>
<thead>
<tr>
<th>Patient 1</th>
<th>Thoracic (°)</th>
<th>Lumbar (°)</th>
<th>DVCP (mm)</th>
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<tr>
<td>Patient 1 Sim a</td>
<td>47</td>
<td>31</td>
<td>96</td>
</tr>
<tr>
<td>Patient 2 Thoracic (°)</td>
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<td>na</td>
</tr>
<tr>
<td>Patient 2 Lumbar (°)</td>
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<td>21</td>
<td>0</td>
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<tr>
<td>Patient 2 DVCP (mm)</td>
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<td>80</td>
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<tr>
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<tr>
<td>Patient 2 Sim a</td>
<td>55</td>
<td>32</td>
<td>80</td>
</tr>
</tbody>
</table>

Sim 1 is without the personalized material properties. Sim opt is with the optimal patient personalized properties. Sim a is with the anaesthesia factor. DVCP is the difference in the vertebral centroid positions.

**Fig. 3.** The difference in the vertebral centroid positions (DVCP) versus the anaesthesia factor for the two patients.
slightly with age (Torell et al., 1985). It was the hope that personalizing the material properties with respect to the bending test would take age factors into account but this does not appear to be the case. Ideally, the model could be personalized based on data available at the pre-surgical visit such as age and bending X-rays. It was relatively easy to adjust the prone simulation by changing the anaesthesia factor to obtain the desired prone geometry. However, if the anaesthesia factor is to be a useful tool there must be a way to predict it from the pre-surgical data.

The positioning of the patient in the prone position decreases the deformity in the frontal plane, which is a desired effect, but thoracic kyphosis and lumbar lordosis are also decreased, which can be an undesirable effect since a certain number of scoliotic curves are already hypokyphotic and hypolordotic. Therefore surgeons should take the appropriate measures to counteract this reduction of sagittal curves by; first providing appropriate support of the thoracic area by the pads to increase the thoracic kyphosis and also appropriate position of the lower limbs to increase the lumbar lordosis through hip extension. These precautions should help obtain a more appropriate 3D correction with curve correction in the frontal plane and preservation or improvement of the normal sagittal profile of the spine.

Since the positioning of the patient provides an average of about 37% correction and the total correction is about 57% the positioning of the patient is a significant step in the correction (Delorme et al., 2000). Excessive forces and/or inadequate correction has been reported in some models that simulate the surgical instrumentation (Aubin et al., 2003). This could be partially due to the fact that positioning is not taken into account. It is recommended that all models that simulate the surgical correction should first include a step to simulate the surgical positioning of the patient.

Even with the personalized material properties, the simulation of the gravitational forces and the simulation of the anaesthesia there is still a discrepancy between the model simulations and the actual geometry measured from the intra-operative X-rays. There are still some factors that are not taken into account in the model. One limitation is that the muscles were not directly included in the model. The muscles are indirectly taken into account as the segmental stiffness of the spine was adjusted based on the bending tests which incorporated the muscles. Another limitation that could play a role is the position of the legs as this influences the lordosis. Future simulations will analyse the effect of the positioning of the pelvis in an attempt to take the positioning of the legs into account. The positioning of the head and the arms, which was not simulated, could also slightly alter the trunk geometry. The discrepancy, between the simulated and actual geometry, highlights the limitations of using a model to simulate individual patients. However, the model is still useful to show the influence of different factors, specifically gravity and anaesthesia, involved in patient positioning.

Since passive prone patient positioning reduces the magnitude of scoliotic curves significantly, this study suggests that further correction and improvement could be obtained by modifying the placement of the various support areas and by adding additional passive and active correction forces. Surgeons should be aware of this possibility and modify support areas accordingly to maximise 3D curve correction during patient positioning. Active correction forces applied to the trunk during surgery may also add additional 3D curve correction in order to facilitate instrumentation of the spine. Using this finite element model as a base the location of the different support cushions and active corrective cushions could be virtually modeled to see their effect on 3D correction.

5. Conclusion

A finite element model of the spine, ribcage and pelvis was used to simulate the position of patients during scoliosis surgery. First, the material properties of the model were personalized based on the pre-operative side bending X-rays of the patient. The results showed that there was significant correction due to the positioning and that this correction was increased by taking into account patient personalized material properties and anaesthesia. The positioning of the patient, as well as the anaesthesia, provides significant correction of the spinal deformity even before surgical instrumentation is fixed to the vertebra. This study shows that the positioning of the patient has a biomechanical effect on the scoliotic deformity and should therefore be included in any computer simulations of scoliosis surgery. Surgeons should take extra care when positioning the patient and modify support areas to maximise 3D correction during patient positioning.

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