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THE DESIGN AND BIOMECHANICAL ANALYSIS OF A DYNAMIC POSITIONING FRAME FOR SCOLIOSIS SURGERY

KAJSA DUKE

INSTITUT DE GÉNIE BIOMÉDICAL
ÉCOLE POLYTECHNIQUE DE MONTRÉAL

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Cette thése intitulée:

THE DESIGN AND BIOMECHANICAL ANALYSIS OF A DYNAMIC POSITIONING FRAME FOR SCOLIOSIS SURGERY

présenté par: DUKE Kajsa

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a été dûment acceptée par le jury d'examen constitué de:

Mme. VILLEMURE Isabelle, Ph.D., présidente

M. DANSEREAU Jean, Ph.D., membre et directeur de recherche

M. AUBIN Carl-Éric, Ph.D., membre et co-directeur de recherche

M. DELORME Sébastien, Ph.D., membre

M. OUELLET Jean; MD., examinateur externe

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Résumé

La scoliose est une déformation tridimensionnelle (3-D) de la colonne vertébrale et de la cage thoracique. Pour les cas sévères de scoliose, une chirurgie est requise. Le positionnement est une étape importante dans la procédure chirurgicale et les études ont démontré une correction spontanée d'environ 37% pendant la chirurgie et une correction finale de 57% avec l'instrumentation. Pour étudier les aspects de la scoliose, les modèles biomécaniques furent développés mais l'effet du positionnement n'a pas été étudié.

L'objectif global de ce projet est de développer un nouveau système de positionnement dynamique (SPD) qui sera utilisé pendant la chirurgie pour améliorer le positionnement des patients scoliotiques. L'hypothèse posée est que l'utilisation de SPD peut améliorer la correction de la déformation scoliotique comparablement au système Relton-Hall présentement utilisé. En parallèle, la deuxième hypothèse est que les structures osseuses du patient, l'effet de la gravité, l'anesthésie et la position chirurgicale peuvent être modélisés et que ce modèle peut recommander les ajustements du SPD.

Un modèle par éléments finis de la colonne vertébrale, de la cage thoracique et du bassin de deux patientes, a été construit à partir des radiographies. Les propriétés mécaniques du modèle étaient personnalisées avec les radiographies d'inflexions latérales. Pour simuler la position décubitus ventral, les conditions limites furent appliquées pour fixer le bassin et six côtes dans le plan coronal. La gravité a été simulée par application des efforts distribués sur chaque vertèbre. Un facteur d'anesthésie a été appliqué aux tissus mous. Les simulations ont été validées à l'aides des radiographies de flexion latérales et les radiographies prises durant l'opération. Les résultats montrent que le comportement du modèle s'améliore quand les propriétés mécaniques sont personnalisées Avec le facteur d'anesthésie, l'angle de Cobb de la première patiente été corrigé de 62° à 47° et celui de la deuxième patiente a été corrigé de 70° à 55°.

Le SPD a été conçu et fabriqué pour améliorer le positionnement des patients pendant la chirurgie. En bref, le SPD est composé de quatre coussins concaves pour mieux épouser la forme du tronc et qui pourraient être facilement déplacés sur les rails. Il y a aussi des coussins externes qui peuvent être attachés aux rails de la table pour aider à corriger la déformation du tronc.

Pendant l'étude clinique, onze patients scoliotiques ont été positionnés debout, sur le système de positionnement Relton-Hall, sur les coussins de base du SPD et sur les coussins de base du SPD avec l'application des efforts manuels. Les mesures de la géométrie du tronc et les mesures de la pression entre les patients et les coussins ont été acquises. La longueur de la colonne était significativement plus grande sur le SPD que sur le système Relton-Hall. La cyphose était moins aplatie sur le SPD et la torsion de la colonne améliorée. Les pressions étant plus élevées sur le SPD par rapport au système Relton-Hall, des modifications aux coussins ont été apportées avant de procéder en salle d'opérations.

Suite aux modifications des coussins, trois patients ont été positionnés sur le SPD pendant une chirurgie pour la scoliose. Pour des raisons de sécurité, les mesures de pressions ont été comparées entre le SPD et le système Relton-Hall le jour avant la chirurgie. Les pressions pour les trois patients étaient égales ou moindre sur le SPD avec les coussins modifiés en comparaison au système Relton-Hall. Pendant la chirurgie, les pressions plus élevées ont été notées; celles-ci correspondaient avec les rougeurs sur la crête iliaque droite de deux des patients, mais aucune complication majeure n'a été observée. La correction moyenne pendant la chirurgie était de 40% et 68% pour les courbes thoraciques et de 75% pour les courbes lombaires. Les résultats des trois premiers cas sont encourageants et démontrent la fonctionnalité de ce nouveau système. Une étude plus étendue est en cours pour déterminer s'il y a une amélioration significative de la correction.

Pour l'étude finale, le modèle par éléments finis présenté au chapitre 3 a été modifié. Six paramètres du modèle ont été modifiés dans un plan d'expérimental Box-

Hunter pour recommander l'emplacement des coussins. Les paramètres ont inclus : l'inclinaison du bassin, le positionnement et la hauteur des coussins de la poitrine, la force thoracique latéral, la force lombaire latéral et la force sur la gibbosité. Dix indices cliniques ont été mesurés sur le modèle géométrique. Les indices cliniques ont été optimisés individuellement et simultanément. Les résultats ont démontré que plusieurs paramètres du modèle avaient un effet sur les indices géométriques. L'optimisation d'un seul indice géométrique à la fois a souvent causé une détérioration majeure des autres indices géométriques. La correction globale est de 75% sur le SPD comparablement à 55% sur le système Relton-Hall. Le modèle a démontré que le positionnement du patient pendant la chirurgie est une étape très importante qui pourrait être exploitée pour améliorer la géométrie 3-D du patient.

L'objectif global de ce projet était de développer un nouveau SPD pouvant être utilisé pendant la chirurgie pour améliorer le positionnement des patients scoliotiques. L'étude clinique a démontrée que quelques indices de la géométrie du tronc sont améliorés quand le patient est positionné sur le SPD. En parallèle, un modèle par éléments finis a été modifié pour simuler le positionnement per-opératoire des patients scoliotiques et pour recommander les ajustements du SPD. La méthode novatrice développée avec ce modèle pourrait être utilisée pour prescrire la localisation des coussins d'un SPD qui optimisera la position couchée du patient scoliotique pendant la chirurgie afin d'améliorer et/ou faciliter la correction de la scoliose.

Abstract

Scoliosis is a three dimensional deformity of the spine and trunk. In severe cases of scoliosis, surgery is required. Positioning of patients is an important step in the surgical procedure and has been shown to correct the scoliotic curve by 37% prior to the instrumentation which then fixes the final correction at 57%. Biomechanical models have been used to help study scoliosis but these models have not been used to study patient positioning.

The general objective of this work is to develop and validate a dynamic positioning frame (DPF) that will be used in the operating room to improve the surgical correction of spinal deformities. It is hypothesised that the utilization of the DPF can improve the correction of the spinal deformity and the overall trunk geometry compared to the Relton-Hall frame used in the conventional approach. In parallel, it is hypothesised that the osseo-ligamentous structures of the patient, the effect of gravity, anaesthesia and the surgical position can be modeled, and such computer models can be used to recommend placement of the DPF components on a patient for correction of the spinal deformities.

A finite element model of the spine, ribcage and pelvis was created from the 3-D standing geometry of two patients. The material properties of the model were personalised 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 3-D geometry was compared to x-rays taken intra-operatively. Finally an anaesthesia factor was added to the model to relax all the soft tissues. Results show that the behaviour of the model improved once the material properties were personalised. With the anaesthesia factor patient 1 corrected from 62° to 47° and patient 2 corrected from 70° to 55°.

In order to maximise the amount of correction observed in the prone position a DPF was designed. The DPF is composed of a base of four cushions that contour to the sides of the patient that can be easily displaced on rails. There are also external correction cushions that can be used to apply force to the trunk deformities. A clinical evaluation was performed of 11 unanesthetised scoliotic patients; standing, lying prone on the Relton-Hall frame, prone on the DPF and prone on the DPF with external manually applied forces. Trunk geometry and pressure measurements were recorded. The results showed a statistically significant lengthening of the spine, improvement in the torsional deformity (delta hump) and less of a reduction in kyphosis when the patient was lying prone on the DPF compared to the Relton-Hall frame. Higher pressures were recorded on the DPF when compared to the Relton-Hall frame necessitating modifications to the DPF cushions.

Modifications were made to the cushions and the improved DPF was tested on three patients in the operating room. As a safety precaution, pre-operative pressures were measured the day before with the patient lying on both the Relton-Hall frame and the DPF. The pre-operative pressures on the DPF were equal too or less than the pressure on the Relton-Hall frame. During the surgery higher pressures were recorded, which corresponded to reddening of the patient's skin, but no complications due to positioning were noted. Average correction of the both Cobb angles was 40% during surgery and 68% and 75% for the thoracic and lumbar curves respectively post-operatively. This study showed the feasibility of using the DPF during scoliosis surgeries but a larger clinical trial is underway to determine if there is a significant improvement in correction achieved with this system.

The final study involved modifications to the finite element model to simulate the DPF and test various adjustment parameters and recommend placement of the DPF components prior to, and during spine surgery. The prone position of one scoliotic patient was simulated using a patient personalised finite element model. Six DPF model parameters were modified in a controlled Box-Hunter experimental design while ten geometric measures were recorded. The parameters include: pelvic inclination, chest cushion location, chest cushion height, rib hump force, lateral thoracic force and lateral lumbar force. The geometric measures were individually and simultaneously optimised, while corresponding model parameters were noted. The results showed that every model parameter had a significant effect on at least five of the geometric measures. Optimising a single measure individually often resulted in the deterioration of other measures. Simultaneous optimisation resulted in improved overall correction of the patient's geometry by 75% compared to only 55% when positioned on the Relton-Hall frame. This study confirms that the positioning of the patient is an important step that can be exploited

The general objective of designing a positioning frame that can be used to improve the positioning of the patient during surgery is a novel concept. The clinical study showed that the trunk geometry could be improved while the study in the operating room showed the feasibility of the DPF. In parallel, finite element models were modified to simulate patient positioning and recommend adjustments of the DPF components. The innovative methods developed with the model can be used to prescribe cushion adjustments of the DPF which will optimise the position of the scoliotic patients during surgery.

Condensé en français

Revue des connaissances

La scoliose est une déformation tridimensionnelle (3-D) de la colonne vertébrale et de la cage thoracique. Même si la scoliose est connue comme une déformation 3-D, le diagnostic est fait quand l'angle de Cobb mesuré sur une radiographie postero-antérieure est plus grand que 10° (Cobb 1948).

Pour les cas sévères de scoliose (angle de Cobb > 45°), une chirurgie est requise. Pendant une chirurgie typique, la patiente est positionnée en décubitus ventral, une incision est faite le long de son dos et l'instrumentation est fixée aux vertèbres pour redresser la courbure et encourager la fusion osseuse. Les systèmes de positionnement plus fréquemment utilisés sont basés sur un modèle de quatre coussins conçus par Relton et Hall (Relton et al. 1967). L'idée principale de ces systèmes est de supporter les patients sans exercer de pression sur l'abdomen pour diminuer les pertes de sang.

Les études précédentes ont démontré que, pendant la chirurgie, il y a une correction spontanée de l'angle de Cobb d'environ 37% avant même que le rachis soit instrumenté. Cette correction est attribuée au positionnement (gravité), à l'anesthésie et à l'ouverture. Après la chirurgie, l'angle de Cobb est corrigé de 57% environ (Delorme et al. 2000). Dans le plan sagittal, la cyphose, et surtout la lordose, sont influencées par le positionnement des jambes. En général, une lordose équivalente à la lordose en position debout est maintenue quand les jambes sont en extension ou légèrement fléchies de moins de 30° (Peterson et al. 1995).

Pour étudier les aspects biomécaniques de la scoliose, des modèles par éléments finis et par multi-corps flexibles furent développés (Aubin et al. 2003, Gardner-Morse et al. 1994). Pour personnaliser les propriétés mécaniques de ces modèles, des radiographies en flexion latérale ont été utilisées (Lafage et al. 2004, Petit et al. 2004). Les limitations de ces modèles sont les effets de la gravité, et de l'anesthésie, sur le positionnement du patient qui n'ont pas été modélisés. Les modèles par éléments finis ont aussi été utilisés pour étudier et optimiser les effets biomécaniques des corsets (Gignac et al. 2000).

Objectifs et hypothèse

L'objectif global de ce projet est de développer un nouveau système de positionnement dynamique (SPD) qui va être utilisé pendant la chirurgie pour améliorer le positionnement des patients scoliotiques. En parallèle, pour valider et recommander des ajustements au SPD, un modèle par éléments finis va être modifié pour simuler le positionnement per-operatoire des patients scoliotiques. Pour satisfaire cet objectif global, deux hypothèses ont été posées.

H1 : L'utilisation du SPD peut améliorer la correction de la déformation scoliotique comparable au système Relton-Hall présentement utilisé.

H2 : Les structures osseuses du patient, l'effet de la gravité, de l'anesthésie et de la position chirurgicale peuvent être modélisés et ce modèle permet de recommander des ajustements au SPD.

Pour vérifier ces hypothèses, la réalisation de quatre objectifs est présentée.

Objectif 1 : Raffiner un modèle par éléments finis en incorporant les propriétés personnalisées pour simuler le positionnement et l'anesthésie du patient.

Objectif 2 : Déterminer les critères de design et les incorporer dans le design et la fabrication du SPD pour la salle d'opération.

Objectif 3 : Tester le SPD sur un groupe de patients scoliotiques tout d'abord en milieu clinique, puis en salle d'opération.

Objectif 4 : Modifier le modèle conçu dans l'objectif 1 pour simuler l'utilisation du SPD en testant l'ajustement des paramètres et recommander le placement optimal des coussins avant et pendant la chirurgie à partir des résultats de la simulation.

Pour résoudre ces objectifs, trois articles et quelques sections supplémentaires sont présentés dans cette thèse. Les trois sections suivantes résument les chapitres 3, 4, et 5 de ce travail.

Chapitre 3 : Analyse biomécanique des patients scoliotiques dans la position décubitus ventral.

Un modèle par éléments finis (Ansys) de deux patientes (P1 et P2) a été reconstruit à partir des radiographies. Les patientes sélectionnées étaient des candidates pour la chirurgie (Cobb thoracique 62° (P1) 70° (P2). Pour simuler l'interaction entre la patiente et les coussins Relton-Hall, les conditions limites ont été appliquées pour fixer le bassin et six côtes dans le plan coronal. La gravité a été simulée par l'application des efforts distribués sur chaque vertèbre. Les efforts ont été appliqués dans la direction postéro-antérieure et aussi dans la direction crânienne. Les efforts dans la direction crânienne compensent l'effet de compression de la gravité, présent quand la patiente est debout. Finalement, un facteur d'anesthésie a été appliqué aux tissus mous. Les simulations étaient validées en comparaison avec les radiographies en flexion latérale et aussi les radiographies prises en per-opératoire.

Les résultats montrent que le comportement du modèle est amélioré lorsque les propriétés mécaniques sont personnalisées. Avec le facteur d'anesthésie, l'angle de Cobb de la première patiente a été corrigée de 62° à 47° et la deuxième patiente a été corrigée de 70° à 55°.

Cette étude a démontré avec un modèle biomécanique que la position du patient a un effet significatif sur la correction de l'angle de Cobb avant l'instrumentation même. Les chirurgiens pourraient profiter de la correction obtenue par le positionnement des patients. Ils pourraient ajuster les coussins des supports de façon à maximiser la correction 3-D. Le positionnement est une étape importante de la correction chirurgicale du rachis. Elle devrait être incluse dans les modèles qui simulent la correction chirurgicale de la scoliose.

Chapitre 4 : Un système de positionnement dynamique pour la chirurgie du rachis

Le SPD a été conçu et fabriqué pour améliorer le positionnement des patients pendant la chirurgie. Une figure du SPD est présentée dans l'article 2, figure 2. En bref, le SPD est composé de quatre coussins de base qui peuvent être déplacés sur les rails dans les directions x, y, et z. De plus, les coussins peuvent aussi tourner et la forme est concave pour mieux épouser la forme du tronc des patients. Il y a aussi des coussins externes qui peuvent être attachés aux rails de la table et aider à corriger la déformation du tronc.

Onze patients scoliotiques étaient positionnés debout, couchés sur le système de positionnement Relton-Hall, couchés sur les coussins de base du SPD puis sur les coussins de base du SPD avec application des efforts manuels. (Section 4.3 article 2) Les mesures de la géométrie du tronc ont été prises avec un système optoélectronique (Polaris, Northern Digital inc., Canada) et les mesures de la pression entre les patients et les coussins ont été prises. Sur le SPD, la longueur de la colonne était significativement plus grande sur le SPD que sur le système Relton-Hall. La cyphose était moins aplatie sur le SPD et la torsion de la colonne était plus faible. Les pressions étaient plus élevées sur le SPD par rapport au système Relton-Hall. En conclusion, la géométrie du patient est meilleure sur le SPD mais des modifications à apporter aux coussins sont nécessaires avant de procéder en salle d'opération.

Les modifications apportées aux coussins pour les élargir ont eu pour effet de réduire les pressions. Ensuite, trois patientes étaient positionnées sur le SPD pendant une chirurgie de la scoliose. Le jour précédent la chirurgie, les mesures de pression du SPD et du système Relton-Hall étaient comparées. Les pressions pour les trois patients étaient égales sur les deux systèmes ou plus faibles sur le SPD. Pendant la chirurgie, des pressions plus élevées ont été notées. Deux patientes avaient des rougeurs sur leur crête iliaque droite, où la pression maximale a été enregistrée, mais aucune complication majeure n'a été observée. La correction moyenne pendant la chirurgie était de 40% et de 68% pour les courbes thoraciques et de 75% pour les courbes lombaires. À partir de ces résultats préliminaires, nous espérons que le SPD aura la capacité de faciliter les manœuvres d'instrumentation du rachis et d'améliorer sa correction. Les résultats des trois premiers cas sont encourageants et démontrent la fonctionnalité de ce nouveau système. Une étude plus étendue est en cours pour déterminer s'il y a une amélioration significative de la correction.

Chapitre 5 : Ajustement optimale d'un système de positionnement dynamique pour la chirurgie du rachis.

Le modèle par élément finis présenté dans le chapitre 3 a été modifié. Six paramètres du modèle ont été modifiés dans un plan expérimental Box-Hunter pour déterminer les placements optimaux des coussins. Ces paramètres incluent, l'inclinaison du bassin, le positionnement et la hauteur des coussins supportant la poitrine, et trois force de correction entre 10 N et 150 N. Ces efforts étaient situés en position thoracique latéral, lombaire latérale et aussi sur la gibbosité. Dix indices cliniques ont été mesurés sur le modèle géométrique. Les indices cliniques ont été optimisés individuellement et simultanément.

Les résultats ont démontré que plusieurs paramètres du modèle avaient un effet sur les indices géométriques. L'optimisation d'un indice géométrique à la fois a souvent causé une détérioration majeure des autres indices géométriques. La correction globale est de 75% sur le SPD par rapport à 55% sur le système Relton-Hall. Le modèle a démontré que le positionnement du patient pendant la chirurgie est une étape très importante qui pourrait être exploitée pour améliorer la géométrie 3-D du patient.

Discussion et conclusions

L'objectif global de ce projet était de développer un nouveau SPD qui pourrait être utilisé pendant la chirurgie pour améliorer le positionnement des patients scoliotiques. En parallèle, un modèle par élément finis a été modifié pour simuler le positionnement per-opératoire des patients scoliotiques avec pour but de valider et recommander des ajustements du SPD.

La première étude a démontré que le positionnement du patient et son anesthésie peuvent être simulés avec un modèle personnalisé. Dans la deuxième étude, le SPD a été conçu, fabriqué et testé sur un groupe de patients scoliotiques. Cette étude a démontré que quelques indices de la géométrie du tronc sont améliorés quand le patient est positionné sur le SPD. Les pressions entre le patient et les coussins étaient plus élevés sur le SPD que sur le système Relton-Hall. Suite aux modifications effectuées sur les coussins pour améliorer les pressions, un test sur trois patients en salle d'opération a démontré la faisabilité du système. Dans la dernière étude, le modèle par éléments finis a été exploité pour optimiser les ajustements des coussins de SPD. Avec un ajustement optimal, la correction globale du patient était de 75%. La méthode novatrice développée avec ce modèle pourrait être utilisée pour prescrire la localisation des coussins d'un SPD qui optimiserait la position couchée du patient scoliotique pendant la chirurgie afin d'améliorer et/ou de faciliter la correction de la scoliose.

Introduction

Scoliosis is a three dimensional deformity of the spine and trunk affecting between 1.5% and 3% of the population. Though there are different curve types it is commonly recognised as an 'S' shaped curvature when looking at the back. Scoliosis often occurs secondary to some neuromuscular disease such as cerebral palsy; however, the majority of cases (80%) are classified as adolescent idiopathic scoliosis (AIS) and develop in otherwise healthy children with no known cause (Lonstein 1994). The most common treatment of scoliosis involves observation, bracing and in severe cases surgery. It has been found that, depending on the form of treatment, between 7 to 28% of the scoliotic patients require surgery (Weiss et al 2002, Goldberg et al 2001). The most common surgical procedure is posterior fusion where the patient is placed prone on the operating table, an incision is made down the center of their back and various hooks and/or screws are fixed to the vertebrae to stabilize the spine and promote fusion.

Proper positioning on the surgical table is an essential step in the surgical procedure. There are various positioning frames available for scoliosis but most are still based on the principals introduced by Relton and Hall in 1967 whom recommend that the abdomen must remain pendulous to reduce blood loss (Relton and Hall 1967). To date, the prime focus of the positioning of the patient is still to reduce bleeding, prevent complications and facilitate the posterior approach. A recent review by Schonauer et al. highlights the advantages and disadvantages of various frames and positions currently used for posterior spine surgery. Safe options for positioning range from inexpensive chest roles to the OSI Jackson frame (OSI, Union City, CA, USA) (Schonaur et al 2004).

The effect of operative position on lumbar lordosis has been widely studied. In particular, Marsicano et al. looked at the thoracic, thoracolumbar and lumbar alignments in AIS patients undergoing posterior fusion on the OSI frame. They found that on the OSI frame kyphosis was decreased while total lordosis was maintained (Marscicano et al. 1998).

The effect of positioning on the sagittal curves shows that, in general, the standing lordosis is preserved if the position allows for hip extension.

The effect of positioning on the scoliotic deformity has only been noted by a few groups. Delorme et al. (2000) observed spontaneous correction of the scoliosis Cobb angles of an average of 37% due to positioning, anaesthesia and exposure but before instrumentation. Mac-Thiong et al. looked at the effect of positioning on the external trunk geometry and found that trunk geometry is more influenced by the positioning on the operating table than by the magnitude of the patient's spinal deformity (Mac-Thiong 2000). They later found that operative positioning using the Relton-Hall frame does not lead to a reduction in the external scoliotic deformity and that modifications to the frame are required to obtain optimal pre surgical correction (Mac-Thiong 2002).

Computer models have been used to study scoliosis. The preliminary models were limited in that they only provided a two dimensional analysis (Viviani et al 1986). One of the largest advancement in scoliotic models came with a collaboration of Ecole Polytechnique (Montreal, Canada) and ENSAM (Paris, France). Their techniques provided a patient personalised, 3-D, finite element model of scoliotic patients (Aubin et al 1995, Descrimes 1995). Using this model as a base, over 15 studies have been conducted simulating brace and surgical treatments as well as integrating muscles and simulated growth. This model is well suited to be used to study the biomechanics of patient positioning.

The global objective of this project is to design and test a dynamic positioning frame (DPF) to improve the position of scoliotic patients during surgery. Personalised finite element modeling of the scoliotic spine is used in order to better understand the biomechanics of patient positioning and to provide a tool to help recommend adjustments of the components of the DPF.

This thesis is divided into seven chapters. The first chapter provides a review of the pertinent literature. In the second chapter, the hypothesis and the main objectives of this project are presented. The first article, presented in Chapter 3, addresses the simulations of patients lying prone on the currently used Relton-Hall frame. Chapter 4 begins by presenting the second article which documents initial design of the Dynamic Positioning Frame (DPF) and a small clinical study to show its proof of concept. This section is followed by a discussion of pressure measurements and design details of new cushions to provide reduced pressure and external cushions to provide more correction. Further clinical validation of the DPF is presented by three clinical cases on the DPF in the operating room at the Sainte Justine Hospital. This shows the feasibility of its use in the operating room but it is still not being used to its optimal potential. Chapter 5 presents a sensitivity and optimisation study of the DPF simulated on a patient personalised finite element model. Using the new DPF and the tools designed with the finite element models the optimal cushion placement for a particular patient can be determined. Finally, a general discussion is presented in Chapter 6 followed by conclusions and recommendations in the Chapter 7.

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List of Abbreviations

2-D	two dimensional
3-D	three dimensional
α	anaesthesia factor
AIS	adolescent idiopathic scoliosis
AP	anterior posterior
AVR	apical vertebral rotation
AVT	apical vertebral translation
C7PL	cervical 7 plumb line
CSVL	central sacral vertebral line
DLT	direct linear transformation
DPF	dynamic positioning frame
DVCP	difference in vertebral centroid positions
FEA	finite element analysis
FSA	force sensing array
HRL	horizontal reference line
L	lumbar
L3	third lumbar vertebra
LC	lumbar Cobb
LF	lateral lumbar force
MT	main thoracic
MTC	main thoracic Cobb
OR	operating room
PA	position A (prone on R-H)
PB	position B (prone on DPF)
PC	position C (prone on DPF, thigh cushions)

PD	position D (prone on DPF, kyphosis)
PE	position A (prone on R-H reppitition)
PI	position I (standing)
PI	pelvic inclination
PII	position II (prone on DPF)
PIII	position II (prone on DPF with external forces)
PIV	position II (prone on R-H)
PT	proximal thoracic
PTC	proximal thoracic Cobb
RF	rib hump force
R-H	Relton-Hall
S1	first sacral vertebra
T9	nienth thoracic vertebra
TF	lateral thoracic force
TL	thoracolumbar
TLC	thoracolumbar Cobb
US	United States

List of Annex

CHAPTER ONE: REVIEW OF LITERATURE

1.1 Anatomy of the spine, pelvis and ribcage

In order to describe the anatomy of the body it is important to be able to reference oneself with respect to the body. Some commonly used medical terms include the frontal plane, sagittal plane and transverse plane as shown in Figure 1.1. When one is looking at a patient (or an x-ray) from the back this is considered a posterior anterior view. When looking from the side this is the lateral view and when looking from above this is a transverse view. In addition to the medical planes a general Cartesian system, xyz, has been defined by the Scoliosis Research Society (Stokes 1994). The origin is located at the first sacral vertebra, the y direction is towards the left and parallel to the line formed by the anterior iliac crests, the z direction is towards the head and the x direction is orthogonal to these pointing forward.

The vertebral column provides the main structural support for the body and trunk as well as protection for the spinal cord. It is divided into major components as shown in Figure 1.2. At the top is the cervical region which is comprised of seven vertebrae and forms the neck. When viewed from the side, a healthy subject in a relaxed position will have a slight lordosis. The thoracic region is defined by the twelve vertebrae that are attached to the ribs. This region is curved in a mild kyphosis varying between 20° to 40°. The five vertebrae of the lower back, referred to as the lumbar region, are gently curved in lordosis ranging beween 40° to 60°. The section connecting the spine to the pelvis is called the sacrum and the coccyx and is composed of fused vertebrae.

The basic anatomy of a vertebra can be seen in Figure 1.3. The most predominant posterior part of the vertebra is the spinous process. This part can be easily palpated and felt as bumps along the back. The vertebral bodies carry the majority of the mechanical load and these are separated by intervetebral discs. Other important anatomical landmarks include the transverse process, pedicle, the inferior articular facet, superior articular facet, and the costal facets (only in the thoracic region).

The vertebrae are held together, not only by the discs, but also by a complex set of vertebral ligaments. Important ligaments include the supraspinous ligament, interspinous ligament, facet capsulary ligament, ligamentum flavum, intertransverse ligament, posterior longitudional ligament and anterior longitudinal ligament as shown in Figure 1.4. Pertinent pelvic anatomy is shown in Figure 1.5. Important points to note include the iliac crest, sacrum, sacroiliac joint, acetabulum and the ischium.

The ribcage is composed of ribs, cartilage and the body of the sternum, manubrium and xiphoid process as shown in Figure 1.6. There are a total of 12 ribs on each side. Ten attach to the sternum via the costal cartilage and there are two floating ribs. In between the ribs is a tight web of intercostal muscles which are involved in breathing.

1.2 Scoliosis

1.2.1 General overview

Scoliosis is a three dimensional deformity of the spine and ribcage affecting between 1.5 to 3% of the population (Lonstein 1994). Scoliosis is often recognised as an "S" shaped curve but greater emphasis is now being placed on the three dimensional (3-D) analysis of the curve (Aubin et al. 1999, Dubousset 1994). Scoliosis is often secondary to neuromuscular disease but the majority of cases (~ 80%) develop in otherwise healthy adolescence for no known cause. This type of scoliosis is called adolescent idiopathic scoliosis (AIS). The onset and progression of scoliosis is multivariable but is primarily based on the child's maturity and the type and severity of the curve. Larger double curves are more likely to progress then small single curves but it is very difficult for the doctor to predict progression (Lonstein 1994). Treatment varies at different centers but generally includes observation, bracing and, in severe cases, surgery. Table 1.1 summarizes a general guideline for scoliosis treatment.

Table 1.1: Summary of scoliosis treatment options

Cobb Angle (°)	Maturity	treatment
< 20	low maturity	observation
20-30	low maturity	bracing
30-40	growth remaining	bracing
< 45-50	no growth remaining	nothing
> 45-50	any age	surgical fusion

1.2.2 Clinical measures

There are also a number of clinical measures routinely taken during the patients visit as summarised in Figure 1.7. These include rib-hump, decompensation, shoulder height difference. A photo of the patients may also be recorded. From this photo certain cosmetics scores can be calculated such as those used by Raso et al. or the POTSI (Raso et al. 1998, Suzuki et al. 1999). There have also been various groups who have tried to quantify the external trunk geometry using surface techniques (Stokes et al. 1989, Goldberg et al. 2001, Liu et al. 2001). For the patient the external trunk geometry, and cosmetic appearance is often a primary concern. However, to quantify scoliosis it is best to look at the x-rays or analyse the 3-D spine shape.

Although the presence of a 3-D deformity is widely recognised, the clinical definition of scoliosis is a curve over 10° as measured from an AP x-ray using the Cobb method (Cobb 1948). Many clinical measures can be obtained from the anterior-posterior (AP) and lateral x-rays. The common curves measured include the proximal thoracic, main thoracic and thoracolumbar/lumbar Cobb. Other important clinical measures from the AP x-ray include coronal balance (decompensation), pelvic obliquity and apical vertebral translation (AVT). Thoracic AVT is measured from the C7 plumb line and lumbar AVT is measured from the central sacral vertebral line (Figure 1.8). On the lateral x-ray one can measure thoracic kyphosis, lumbar lordosis, sagittal balance and pelvic tilt (Figure 1.9).

In order to measure the true 3-D scoliotic deformity a 3-D reconstruction is necessary. There are two ways to obtain the 3-D geometry, either from multi-view x-rays or from the reconstruction of CT scans. In order to acquire a CT scan of the entire scoliotic spine the patient would have to be subjected to high radiation doses making it an impractical reconstruction method for routine scoliosis assessment.

Geometric reconstruction of vertebrae from x-rays involves a three step process which is explained in detail by Delorme et al. (2003) but will be briefly explained here. First is the computer assisted analysis of radiographic images. The patient stands in a positioning apparatus while three x-rays are taken: postero-anterior, potero-anterior at 20° and lateral. Anatomical landmarks are then digitised on each image. The second step involves the 3-D reconstruction using the direct linear transformation (DLT) method (Abdel-Aziz et al. 1971). The third step involves the free form deformation of the anatomical primitive using dual kriging equations (Trochu 1993). Anatomic primitives of typical vertebrae are gathered using a reconstruction technique. Transformations are applied to each anatomic primitive to deform them in such a way that they match the 3-D personalized geometry of the patient as close as possible. Kriging fits a mathematical function to a specified number of points or to points within a specified radius (Figure 1.10). The three x-ray acquisitions and the results of the kriging on the spine geometry are shown in Figure 1.11 (Delorme et al. 2003, Aubin et al. 1995).

An equivalent of the above mentioned clinical measures can be determined from the 3-D reconstruction. To calculate the Cobb angles a curve is passed through the vertebral centers and points of inflection are found from which the approximate equivalent of the Cobb angle is found (Jeffries et al. 1980). In addition to the standard 2-D measurements detailed 3-D measurements can be calculated such like the curve of the plane of maximum deformity, the orientation of the plane of maximum deformity (OPMD) (Labelle et al. 1995) and vertebral rotation calculated from a method devised by Stokes et al. for use with AP x-rays (Stokes et al. 1986).

1.2.3 Curve classification

In order to compare different clinical cases various methods of classification have been devised. A method developed by King et al in the 1980's was used to help define the appropriate surgical procedure (King et al. 1983). King defined five curves types which are briefly summarised in Table 1.2. The Lenke classification first divides the curves into one of 6 types as summarised in Table 1.3 (Lenke et al. 2002). The primary curve is that with the largest Cobb measurement and is always considered structural. For the other curves they are defined as structural if the curve is ≥ 25° on side bending or if the kyphosis ≥ 20°. A lumbar modifier (A, B, or C) is then selected based on whether or not the pedicles of the apical lumbar vertebra cross the central sacral vertebral line (CSVL). The CSVL is drawn up from the center of S1 perpendicular to the floor. Finally a sagittal modifier is attributed based on the thoracic Kyphosis. The King method is still in use today but the Lenke classification is gaining popularity since it was found to have greater inter- and intra-observer reliability and could classify every deformity. There have also been attempts at 3-D classification but these have yet to gain popular use (Poncet et al. 2001).

Table 1.2 Summary of King curve classification

King Type	General curve information	Flexibility index
Type I	S shaped Lumbar > Thoracic	-ve flexibility
Type II	S shaped Thoracic > Lumbar	+ve flexibility
Type III	Thoracic curve	
Type IV	Long thoracic curve L4 tilts into curve	
Type V	Double thoracic T1 tilted into curve	

Central Sacral line is the line drawn throught the center of the sacrum perpendicular to the iliac crests.
Stable vertebra is that which is most closley bisected by this line
Flexibility index is the %Lumbar correction - % Thoracic correction from bending radiographs

Table 1.3 Summary of Lenke curve classification.

		Primary curve	thoracolumbcar/lumbar	Proximal Thoracic
Curve type	1	Thoracic	non-structural	non-structural
	2	Thoracic	non-structural	structural
	3	Thoracic	structural	non-structural
	4	Thoracic	structural	structural
	5	Lumbar	Primary curve	non-structural
	6	Lumbar	Primary curve	structural
Lumbar modifer	A	CSVL lies between lumbar prdicles upto stable vertebra		
	B	CSVL between concave pedicle and apical vertebral body		
	C	CSVL lateral to the vertebral bodies		
Sagittal Modifier	-	Hypokyphotic	< 10°	
	N	Normal	10°-40°	
	+	Hyperkyphotic	> 40°	
CSVL = central sacral vertical line				
Structural Proximal Thoracic = Cobb>25 on bending and/or T1-5 kyphosis >10				
Structural Thoracic = Cobb>25 on bending				
Structural Thoracolumbar/lumbar = Cobb>25 on bending and/or T10-L2 kyphosis >10				

1.2.4 Surgical Treatment

Posterior instrumentation is the most common form of surgical correction for scoliosis. It is used to stop a curve from progressing and/or minimise the deformity by straightening and stabilizing the spine. Various types of instrumentation can be attached to the vertebrae. A brief summary of some different systems will now be presented.

The Harrington system was the first widely used surgical fixation system used to treat scoliosis surgery (Harrington 1962). The main hardware used are hooks, rod with ratchets and cross members. An outrigger was sometimes used to provide some pre-distraction before the actual distraction rod was implanted. This was believed to allow some initial relaxation and stretching of the ligaments. The main mechanical force is the distraction of the rod on the concave side which is sometimes accompanied by a convex rod in slight compression. The main disadvantages of this system is that it only corrected the deformity in the frontal plane and occasionally resulted in flat back syndrome (Humke et al. 1995).

In the mid 1980's the three dimensional deformity of scoliosis was becoming more widely recognized and surgical systems were aimed at correcting the 3-D deformity. The first system to correct the 3-D deformity was developed by Cotrel and Dubousset (Cotrel et al. 1988). The Cotrel-Dubousset (CD) system also consists of hooks, rods and transverse connectors but also has screws that provide stronger anchorage. The specialized tools include a "translator-introducer", used to translate rod to the spine (or vice versa). As well there is a rod rotator used to rotate the rod. The primary mechanism of the CD instrumentation is the rod rotation. A pre-contoured rod, to account for the deformity and a final kyphosis, is loosely attached to the hooks and then rotated approximately 90° and tightened. The belief is that the deformity in the frontal plane is rotated 90 degrees to provide a natural kyphosis in the sagittal plane. The mechanics of correction involve a combination of translation, rod rotation, distraction and compression. While distraction and compression are used to correct the coronal deformity and hook seating, they have obvious effects on the sagittal plane that must be considered when these forces are utilised.

There are many different types of instrumentation available on the market. Each system has some unique features but they all follow the same principles of stabilizing the spine to promote fusion. To mention a few; the TRSH (Richards et al. 1994) and Colorado 2 (Sofamor Danek) are systems whose primary mechanism is rod rotation followed by distraction and compression. The Luque (Luque 1982) and ISOLA (Boachie Adjei et Asher 1994) systems can also be used with sublaminar wires. The Moss Miami system uses primarily translation and cantilever manoeuvres (Shufflebarger 1998). The type of system used is primarily selected based on the surgeons experience and preference. When comparing the CD instrumentation to the Colorado instrumentation one study noted that there is no significant difference in 3-D correction for the two systems (Delorme et al. 2000a).

Anterior release and interbody fusion is sometimes necessary in conjunction with posterior fusion to prevent the "Crank Shaft" phenomenon. This occurs due to the continued anterior growth in skeletally immature patients. A completely anterior approach and fusion to correct scoliosis was first introduced by Dwyer et al. (Dwyer et al. 1974) and was popularised with the Zielke instrumentation in the 1980's (Zielke 1982). In the past thirty years it has been shown that anterior fusion is a viable option but there are controversies surrounding its use and the posterior instrumentation and fusion is still considered the gold standard (Betz et al. 2001).

1.3 Surgical Positioning

1.3.1 Positioning systems

It was first noted by Relton and Hall, and the system used today still stress, that the abdomen must remain pendulous to help reduce excessive pressure on the abdomen and vena cava and in turn reduce bleeding (Relton et al. 1967). Calahan noted that the three principals to positioning during spine surgery are: the stability of the spine, the required surgical exposure and physiological limitations of the patient and surgeon. (Callahan et al. 1981). The simplest method for positioning the patient in the prone position, while reducing pressure on the abdomen, is with the use of chest roles. These can be as simple as rolled towel bolsters or gel pads.

An array of positioning systems are shown in Figure 1.12. The four post frame, described by Relton and Hall, is still considered the gold standard for spine surgery as seen in Figure 1.12.a. This type of frame is recommended for longer posterior fusions. The lumbar lordosis can be controlled by flexing and extending the legs. The tuck position

(Figure 1.12.b) was on once used for lumbar surgeries but this extreme flexion can lead to complications. One of the most common commercially available surgical spine tables is the Jackson spinal table shown in Figure 1.12.c. To the Jackson table, or other commercially available orthapediac tables, the Wilson frame can be attached Figure 1.12.d. The Wilson frame consists of two arched cushions that run along the lateral sides of the torso. This is a convenient and stable method of maintaining patients in a flexed position for laminectomy, decompression, disc surgery and micro-discectomy procedures. In particular for lumbar surgery the Andrews table can be used where the knee chest position can be easily obtained as shown in Figure 1.12.e. An initial prototype of a positioning system that could help hold the patient in an optimal position was designed (Pazos V 2000) at the Sainte Justine Hospital (Figure 1.12.f). This system provided a good base to test the initial concepts of correcting the positioning of patients but was not suitable for use in the operating room primarily because of radio-transparent issues.

1.3.2 The effect of positioning on lordosis

The effect of patient position on sagittal curves has been examined by various authors. The first of which, was Tan et al, who noted no significant difference between the standing lordosis and when the patient was positioned on simple chest roles (Tan et al. 1994). Various authors all concur, that with the hips flexed there is a significant loss in lordosis but with the legs extended the standing lordosis is preserved (Peterson et al. 1995, Guanciale et al. 1996, Stephens et al. 1996, Benfanti et al. 1997). Even the kneeling position has been found safe at preserving lordosis after surgery but only when short-segment instrumented fusion is used (Tribus et al. 1999).

1.3.3 The effect of lying prone/supine

It is common knowledge that you are taller in the morning then at the end of the day. In the standing position, gravity acts on the body compressing the spine. Studies by NASA have shown that astronauts experience an elongation of the spine when they are subjected to a zero gravity environment (NASA). Relating this phenomenon to scoliosis, various studies have found that there is a spontaneous correction of the scoliotic deformity when patients are x-rayed in the supine position (Klepps et al. 2001, Yazici et al. 2001, Torell et al. 1985, Vedantam et al. 2000, Duval-Beaupere 1996). Looking at the mentioned studies the average correction, observed due to lying in the prone position, ranges from 22% to 29%.

1.3.4 Correction of the scoliotic deformity

From the above mentioned studies one can see that there is significant correction due to the patient lying supine however, the effect of the prone position in the frontal plane has not been as widely studied. Delorme et al have found that significant and important correction to the thoracic and lumbar Cobb angle of 37% is due to prone positioning, surgical exposure and anesthesia (Delorme et al. 2000b). Radiographic analysis found 1/3 of the total correction occurred prior to instrumentation (Behairy Y et al. 2000). Another study found that 31% curve correction is a result of the prone position and that an additional 17% correction occurs due to soft tissue release in preparation for the passage of sub laminar wires. Even more correction is obtained when an anterior release discectomy is first performed and, for a strictly anterior thoracolumbar surgery, almost complete correction of the curve is obtained before the application of any instrumentation (Polly et al. 1998). When looking at the external trunk deformity of normal and scoliotic subjects lying prone on a Relton-Hall type frame, Mac-Thiong et al. noted that the trunk geometry is more influenced by the positioning of each subject then by the magnitude of the deformity (Mac-Thiong et al. 2000). In a subsequent study of scoliotic patients standing and lying prone, recommendations were made to modify the Relton-Hall frame to provide optimal presurgical correction of the trunk deformity (Mac-Thiong et al. 2002).

1.4 Biomechanical modeling techniques applied to the spine

1.4.1 Biomechanical modeling

Biomechanical modeling is used to create a mechanical representation of a biological structure. The primary interest in this review is spine models. Once a model is made it can provide an almost limitless number of simulations. This allows more variables to be tested then could be recreated with in-vitro and especially in-vivo experiments. They are non destructive so they generally reduce the overall cost of an experiment. Also they have the advantage of eliminating sample inconsistencies, or patient differences. Common model types include those using the finite element analysis (FEA) or a mechanical system simulation. A general description of these two types of models will be presented.

The first difference in FEA and mechanical system model can be seen from a theoretical point of view. In using FEA, a complex shape can be broken into smaller pieces, called elements, so that a solution can be found. For each element, the force vector, F, and the displacement vector, Q, can be related by means of a stiffness element matrix, K_e.

$$K_e Q = F \qquad \text{e1.1}$$

For multiple elements, the stiffness matrices for each element are assembled to form the global stiffness matrix.

$$\begin{bmatrix} K_{11} & K_{12} & \cdots & K_{1N} \\ K_{21} & K_{22} & \cdots & K_{2N} \\ \vdots & & & \\ K_{N1} & K_{N2} & \cdots & K_{NN} \end{bmatrix} \begin{Bmatrix} Q_1 \\ Q_2 \\ \vdots \\ Q_N \end{Bmatrix} = \begin{Bmatrix} F_1 \\ F_2 \\ \vdots \\ F_N \end{Bmatrix} \qquad \text{e1.2}$$

The global stiffness matrix, *KG*, is based on the material property, geometry and boundary conditions. In most cases the force vector, *F*, is given and the displacement vector, *Q*, can be solved. Once *Q* has been determined the element stresses and strains can be calculated (Chandrupatla et Belegundu 1997).

For a mechanical system model the approach is based on the equations of motion such as:

$$\sum F = ma \quad \text{and} \quad \sum M = I\alpha \qquad \text{e1.3}$$

Where the sum of the forces (F) is equal to the mass (m) times the acceleration (a) or where the sum of the moments (M) is equal to the moment of inertia (I) times the angular acceleration (α). For both models there are various forms of software available. ANSYS and ADAMS are presently used in the laboratory at the Sainte Justine Hospital. In general, a FEA model is primarily used when the stresses and strains within and object are of interest. Mechanical system models are most useful for dynamic analysis when the position of a part and the reaction forces are of interest to the user. Both models require input geometry, material properties and boundary conditions. The FEA model also requires the definition of element types. For a mechanical system model, the mass and acceleration of the object must also be entered. Output from both models will include information on displacements and reaction forces. A FEA model will also include stress and strain information while the mechanical model will provide more information on acceleration. The major differences in a finite element analysis model and a mechanical systems model are summarized in Table1.4.

Table 1.4: Summary of finite element and mechanical system models.

Model	Finite Element	Mechanical system model
Comercial software	ANSYS, ALGOR, ABAQUS, LUSAS	ADAMS, Solid works
Use	Primarilly static but also dynamic, thermal, electrical...	Primarly dynamic & kinematic
Theory	Matrix based. ie. F=KeQ (structural)	Equations of motion $\sum F = ma$
Input	Geometry, material properties, boundary conditions, definition of element type	Geometry, material properties, boundary conditions, mass, accelerations
Output	Displacement, Stress, strain, reaction forces	Displacement, reaction forces, acceleration

Both types of models can be used for various biomechanical simulations. When the specific detail of the stress and strain is required, a FEA model should be used. A mechanical systems model should be used if the application is dynamic, and the displacement and acceleration of the parts is the prime question. For the case of complex dynamic simulations for example, the stresses in the lumbar discs while walking, a hybrid of the two models could be used. A mechanical system model could first be used to model the lower limbs, upper body and trunk to determine the reaction forces in the lumbar section. A finite element model could be used to model the lumbar section with the calculated reaction forces and the stress and strain in the intevetrabral discs can then be analysed.

1.4.2 Finite element spine models

1.4.2.1 Detailed lumbar models

Some of the first and most detailed models of the spine were models of the lumbar spine or lumbar motion segments. The main characteristics of these models are summarized in Table 1.5. In 1986 Shirazi-Adl et al. created a detailed 715 node model of the L2-L3 segment to study flexion and extension (Shirazi-Adl et al. 1986). Almost ten years later the model was extended to include the entire lumbar spine and now has 3020 nodes (Shirazi-Adl 1994). The mechanical properties of the original model were updated and the section was now analyzed in torsion. They found that the gap junctions, between the facets, and not the facet geometry were the main factor in the resistance to torsion. Coupled motions were also noted which have been observed in vitro studies (Panjabi et al. 1989). Goel et al used ANSYS to create an extensive 3-D model of the lumbar spine from 1mm CT scans and mechanical properties were extracted from literature (Goel et al. 1988). Bilateral decompression surgery was simulated where they observed low stress in cancellous bone and high stresses in cortical bone which is in agreement with clinical observations of screw loosening. The lumbar finite element model of Lavaste et al. was constructed from two x-rays and six parameters were measured per vertebra (Lavaste et al. 1992). The rest of the vertebral geometry was parameterized from these initial 6 parameters. They found that simulation of bending, flexion, extension and torsion tests were similar with those reported in literature. They also observed coupling motion and that disc stress were highest in torsion. Using the same model, Robin et al. performed a sensitivity test (Robin et al. 1994). These tests showed that disc dimension parameters have the greatest influence on the mechanical behaviour of the spine, the width of the vertebral bodies have the greatest influence in lateral bending while length has the greatest influence in flexion and extension. They again confirmed that the gap has a greater influence then the actual facet geometry. These sensitivity findings are promising and important. Since the vertebral bodies and hence, intervertebral discs, are easy to visualize on x-rays it is good to know that they can be easily measured and that they have the greatest influence on then model behaviour. This is important when trying to make a patient specific model. In general, for the analysis of stresses and strains at a particular section in the spine a detailed model similar to those presented above is required. For cases where the sensitivity of particular geometry on the model is required that geometry, such as facets, must be modeled.

Detailed lumbar models have more recently been used to simulate the biomechanical effects of kyphoplasty (Villarraga et al. 2005) total disk replacements

(Rohlmann et al. 2005) and the stability of facetectomy (Lee et al. 2004). For cases where the displacement of the spine as a whole is of interest less detailed models can be used. A summary of some less detailed but more global models of the spine are discussed in the following section.

1.4.2.2 Simplified global models

In the simplified models of the spine the complete geometry and volume of the vertebrae are not modeled. Rather, a few elements are usually representative of the vertebra. Sometimes more detail is added to include the pedicles and articular facets but generally the function, rather then true geometry of the anatomy, is represented. A summary of some simplified global models are presented in Table 1.6. Stokes and Laible created a 336 node model of the vertebra, ribs, sternum and intercostal ligaments (Stokes et al. 1990). The geometric properties were extracted from stereo radiographs using the method of Dansereau (Dansereau et al. 1990). The mechanical properties were taken form literature. Asymmetric growth of the ribs was modeled and resulted in lateral deviation and vertebral rotation similar to that observed in scoliosis. The magnitude of the deformity was only slight so they concluded that there must be some other factors responsible for adolescent idiopathic scoliosis.

The combined work of Aubin et al. and Descrimes et al. was primarily aimed at refining the facet joints of the Stokes model (Aubin et al. 1995, Descrimes et al. 1995, Stokes et al. 1990). The main objective was to develop an intermediate model that could be used for the prediction of orthotic treatments while respecting the available computational capabilities. The geometry of the facet joints was refined by Aubin et al. with the used of 3-D stereo reconstruction (Dansereau et al. 1990) and CT scans of cadaver specimens (de Guise et al. 1988). Descrimes et al. focused on the mechanical aspects of the facet joints, which were found to have a greatest importance in torsion as well as the lumbar region.

A recent non-linear finite element model of the entire spine was developed to evaluate the trunk muscle forces, internal loads and stability margins with and without an external load (Shirazi-Adl et al. 2005). They showed that minimal changes in posture, such as pelvic tilt, substantially influenced the muscle forces and internal loads.

Once a spinal model, such at the ones presented above, has been created and validated it can be used to simulate surgical procedures or orthotic treatments. The following sections will present some models that have been used for scoliosis surgical simulations as well as orthotic treatments.

1.4.2.3 Scoliosis surgical models

A summary of various scoliosis surgical models is shown in Table 1.7. In a study by Ghista et al. the spine was simply represented by a two dimensional (2-D) beam element with the node at the center of each vertebra (Ghista et al. 1988). The dimensions and nodal coordinates were obtained from radiographs. The mechanical properties were adjusted to match traction radiographs of 10 patients. In this study, the patient personalized model was used to predict the amount of corrective force required during Harrington instrumentation. A specially designed distractor and forceps were then used to apply "as best as possible" the prescribed forced during distraction and lateral traction. The error from the simulated model and post op x-ray ranged from 0 to 10%. An interesting final step that would further validate their model, that was not included, would have been to record the forces applied during surgery and re-input those into the patient model to see if the result was closer to the actual surgical outcome. Even though a simple 2-D model of the spine was used, the combination of finite element modeling, patient personalized geometry and mechanical properties and intra-operative force measurement makes this an exceptional study. They were able to conclude that it is possible to predict post operative correction using a finite element model and apply an optimal set of corrective forces using instrumented distractor and forceps. This model was further refined to simulate asymmetrical muscle contraction and surgical correction (Noone et al. 1991). They found that simulations of bi-lateral muscle contractions was able to create a scoliotic model. Following an update in geometry, simulations of surgical corrective forces were applied. Finally, the results of the continuous model were compared to the finite element model and post-surgical curve with good agreement for both models. These 2-D models, though effective, were limited in that they simulated the Harrington instrumentation which focused on the 2-D correction of scoliosis. 2-D models would likely not be as effective for CD instrumentation simulations and it is partly because of this, that the more recent models are 3-D and more complex.

A 3-D simulation of Harrington rod surgery was performed by Stokes and Gardner Morse (1993). They varied the amount of sagittal beam offsets to obtain accurate agreement with the model and post-op x-rays. The amount of offset was not consistent or predictable.

Continuing with their three dimensional model, Garnder-Morse and Stokes were able to simulate the derotation maneuver with CD instrumentation. Derotation was simulated by applying a moment about the vertical axis to the midsection of the rod. The hooks were connected to the vertebra by joints and they were free to rotate about the longitudinal axis of the rod until after 90° of rotation when hook lock was simulated. A 30mm posterior offset of the motion segment beams was required to produce both spinal (65 to 19°) and vertebral (11 to 8°) derotation. With a 30mm posterior offset they were able to show how both spinal derotation and vertebral derotation in the opposite directions can be produced by the rotation of a single rod attached to the spine.

Instead of using a finite element model Poulin et al. chose ADAMS to represent the spine by rigid bodies and flexible elements. On one patient they showed good agreement between the model at various stages of the surgery. They also point out that this modeling reduces computational and convergence problems versus a finite element model. Patient personalised material properties have been added to this model (Petit et al. 2004) and more detailed simulations of posterior instrumentation on three patients have been performed (Aubin et al. 2003).

Simulation of CD surgery was done on a patient specific FEA model (LeBorgne P, et al. 1999). Traction was used during the operation so that was the first step in the simulation. The geometry of the CD rod is imputed followed by displacement of the vertebra towards the rod. Finally the rod is rotated. Intraoperative measurements were made of the vertebral position at various surgical steps (Lecire et al. 1999). Comparison between the simulation and monitored vertebral displacement help to validate the model. This model was recently validated with ten patients and to further evaluate its potential three different clinical strategies were performed on one patient (Lafage et al. 2004).

Stokes et al. investigated the feasibility that biomechanical models could be used as a planning tool for scoliosis surgery (Stokes et al. 1999). They noted some complication with the present models including: a reduction of Cobb angle that occurs due to positioning, anaesthesia and surgical opening that is not accounted for in the model constructed from standing radiographs. The patient specific stiffness in unknown at each anatomical level. Large difference in stiffness as well as large displacements create convergence problems with the model, and finally the surgical instrumentation is not applied in the same manner by each surgeon creating a number of unknown variables.

In addition to spinal instrumentation costoplasty is also used to treat the rib hump deformity. A study by Grealou et al. simulating rib lengthening and/or shortening, was able to support the concept that concave rib shortening or convex rib lengthening provide a useful means of correction of the spinal deformity during the growth stage (Grealou et al. 2002). To the same model longitudinal growth was added and the progression of a scoliotic curve, due to a slight initial perturbation, was demonstrated (Villemure et al. 2002). Combining the costoplasty and growth simulations a novel technique for a less invasive surgery confirmed the potential of long-term correction of spinal curvature resulting from the rib shortening on the concavity (Carrier et al. 2004).

1.4.2.4 Brace models

A summary of some brace models is shown in Table 1.8. In 1974 Andriacchi et al. focused mainly on the interactions between the spine and the ribcage(Andriacchi et al. 1974). They found that the ribcage has a significant role on the bending response of the spine. The ribcage also increases stability in compression. This model provided the foundation for their 1976 study on brace simulations. They found that for a single curve a lateral pad at the curve apex in combination with traction was the most effective.

A hybrid model by Goel and Gilbertson was an innovative combination of a detailed and simple model (Goel et al. 1995). The T11-L1 section was modeled with great detail similar to their previous model (Goel et al. 1988). This high level of detail was necessary to study the stresses on an injured T12 vertebra. The remainder of the spine and ribcage was modeled with simple beam elements in order to conserve processing time. In this model an orthosis and flexion tests were simulated to see the effect of the orthosis on the stresses at T12. They found that a "loose fitting" orthosis reduces the stresses on T12 better than a preloaded orthosis.

Wynarsky and Schultz used various objective functions and were able to show that optimal correction was achieved when the primary force was applied to the ribs on the convex side of the curve (Wynarsky et al. 1991). A counter force was then necessary on the concave ribs 5 & 6 and on the concave side of L2-L4. They also simulated muscle forces and found that they were somewhat more effective then brace forces in correction of the curve.

Using the model created at the Sainte Justine Hospital, Aubin et al looked at the coupling mechanics between the spine and rib cage (Aubin et al. 1997). They observed that an anterior load applied to the rib hump created a reduction of sagittal curves and an increase in lateral shift and frontal curve. Based on this, they proposed a new approach of lateral load on the convex side with a posterior load on the anterior rib hump and finally a mechanical constrain in the posterior rib hump. Gignac et al. simulated the proposed new loading on 20 patient models with positive results (Gignac et al. 1998). In a subsequent study, in order to find even more optimal loads they simulated the same general loads with slight variations in magnitude and location (Gignac et al. 2000). They found that optimal loads were mostly located on the convex side of the curve. The study also demonstrated the feasibility of using an optimization approach to find the most advantageous loads for a particular patient.

More recently the same models have been used for personalized simulations of brace treatment (Perie et al. 2004). The pressures were measured inside the braces with the help of a force sensing array (FSA) (Vista Medical, Winnipeg Canada). Equivalent forces, ranging from 18 to 73N, were recorded inside the braces, applied to the model and the simulated deformation was validated against the real in brace geometry.

As it is clear from some of the above mentioned studies, the research group at the Sainte Justine Hospital has conducted a series of various projects and simulations built on the foundation of the Aubin and Descrimes model (Aubin et al. 1995, Descrimes et al. 1995). In summary, a flow chart of the various studies can be seen in Figure 1.13.

Table 1.5 Detailed Lumbar models

Author	Anatomy modeled	Resolution	Geometric Properties	Mechanical Properties	Boundary Conditions	Results	Validation
Shirazi-Adl, Ahmed, & Shirvastave (1986)	L2-L3 Detailed Facet articulations, disk nucleus and annulus modeled	Nonlinear Finite element 715 nodes	in vitro measurements	Literature including Betsch '80, Harkness '61, Haut '72, Sanjeevi '82 and others	Not stated but hyperflexion and hyperextension were simulated	>Stiffer with increasing motion. Stiffer in extension than flexion. >Flexion high disk pressure, extension negative disk pressure. >Ligaments transfer load in flexion, posterior bone transfer's in extension.	Compared to literature. Panjabi '82, Adams '80, Farfan '73 and others
Shirazi-Adl (1994)	L1-S1, extension of 1986 model including various ligaments	Nonlinear Finite element 3020 nodes	Ct scan of cadaver specimen	From 1986 paper, updated Hukins 1990, Myklebust '88, Neuman '92, Pintar '91	S1 fixed	Gap junction and not facet articulation geometry is the main factor in the resistance to torsion. Coupled motions were also noted.	Compared to literature. Panjabi '89, Oxland '92, Ahmed '90 and others
Goel, Kim, Lim & Weinstein (1988)	Extensive 3-D model of lumbar motion segments	ANSYS used resolution not specified	From 1mm ct scan using midsagital symmetry	Literature	Inferior surface of inferior vertebra fixed	Interbody bone graft transmits 80% of loads compared to 96% for an intact disc. Low stress in cancellous bone, high stresses in cortical bone around screw.	Low cancellous stress and high cortical stress supported by clinical screw loosening observations.
Lavaste (1992)	Lumbar spine, Vert. (volume elem.) Ligaments (cable elem.) Facets (3-D contact elem.)	ANSYS Vertebra 777 nodes. Disc 365,144 Ligaments 26, Artic 18 gap	6 parameters digitised per vertebra. Other parametric variables.	Literature Goel '88, Shirazi '84, and in-vitro experiments	Same boundary conditions used by Martie et al 1990	Simulation of bending, flexion, extension and torsion consistent with literature and experiments. Coupling motion consistent. Disk stress highest in torsion	Dry specimens. Stiffness compared to various experiments Farfan '70, Markolf '72, Panjabi '76, and others
Robin, Skalli, Lavaste (1994)	Same as Lavaste '92	Same as Lavaste '92	Same as Lavaste '92. Orientation of facets, height of disks can be modified	Same as Lavaste '92	Lower part of segment immobilized	Disk dimension parameters have greatest influence. Width of vert. bodies has greater influence in lateral bending. Length has greater influence in flexion/extension. Gap between facets has more influence then geometry.	Sensitivity study model was validated by Lavaste '92
Totoribe, Chosa & Tajima (2004)	L4-L5 motion segment	Same as Lavaste '92	Ct scan, 29y old male, at 2mm increments	Literature including, Burstein '76, Goel '95, Lindhal '76, Schultz '79, Shirazi '84, and	L5 inferior fixed. Axial compressive forces and flexion extension	Poster lateral fusion, posterior fusion and facet fusion was simulated. A more solid fixation is obtained, especially in posterior fusion, if facet fusion is performed.	Compared to literature. Shirazi '84 & '86, Brown '57, Goel '89, Markolf '72 and others

Table 1.6: Simplified global models

Author	Anatomy modeled	Resolution	Geometric Properties	Mechanical Properties	Boundary Conditions	Results	Validation
Stokes & Laible ('90)	Vertebra (bodies, sp, tp) Ribs 11 points, Sternum, Intercostal ligaments, CT CV.	Algor software used with 336 nodes	Taken form sterio radiogrraphs (Dansereau '88)	Literature inculding Roberts '70, Panjabie '76.	L5 flexion only, T1 free vertical translation and flexion. Minimally constrianed L5 All 3 rotations permitted, T1 Constrained only in horizontal translation	Asymetric growth of ribs caused lateral deviation usually coupled with vertebral rotation	Rib flexibility and CV CS articulation compared to Schultz et al '74. Global thorax compared to Agostoni '66 and Nahum '70.
Goel and Gilbertson ('95)	review and presentation of a "combined" model	Detailed T11 L1 segment in a simple beam element model of spine and ribcage	Not specified	Not specified	Not specified	loose fitted orthosis reduces stresses by 44% on the injured T12 vertebra. Preload orthosis reduces by only 25%	Not mentioned
Descrimes et al Aubin et al ('95)	Vertebral bodies, pedicle, SP, Ribs, cartelige, sternum, disk, ligaments, articular facettes	ANSYS 5.0 1376 Nodes 2800 elements Bones (rigid beams), Disk (beam + torsional spring), ligaments (linear springs), facetts (point surface contact elements)	Reconstruction form steriodgraphic (Dansereau '90) CT scan (de Guise '88)	Literature from Lavaste '90 Roberts and Chen '72 or in vitro experimental tests	L5 fixed T1 free for vertical translation only	Anterior bony part very important 67-97% except for torsion in the lumbar area which is dominated by posterior section. Posterior ligaments play no mechanical role	In vitro tests on functional units.
Shirazi, El-Rich, Pop, & Parnianpour (2005)	T1-S1 spine	T1-T12 one rigid body, T12-S1 vert rigid, T12-S1 discs deformable beams, 46 muscles.	Keifer ('97)	Muscles literature, Bergmark '89,Dumas '91 and others	Not specified. S1 fixed?	Minimal changes in posture (pelvic tilt) influence muscle forces.	Muscle forces consistent with EMG (Parnianpour '94)

Table 1.7: Scoliosis surgical models

Author	Anatomy modeled	Resolution	Geometric Properties	Mechanical Properties	Boundary Conditions	Surgical simulated details	Results	Validation
Ghista, Viviani, Subbaraj, Lozada, Srinivasan & Barnes ('88)	Vertebral bodies	2-D model, 1 beam element with node at vertebral body center	dimensions and nodal coordinates measured from x-ray	Adjusted until model matched traction x-rays at 20,35 and 50 % body weight	L5 fixed in y,z but free to rotate. T1 fixed y, free to rotate	Distraction force (90-270 N) and Total Lateral traction force (220 to 720 N)	Patient stiffness 100-6000 kg/cm2. Max error 10%. Possible to predict post op correction and apply optimal corrective forces using instrumented distractor & forceps.	good agreement between model and post-op geometry. Forces applied in the surgery are monitored.
Noone, Mazumdar, Ghista ('91)	spinal model of vertebra, rectus abdominus, psoas	Continuous 2-D simple beam-column model,	Node at vertebral centriods	Adjusted to reflect the stiffness of a scoliotic curve under traction force	Roller supported, superior end. No disp. at inferior end and torsional spring.	First asymmetrical rectus abd. and psoas contractions. Surgical correction using Distraction and lateral traction forces	Simulations of the rectus abdominus contraction followed by a reflexive psoas contraction resulted in a scoliosis. Surgical correction with D&LT forces applied to model simulated correction	Comparing the continuous model to a finite element model (Subbaraj '89) and a post surgical case was consistent
Stokes and Gardner-Morse ('93)	Stokes and Laible "90	Stokes and Laible "91	Taken form stereo radiographs (Dansereau '88)	Literature including Roberts '70, Panjabie '76.	Sacrum free to rotate in flexion and extension. C7 free to translate vertically and rotate.	Upper hook free to rotate, lower hook fixed. Displacement applied to hooks to simulate distraction	Overall agreement better in the frontal plane. Simulation and post-op results in agreement but with varying offsets (0-30mm). Amount of motion segment offset not consistent or predictable.	Base model was previously validated (Stokes and Laible '90).
Gardner-Morse & Stokes ('93)	7 vertebra, 4 CD hooks and one 7mm concave rod	ANSYS model	Taken form stereo radiographs (Dansereau '88)	Panjabi '76	End vertebrae free to rotate and displace vertically. Apical vertebra's vertical disp, constrained	Derotation manoeuvre with moment applied to midsection of rod. After 90° rotation hook lock-up by increasing hook-rod joint stiffness.	Cobb 65 to 29 deg, Rotation worsened from 11 to 19 deg. With a 30mm posterior offset applied to the derotation manoeuvre the Cobb corrected from 65 to 19 degrees and the rotation from 11 to 8 deg.	The motion segment behaviour matched Panjabi.

Table 1.7 : Scoliosis surgical models (Continued)

Author	Anatomy modeled	Resolution	Geometric Properties	Mechanical Properties	Boundary Conditions	Surgical simulated details	Results	Validation
Poulin, Aubin, Stokes, Gardner-Morse, & Labelle ('98)	Vertebra (rigid bodies), Disks, Ligaments, Facets (flexible elements)	ADAMS model used resolution not specified	Vertebra centers found from stereo radiographs (Dansereau '90)	From literature including Gardner-Morse '90 and Panjabi '76.	L5 blocked except sagittal rotation. T1 constrained transverse direction, sagittal rotation allowed.	Hook-vertebra spherical joint. Screw-vertebra fixed. Rod-instrumentation cylindrical joint. Simulate hook instillation, hook anchorage and rotation.	Reduces computational problems of convergence over FEA models. Patient Cobb 32.5 to 30.4 to 8.7 Model Cobb 32.5 to 29.8 to 14.8 Good agreement between model and patient	Validated against a physical mode and one patient case. Computer model within 80%
LeBorgne, Skalli, Lecire, Dubousset, Zeller, Lavaste ('99)	Vertebra, discs, ribs, sternum, ligaments, articular facets	3-D FEA	Vertebra centers found from stereo radiographs (Dansereau '90)	Descrimes '95 model, soft tissues altered locally until model matches bending tests.	Traction load input as a boundary condition.	4 step surgical simulation. 1-Model of patient prone with traction 2-Geometry of CD rod 3-Distraction 4-Rotation	Simulation of one patient is encouraging with the maximum difference being 12 mm.	Surgical steps simulated and compared to pre, intra, and post-op data. (Lecire '99)
Aubin, Petit, Stokes, Poulin, Gardner-Morse, & Labelle ('03)	Vertebra (rigid bodies), Disks, Ligaments, Facets (flexible elements)	ADAMS model used resolution not specified	Vertebra centers found from intra-op stereo radiographs (Dansereau '90)	From literature including Gardner-Morse '90 and Panjabi '76.	lowest vert. fixed, highest vert. free, highest implanted vert. constrained in disp.	3 patients, translation of hooks to 1st rod, rotation, lock-up of hooks.	Reaction loads at vertebra implant link mostly below 1000N. Realistic representation of the surgery.	Final simulation compared to post-instrumentation x-ray.
Lafage, Dubousset, Lavaste, Skalli ('04)	Vertebra, discs, ribs, sternum, ligaments, articular facets	3-D FEA 1025 nodes, 1734 elements.	As in LeBorgne ('99)	Descrimes '95 model and then soft tissues altered locally until model matches bending tests.	Traction load input as a boundary condition.	10 patients plus 3 strategies on one patient. Traction, facet resection, hook placement, 1st rod placement, rod rotation, second rod, compression and distraction, return to standing.	Max difference 13 deg & 12 mm, intermediate steps consistent with literature. A one level change in instrumentation can have a significant effect on spine alignment.	Compared to post operative 3-D reconstruction
Rohlmann, Richter, Zander, Klockner & Bergman ('05)	T1-T12	3-D FEA, Volumetric vertebra. Ligaments, facet joints and ribcage spring elements	"Geometrically simplified" of scoliotic spine, Cobb angle 61 deg	Intersegmental stiffness adapted iteratively but within range of Panjabi ('76)	T12 rigidly fixed. Follower load of 280 N applied.	Ventral derotation spondylodesis anterior (Zielke) surgery. 7 different strategies simulated. Measure screw pullout force.	Rish of screw pullout greatest at cranial end of vertebra. Stiffer implant has better correction but higher screw forces.	Not validated. But simulations justify surgeon's intuitive results

Table 1.8: Brace models

Author	Anatomy modeled	Resolution	Geometric Properties	Mechanical Properties	Boundary Conditions	Brace forces	Results	Validation
Andriacchi, Schultz, Belytschko, Galante ('74) Andriacchi, Schultz, Belytschko, Dewald ('76)	Vertebra, Ribs, Sternum	39 Rigid bodies, 236 Springs, 59 beams	vertebra 16 points, CV CT articulations 4 points, Sternum 14 points, Ribs 8 points Schultz 1974	8 spring elements, bilinear spring compression 10x Tensil. Modified to agree with Schultz 1974	Sacrum fixed, T1 free to rotate and z disp.	Forces applied individually and in combinations. Traction on T1, right lateral thoracic pad force, left lumbar pad force, left shoulder sling force.	In bending, rib cage has a sig. role. Interaction between right & left sides. Resection of 2 ribs, little effect on bending. Ribs increase buckling load by 3-4 times. Rib cage stiffend scoliotic spine. Brace correction predicted in 81% of patients. Combo lateral pad & traction best for single curves	Lateral rib gace loading compared with Agostoni '65, Posterior loading of sternum (Nachum 1970 Patric 1065)
Wynarsky & Schultz ('91)	Thoracic and lumber vertebra, sacrum, ribs, costal cartilages ligaments	39 Rigid bodies, 236 Springs, 59 beams	Literature, Andriacchi '74, Belytschko '73, Schultz '73 & '74, Takashima '79	Schultz 1974, Takashima 1979	Sacrum fixed	Primary force on convex side of curve ribs 7-10. Counter force on concave ribs 5 & 6. Counter force at concave side L2-L4 vert.	Both brace & muscle forces are capapble of correction. Muscle forces are more effective.	Qualatitive agreement with clinical outcomes.
Gignac, Aubin, Dansereau & Labelle (2000)	20 patient specific models of spine and ribcage (Aubin '95)	1411 Nodes & 3011 elements	From sterio radiagraphs (Dansereau '90)	Taken from literature (Descrimes '95)	Not stated but reaction forces shown on L5 and T1.	Two thoracic forces on the 5, 7 or 9th rib. One lumbar force.	Optimal thoracic forces were mostly located on the convex side of the curve. Optimal lumbar force located at about 190º from the x axis.	Results similar to frontal correction achieved by Boston brace.
Perie, Aubin, Petit, Labelle & Dansereau (2004)	3 patient specific models of spine and ribcage (Aubin '95)	1356 nodes & 2801 elements	From sterio radiagraphs (Dansereau '90)	Taken from literature (Descrimes '95)	Not stated but reaction forces shown on L5 & T1.	Pressure forces between brace and patient measured. Equivalent forces applied	Differences less then 9.8 mm and 7.7 deg. Forces of 18-73 N meausred inside the braces.	Validated against in brace x-rays.

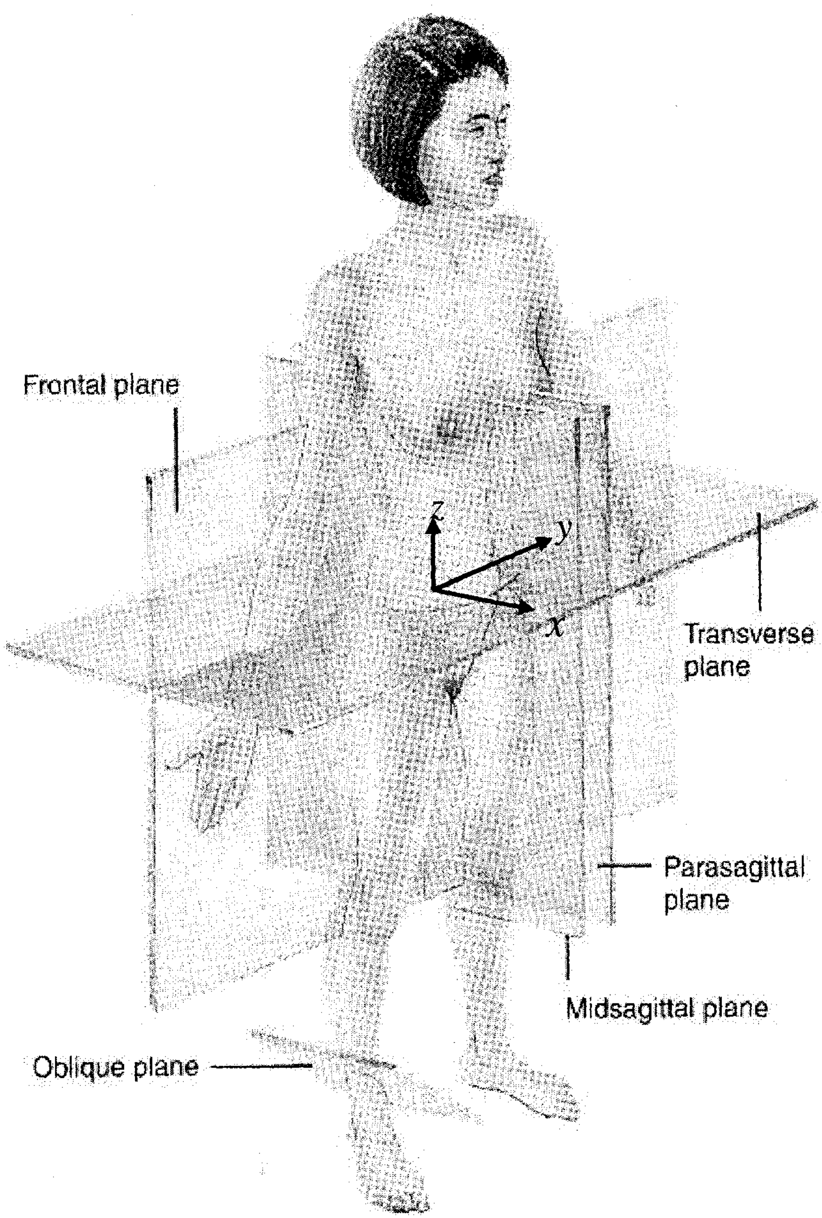
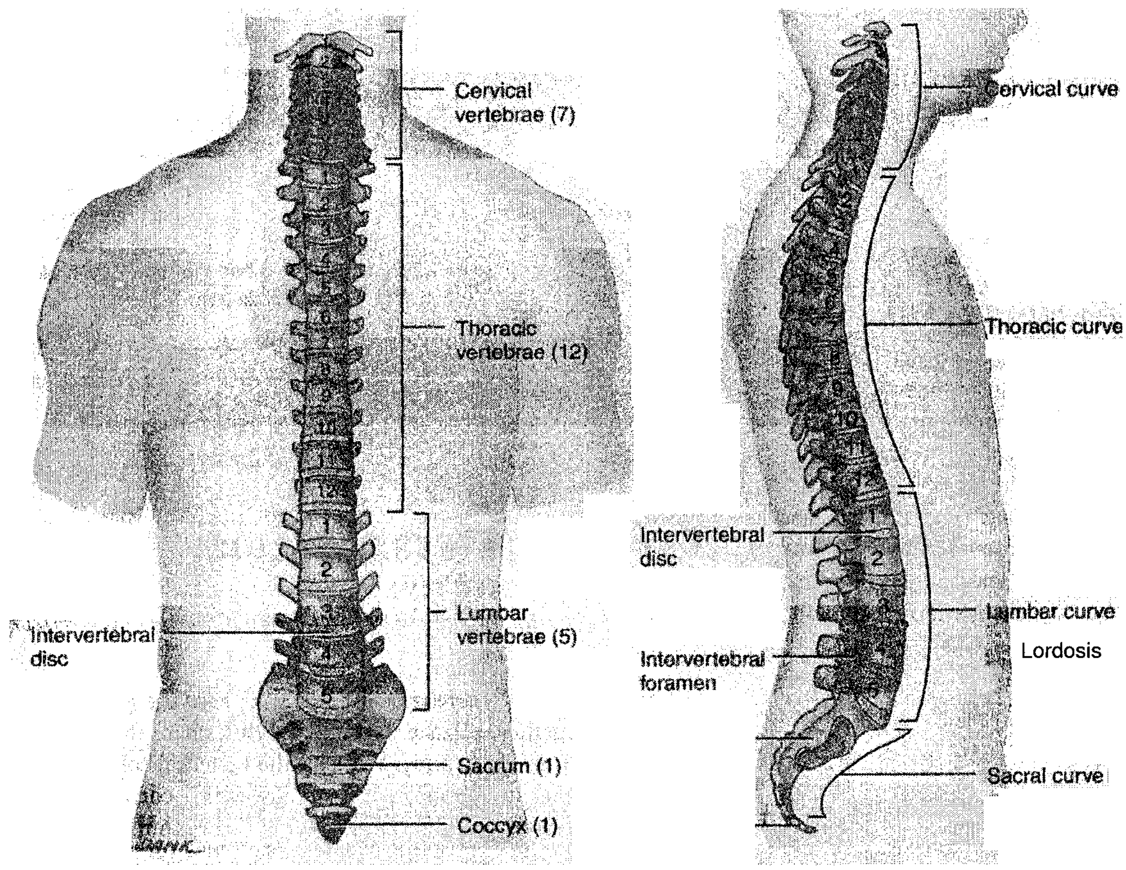


Figure 1.1: The Cartesian co-ordinate system as oriented with respect to the spine.
(Modified from Tortora, 2000)



a) Anterior view   b) Sagittal view

Figure 1.2: Anatomy of the vertebral column. (Modified from Tortora, 2000)

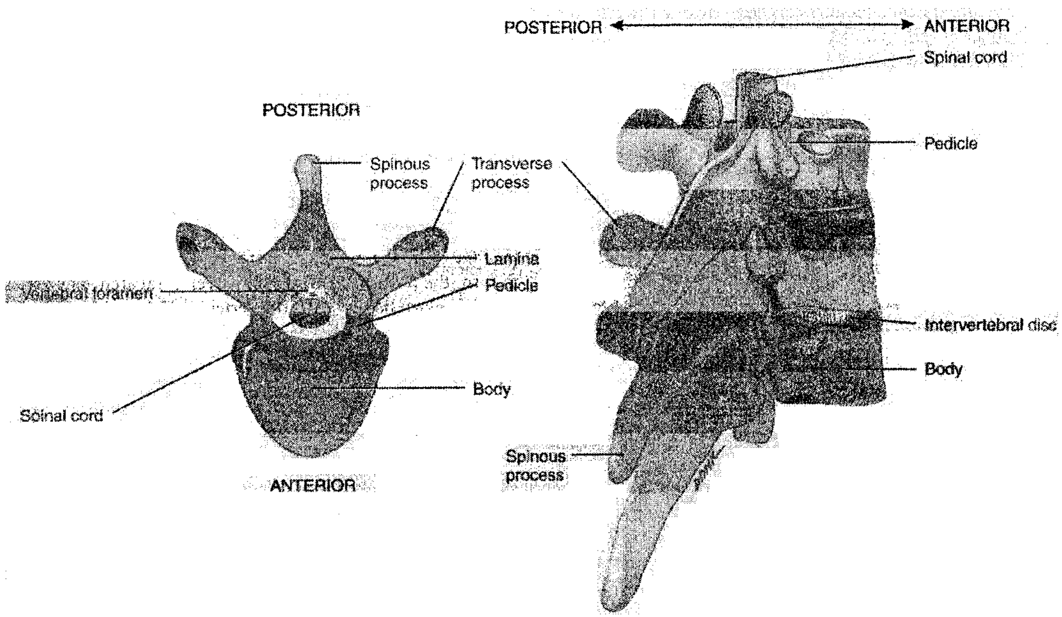


Figure 1.3 Vertebra anatomy. (Modified from Tortora, 2000)

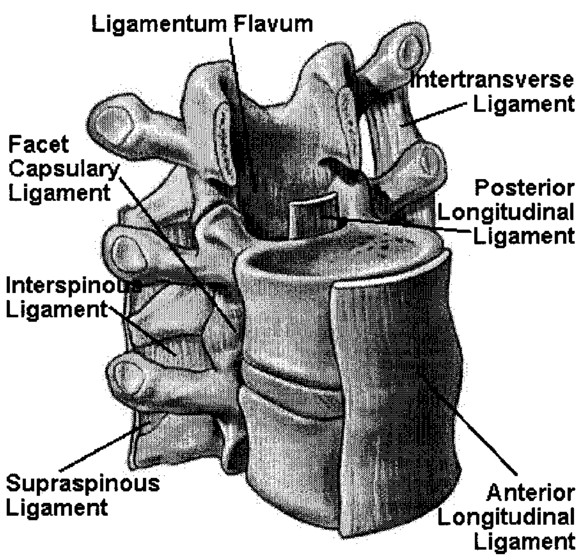


Figure 1.4 : Vertebral ligaments. (Modified from http://www.spineuniverse.com/displayarticle.php/article2001.html)

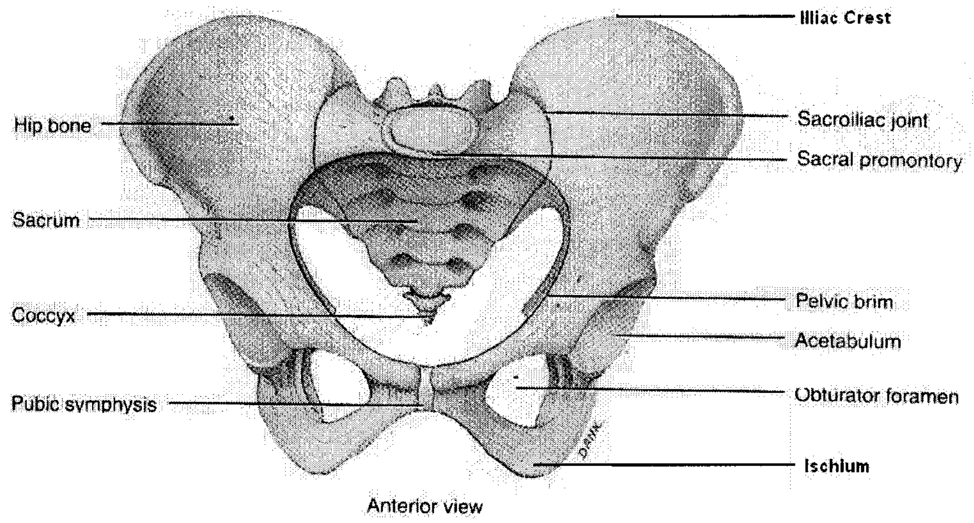


Figure 1.5 Pelvic anatomy. (Modified from Tortora, 2000)

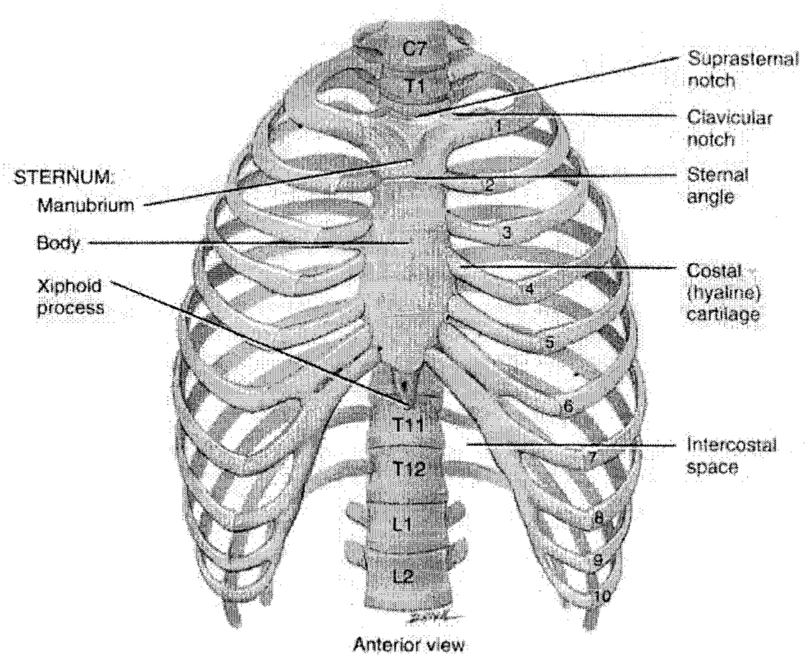


Figure 1.6 Anatomy of the ribcage. (Modified from Tortora, 2000)

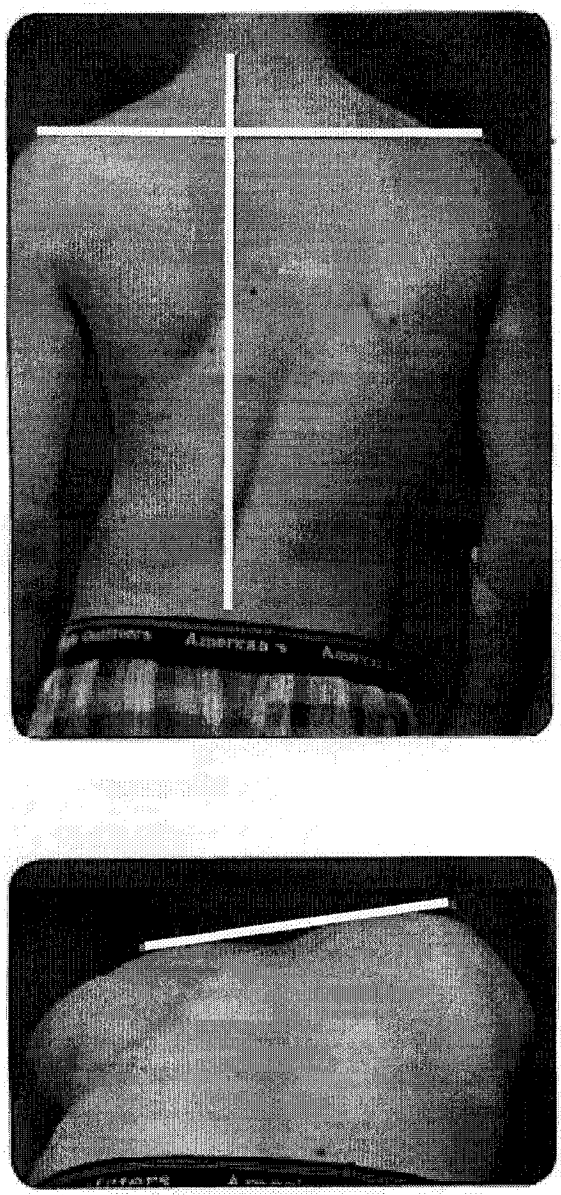


Figure 1.7: Clinical measures taken directly from the patient. (Modified from O'Brian et al. 2004)

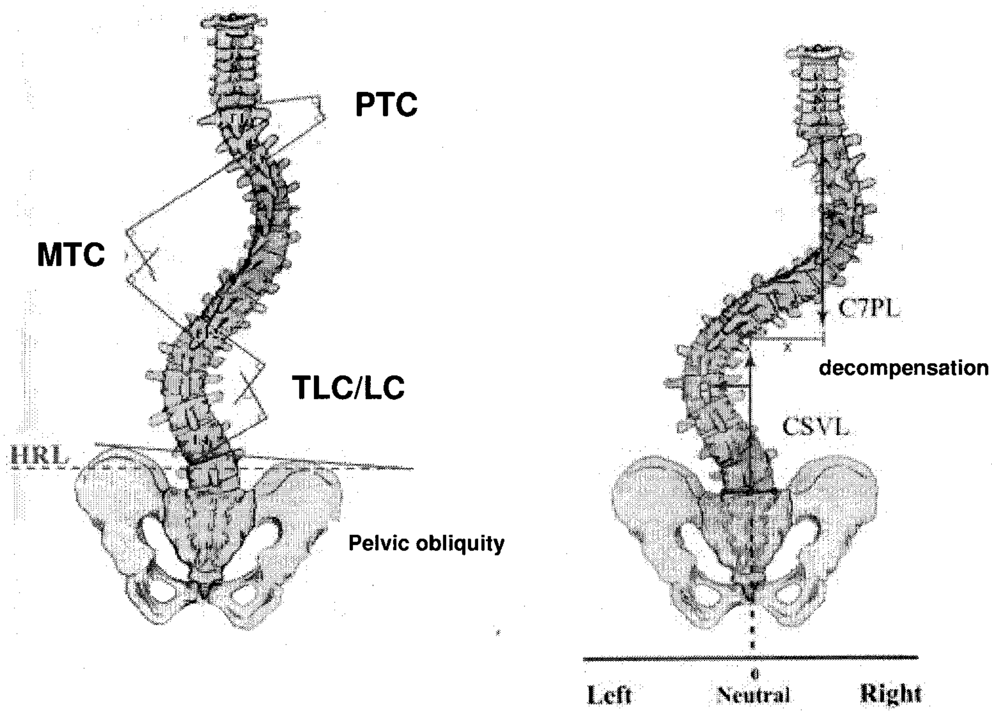


Figure 1.8: Clinical measures taken from the AP x-ray Proximal thoracic Cobb (PTC), Main thoracic Cobb (MTC), thoracolumbar Cobb (TLC), lumbar Cobb (LC), horizontal reference line (HRL), cervical 7 plumb line (C7PL), central sacral vertebral line (CSVL) and decompensation (shown by the x). (Modified from O'Brian et al. 2004)

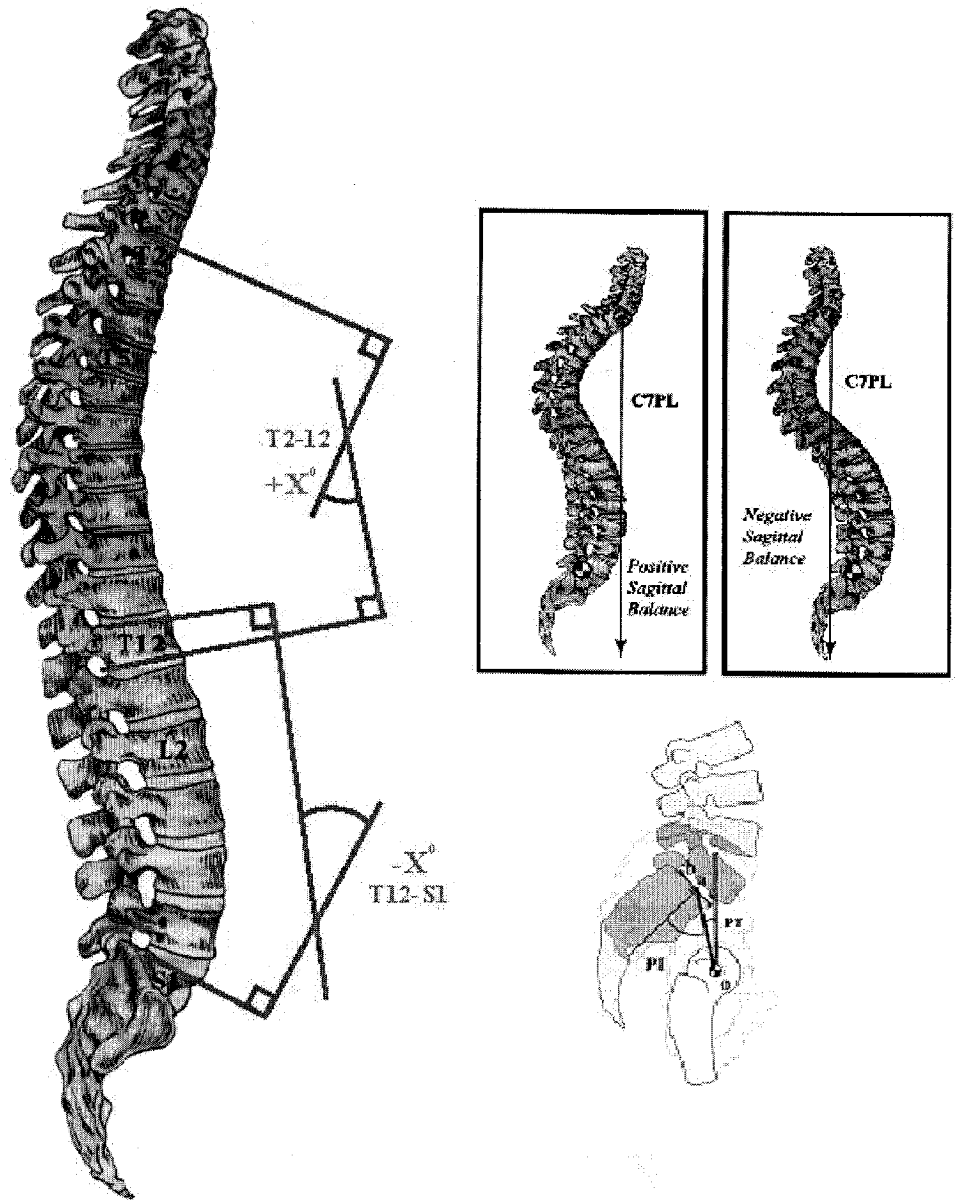


Figure 1.9: Clinical measured taken from the lateral x-ray (Modified from O'Brian et al. 2004)

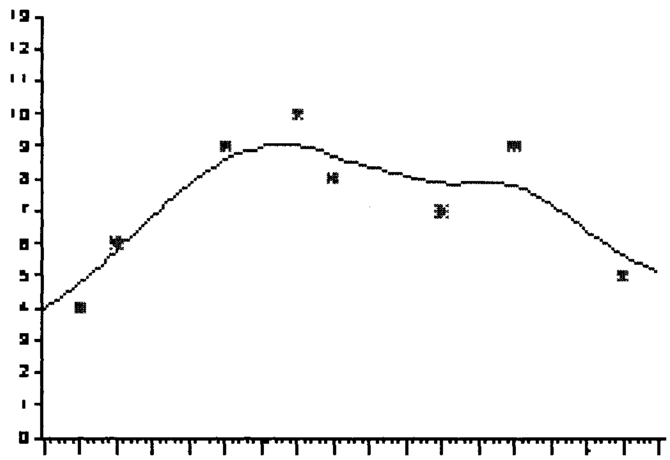


Figure 1.10: Graphical representation of the Kriging with nugget effect (Modified from Trochu 1993)

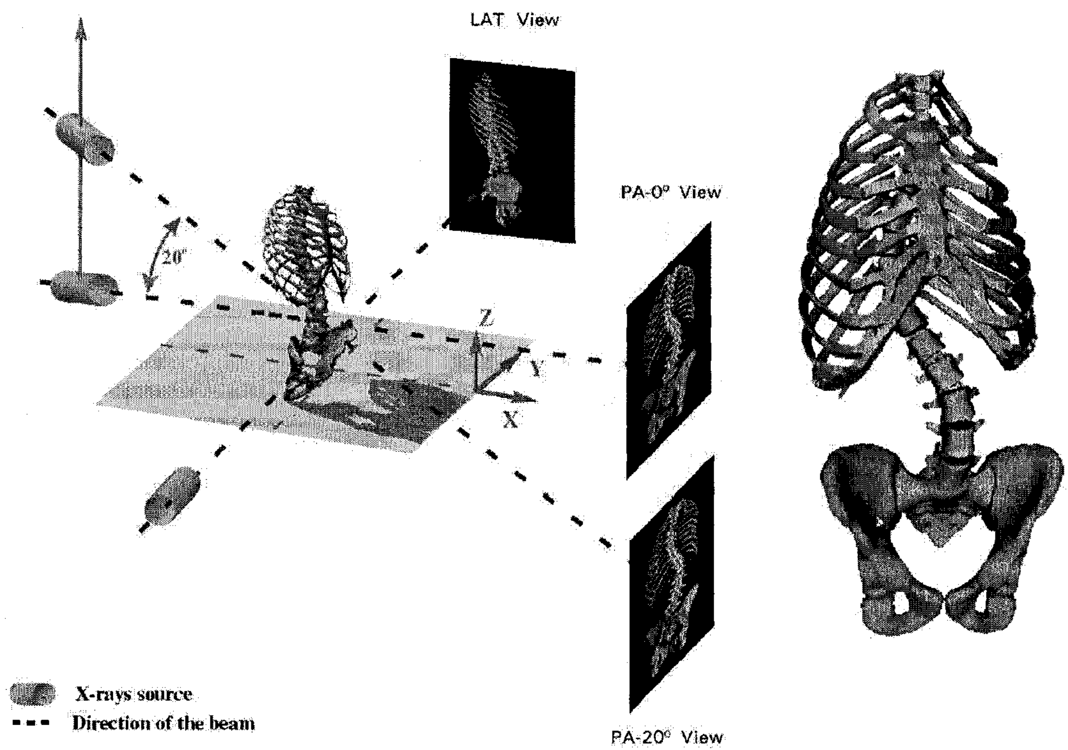


Figure 1.11 : Reconstructed 3-D spine geometry. (Modified from Delorme et al. 2003)

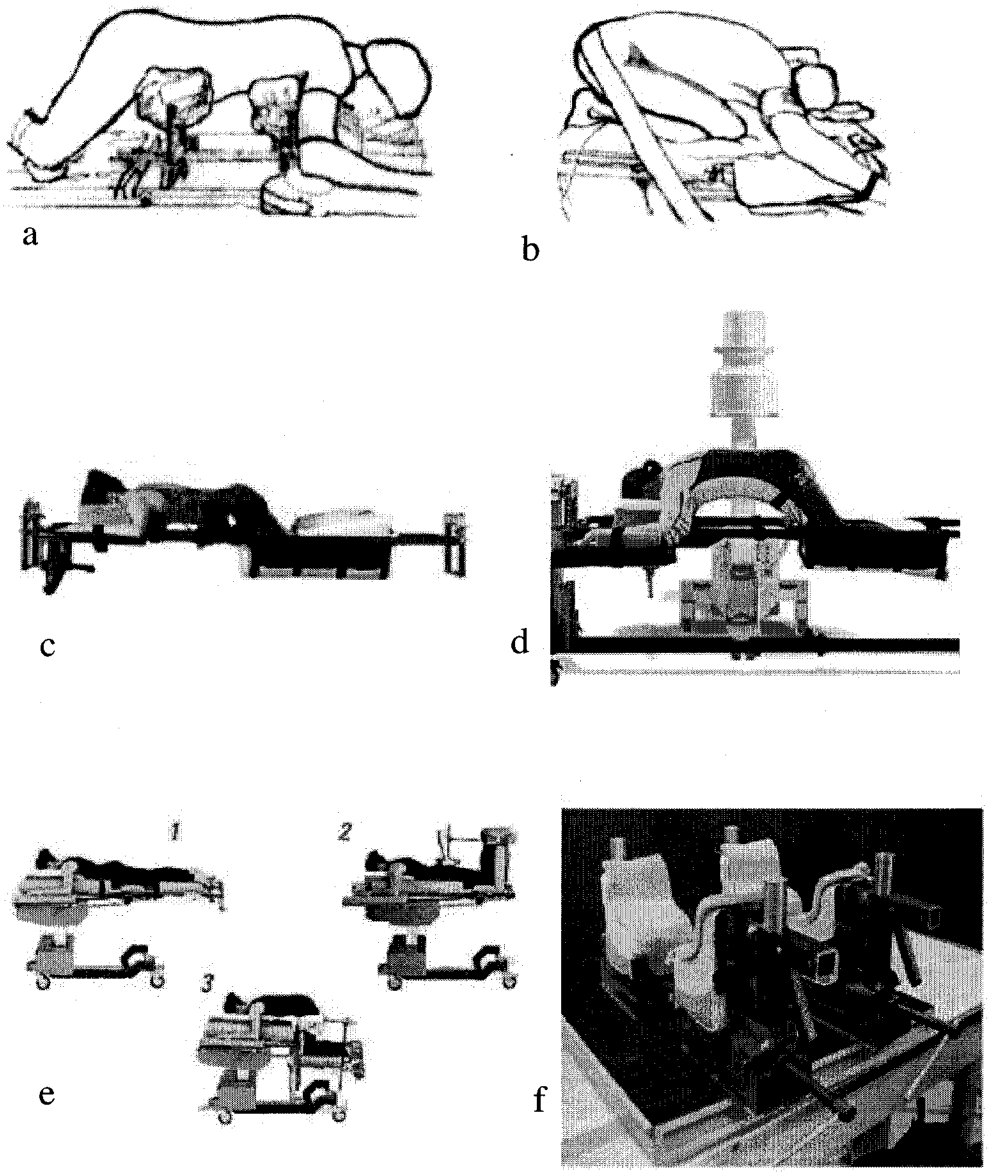


Figure 1.12: Various positioning systems (Modified from Schonauer et al. 2004)

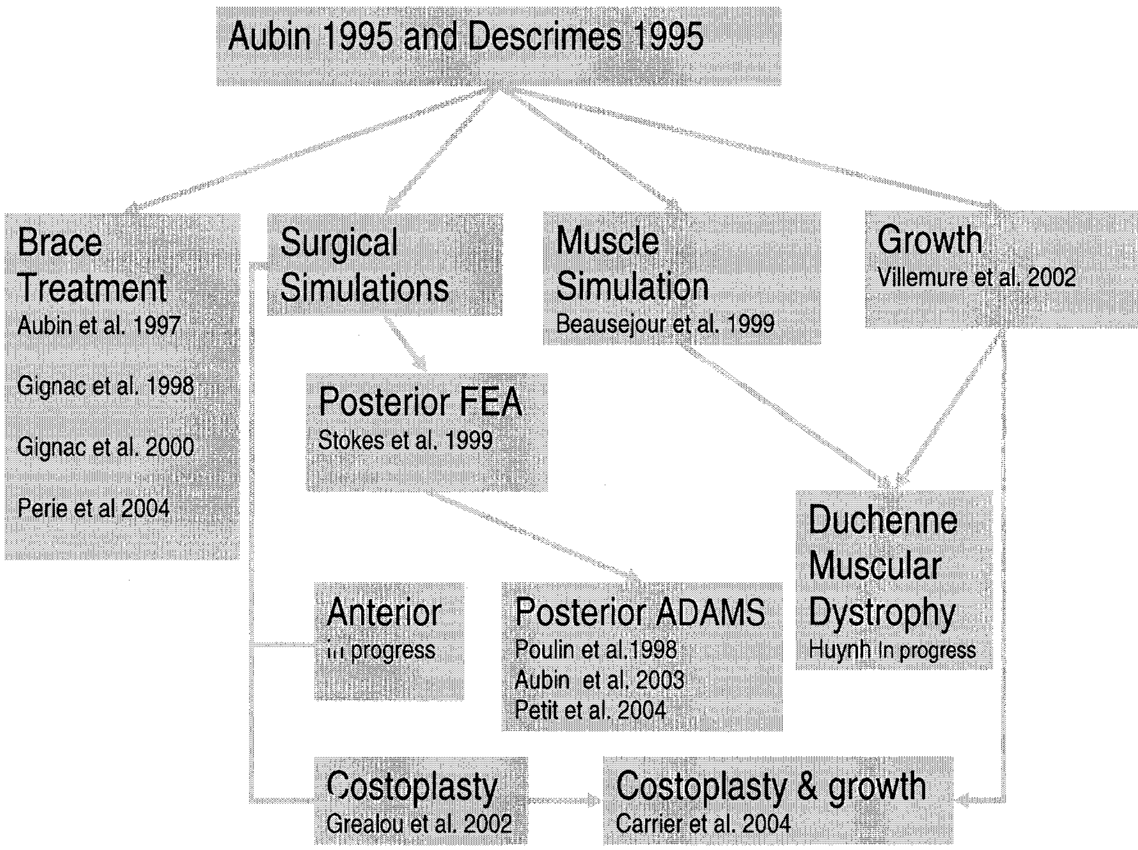


Figure 1.13: Summary of the models at the Sainte Justine Hospital

CHAPTER TWO: HYPOTHESES AND OBJECTIVES.

In summation of the review of literature the following points can be extracted that have a direct link to this research project:

- Scoliosis is a three dimensional deformity of the spine and trunk;
- Severe cases of scoliosis require surgical correction;
- Various positioning tables exist for spinal surgery with the prime focus being on patient stability and the reduction of blood loss;
- Significant correction of the scoliotic curves is observed when the patient is placed prone on the operating table;
- Many different biomechanical models have been developed to represent the spine and simulate surgical instrumentation;
- There is no biomechanical model that takes into account the effect of gravity and anaesthesia on the positioning of the patient.

Taking into consideration the above points, the general objective is to develop and validate a dynamic positioning frame (DPF) that will be used in the operating room to improve the surgical correction of spinal deformities. In parallel, as part of the validation and to recommend DPF adjustments, a finite element model will be created to simulate intra-operative prone positioning. In order to satisfy this main objective the following two hypotheses were created.

H1: The utilization of a dynamic positioning frame can improve the correction of the spinal deformity and the overall trunk geometry compared to the Relton-Hall frame used in the conventional approach.

H2: The osseo-ligamentous structures of the patient, the effect of gravity, anaesthesia and the surgical position can be modeled, and such computer models can be used to recommend placement of the positioning frame components on a patient for correction of the spinal deformities.

To embark upon the above hypotheses the project has been subdivided into 4 smaller objectives. A flow chart depicting the thesis layout and when the various objectives were satisfied can be found in Figure 2.1.

Objective 1: To refine a finite element model such that it is able to simulate the influence of position and anaesthesia while incorporating patient personalised material properties.

As seen in the review of literature, there is significant correction observed due simply to the prone positioning and the existing computer models do not take this correction into account. The models either, ignore this correction and rely solely on the instrumentation manoeuvres to correct the curve or the models are based on prone intra operative x-rays which renders them useless for pre surgical planning. There are some models that do take the positioning into account but only if traction is used. The first objective is addressed in the first article, *Biomechanical simulations of scoliotic spine correction due to prone position and anaesthesia prior to surgical instrumentation*, which is presented in Chapter 3.

Objective 2: To set out the design criteria and to incorporate these specifications into the design improvement and fabrication of a prototype of the DPF that could be used in the operating room.

Since there is significant correction observed due to prone positioning, the goal is to take advantage of an observed effect and make a positioning system that can potentially provide even more correction. Two different clinical assessments were done of the DPF bringing us to our third objective:

Objective 3: To test the DPF on a group of scoliotic patients first, in a clinical setting and then, in the operating room.

Before taking the DPF into the operating room it was tested on a group of scoliotic patients in a clinical setting. The design of the DPF and this preliminary test on a group of patients is presented in the second article found in Chapter 4 section 4.3 and titled A *Dynamic Positioning Frame for Spinal Surgery.*

There were some modifications made to the Dynamic Positioning Frame which are discussed in Section 4.5. Once the corrective positioning system was built and validated in the clinical setting it was time to move into the operating room. Section 4.6 addresses the second part of the third objective by presenting the preliminary results of 3 patients who were positioned on the DPF.

Objective 4: To modify the model created in objective 1 to simulate the DPF and test various adjustment parameters and recommend placement of the positioning system components prior to, and during surgery based on the results of the computer simulations.

Chapter 5 addresses this fourth objective and includes a third article titled *Optimal cushion placement of a dynamic positioning frame for spine surgery.* This chapter is an amalgamation of the model defined in Chapter 3 combined with the DPF designed in Chapter 4.

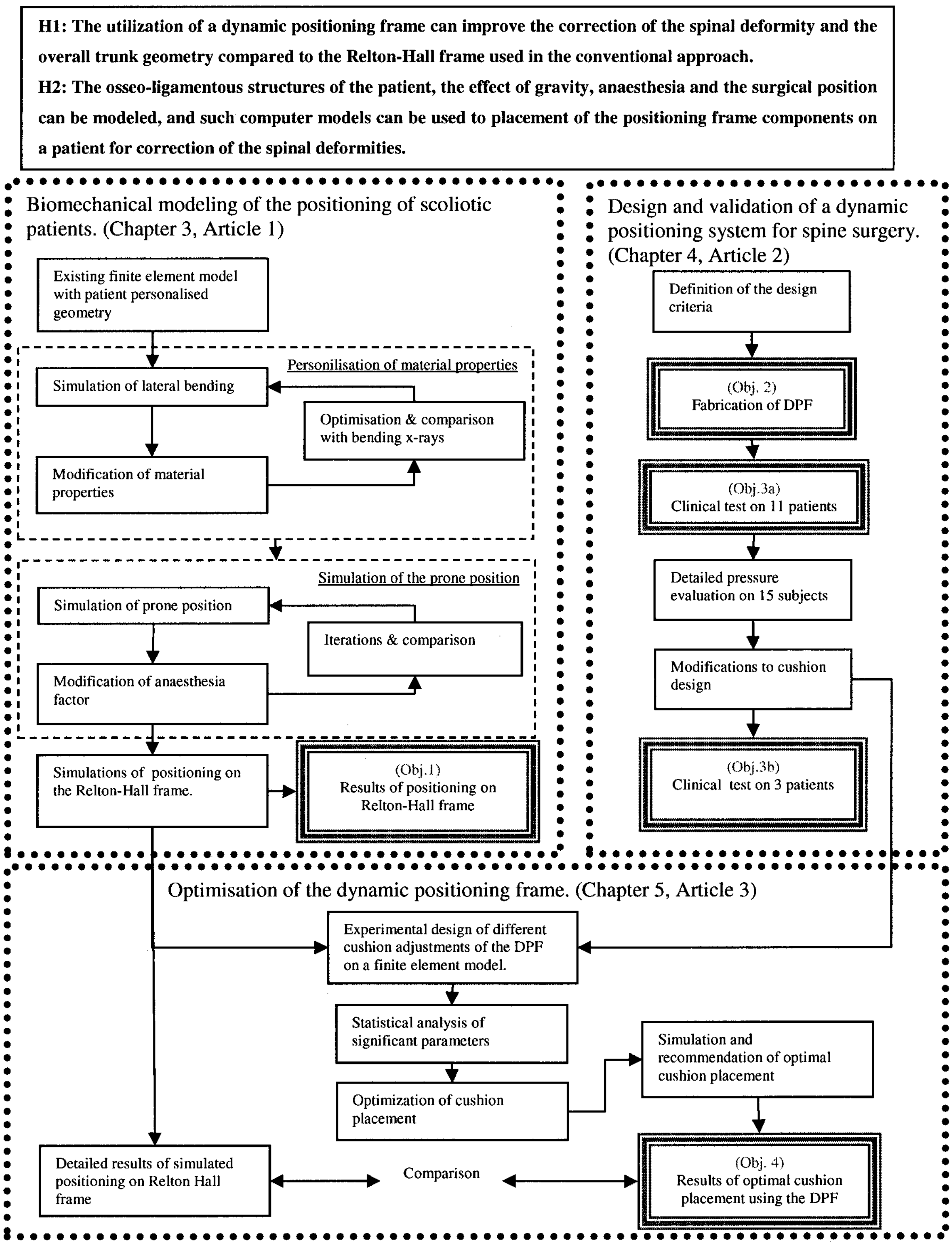


Figure 2.1: Thesis layout. Highlighted boxes represent a satisfied objective.

CHAPTER THREE: BIOMECHANICAL MODELING OF SCOLIOTIC PATIENTS IN THE PRONE POSITION

3.1 Situation of the first article

The realisation of the first objective of this project is presented in this chapter with the help of a single article. To reiterate, the first objective is to refine a finite element model such that it is able to simulate the influence of position and anaesthesia while incorporating patient personalised material properties.

The geometry of the model was obtained with the help of 3-D reconstruction techniques (Delorme et al. 2003). From this geometry a patient personalised finite element model of the spine was created (Aubin et al. 1995). The material properties were personalised from the patients bending x-rays. Supine bending x-rays are routinely taken at the Sainte Justine Hospital to characterise the flexibility of the patient prior to surgery. Gravity was simulated by applying distributed loads on each vertebra representative of the mass of each trunk slice. One of the novel methods applied to this model is the simulation of gravity. 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 caudal direction to each vertebra. In addition anaesthesia was simulated by applying a relaxation factor to the soft tissues. For validation purposes the simulation results were compared to the patients intra-operative x-rays.

This study is presented with the following article entitled "Biomechanical simulations of scoliotic spine correction due to prone position and anaesthesia prior to surgical instrumentation." for which the contribution of the first author is considered to be 90%. The authors are the following:

Kajsa Duke, Carl-Eric Aubin, Jean Dansereau, Hubert Labelle.

3.2 Article 1: Biomechanical simulations of scoliotic spine correction due to prone position and anaesthesia prior to surgical instrumentation

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

Kajsa Duke MSc. [a,b,]
Carl-Eric Aubin PhD. [a,b*]
Jean Dansereau PhD. [a,b]
Hubert Labelle MD [a]

[a] Research Centre, Sainte-Justine Hospital, Montreal, Canada
[b] Department of Mechanical Engineering , École Polytechnique de Montreal, Canada

*Corresponding author.

Address: Canada Research Chair "CAD Innovations in Orthopedic Engineering"
Department of Mechanical Engineering,
École Polytechnique, P.O. Box 6079, Station "Centre-ville"
Montréal Québec Canada
H3C 3A7

Fax: 01-514-340-5867.

E-mail: carl-eric.aubin@polymtl.ca

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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 personalised material properties.

Methods: A finite element model of the spine, ribcage and pelvis was created from the 3-D standing geometry of 2 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 intervetebral disks were personalised 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 3-D 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 personalised. 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 1corrected from 62° to 47° and 43° to31°. 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 3-D 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.

Introduction

Scoliosis is a three-dimensional (3-D) 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 3-D 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 emphases 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 then 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 Y 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 K 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 K 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.

Methods

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 et al. 1983). A summary of the patient's clinical information can be found in Table 1.

Model description

The 3-D geometry of the patients was obtained from a bi-plainer reconstruction technique (Delorme et al. 2003). From this geometry a patient personalised 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, intervetebral disks, ribs, sternum, pelvis and cartilage are represented by 3-D 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 intercostals elements are represented by 3-D 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 personalised properties. To personalise the material properties, supine pre-operative bending simulations are performed with different material properties and the results are compared and optimised 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 (α) until the model behaviour best matches the intra-operative x-ray.

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 4 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 4 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 optimized were the proximal thoracic (PT), the main thoracic (MT) and the lumbar (L). The model is then optimized 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 3-D geometry was minimised. The difference was calculated from equation 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.

Equation 1:

$$\text{Difference} = 0.1*(PTr_{\text{Actual-Simulated}})^2 + 0.3*(MTr_{\text{Actual-Simulated}})^2 + 0.1*(Lr_{\text{Actual-Simulated}})^2 + 0.1*(PTl_{\text{Actual-Simulated}})^2 + 0.1*(MTl_{\text{Actual-Simulated}})^2 + 0.3*(Ll_{\text{Actual-Simulated}})^2$$

Where PTr is equal to the proximal thoracic Cobb angle during right bending. Greater weighting was given to the main thoracic curve (MTr) during right bending, as well as to the lumbar curve (Ll) 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 personalised material properties.

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 figure 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 apposed 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 3-D 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 x 43 cm (14 x 17in) 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 3-D 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 3-D 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 personalised material properties obtained from the bending tests. Finally, a relaxation factor (α) 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 3-D intra-op geometry and the 3-D simulation geometry. A local minimum was found, and further iterations were performed around that local minimum.

Results

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° to 33° and 18° to 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 equation 1, was reduced from $263mm^2$ to $170mm^2$.

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 equation 1, was reduced from $113mm^2$ to $89mm^2$.

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.

Simulation of the intra-operative prone position

Figure 2 represents the 3-D reconstruction of the vertebral body line for the patients standing (x-ray), initial prone simulation, prone simulation with personalised material properties and anaesthesia factor, and the 3-D 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° to 46°) and 37% (43° to 27°) for the thoracic and lumbar curves respectively. The simulation with the original material properties was only able to provide 17% (62° to 52°) and 22% (43° to 33°) correction for the thoracic and lumbar regions. By using the optimised intervetebral 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°. Figure 3 shows the results of the iterations used to determine the anaesthesia factors for both patients. The total difference in the vertebral centroid positions was originally 167mm and reduced to 97mm and 96mm for the personalised 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 80mm and reduced to 76mm and 30mm for the personalised material properties and the anaesthesia simulations respectively. Although there is still some discrepancy in the Cobb angles, visual inspection on Figure 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.

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 the supine simulation to the work of Torell et al. (1985) who observed an average of 9° ± 6° correction when lying supine, both patient simulations appear to fall within this range.

It is difficult to know the exact contribution of the anaesthesia on the spine curve modification. When Peterson et al. (1995) compared their results on anesthetized subjects with the results of Tan et al. (1994) on awake volunteers they concluded that anaesthesia does not appear to affect lordosis. In looking at various studies analyzing Cobb angle in the supine position of nonanesthetized patients the average correction is about 24% (Klepps et al. 2001; Torell et al. 1985). Comparing those studies to those of Behairy Y 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 anaesthesia), 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 Figure 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 know 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 slightly with age (Torell et al. 1985). It was the hope that personalising 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 personalised 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 hypolorditic. 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 3-D 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, 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 personalised 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 3-D curve correction during patient positioning. Active correction forces applied to the trunk during surgery may also add additional 3-D 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 3-D correction.

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 personalised 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 personalised 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 3-D correction during patient positioning.

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Table 3.1: Article 1 Table1: Clinical data for both patients

	Clinical Information		Cobb angles (º)	Standing	Left Bending	Right Bending	Prone Intra-op
Patient 1	Age	18	Proximal thoracic	45	22	42	*
	King Curve Type	II	Main thoracic	62	59	35	46
	Weight (kg)	58.6	Lumbar	43	11	42	27
Patient 2	Age	11	Proximal thoracic	39	11	33	25
	King Curve Type	III	Main thoracic	70	73	53	46
	Weight (kg)	36.8	Lumbar	40	16	33	21

* Data not available

Table 3.2: Article 1 Table 2: Results of the supine and bending simulations.

Sim 1 is without the personalised material properties. Sim Opt is with the optimal patient personalised properties.

		Standing	Supine Simulation		Left Bending			Right Bending		
		X-ray	Sim 1	Sim Opt	X-ray	Sim 1	Sim Opt	X-ray	Sim 1	Sim Opt
Patient 1	Thoracic	62	50	47	59	53	50	35	49	44
	Lumbar	43	35	33	11	30	23	43	48	50
Patient 2	Thoracic	70	61	59	73	66	68	53	59	57
	Lumbar	40	37	38	16	17	19	33	42	41

Table 3.3: Article 1 Table 3: Results of the prone simulations.

Sim 1 is without the personalised material properties. Sim Opt is with the optimal patient personalised properties. Sim alpha (α) is with the anaesthesia factor. DVCP is the difference in the vertebral centroid positions.

		Standing	Prone Position			
		X-ray	X-Ray	Sim 1	Sim Opt	Sim α
Patient 1	Thoracic (º)	62	46	52	48	47
	Lumbar (º)	43	27	33	32	31
	DVCP (mm)	na	0	167	97	96
Patient 2	Thoracic (º)	70	46	63	63	55
	Lumbar (º)	40	21	37	37	32
	DVCP (mm)	na	0	80	76	30

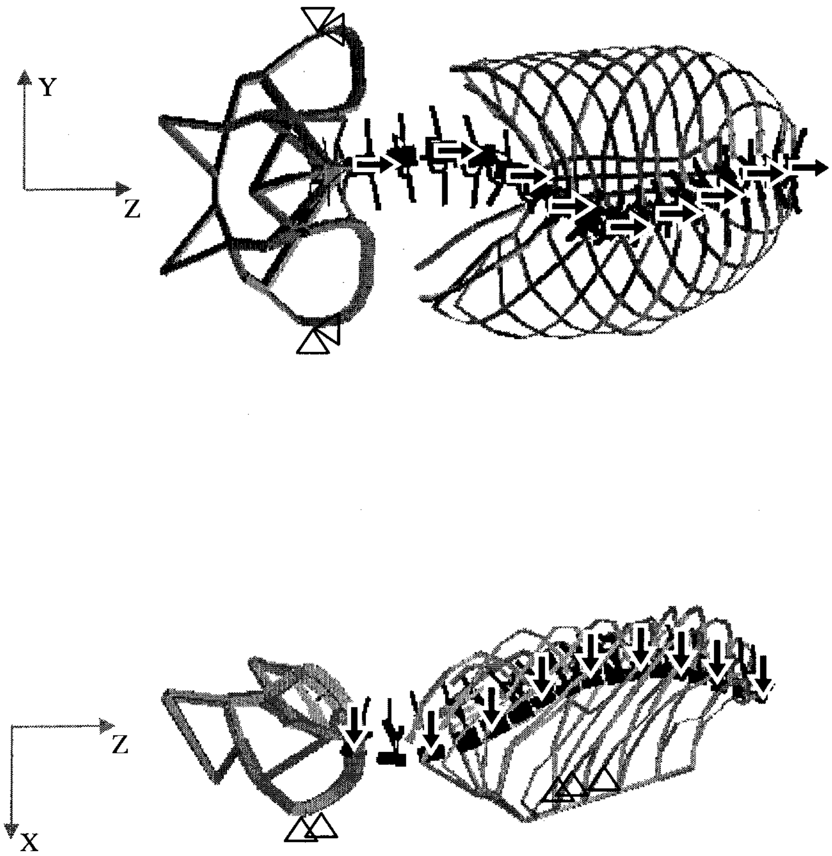

Figure 3.1: Article 1 Figure 1: Finite element model of the scoliotic patient showing the pelvis, spine and rib cage.

Boundary conditions and gravitational forces are also shown. The posterior view shows the boundary conditions in the y and z direction as well as the "traction type" load in the positive z direction. The sagittal view shows the boundary conditions in the x direction as well as the gravitational load in the positive x direction. For clarity, the gravitational forces are only shown on every other vertebra.

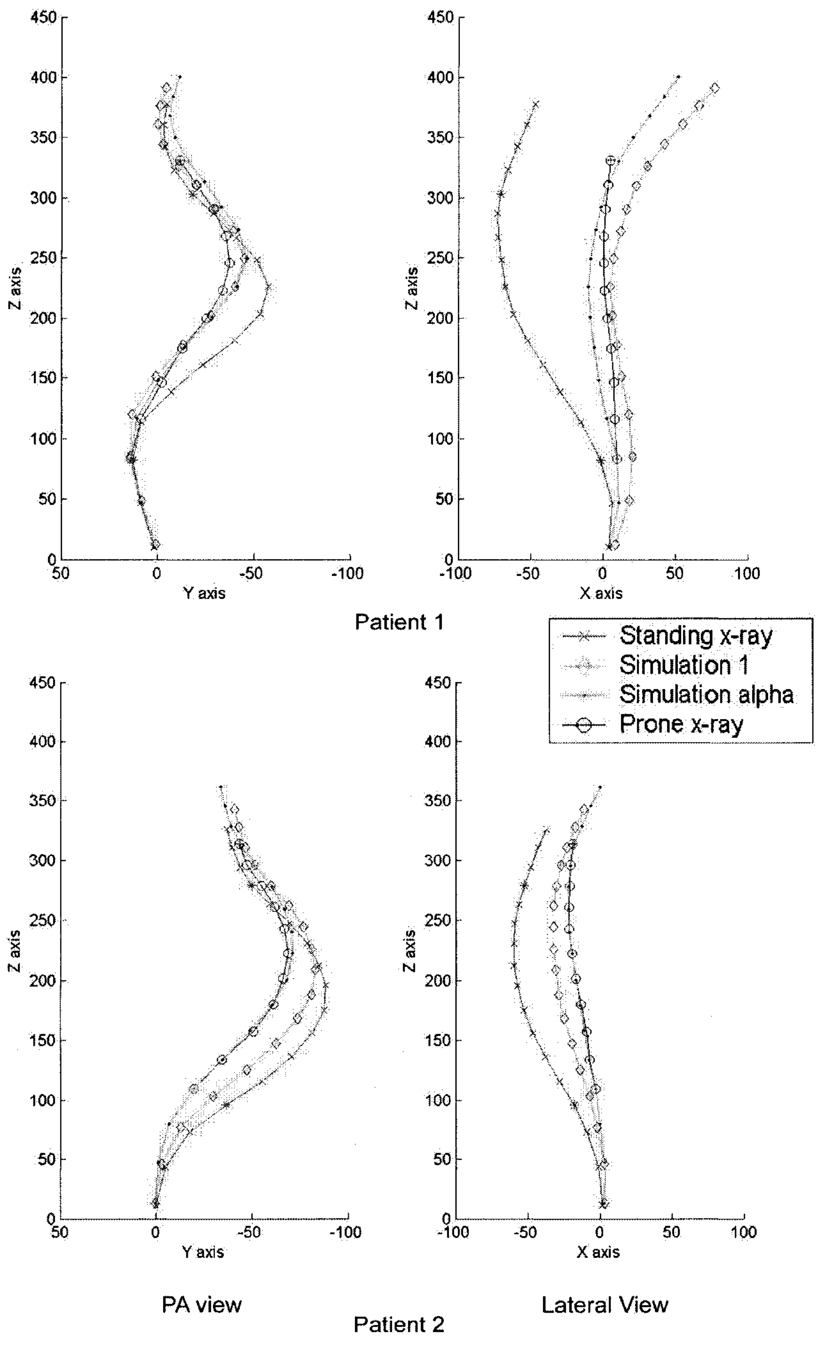

Figure 3.2: Article 1 Figure 2: Spine curves of the two patients.
Obtained from the radiographic acquisitions standing and prone, and from the finite element model simulations. All data are in mm.

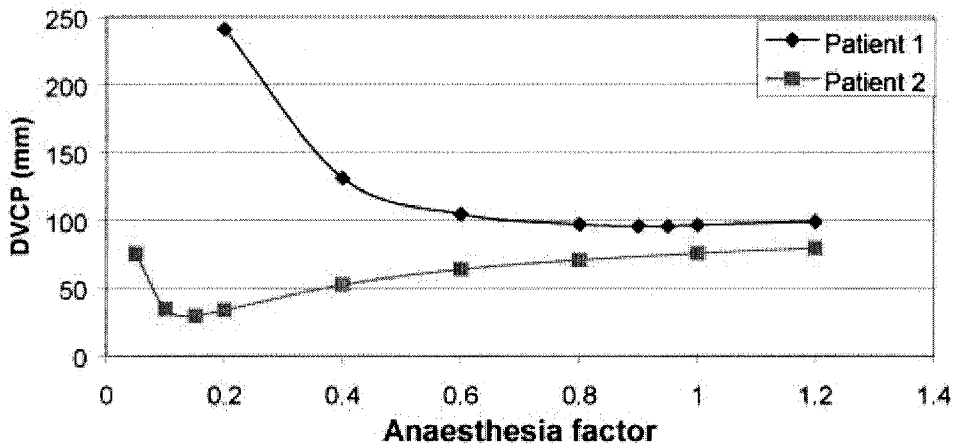

Figure 3.3: Article 1 Figure 3: The difference in the vertebral centroid positions (DVCP) versus the anaesthesia factor for the two patients.

CHAPTER FOUR: A DYNAMIC POSITIONING FRAME FOR SPINE SURGERY

The realisations of the second and third objectives of this project are presented in this chapter with the help of a single article and additional complementary sections. To reiterate, the second objective is to set out the design criteria and to incorporate these specifications into the design improvement and fabrication of a prototype of the DPF that could be used in the operating room. The third objective is to test the DPF on a group of scoliotic patients first, in a clinical setting and then, in the operating room.

4.1 Definition of the design criteria.

Before beginning with the design of the dynamic positioning frame (DPF), a clear definition of the design criteria was required. In order to establish such design criteria, the existing Relton-Hall system was analysed as well as the first clinical prototype shown in Figure 1.12.f (Pazos 2001).

One of the first restrictions to the design of the DPF was that the abdomen was to remain pendulous in order to reduce the amount of bleeding during the surgery. It was also decided that the DPF would be able to support a 100kg patient. The dynamic positioning frame is designed to be used at the Sainte Justine paediatric hospital. By setting a limit of 100kg the system will be able to support the 99 percentile woman who is 98.9 kg (Tilley 2002).

The next set of design criteria focus around the cushion placement and that the system is able to be adjusted for various different sized patients. First, it was decided that the design would allow for movement of the cushions in three directions x, y, and z, without having to raise the patient off the cushions. The x direction runs anterior to posterior, y direction runs from the right to the left hand side of the patient, and the z direction points towards the head. These directions are defined by Stokes and were already presented in Figure 1.1 with reference to the patient's body. These directions are defined with respect to the table as shown in Figure 4.1 on the DPF. The existing Relton-Hall style cushions are fixed with Velcro. This requires that two people are present to move one cushion, first the patient must be raised and then the cushion is ripped off the base and repositioned. Minima and maxima had to be determined for the range of motion for the cushions with the help of the anthropometric data provided by Tilley (2002).

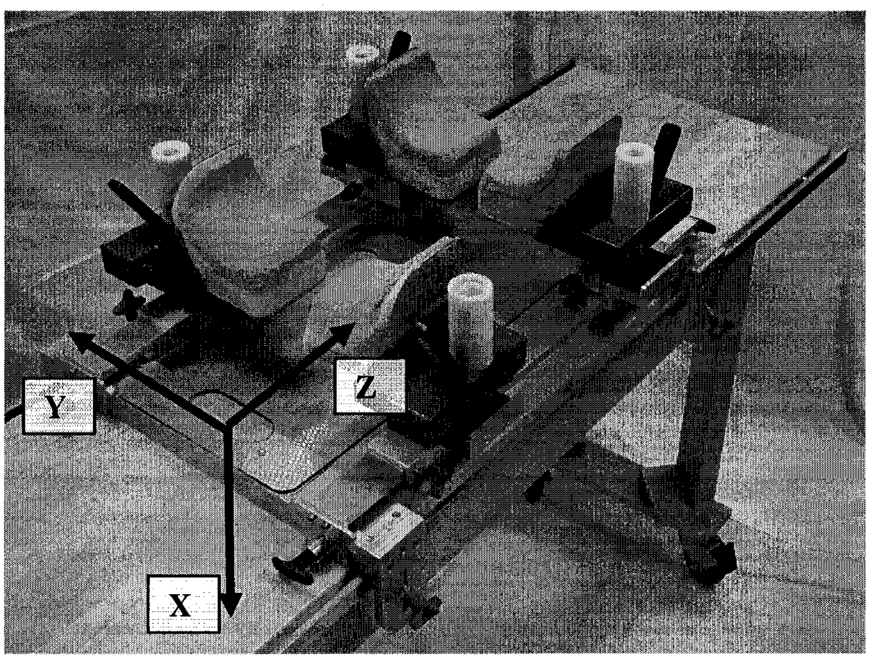

Figure 4.1: The Dynamic positioning frame and coordinate system

The cushions must have an adjustable length in the z direction based on torso lengths. For the torso lengths the distance from the hip joint to the shoulder joint are 332 mm for the 8-year-old and 521 mm for the 99 percentile man (Tilley 2002). Given these measures, the design criterion was set to 300 mm and 530 mm. For a very small patient there is a risk that the cushions will touch constricting the abdomen, a problem that is already present with the Relton-Hall cushions. The length of the table can easily accommodate the length of the positioning system so there are no constraints imposed for the total length.

The cushion must also be adjustable in width along the y axis. In order to accommodate the hips of an 8-year-old child and the 99 percentile woman the minimum was set to 200 mm and the maximum was set to 430 mm. These measurements are crucial in the table design as a very small patient will be limited when the cushions touch and with a large patient the amount that the cushions extend beyond the width of the table is a prime concern for the surgeon.

Another design criterion is that the height of the cushions is adjustable by at least 75 mm. The purpose of this height adjustment is to be able to alter the patient's balance by raising the chest cushions relative to the hips, to potentially alter the kyphosis or to potentially alter the rib hump by raising the left side while lowering the right side. It was also requested that the cushions can be rotated around the x axis. This rotation will allow the cushions to better conform to the patients trunk and latissimus dorsi musculature.

The cushions of the Relton-Hall system are convex and cushions that better contour the sides of the patient was an added criteria. In order to facilitate patient transfer it was requested that the height of the system on the table be minimized and a height of less then 200 mm was set as a guide line.

The positioning frame must be adaptable to various surgical tables as the table used in the clinic is slightly different from that in the operating room. As well, since the design will involve the addition of external correction forces, it was requested that the frame locks securely to the table. Ideally the system would not extend beyond the width of the table which is 55 cm from rail to rail.

The system must be easy to clean and ideally it could be autoclaved. There were also considerations taken into account to facilitate the ease of use of the system by the surgeons and technicians installing the frame on the table.

Another major area that must be considered in the design of the positioning frame is the clarity and accessibility to the radiograph. For a clear, less distorted image, the distance between the patient's stomach and the radiograph should be minimised. There must be sufficient space for a 28 x 36 x 2 cm (11 x 14 x 0.5 in) cassette to fit under the patient so that an AP x-ray can be taken during the surgery to verify screw placement. Also, it is imperative that the complete spine is visible during the acquisition of both AP and lateral x-rays. It is also desired that the ribcage and pelvis are also visible for both x-rays.

The established design criteria were tabulated and given weightings as shown in Table 4.1. The existing Relton-Hall frame, the clinical prototype and the dynamic positioning frame are all compared given these specific design criteria. It is interesting to note that the Relton-Hall system ranks the highest in terms of the x-ray criteria as it is only made of wood, Velcro, foam and gel pads. The clinical prototype is strong for the cushion placement components but it was not conceived for use with x-rays and so it is therefore, weaker in that respect. The DPF ranks the highest of the three, and will be described in the following sections.

Table 4.1: Design criteria and weightings.

Design Criteria	Weighting	Points	Score	Relton-Hall	Clinical Prototype	DPF
Abdomen pendolus	5	4 post frame	100	5	5	5
		Sides of patient supported	80			
		No	0			
Able to support 100 kg	10	100 kg	100	10	10	10
		80 kg	75			
		60 kg	50			
		Less than 60 kg	0			
Dynamic and asymetrical	15	3D with the patients weight	100			15
cushion displacement		2D with the patients weight	80		12	
		1D with the patients weight	50			
		Must lift the patient to move cushion	20	3		
Adjustable length	5	From 300 to 530 cm	100	5	5	5
		Shortened by less than 10 cm	80			
		Shortened between 10 to 20 cm	40			
		Shortened by more than 20 cm	0			
Adjustable width	5	From 200 to 430	100			5
		Shortened by less than 10 cm	80	4	4	
		Shortened between 10 to 20 cm	40			
		Shortened by more than 20 cm	0			
Adjustable height	5	By 75 mm	100			
		From 50 to 75 mm	80		8	
		From 25 to 50 mm	60			6
		Less than 25 mm	0	0		
Rotation of cushions	5	90 or more	100		5	5
		45 degrees	60			
		No rotation	0	0		
Cushions contour patient	5	Yes	5		5	5
		No	0	0		
Height of system on table	5	Less than 200 mm	100	5		
		Between 200 to 300 mm	80			4
		More than 300 mm	0		0	
Adapatable to various surgical	5	Yes and locks securely	100			5
tables		Yes	80	4	4	
		No	0			
Width exceeds current table	5	From 0 to 2 mm	100	5		
		2 to 20 mm	90			
		20 to 40 mm	70			3.5
		40 to 60 mm	40			
		more than 60 mm	0		0	
Easy to clean	5	Can be autoclaved	100			
		Impermeable and can be washed	80	4	4	4
Distance between patient's	5	0-50 mm	100			
stomach and radiograph		50-100 mm	80	4		
		100-150 mm	50			2.5
		150 mm or more	0		0	
Accessibility to radiograph	10	Without special manipulation	100	10	10	10
11 in by 14 in		With the use of a pallet	80			
		Cushions must be displaced	0			
Clarity of radiograph	10	Spine, pelvis and ribcage AP & Lat	100	10		
		Spine, pelvis and partial ribcage	70			7
		Spine, pelvis AP & Lat	50			
		Spine, pelvis AP only	20		2	
	100			69	74	92

4.2 Situation of the second article

A second article titled “A dynamic positioning frame for Spine Surgery” is presented in order to describe the design of the DPF and a preliminary study that tested the DPF on a group of scoliotic patients in a clinical setting. For this study the contribution of the first author is considered to be 75%. The authors are the following:

Kajsa Duke, Carl-Eric Aubin, Annick Koller, Jean Dansereau, Hubert Labelle.

4.3 Article 2: A Dynamic Positioning Frame for Spine Surgery

A Dynamic Positioning Frame for Spine Surgery.

Kajsa Duke MSc. [a,b]
Carl-Eric Aubin PhD. [a,b*]
Annick Koller HES [a,b]
Jean Dansereau PhD. [a,b]
Hubert Labelle MD [a]

[a] Research Centre, Sainte-Justine Hospital, Montreal, Canada
[b] Department of Mechanical Engineering , École Polytechnique de Montreal, Canada

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[*]Corresponding author
Address: Canada Research Chair "CAD Innovations in Orthopedic Engineering"
Department of Mechanical Engineering,
École Polytechnique, P.O. Box 6079, Station "Centre-ville"
Montréal, Québec Canada
H3C 3A7
Fax: 01-514-340-5867
E-mail: carl-eric.aubin@polymtl.ca

Abstract The positioning of the patient is an important step in scoliosis surgery. Present day positioning frames focus on keeping the abdomen pendulous to reduce blood loss. When the patient is lying prone, there is already some spontaneous correction of the scoliotic deformity due to gravity. It has also been shown that the sagittal alignment of the spine can be altered due to leg position. However, the design of the positioning frame has not been exploited to help improve the scoliotic deformity in the frontal plane. This study describes and provides a preliminary evaluation of a Dynamic Positioning Frame (DPF) for spinal surgery. The trunk geometry and pressure was measured for eleven unanaesthetised patients in various positions. The patients were standing, lying prone on the DPF, lying prone on the DPF with the addition of external forces and finally, lying prone on the existing Relton-Hall (R-H) frame for comparison purposes. A statistically significant lengthening of the spine was recorded when the patient was lying prone on every frame compared to the standing position. There was an improvement in the torsional deformity (delta hump) of the patients lying on the DPF with external forces. The decrease in kyphosis is less when lying on the DPF with external forces. Overall, the DPF provides a dynamic way of modifying the patient's position before and during surgery. The trunk geometry of the patient was improved when lying on the DPF with external forces, however, further investigation is required to determine the optimal cushion placement. Higher pressures were recorded on the DPF when compared to the Relton-Hall frame. Cushion modifications on the DPF are necessary before application in the operating room.

Keywords: Surgery, scoliosis, prone, positioning frame, trunk geometry.

Introduction

For severe cases of scoliosis, where the Cobb angle is greater than 45°, surgical correction is often required to straighten and stabilise the curve. During the most common surgical procedure, the patient is 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 of the Cobb angle of approximately 57% is obtained after surgery [4].

The positioning of patients during spinal surgery is a critical step. A recent review of positioning on the surgical table highlighted that advantages and disadvantages of various frames focusing mainly on blood loss [21]. At most hospitals chest roles or four post frames, like the Relton-Hall (R-H) positioning frame shown in Figure 1, are used to position the patients and keep the abdomen pendulous during the surgery to minimize blood loss [20]. Other tables used for spine surgery include the Jackson, Wilson and Andrews tables (OSI, Union City, CA, USA). Traction has also been used during scoliosis surgery but a recent study found it does not create an advantage in terms of post-operative correction [12].

During the surgery, routine x-rays are taken to orient the surgeon to the vertebral level as well as to verify instrumentation placement. A PA x-ray is taken from directly above the patient and a lateral x-ray is taken from the side of the patient. To obtain the best quality of x-rays, the distance between the patient and the x-ray film should be minimized.

Complications resulting from improper positioning must be taken into account when designing a positioning frame for surgeries. Reddening of the skin and, in severe cases, pressure sores can occur due to the length of time of the surgery and the fact that the patient is positioned on four small cushions. Another reported complication is vision loss due to improper positioning of the head. It is recommended that facial and ocular compressions are avoided [8,16]. As well, some neural damage can occur due to improper positioning of the arms [22]. There have also been reported cases of femoral artery ischemia or injury to the lateral femoral cutaneous nerve due to improper positioning of the legs [15,28]. Though these complications are rare, it is important to note that they have been recorded and that every effort should be made by the surgeon, anaesthesiologist and nurse to reduce the risk of any complications when positioning the patient regardless of the positioning frame.

The positioning of the patient's legs during spinal surgery can affect the lordosis. It is important that normal standing lumbar lordosis is maintained after surgery. In general, a lordosis, similar to that found in standing, is maintained during and after surgery when the hips are flexed less then 30° [14,17,23,24,27].

A few studies have shown spinal deformation in the frontal plane can decrease prior to the insertion of posterior spinal instrumentation [2,4,10]. This improvement can be attributed to prone positioning of the patients under anaesthesia and surgical exposure. Other studies have shown the trunk deformity of patients with adolescent idiopathic scoliosis can be modified while the patient is lying on the surgical table [5,13].

While positioning of the patient is recognized as an important step in surgery, design of the surgical table or positioning frame has not been exploited to help improve surgical correction. The purpose of this paper is to present the design of a dynamic positioning frame (DPF) that could provide improved correction of the patient's deformity and report a pilot study that tests this frame on scoliotic patients before its use in the operating room.

Material and Methods

Dynamic Positioning Frame (DPF) Description

The DPF will be described following the installation procedure from a bottom-up approach (US patent Docket No. 15490-7US SC/ip, international patents pending).

It is composed of crossbars that extend over the existing surgical table and are locked into position on the existing table rails. On top of the crossbars is the Lexan® sheet that serves to protect the x-ray from a pendulous abdomen (Figure 2a). Two transverse rails are positioned on the crossbars and are locked into place with the help of a spring mechanism (Figure 2a). There are three positions for the rails, small, medium and large depending on the size of the patients. The setting is selected before positioning the patient and finer independent adjustments are later made to each cushion with the Delrin® lateral slider. The frame can accommodate patients from 19 cm to 42 cm at the hips therefore it would accommodate the average 8 year old child to the 99 percentile woman [26].

Aluminium carriages are placed on the transverse rails (Figure 2b). There are four aluminium carriages, two for the chest cushions and two for the hip cushions. On top of the aluminium carriage is the lateral slider which is made of Delrin® to provide radio-transparency. Placed on the Delrin® slider is the support post for the cushion holder (Figure 2c). The support post is made of a combination of nylon and aluminium. The base is aluminium to provide strength and a high friction surface for the Manfrotto® grip to attach to. The upper section made of nylon is to provide direction for the Manfrotto® grip while still being radio-transparent.

The Manfrotto® grip serves two functions. First it is used to move the cushion up and down and second the cushions can be turned to better contour the sides of the patient. The Manfrotto® grip is housed within the cushion support (Figure 2-D).

The cushion support is made from bent and welded steel, designed to be low profile so that it will not occlude the spine during the lateral x-rays. The cushions are made with Lexan® folded at 90° and reinforced with an aluminium plate at the back to secure its attachment onto the cushion support (Figure 2-D). The width of the cushions are smaller then the existing Relton-Hall type cushions to better fit paediatric patients. The shape of the cushion was designed to be concave to better conform to the curves of the patient. The cushions are made with various layers of Tempur® foam, which is an improvement over our Relton-Hall cushions composed of standard foam. The upper layer is made of gel. The cushion is then covered with an impermeable flexible material. In order to better follow the natural curvature of the legs, and to better preserve the lordosis, the distal edge of the hip cushions are angled at approximately 30°.

The utility of the DPF is a combination of the movements permitted by the cushions and their overall shape. The four independent cushions, set on sliders and rails, can be adjusted to accommodate different sized patients with varying asymmetries. To correct for decompensation, the chest cushions can be slightly offset from the hip cushions to help centre the top of the patient. Both chest cushions can be raised to promote more kyphosis. By raising the left chest cushion more then the right chest cushion, the rib hump of a right thoracic curve can be reduced. The cushions are shaped to contour the sides of the patient. This helps secure the pelvis, and also provides some resistance along the sides of the patient's chest when the external forces are being applied. The 30° angle at the distal edge of the hip cushions allows for better positioning of the lower extremities. It is also expected to reduce the risk of pressure sores as well as the risk of damage to the femoral cutaneous nerve as reported with the Relton-Hall cushions.

Pilot Study

Eleven patients (nine females and two males) with AIS, who were potential candidates for surgery but with no previous surgery, were recruited from the scoliosis clinic at Sainte Justine's Hospital. The average age was 14.3 years (11 to 17) and all patients had a right thoracic Cobb angle that averaged 52° (35° to 64°).

The trunk geometry was measured in four positions using an optoelectric system (Polaris, Northern Digital Inc. Canada) and a protocol similar to that used in a previous study [11]. Ten passive markers were placed on the patients back as shown in Figure 3. From these markers various indices were calculated including shoulder angle, decompensation distance, balance angle, spine height, rib hump, kyphosis height, lordotic hump and delta hump. The triad of markers placed on S1 served as a reference where the origin (S1) was the centre of the triad, the y direction was towards the left and parallel to the line formed by 8a and 8b, the z direction was towards the head in plane with all three markers and the x direction was orthogonal to these pointing forward. The shoulder angle is measured between points 2 and 3 in the frontal and transverse planes. The decompensation distance, balance angle and spine height were all measured using points 1 and 8. The rib hump was measured between points 4 and 5 while the lordotic hump was measured between points 6 and 7. The kyphosis height was calculated as the distance from the midpoint of points 4 and 5 to the line used to measure the spine height as projected onto the sagital plane. The delta hump is the rib hump minus the lordotic hump and provides a measure of the torsional deformation of the patient's trunk.

The first acquisition was in the standing position (PI). The other acquisitions were in the prone position, with the patients being placed on the cushions in order to reproduce the operative position. The anterior superior iliac crests prominences were placed in the centre of the hip cushions and the chest cushions were placed under the axilla supporting the rib cage. The hips were slightly flexed and the legs were supported. The arms were supported as well as the head as shown in Figure 1. For each prone position, the pressure at the patient cushion interface was measured using a force sensing array (FSA) pressure mapping device with 256 captors and a maximum pressure of 300 mmHg (Vista Medical, Winnipeg, Canada). The pressure was recorded for 30 seconds and averaged to reduce the effects of breathing. Position II (PII) was prone on the DPF with no specific adjustments to the cushions. For position III (PIII) the patient was prone on the DPF and the cushions were adjusted and external forces were applied. A previous study looked at applying forces independently and determined that kyphosis could be increased by raising the chest cushions and the best correction was achieved when simultaneous forces were applied to the thoracic and lumbar regions (Duke 2002). In this study, the cushions under the chest were adjusted to help increase kyphosis, reduce the rib hump and then simultaneous forces were applied to the thoracic and lumbar regions to straighten the curve and further reduce the humps. These cushion adjustments and forces were applied based on visual observations and at the discretion of the same clinician for each patient.

Finally, the patient was placed on a conventional Relton-Hall type frame for comparison purposes. Paired student T-tests were performed for all the trunk geometry indices using the standing position (PI) as well as PIV, on the Relton-Hall cushions, as a reference. For the pressure measurements only PIV was used as the reference. Ethics approval was obtained from the Sainte Justine ethics committee and all patients and parents read and signed the information and consent forms.

Results

In Table 1 the average results for the trunk geometry as well as the average peak pressure recorded on each of the four cushions are shown. Though the shoulder angle changed for various patients, as a group there was no significant difference for any of the prone positions. The decompensation changed very little on PII and PIV but once force was applied, PIII, it changed from 0.8 cm to the right to 1.2 cm to the left. However, this difference was not statistically significant. The balance also changed but the difference was not significant. There were however significant changes in the spine height which increased in all prone positions by, 5.3 cm, 5.7 cm and 3.8 cm for PII, PIII and PIV respectively. The spine height of the 47.6cm for PIII was significantly greater then the spine height of 45.7 cm for PIV. The changes in rib hump are not significant, but on average it decreases from 6.1° while standing (PI) to 1.5° while prone on the DPF with corrective forces (PIII) and a positive trend is observed. The kyphosis height decreases significantly on the PIV and PII. However, when the cushions are adjusted (PIII) this decrease is less, and is not significant. For this parameter, no significant change from the standing position is the desired effect. Unfortunately, there was no significant change in lordotic hump. The change in delta hump was significant for the patients lying in PIII as it reduced from 8.3° to 3.7°.

Using the PIV on the Relton-Hall positioning frame as a reference for all of the pressure measurements, no significant differences were found with the exception of the left shoulder, which was significantly higher in PII and PIII on the DPF. However, it should be noted that the pressure was generally higher on the DPF.

Discussion

During the treatment of scoliosis one must not only correct the spinal deformity but the trunk deformity is also a major concern [7,19,25]. It was theorized that by optimising the presurgical trunk geometry, less energy would be required to correct the spinal deformity [11]. Until this time, no spinal positioning frame was available with the goal of optimizing the trunk deformity. The goal of this study was to present such a frame as well as some preliminary results of the trunk geometry of patients lying on the frame.

There was no significant difference in the shoulder angle in either the frontal or transverse planes. Mac-Thiong et al. also found no significant differences for the shoulders while patients were positioned on the Relton-Hall positioning frame [11]. Neither the Relton-Hall positioning frame, nor the DPF, provided any direct constraint to the shoulders. The decompensation distance and balance angle slightly changed showing no significance.

The increase in spine height measured from S1 to C7 was significant in all prone positions. This agrees with the fact that the absence of gravitational forces creates a straightening of the scoliotic and sagittal curves [4,9,14]. Though the scoliotic curves were not directly measured, the fact that PIII, on the DPF with corrective forces, has the greatest spine height, suggesting that it is not just gravity acting on the spine but the adjustments to the cushions and the external forces creating even more lengthening in the scoliotic plane.

The kyphosis height decreased significantly on for PII and PIV creating more of a flat back. Once the cushions were adjusted, PIII, this decrease was not significant. Creating a normal thoracic kyphosis, and avoiding the flat back, is a concern of scoliotic surgery [3]. The positioning of the patient, not just the instrumentation, could be partly responsible for this complication. A previous study analysed the applied forces on the DPF in independent steps, showed that by elevating the two chest cushions the kyphotic height increased significantly [5]. The design of the DPF provides greater flexibility of the cushion placement and aims to improve the deformity in both the sagittal and the frontal planes.

Even though the correction in rib hump is not significant the application of external forces (PIII) tend to provide some correction to the trunk deformity. There was no significant change in lordotic hump but this could be partly due to the fact that the standing lordotic hump was relatively small (2.2°) in this group of patients. However, the delta hump does show a significant improvement for PIII when external forces are applied. This improvement in delta hump was not observed on the Relton-Hall frame.

Analysis of scoliotic surgery should also include surface data, since even though there is a reduction in the scoliotic curve, there is insufficient improvement in trunk deformity [18]. The positioning of patients on the DPF with external forces appears to address the trunk deformity but the post-op results after instrumentation are yet to be determined.

The pressures were higher while the patients were lying on the DPF. This can be explained due to the reduced surface area of the cushions and the addition of external forces. The cushions are made with various layers of Tempur® foam and the upper most layer is made of gel commonly used to provide cushioning during surgeries. An ongoing study is looking at modifying the cushion design to further reduce the pressures. Also, since the pressures were especially high during the application of external forces it is not suggested that they be applied during the entire procedure. It is suggested that the external forces be applied after the hooks and screws have been implanted, and prior to rod placement, and kept in place for about 20 minutes while the rods are being rotated and/or translated and secured. The pressure was especially high on the left chest cushion compared to the right chest cushion for PIII. The high pressures measured on the left cushion are due to the reaction forces and show that the corrective forces, applied in the right thoracic region of the patient, are being transferred through the patient.

It is difficult to optimize the entire patient to minimize every parameter of the trunk geometry and not apply excessive forces resulting in excessive cushion interface pressure. For example, an improvement in rib hump may result in a deterioration of shoulder angle and excessive force on the right chest cushion. Finite element models and computer optimization have been used to study scoliosis biomechanics including braces and surgical instrumentation [1,6]. These tools will be used in a future study to help determine the cushion locations and applied forces to optimize the patient position.

Conclusion

The DPF provides the flexibility of adjusting the cushions and applying external forces that are not presently available with traditional frames. No significant improvement was observed in shoulder angle, however, there was improvement in the spine height and delta hump. Further investigation into the optimal placement of the cushions is required to help improve the correction of shoulder imbalance. Overall, the trunk geometry improves using the DPF compared to the standing position. There was a greater increase in spine height, less of a flat back, improvement in the rib hump and torsional deformity on the DFP with corrective forces compared to the Relton-Hall frame. Unfortunately, the pressures were generally higher on the DPF. Further modifications to the cushion design are underway to reduce the pressures on the patients while still providing correction to the trunk geometry.

Acknowledgements

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Table 4.2 Article 2 Table 1: Average results for the trunk geometry and average peak pressures

Average & (Standard Diviation)	Standing	DPF	DPF + Forces	Relton-Hall
	I	III	IV	II
Shoulder angle (deg)	-2.8 (3.3)	-1.3 (6.7)	-4.7 (5.3)	-2.4 (5.7)
AP Shoulder angle (deg)	-3.5 (8.1)	-2.6 (7.1)	1.2 (9.9)	-1.7 (4.9)
Decompensation Distance (cm)	0.8 (2.8)	0.8 (5.8)	-1.2 (5.7)	-0.1 (3.4)
Balance Angle (deg)	-18.7 (6.9)	-13.3 (5.1)*	-10.0 (6.1)	-11.9 (3.3)*
Spine Height (cm)	41.9 (3.0)	47.2 (5.3)*	47.6 (2.1)*†	45.7 (2.3)*
Rib Hump (deg)	6.1 (4.3)	6.1 (8.3)	1.5 (6.7)	5.4 (6.3)
Kyphosis Height (cm)	8.1 (3.7)	5.3* (2)*	5.9 (1.7)	5.3* (2.6)
Lordodic Hump (deg)	-2.2 (2.4)	-2.1 (5.1)	-2.2 5.9)	-2.2 (4.0)
Delta Hump (deg)	8.3 (3.6)	8.2 (6.8)	3.7 (6.0)*	7.6 (5.3)
Right Chest Pressure (mmHg)	-	67.7 (22.9)	65.3 (40.7)	59.5 (24.9)
Left Chest Pressure (mmHg)	-	75.2 (16.1)†	113.2 (50.8)†	61.5 (15.9)
Right Hip Pressure (mmHg)	-	124.2 (52.5)	122.8 (75.5)	92.6 (35.0)
Left Hip Pressure (mmHg)	-	117.1 (25.3)	121.1 (43.0)	94.6 (38.3)

* Statistically significant difference when compaired to the standing position
† Statistically significant difference when compaired to the prone Relton-Hall position

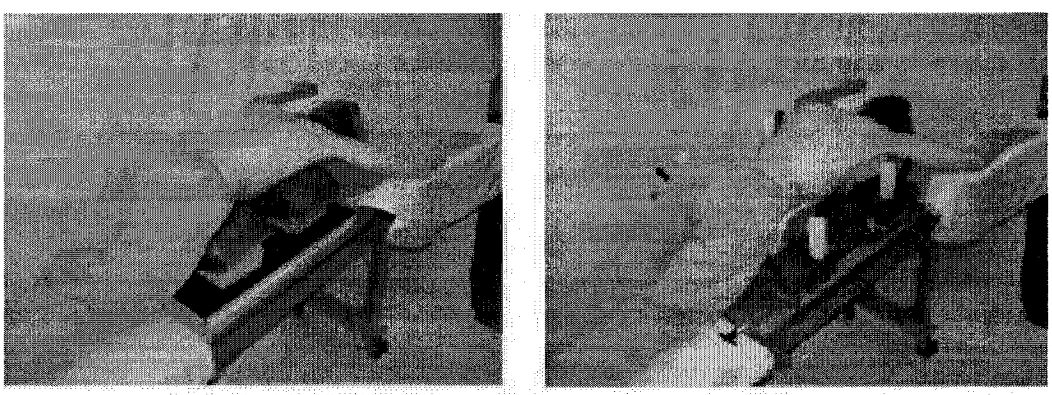

Figure 1: Figure (a) shows the Relton-Hall types positioning frame currently used at the Sainte Justine Hospital. (b) shows the dynamic positioning frame.

Figure 4.1: Article 2 Figure 1: Relton-Hall & DPF

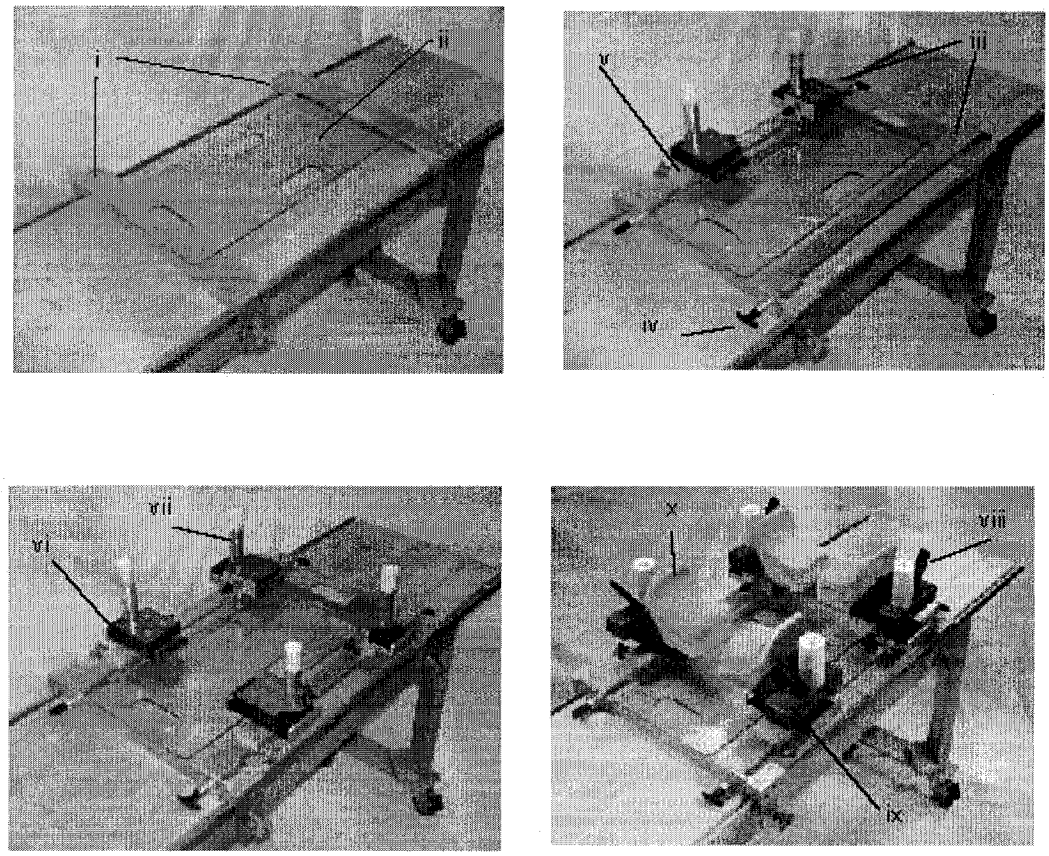

Figure 2: a) Showing the cross bars (i) and the lexan sheet (ii)
b) Showing the transverse rails (iii) the locking mechanism (iv) and the aluminum carraige (v)
c) Showing the delrin sliders (vi) and the support posts (vii)
d) Showing the Manfrotto handle (viii) the cushion support (ix) and the cusnions (x)

Figure 4.2: Article 2 Figure 2: Details of DPF

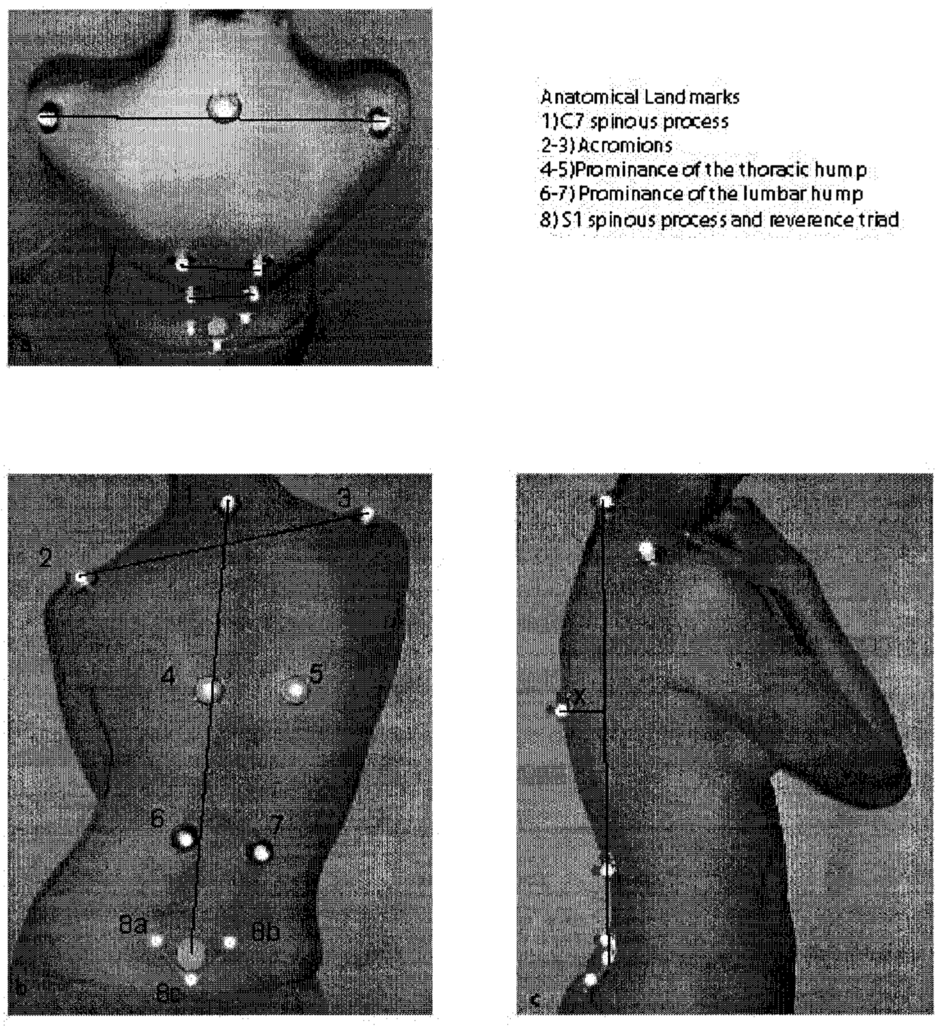

Figure 3.Measurement of the trunk geometry. Figure 3a, in the top view the Shoulder angle, rib hump, lumbar hump and delta hump are measured. Figure 3b ,in the AP view the AP shoulder angle and the sagital balance are measured. Figure 3c, in the sagital view the balance and kyphosis height are measured

Figure 4.3: Article 2 Figure 3: Trunk geometry measures

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4.4 Pressure measurements

Because there were higher pressures observed when the patients were laying on the DPF a more extensive pressure study of normal subjects lying on the Relton-Hall frame and the DPF was performed. This study was done with the help of Emanuelle Margues a visiting orthopaedic fellow from France and Guillaume Coulomb, a summer student from the department of kinesiology at the University of Ottawa.

The results of the study are in preparation for publication however, a brief summary is presented here as the results are important to the context of further design modifications and the protocol for the clinical study in the operating room.

The purpose of this study is to compare mean and peak interface pressure between the Relton-Hall frame and the DPF. The ultimate goal is to determine if the DPF has interface pressures that are lower then, or similar to, the Relton-Hall frame to ensure that the risk of pressure related complications are not increased.

4.4.1 Pressure measurements: Material and methods

Fifteen healthy subjects (5 males and 10 females) were recruited on a volunteer basis at the LIS 3-D lab of the Sainte Justine Hospital. The mean subject's age averaged 26.7 years (20-41). Pressure was measured with a force sensing array (FSA) pressure mapping device containing 512 sensors and a maximum pressure reading of 300 mmHg (Vista Medical Ltd, Winnipeg, Canada). The subjects were placed in the prone position on the Relton-Hall frame and on the DPF. They were asked to lay relaxed while data was collected, and later averaged for a 30 second period in each position. The positioning on both frames was similar to the positioning currently used during scoliosis surgery, with a head rest, arm rests and support cushions under the legs of the subjects. In addition, three different combinations of cushion placement were tested on the DPF. The positions are as follows:

A. Prone on the Relton-Hall frame with the anterior superior iliac crest prominences placed in the middle of the hip cushions and the chest cushions were placed at the infraclavicular chest areas of the subject;
B. Prone on the DPF, again with the anterior superior iliac crest prominences placed in the middle of the dynamic frame hip cushions and the chest cushions were placed at the infraclavicular chest areas of the subject;
C. Prone on the DPF, however the anterior superior iliac crest prominences were placed at the proximal end of the dynamic frame hip cushions and the shoulder cushions were turned approximately 30° to better follow the latissimus dorsi musculature of the patient's trunk. Wedge shaped support cushions were also placed under the thighs of the subject;
D. Prone on the DPF as described in position III except that the chest cushions were elevated 3.5 cm in order to provoke more of a kyphosis;
E. Prone on the Relton-Hall frame to serve as a repetition of the first position.

Statistical analysis was performed by using a paired student-t test to determine significant differences ($p \geq 0.05$) between the various positions using the position I as a base.

4.4.2 Pressure measurements: Results

The results for the mean peak and average pressures are shown in Table 4.3 and Table 4.4 respectively. First it is important to note that peak pressures of 300 mmHg were recorded on each of the 5 positions. When comparing position B to position A higher peek and mean pressures were observed on every cushion. The mean and the peak chest cushion pressures decreased when the cushions were turned to better contour the patient's body in position C. Subsequently raising the cushions in Position D, does not appear to significantly alter the pressure. Comparing position E to position A there was a significant difference in the chest cushion pressure on the right side.

Table 4.3: Mean peak pressures for 15 subjects

* denotes statistically significant compared to PA

(mmHg)	PA	PB	PC	PD	PE
Right chest	158	300*	278*	298*	114*
Left chest	184	291*	281*	287*	168
Right hip	285	300*	300	296	289
Left hip	291	300	299	291	296

Table 4.4: Mean average pressures for 15 subjects

* denotes statistically significant compared to PA

(mmHg)	PA	PB	PC	D	PE
Right chest	40	87*	68*	67*	35*
Left chest	45	87*	72*	69*	42
Right hip	57	84*	62	64	53
Left hip	58	84*	74*	70*	55

4.4.3 Pressure measurements: Discussion and conclusion

The main reason for the increased pressure, when comparing the DPF to the Relton-Hall, is the fact that the cushions of the DPF are smaller and the patient is resting on less surface area. The surface area for the Relton-Hall cushions is approximately 200 cm^2 compared to 120 cm^2 for the DPF. The mean average pressures were 50mmHg for PI and 86 mmHg for PII. Since pressure is equal to force over area the estimated force can be calculated at 133 N and 137 N for each cushion on the Relton-Hall and DPF respectively. By doing this calculation it is confirmed that the force has not changed since this is related directly to the patients mass. One can see that the smaller cushion surface area is directly responsible for the increased pressure. By slightly turning the chest cushions to better contour the patient's sides the chest pressure was improved but the pressures were still significantly higher and therefore too high. The applied force is related to the mass of the patient and the influence of gravity while the contact area is related to the size of the cushion and any boney prominences. Although efforts were made to improve the contact area by using several layers of high quality Tempur foam and an upper layer of gel, the overall reduced area of the cushions is the limiting factor. There was surprisingly a significant difference on the right chest cushion when comparing position A to position E and the reproducibility in repositioning the patients on the same system appears to be about ± 20 mmHg. This is slightly worse then the accuracy of the FSA, reported by Vista Medical is less than 10%.

In conclusion, the pressures are higher on the new system and it is recommended that modifications be made to the cushions to increase their surface area before proceeding with further tests on anaesthetised patients in the operating room. It is also recommended that the FSA be used to compare and evaluate the pressures of the patients lying on the Relton-Hall cushions to any new cushion design pre-operatively and that the pressures of the patients lying on the new system is monitored in the operating room.

4.5 Detailed cushion design

4.5.1 Modifications to existing base cushions

As a first step in determining the new base cushion design the contact surface area of the Relton-Hall Cushions was measured at approximately 200 cm^2 while the DPF cushions were only 120 cm^2 each. In order to increase the surface area of the cushions, add-ons were designed and fabricated as shown in Figure 4.4. For the hip cushions a 2.5 cm^2 border was border added increasing the contact surface area to approximately 210 cm^2. Because the pressure study confirmed that it was better to gently angle the chest cushions to follow the latissimus dorsi musculature an asymmetrical cushion add-on had to be designed. This design was selected as it better contours the shape of the patient's chest and still allows space for a pendulous abdomen. This asymmetrical design is cut up to follow the lines of the patient's lower ribs and extends on the patients chest towards the manubrium. The total contact surface area of the chest cushions is now at approximately 230 cm^2 each.

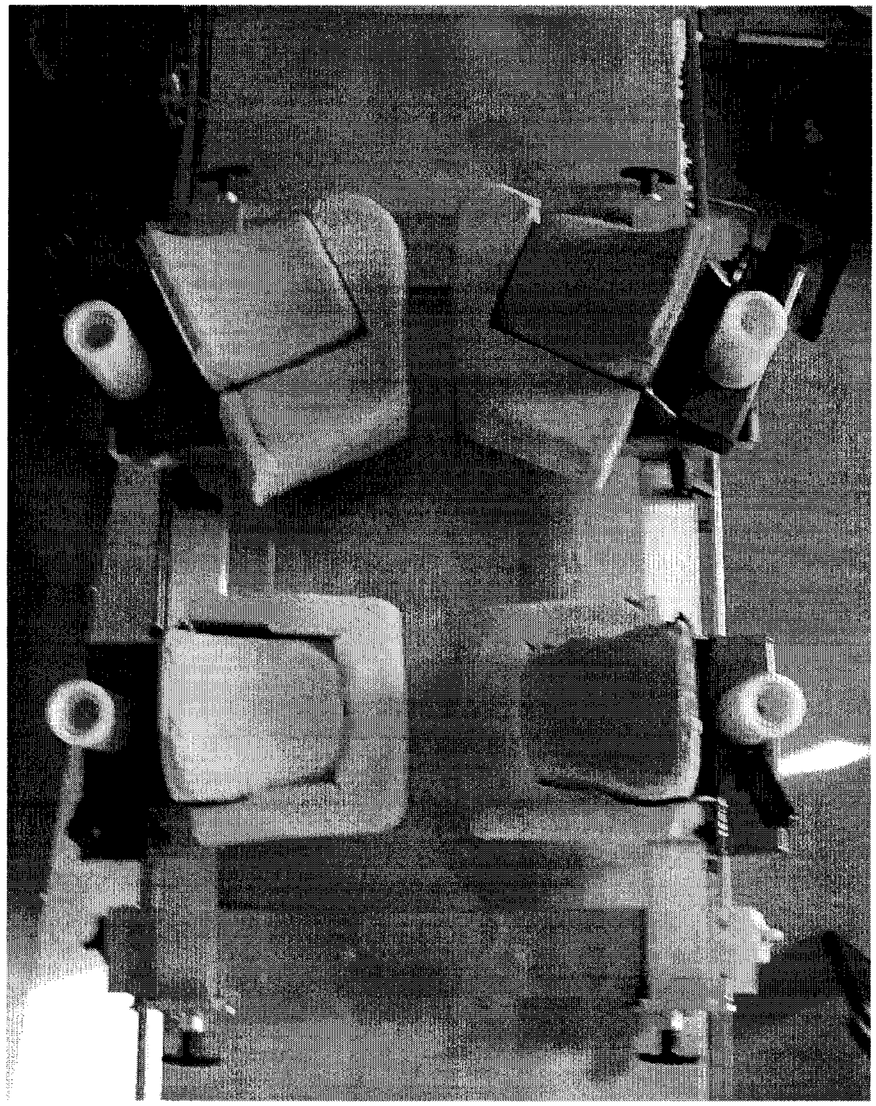


Figure 4.4: Top view of the DPF with base cushions and redesigned add-ons.

4.5.2 External correction cushions

The corrective forces that were applied in the initial clinical study, described in section 4.3, were applied manually and at the discretion of the clinician. These types of forces are often applied to the patient's trunk during the surgery by the resident or the surgeon's assistant while the rod is being inserted and/or rotated. In order for the DPF to be able to provide optimal correction, external correction cushions were designed. Because of the high forces observed, when applying the manual corrective force, it became apparent that the cushion could not be applied during the entire procedure and should only be placed just before rod insertion and rotation. This constraint leads us to the first design criterion which states that these cushions must be able to be sterilised. The cushions were also designed so that the surgeon had a sense of feel while they were pushing on the cushion and the patient. Ideally the cushion would automatically lock into place as the surgeon was applying the desired force. The cushions must of course be as easy as possible to use and attach to the rails of the standard operating table. One problem that arises when working in the sterile field is that anything below the level of the surgeons waste is considered not sterile even though it is covered with sterile sheets. Because of this, a mechanism that can be secured onto the rails, through the sterile sheets, without lowering the hand below waste level must be used. Another constraint is that there are no long components that extend beyond the edge of the table. Looking at the size of the patients, in order to apply lateral force the travel length for a cushion should be a minimum of 11.5 cm.

Two types of cushions were designed one to push on the patient's rib hump and the other to apply lateral corrective force to correct the curve. Both cushions attach to the table rails using the same principle base post. This base was modeled after designs by Thompson elite rail (Thompson Surgical instruments, Traverse City, MI) used to hold retractors during abdominal surgeries.

The rib hump cushion, shown in Figure 4.5, is composed of two square tubes that slide over the base posts and can be raised and lowered 15 cm. A cross bar, with the same dimensions as a standard table rail, then connects the two square tubes. To this cross bar a standard OR direct rail clamp and single circular cushion are attached (The OR Group inc. Acton, MA). The circular cushion has 10 cm of travel in the z direction when fixed to the clamp. There is also 18 cm of travel in order to push on the rib hump. The surgeon can apply the corrective force to the rib hump but must then use one hand or have an assistant lock the yellow leaver of the cushion into place.

This design could not be used for the lateral correction cushion since there was the 18 cm long component that would extend beyond the edge of the table and be in the way of the surgeon. Some telescopic self locking mechanism had to be devised. The final design is show in Figure 4.6. It is composed of a base post which permits 15 cm of height adjustment. Onto this a slider is placed which allows for 4 cm of longitudinal adjustment. The telescopic wedging mechanism is then attached which allows for 13 cm of lateral length adjustment. On the end of this telescope is a ball joint and a 10 cm round cushion. The details of the telescopic wedging mechanism are shown in Figure 4.7.

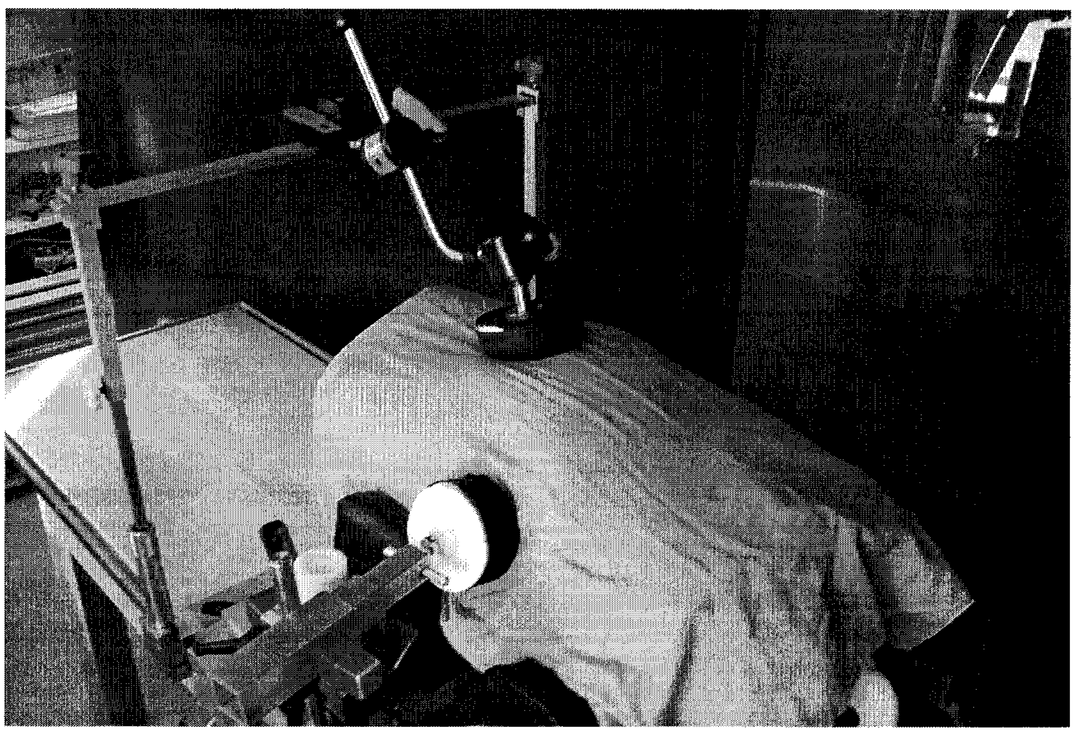


Figure 4.5: Rib hump cushion pushing on a mannequin's torso.

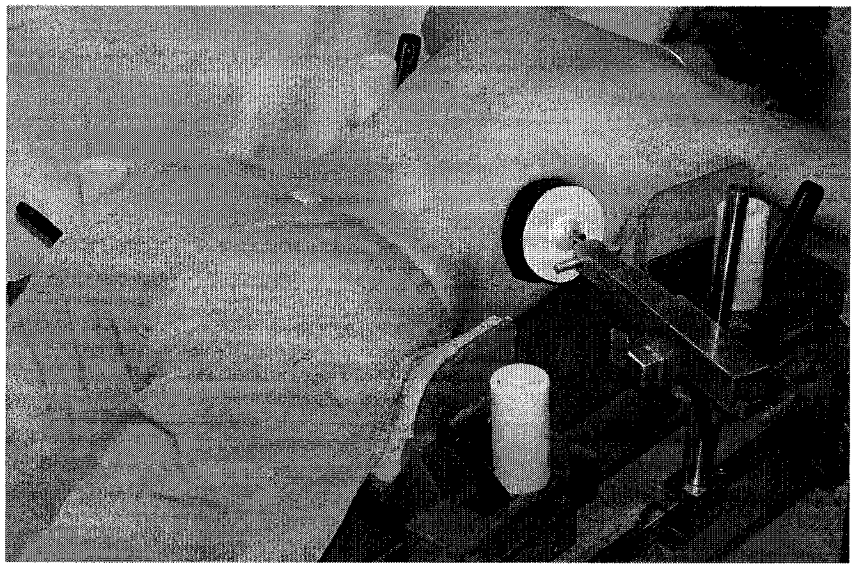


Figure 4.6 lateral correction cushion pushing on a subject.

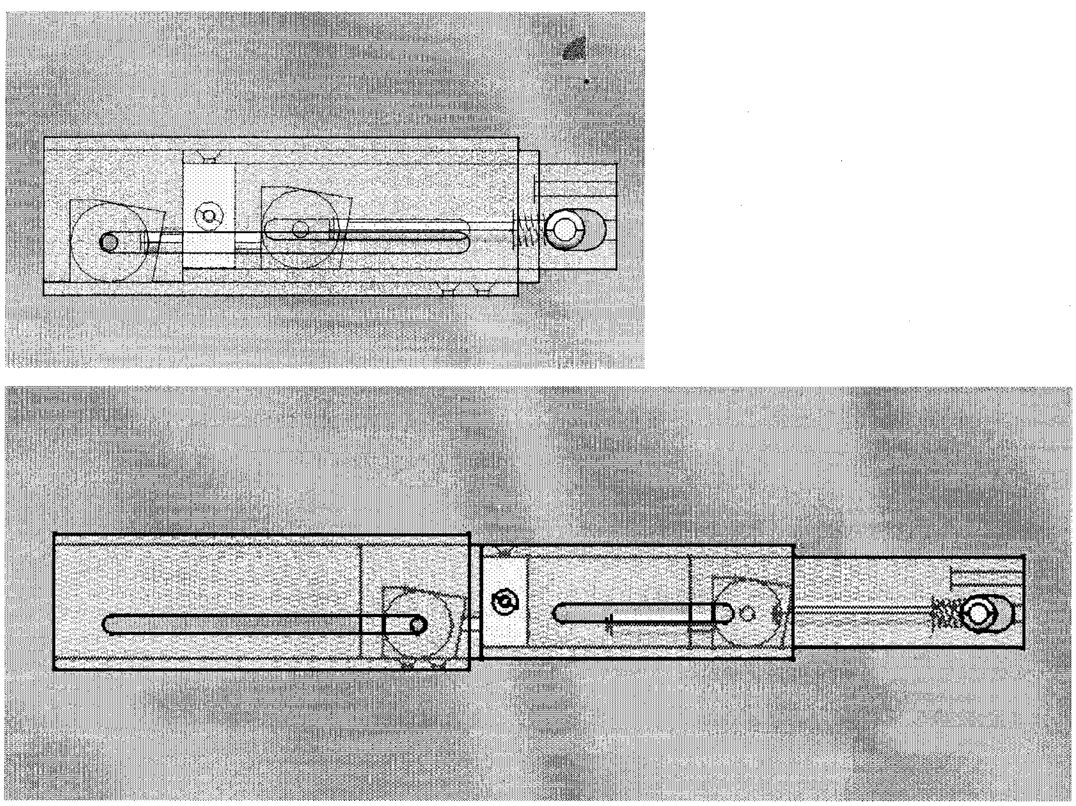


Figure 4.7: Telescopic, self locking, slider shown in full contraction and full extension.

4.6 Summary of the innovations of the dynamic positioning frame

A first prototype of a positioning frame that could be used in a clinical setting, but not in the operating room, was constructed by Pazos (2000). Analyzing the advantages and disadvantages of this design was crucial in developing the DPF presented in this study. The design of the DPF was done with equal contributions from Annick Koller, an industrial designer, and the author. The first major change was in the selection of the materials since the system had to be sufficiently radiotransparent to provide exposure of the spine during an AP and a lateral x-ray. The clinical prototype was attached with Velcro to a wooden base and this was replaced with rails providing easy translation in the y direction. The wooden base was replaced with the aluminium cross bars and a locking mechanism to secure it to the table. On top of this, the Lexan sheet protects the x-ray cassette from a pendulous abdomen. The heavy steel x translation sliders were replaced with compact Delrin sliders and dimensions of the system were minimized so that there was less overhang from the side edges of the table while still accommodating large patients. The overall height of the system was also reduced to facilitate patient transfer but this is still a problem. A low-profile cushion support provided an aesthetic and ergonomic solution to house the Manfrotto handle and hold the cushion, while allowing for x-ray clearance of the spine from both the AP and lateral views. The general side profile of the base cushions is unchanged yet the caudal side of the hip cushions was slightly sloped to accommodate the flexed legs. In addition the shape, as seen from above, was modified and enlarged to account for the high pressures and to better contour the patient's chest and rib cage. The correction cushions are also a new contribution and the self locking mechanism was designed and fabricated with the help of Demetri Giannitsios, a biomechanical engineer.

4.7 Validation of system in the operating room (Case study with 3 patients)

After adjustments were made to the existing base cushions of the DPF tests, similar to those described in section 4.4, were preformed on three subjects lying on the Relton-Hall frame, the DPF and the DPF with the cushion add-ons. The pressures measured, while the subjects were lying on the DPF with the add-ons, was less then or equal to the pressures on the Relton-Hall frame. Confirming this, the DPF was ready for use in the operating room. The objective of this section of the project is to test the functionality and safety of a new positioning system for posterior spinal surgery.

4.7.1 Validation of system in the operating room: materials and methods

Three scoliotic patients, which were scheduled for posterior surgery, were selected for this study. On the day prior to surgery the patients were placed on the Relton-Hall frame and the base cushions of the DPF to compare the pressures at the patient cushion interface. The pressures were again measured with a FSA containing 512 sensors and a maximum pressure reading of 300 mmHg (Vista Medical Ltd, Winnipeg, Canada). The patients were asked to relax and breath normally for approximately 1 minute and during that time a 30 second acquisition was recorded. The pressures over this time were averaged to negate the effects of breathing, which cause the pressures to fluctuate. In the operating room, the patient was positioned on the DPF and pressure data was collected during the entire length of the surgery with the exception of when the electro cautorizor was being used or when electric potential were recorded as these interfere with the FSA. The corrections cushions were not used for these first three patients. The introduction of new material into the operating room (OR) environment must be done in a conscientious manner. For this reason, only the base cushions of the DPF and the FSA were utilised at this point until the surgeon and OR staff are familiar with the new equipment.

4.7.2 Validation of system in the operating room: results

The clinical data for the three patients is summarised in Table 4.5 and the x-rays are shown in Figure 4.9. The first patient was a 14.4 year old female with a 64° thoracic curve (T4-T11) and a 42° lumbar curve (T11-L4). The second patient was a 13.8 year old female with a 62° thoracic curve (T6-T12) and a 50° structural lumbar curve (T12-L4). The third patient was a 14.2 year old female who had a 30° thoracic curve (T5-T10) and 42° thoracolumbar curve (T10-L3). When positioned on the operating table the curves corrected to 40°, 27°; 36°, 32° and na, 22° for the thoracic and lumbar curves for each patient respectively. The intra-operative thoracic curve is not available for the third patient since the x-ray cassette was placed quite low in order to capture the lumbar curve. Standing post-op x-rays showed 17°, 21°; 16°, 8° and 13°, 3° for the thoracic and lumbar curves for each patient respectively.

Table 4.5: Summary of the clinical data for the three patients.

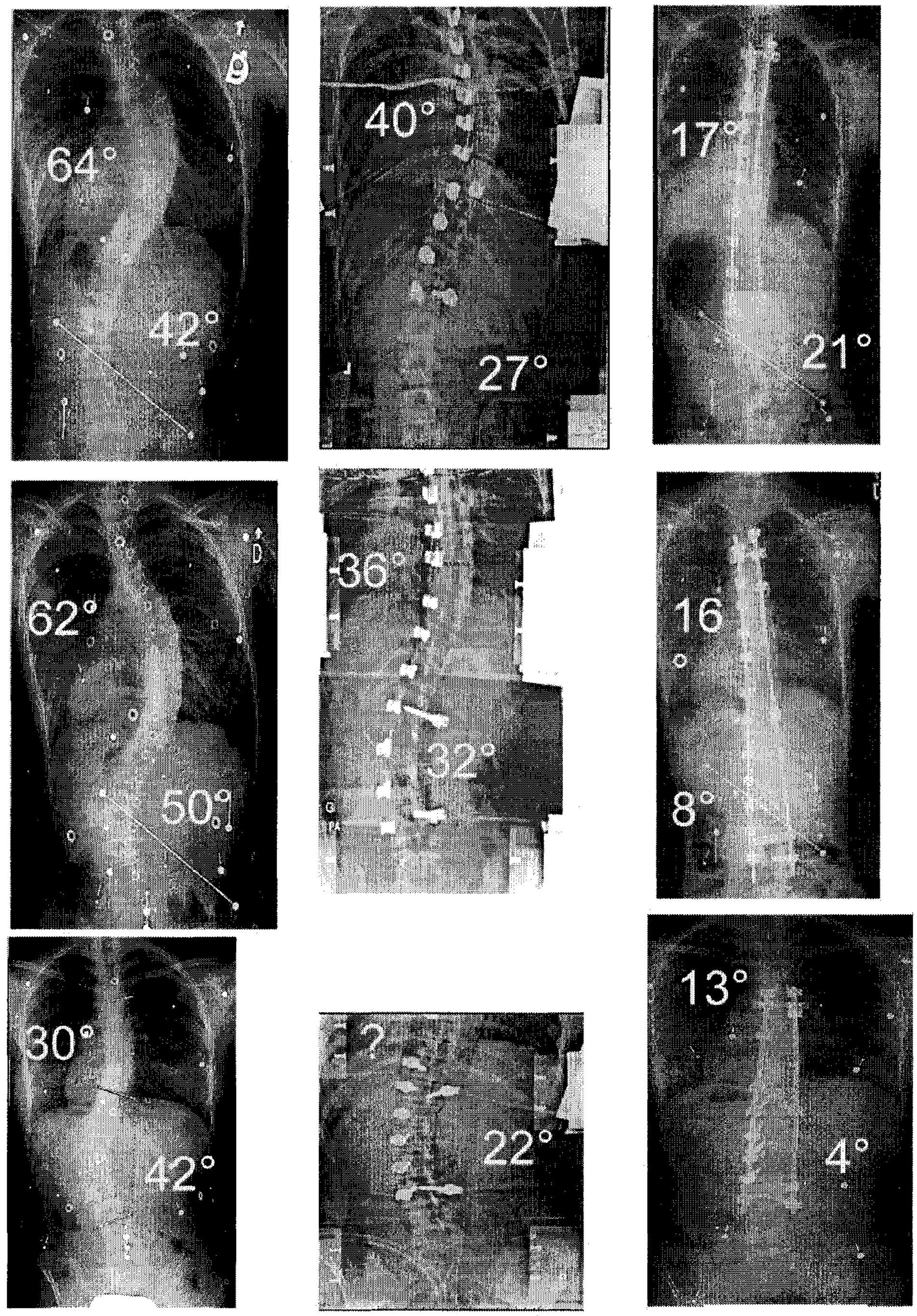
		Patient 1		Patient 2		Patient 3		Average	
		Main thoracic Cobb	Lumbar Cobb	Main thoracic Cobb	Lumbar Cobb	Main thoracic Cobb	Lumbar Cobb	Main thoracic Cobb	Lumbar Cobb
Standing Pre-op	Cobb angle (°)	64	42	62	50	30	42	52	45
Prone Intra op	Cobb angle (°)	40	27	36	32	na	22	38	27
	% correction	38	36	42	36	na	48	40	40
Standing Post-op	Cobb angle (°)	17	21	16	8	13	4	15	11
	% correction	73	50	74	84	57	90	68	75

An analysis of the pressure measurements in Table 4.6 shows that both systems are similar but there are higher pressures on the DPF during the operating room. In order to standardise the acquisition taken in the operating room, 30 seconds was selected when the x-ray was taken. For the first patient the average (and maximum) pressures were 21 (63), 15(35) and 26 (300) mmHg on the R-H, new pre-op and new per-op respectively. The pressures were 21 (73), 14 (47) and 24 (300) mmHg for the second patient. Finally, for the third patient the pressures were 17 (151), 17 (43) and 44 (222) mmHg.

Patient 1

Patient 2

Patient 3



Pre-op

Intra-op

Post-op

Figure 4.8: Pre-op, intra-op and post-op x-rays for the three patients

Table 4.6: Summary of the pressure measurements for the three patients (mmHg).

		Patient 1	Patient 2	Patient 3
R-H Pre-op	average	21	21	17
	maximum	63	73	151
DPF Pre-op	average	15	14	17
	maximum	35	47	43
DPF Intra-op	average	26	24	44
	maximum	300	300	222

Visual representations of the pressures are shown in Figure 4.9. The first two patients had slight reddening on the chest and pelvis and severe reddening of their right iliac crest. This reddening corresponded to the maximum pressures observed on the FSA. The third patient had slight reddening on her chest and pelvis but no one area that stood out. For all three cases there were no major complications from the positioning or any other complications.

4.7.3 Validation of system in the operating room: Discussion and Conclusion

The results of the first three trials were promising and showed the feasibility and security of this new positioning system. As these were the first few trials the external correction cushions were not utilised. Even more correction is expected once these cushions are used. Already, the preliminary Cobb angle results show a trend towards improved correction. A larger clinical trial is underway to determine if there is a significant improvement in the amount of correction achieved with this system.

The pressures on the DPF were less then or equal to the Relton-Hall frame the day before surgery. There were, however, higher pressures during the surgery once the patient was anesthetised. This is likely due to that anaesthesia and the fact that the patient's muscles are lax during the surgery. Because the patient is asleep and positioned by someone else, they are unable to adjust their position to move away from a potential area of higher pressure. It is recommended that the pressure be monitored with the FSA and that the surgeon uses it as a tool to better reposition the patients.

There were some other general comments and critiques that came out of the use of the system. For the first case, the patient was positioned slightly high on the system such that her first thoracic vertebra was directly above the cross bar and not visible on the x-ray. Between the crossbars of the DPF there are 57 cm in which to place the 36 cm (14 in) x-ray. Care must be taken when positioning the patient and ensure that the region that needs to be x-rayed falls within the 57 cm space between the crossbars. There were also some critiques when positioning the patient. Because of the shape of the cushions, that better conform to the sides of the patient, the patient must be lifted over the sides of the cushions to be placed on the frame. The white nylon posts that support the cushions and allow for rotation and translation were also found to be cumbersome and had to be covered with a gel pad during patient transfers. Positive comments include Dr. Poitras who found that the DPF seamed to hold the patients in a more stable position. Dr. Labelle, who is generally pleased with the DPF, has agreed to use it on all subsequent adolescent idiopathic cases that are willing to participate in the ongoing clinical study.

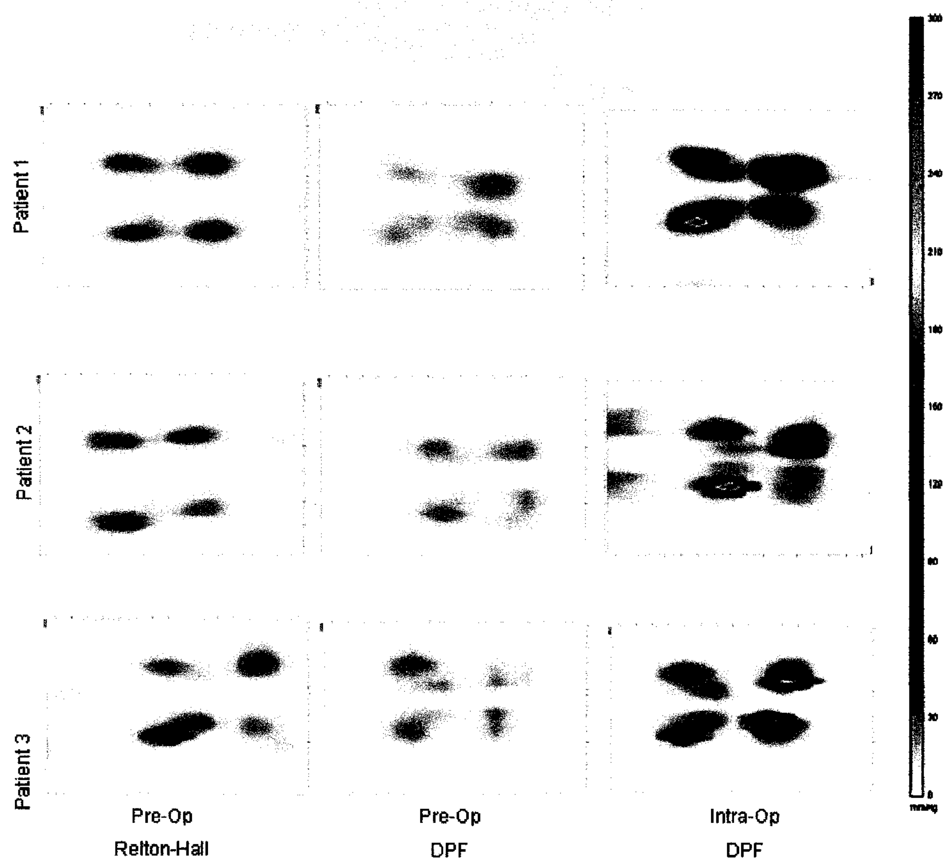


Figure 4.9 : Pressure measurement output for the three patients. The mannequin shows how the patients were oriented on the DPF.

CHAPTER FIVE: OPTIMAL ADJUSTMENT OF DYNAMIC POSITIONING FRAME FOR SPINE SURGERY

5.1 Situation of the third article

In this chapter, the realisation of the fourth objective of this project is met with the help of a third article. As a review, the fourth objective was to modify the model created in objective 1 (Chapter 3) to simulate the Dynamic Positioning Frame, test various adjustment parameters and recommend placement of the positioning system components prior to, and during surgery based on the results of the computer simulations.

In both clinical studies, presented in Chapter 4, the cushion placement was selected based on the clinicians experience and upon visual inspection of the patient's trunk. Because it is difficult to directly correlate what is happening to the patient's trunk to what is happening inside to their rib cage and spine, it is useful to look at models of the internal skeletal geometry.

By combining the finite element model presented in Chapter 3 with the dynamic positioning frame fabricated and clinically tested in Chapter 4 we can better understand the biomechanics of patient positioning. In order to simulate the dynamic positioning frame the boundary conditions and various external forces were added to the model as described in detail in the following section.

The simulations that were done on a scoliotic patient help to show the sensitivity of the model to cushion placement. By structuring the simulations in a Box-Hunter experimental design a detailed statistical analysis was possible. From this analysis, optimization of different geometric measures was performed in order to recommend the positioning of the system components prior to and during the surgery. Greater details as to the steps required to simulate and optimise the cushion placement are presented in Appendix 1.

This study is presented with the following article titled "Optimal adjustment of a dynamic positioning frame for spine surgery" for which the contribution of the first author is considered to be 90%. The authors are the following:

Kajsa Duke, Carl-Eric Aubin, Jean Dansereau, Hubert Labelle

5.2 Article 3 Optimal adjustment of a dynamic positioning frame for spine surgery.

Optimal adjustment of a dynamic positioning frame for spine surgery.

Kajsa Duke MSc. [a,b,]

Carl-Eric Aubin PhD. [a,b*]

Jean Dansereau PhD. [a,b]

Hubert Labelle MD [a]

[a] Research Centre, Sainte-Justine Hospital, Montreal, Canada

[b] Department of Mechanical Engineering , École Polytechnique de Montreal, Canada

Submitted to Spine.

[*]Corresponding author.

Address: Canada Research Chair "CAD Innovations in Orthopedic Engineering"

Department of Mechanical Engineering,

École Polytechnique, P.O. Box 6079, Station "Centre-ville"

Montréal Québec Canada

H3C 3A7

Fax: 01-514-340-5867.

E-mail: carl-eric.aubin@polymtl.ca

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Abstract

Study Design. A finite element model of the scoliotic spine simulated a new dynamic positioning frame (DPF) and recommend optimal cushion placement.

Objectives. To simulate a DPF and test various adjustment parameters and recommend placement of the DPF components prior to, and during spine surgery based on the results of finite element simulations.

Summary of Background Data. Studies have shown scoliosis curves correct when patients are positioned on the operating table prior to instrumentation. However, biomechanical aspects of positioning have not been widely studied.

Methods. One scoliotic patient was simulated using a finite element model and six DPF model parameters were modified while ten geometric measures were recorded. Statistical analysis determined which model parameter had a significant effect on the geometric measures. Geometric measures were individually and simultaneously optimised, while corresponding model parameters were noted.

Results. Every model parameter had a significant effect on at least five of the geometric measures. When optimising a single measure others often deteriorated. Simultaneous optimisation resulted in improved overall correction of the patient's geometry by 75% however ideal correction was not possible for every measure.

Conclusions.

Finite element model simulations of a new DPF enabled optimisation of ten geometric measures. Positioning of scoliotic patients is an important step in the surgical procedure that should be exploited to achieve maximum correction.

Key Words: Scoliosis, prone position, surgery, finite element modeling, biomechanics, experimental design

Key Points

- Biomechanical simulations and optimisation of a new dynamic positioning frame was performed.
- Many model parameters had significant effect on the clinical geometric measurements.
- Caution should be used as focusing on optimisation of a single clinical geometric measure can cause deterioration of other measures.
- The overall geometry improved by 55% when positioned on the Relton-Hall frame and by 75% when positioned on the dynamic positioning frame.
- Box-Hunter experimental designs provide an efficient way of optimising finite element models.

Mini Abstract

One scoliosis patient was simulated on a new dynamic positioning frame using a finite element model. Six parameters were modified while ten geometric measures were recorded. Simultaneous optimisation resulted in improved correction of the patient's geometry (75%). Positioning is an important surgical step that should be exploited to achieve maximum correction.

Introduction

Patient positioning is an important step in spine surgery. The majority of positing frames are based on the Relton-Hall four post principal[23]. The main focus of such frames is on keeping the abdomen pendulous to reduce blood loss[24]. It has been recognised that positioning of the patients legs can have an affect on lumbar lordosis[18,21,28]. More recently noted, the scoliotic spinal deformity can decrease due to positioning prior to surgical instrumentation[6].

To capitalise on the reduction in deformation prior to instrumentation, a dynamic positioning frame (DPF) has been designed (Figure 1) and the details are presented else ware (US patent Docket No. 15490-7US SC/ip, international patents pending). In summary, base cushions can be adjusted at the start of surgery to adapt to patients of various sizes and also modify kyphosis and lordosis. Positioning of the legs can be adjusted based on supports placed under thighs. As well, there are corrective cushions that can be applied directly to the patient's trunk at the prominence of the deformity. This system has been tested on a group of unanestethesied scoliotic patients[10].

Because it is difficult to test different treatments on the same patient, computer models have been created. These computer models simulated various scoliosis treatments including bracing[20], instrumentation[2,15] and thorocoplasty[5,12]. Optimisation techniques were applied to determine patient personalised material properties [9,15,22] and brace treatments[11].

The objective of this study is to simulate, with a finite element model of the trunk, the dynamic positioning frame and test various adjustment parameters and recommend placement of the positioning frame components prior to, and during spine surgery based on the results of the computer simulations.

Materials and Methods

A flow chart is provided to summarise the materials and methods used in order to optimise patient positioning on a dynamic positioning frame (DPF) and compare that to the position on a standard Relton-Hall (R-H) type frame (Figure 2).

Patient

The patient, used in this study, was classified King II [14] or Lenke 2C+ [17]. This patient is representative of one of the more common types of adolescent idiopathic scoliotic curves[16]. Various geometric measures were taken from the 3-D reconstruction of the patient's standing calibrated x-rays; lateral, anterior posterior (AP) and tilted AP at 20° [7] (Table 1).

Model

A personalised finite element model was created using ANSYS 8.0 from the 3-D geometry of the above scoliotic patient [1,7]. In summary, this model contains 2974 elements and 1440 nodes representing the osseo-ligamentous structures of the pelvis, spine, ribcage, and sternum. Material properties were obtained from literature and experimental trials from cadaver specimens [8]. These material properties were personalized to the bending x-rays and the standard prone position was simulated on a Relton-Hall type frame, as described in another study [9].

The model was defined in the prone position using the application of gravity as a distributed load and specific boundary conditions [9]. To simulate the DPF six model parameters were tested at different extremities (Figure 3). Pelvic inclination (-15 and 15°), chest cushion location (under ribs 3 to 6 or under ribs 6 to 9) and chest cushion height (0 and 3.5 cm) are boundary conditions which represent adjustments to the base cushions of the DPF. Though the legs were not simulated in the model, inclination of the pelvis represented flexion and extension of the legs. Rib hump force (posterior node of 9th rib on the right hand side), lateral thoracic force (lateral node of same rib) and lateral lumbar force

(L3 on the left hand side), involve adjustment of external correction cushions which all varied between 10 N to 150 N. The upper limit of 150 N was selected based on a quick test in the lab where a clinician was asked to push on the trunk of a subject as done in the operating room while a the applied pressure was measured using a force sensing array (Vista Medical, Winnepeg, Canada).

Geometric Measures

Ten geometric measures were taken from the 3-D geometry of the model using the standing geometry as a base. The reference plane used in the model was defined by Stokes [26] as the global (body) coordinate system and is shown on the model in Figure 3. Whenever possible, clinical measures outlined in the Spinal Deformity Study Group Radiographis Measurements Manuel were used to assess the deformity [19].

Details of the clinically relevant geometric measures are now given. Decompensation was the distance between T1 to L5 projected onto the AP plane, positive to the left and negative to the right. Balance was the distance between T1 to L5 projected onto the sagittal plane and was positive to the front and negative to the back. Note that L5 was used instead of S1 because the geometric measures were based solely on the spine geometry because the in house software used to calculate these parameters does not include the pelvis. Main thoracic Cobb was the difference of the tilt of the vertebra projected onto the AP plane between T6 and T12. Lumbar Cobb was measured between T12 and L4. These end vertebral levels were determined at the inflection points of the curve[13], corresponded to the most inclined vertebra, were determined from the standing geometry and were maintained for all subsequent AP Cobb angle measurements. Apical vertebra translation (AVT) was measured as the projection onto the AP plane of the most laterally deviated vertebra (T9 thoracic, L3 lumbar). Apical vertebral rotation (AVR) was calculated at T9 based on the Stokes method[27]. Kyphosis was calculated between T2 and T12 and lordosis between L1 and L5 as the difference in tilt of the vertebra projected onto the sagittal plane. Finally, rib hump was the average angle of the double tangent lines across ribs 8, 9 and 10.

Experimental Design

Model parameters were manipulated in a controlled manner as part of a Box-Hunter experimental design with six manipulated variables, at two extreme modalities for a total of 32 runs. Results for the 32 simulations were entered into Statisica (StatSoft, USA) where an analysis of significant parameters was performed by looking at Paretto graphs and linear regressions coefficients. Ten equations were obtained representing the ten geometric measurements as a function of the six model parameters. These equations were entered into a cost function (Equation 1) and optimised in Matlab (MathWorks, USA).

Equation 1

$$\text{Cost Function} = w_1(a_s - a_m)^2 + w_2(b_s - b_m)^2 + w_3(c_s - c_m)^2 + w_4(d_s - d_m)^2 + w_5(e_s - e_m)^2 + w_6(f_s - f_m)^2 + w_7(g_s - g_m)^2 + w_8(h_s - h_m)^2 + w_9(i_s - i_m)^2 + w_{10}(j_s - j_m)^2$$

Where:

w_i = weighting	e=Thoracic AVT
$_s$ = simulated geometric measure	f=Lumbar AVT
$_m$ = desired geometric measure	g=Thoracic AVR
a = Decompensation	h = Kyphosis
b= Balance	i = Lordosis
c= main thoracic Cobb	j = Rib Hump
d=Lumbar Cobb	

Each geometric measurement was individually optimised by attributing a normalised weighting of 0.9991 while the other nine geometric measurements were given a weighting of 0.0001, for a total of 1.0. Optimisation was also performed with every geometric measure equally weighted (0.1). The sum of this equally weighted cost function was used to quantify the overall correction.

What are the desired geometric measurements attainable by properly positioning the patient? This is a difficult question and no two surgeons would likely give the same answer. Realistically, if a positioning frame could correct 50%, it would be an improvement over existing frames which are able to correct thoracic and lumbar Cobb angles between 33% to 37% [4,6]. An acceptable range of correction was divided into 3 categories; ideal correction (> 50%), realistic correction (20-50%) and unsatisfactory correction (< 20%). Ideal kyphosis was selected at 25° with a range of 10° to 40° and the ideal lordosis is 50° ranging from 40° to 60° [19]. Decompensation and sagittal balance are important [3] so the ideal was set to zero. Considerable variability is observed in balance, even in normal adolescents, so the desired range has a minimum of ±10 mm and a maximum of ±35 mm[29].

The equations, built from the regression coefficients, are in effect a simplified mathematical model. The optimised model parameters were entered into the equations giving estimated geometric measures and re-tested in ANSYS giving optimal simulated geometric measures. Comparing these measures the robustness of the optimisation is tested using the initial simulations (n=32) as a training set and the optimal simulations (n=11) as a test set.

Results

Though the Cobb angle results of the initial prone simulation are presented elsewhere[9], a more detailed 3-D analysis of ten geometric measures is presented here (Table 1). The initial prone simulation shows that decompensation moves further to the right (-5mm to -10mm). Balance moves forward, improving from -43mm to -23mm. Main thoracic Cobb and lumbar Cobb correct from 62° to 47° and 42° to 34° respectively. Thoracic apical vertebral translation improves from 46 mm to 32 mm while lumbar apical vertebra translation worsens slightly (8.5 mm to 9.2mm). Thoracic apical vertebral rotation remains unchanged at -40°. Kyphosis and lordosis both decrease from 45° to 38° and -37° to -32° respectively. The rib hump improves from -5.0° to -4.4°.

Looking at the minima and maxima obtained for the 32 simulations one can see the variability and determine if the geometric measures fall within the acceptable range (Table 1). The simulated range for decompensation, balance, thoracic apical vertebral translation, apical vertebral rotation and rib hump all contain the desired ideal. Main thoracic Cobb, lumbar Cobb, lumbar apical vertebral translation, kyphosis and lordosis overlap with the realistic range but ideal correction (> 50%) was never obtained in the tested area.
Statistical analysis determined which model parameter has a significant effect on the geometric measures. Table 2 shows only those parameters that have a significant ($p < 0.05$) effect on the geometric measures and lists their importance as represented from the Paretto charts. Looking at kyphosis, for example, chest cushion location, pelvic inclination, chest cushion height and rib hump force all have a significant effect and act in that order of importance. One can also see how often a model parameter is significant. For example, rib hump force and lateral thoracic force were significant for six and eight of the geometric measures respectively. Each of the model parameters had a significant effect on at least 5 of the geometric measures meaning that all of the model parameters need to be considered when positioning the patient.

From the regression coefficients different equations to represent the geometric measures with respect to the model parameters were determined. There were 22 regression coefficients for each of the ten equations representing the geometric measures for a total of 220. As an example, Equation 2 is given to show the main thoracic Cobb as a function of the model parameters. The equation was simplified to show only the 4 most significant model parameters and no interactions. In order to minimise the main thoracic Cobb, pelvic inclination, rib hump force and lateral lumbar force should be minimised while lateral thoracic force should be maximised.

Equation 2

$$MTC = 48.218 + 0.068\,(PI) + 0.024\,(RF) - 0.072\,(TF) + 0.008\,(LF)$$

Where: MTC = Main Thoracic Cobb
PI = pelvic inclination
RF = rib hump force
TF= lateral thoracic force
LF = lateral lumbar force

The geometric measures were individually optimised using the cost function and weightings described in the materials and methods section and the resulting model parameters are shown in the left portion of Table 3. Using the main thoracic Cobb, as an example, the pelvic inclination should be -15°, the chest cushions placed under ribs 6 to 9 and raised 3.5 cm, rib hump force, lateral thoracic force and lateral lumbar force should be 10N, 150N and 10N respectfully. There is a large range for each model parameter varying between its minimum and its maximum tested range depending on what geometric measure is optimised. For example, the main thoracic force it is optimised at 150N for the main thoracic Cobb and only 10N for the lumbar apical vertebral translation.

In addition, it is important to note how the other 9 geometric measures change when a single measure is optimised as the other measures may deteriorate (right hand portion of Table 3). When the main thoracic Cobb was optimised, a correction to 35° was attained, however, other geometric measures worsened such as the decompensation which increased to -11mm, apical vertebral rotation worsened by 1° and lordosis decreased to -25°.

Using the equally weighted (0.1) optimisation it was impossible to achieve an ideal correction for every parameter as shown in the last line of Table 3. The only parameter that attained ideal correction was thoracic apical vertebral translation (-20 mm). Balance (-3 mm), main thoracic Cobb (43°), lumbar Cobb (-33°), apical vertebral rotation (-24°) and kyphosis (40°) were all with the desired ideal range. Unfortunately, the decompensation (-13mm), lumbar AVT (12mm) and lordosis (-32°) all worsened compared to their initial standing values. As an analysis of overall correction, the cost function was 479 for the initial standing position, 216 for the original prone position and 122 for the simultaneous optimised DPF. To obtain this correction it is recommended that the pelvis is inclined 12°, the chest cushions should be placed under ribs 4 to 7 and should not be raised. For the external corrective forces the rib hump force, lateral thoracic force and lateral lumbar force should be 68N, 97N and 10N respectively. It is recommended that these forces are applied after the hooks and screws are in place but before rod insertion. They should be removed once the rods are secured to the hooks and screws. It should also be noted that for the chest cushion location the optimal value sometimes fell between one of the ribs. Since the ribs are represented by integer numbers, rib 3.6 does not exist, so this was rounded and ribs 4 through to 7 were constrained.

The calculated absolute difference of the estimated geometric measures compared to the simulated geometric measures is shown in Table 4. The largest difference for the training set (n=32) was found in the balance at 5.4 mm ± 3.8 mm when looking at displacement and the rib hump 1.9° ± 1.6° when looking at angles. For the test set (n=11) the largest difference was again found in the balance at 7.8 mm ± 4.8 mm. For the angles the largest difference was the apical vertebral rotation at 2.8° ± 1.3°.

Discussion

All the corrections simulated in this study are due only to the positioning of the patient and not the instrumentation. Ideally, if the patient were placed in an optimal position, it is hypothesized that the act of inserting the instrumentation would be facilitated and final correction after instrumentation could be improved, compared to the standard positioning frames. Optimal positioning of the patient could yield better post-operative correction and/or make the surgery easier by reducing the necessary applied forces and potentially reducing operative time and blood loss.

In a clinical study measuring the trunk geometry of various patients lying on the DPF various parameters were tested including chest cushion height, lateral lumbar force and a combined thoracic/rib hump force [10]. These simulations showed that decompensation was sometimes over corrected or negatively affected as was also the case in the clinical study. In both studies raising the chest cushions increase the kyphosis. However, in these simulations the patient was slightly hyper kyphotic at 45°, so in order to decrease the kyphosis towards the desired 25°, the chest cushions were not raised but left at 0 cm.

The effect of leg position on lordosis has been extensively studied [18,21,28]. These simulations support clinical studies that show that the lordosis increases when the legs are extended (pelvis inclines towards +15°) and decreases when the legs are flexed (pelvis inclines towards -15°). The original standing lumbar lordosis was only -37° and further worsened to -32° on prone positioning. The simulations showed that lordosis could be regained to -42° with the pelvic inclination at +15°. However, since there is already loss in lordosis due to prone positioning, it would be difficult to expect to correct a very hypolordotic curve and maintaining the standing lordosis or increasing it by a few degrees would be a realistic correction.

This is the first study that has simultaneously optimising so many different geometric measurements. The most similar study is that of Gignac et al. whom optimised five geometric measures to determine the best type of brace treatment in adolescent idiopathic scoliosis[11]. It is interesting to see the effect (sometimes negative) that optimising a single geometric measures has on the other nine measures. This makes the task of optimising all ten simultaneously difficult. When looking at the overall correction and optimisation it is important to evaluate the cost function. The patient positioned on the DPF had a cost function of 122 compared to 216 on the standard and 479 when standing. In other words, there was an overall correction of 55% when the patient was simulated on the Relton-Hall frame and 75% correction when the patient was positioned on the DPF.

Equations were obtained for each geometric measure and a simplified mathematical model was created. For this simplified model to be a useful tool, the equations must accurately represent the finite element analysis. The greatest errors were for balance and apical vertebral rotation. However, even with the test set the difference in Cobb angles, kyphosis and lordosis were all under 2° which is below intra-observer variability which is around 3° [25]. Though the results are not presented here, the option exists to ignore the interactions when performing statistical analysis (Statisica, StatSoft, USA). Less accurate results were obtained since some interactions were significant, and should therefore be taken into account. A limitation of the Box-Hunter experimental design is that it tests at extreme limits not taking into account any non linearities in the model. The accuracy of the simplified model could be improved by increasing the experimental design to three modalities (Box-Benkhen) to account for non linearities. However, to obtain the same information regarding the interaction would require 243 simulations and not 32. The average time for the simulations was about 22 minutes using a Pentium III 557 MHz processor with 512 Mega bytes of RAM.

The equations were originally used to optimise cushion placement for desired geometric measures. However, these equations can be used in the reverse sense. This that the simplified model of equations, represents the patient personalised finite element analysis, and can be used to test various positioning configurations. Using this experimental design method to find the optimal configuration provides an advantage of reduced computational time over using an optimisation algorithm which calls the finite element software at each iteration.

Even without a dynamic positioning frame, finite element simulations and optimization techniques clinical relevance can be extracted from this study. This biomechanical study showed that the patient's geometry varied depending on the DPF cushion placement. Some of the results were obvious and well known such as flexing the legs, and hence applying anterior rotation to the pelvic, will reduce the lordosis. External corrective forces applied to the apical vertebra and rib hump produced the desired correction but optimal rib hump correction was not observed at the maximum force. To simultaneously optimise the geometric measures, maximum corrective forces are not recommended. There were some surprising results in that a large number of model parameters had a significant effect on the geometric measures (Table 2). Out of the ten geometric measures eight of them were significantly dependent on three or more of the model parameters. Most interesting was to note the effect that the location of the chest cushions has on the patient's kyphosis and balance. It was surprising to find that the kyphosis was heavily influenced by the chest cushion location and less, but still significantly, influenced by the chest cushion height. Balance was also significantly influenced by chest cushion location. Chest cushions are normally placed as high as possible without interfering with the axilla and arms. This practice could be altered depending on the desired balance and kyphosis.

There are always certain limitations to modeling. One limitation of this study is that the muscles are not directly simulated in the model. Indirectly, the muscles are taken into account as the segmental stiffness of the spine was adjusted based on the patients bending x-rays which incorporate the muscles[9]. The fact that the intervertebral discs are modeled as linear beams when in fact the discs are non-linear and visco-elastic is another limitation. In order to determine if this is a valid simplification it is useful to look at the overall change in disc height and see if this is reasonable. In the model the L1 L2 disc shows the greatest elongation with an increase of 6.8%. This elongation appears realistic compared to Kurutz et al. (2005) who measures disc elongation in patients subjected to traction at 10% to 18%. Another limitation that the head and neck of the patient was not simulated and no constraints were imposed at T1 so its displacement may be slightly exaggerated. During surgery the head, which is gently positioned on a cushion, provides a slight constraint to the top of the spine. To rigidly fix T1 would over constrain the model and the response of balance and decompensation would not be observed. As well, the patient's limbs are not modeled. Another limitation is that this model was not directly validated against a particular patient. For ethical reasons it is not permitted to subject scoliotic patients to additional x-rays in order to determine the internal 3-D geometry while subjected to various positioning cushion locations and forces. However, the base model of patients positioned on the standard Relton-Hall type positioning system was validated against standard x-rays taken during surgery, prior to the instrumentation[9]. The three patients that were positioned on the DPF during surgery were unfortunately not simulated because their intra-operative x-rays were not taken with the calibration object and a 3-D spinal reconstruction could not be created. A final limitation that, at present, is preventing this from becoming a useful clinical tool is the length of time and different programs involved in the data analysis. Ideally, the results of the ANSYS simulations could be entered into one software program where the entire statistical analysis and optimisation is performed and displayed on a surgeon's Palm® with a graphical user interface for the surgeon to enter his desired geometric measures or DPF parameters and see the results in "real time".

Conclusion

Modifications to an existing finite element model were done such that six parameters of a new DPF could be simulated and ten geometric measures were optimised. Every model parameter had a significant effect on at least five of the geometric measures. Individual optimisation of the geometric measures was possible but often resulted in a deterioration of other geometric measures. Simultaneous optimisation of all ten geometric measures is difficult but better overall correction was achieved with the DPF (75%) compared to the standard prone position (55%). This study suggests that the positioning of the patient is a very important step in the surgical procedure and that it should be exploited to achieve maximum correction.

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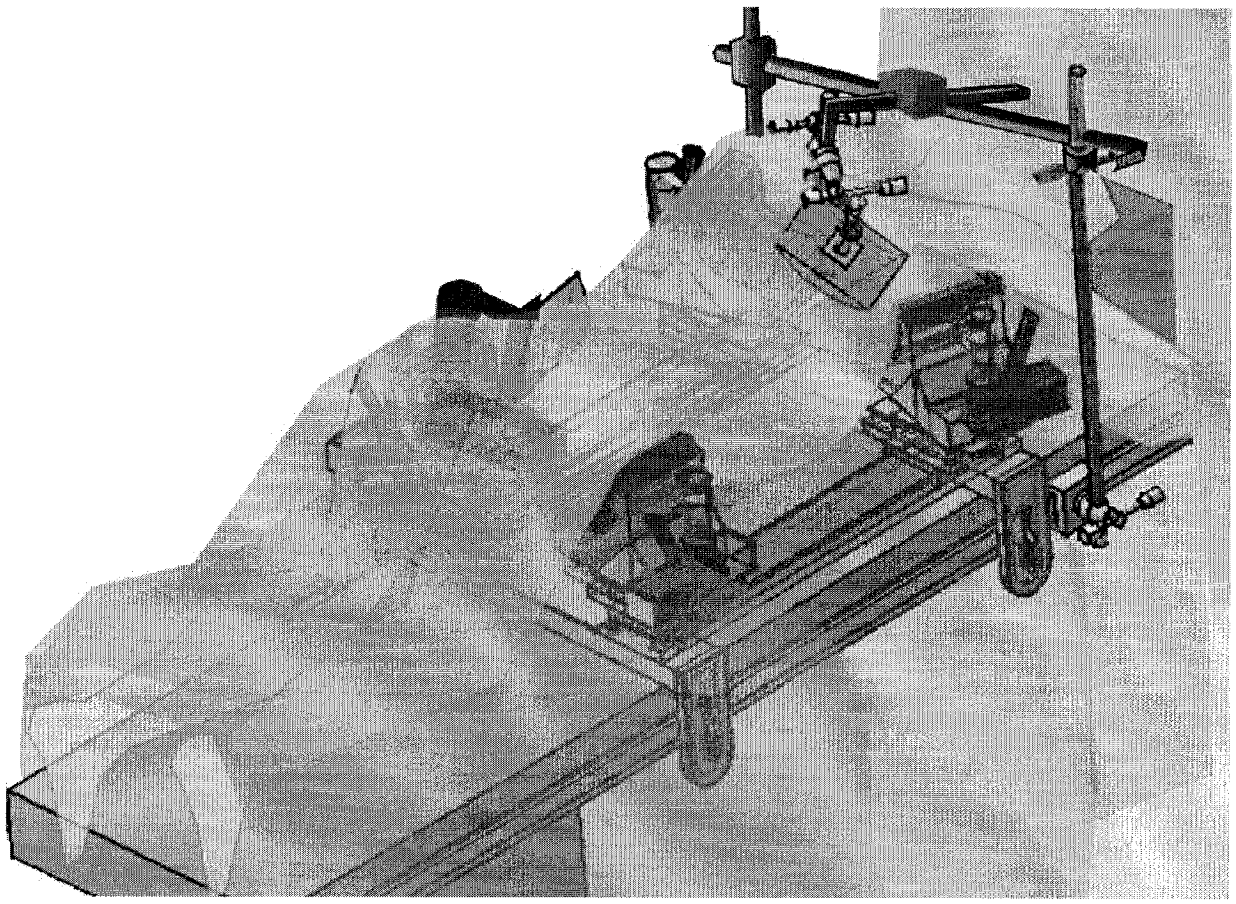

Figure 5.1: Article 3 Figure 1: The Dynamic positioning frame.
Shown with a patient positioned on the base cushions and the corrections cushions pushing on the lumbar and thoracic deformities.

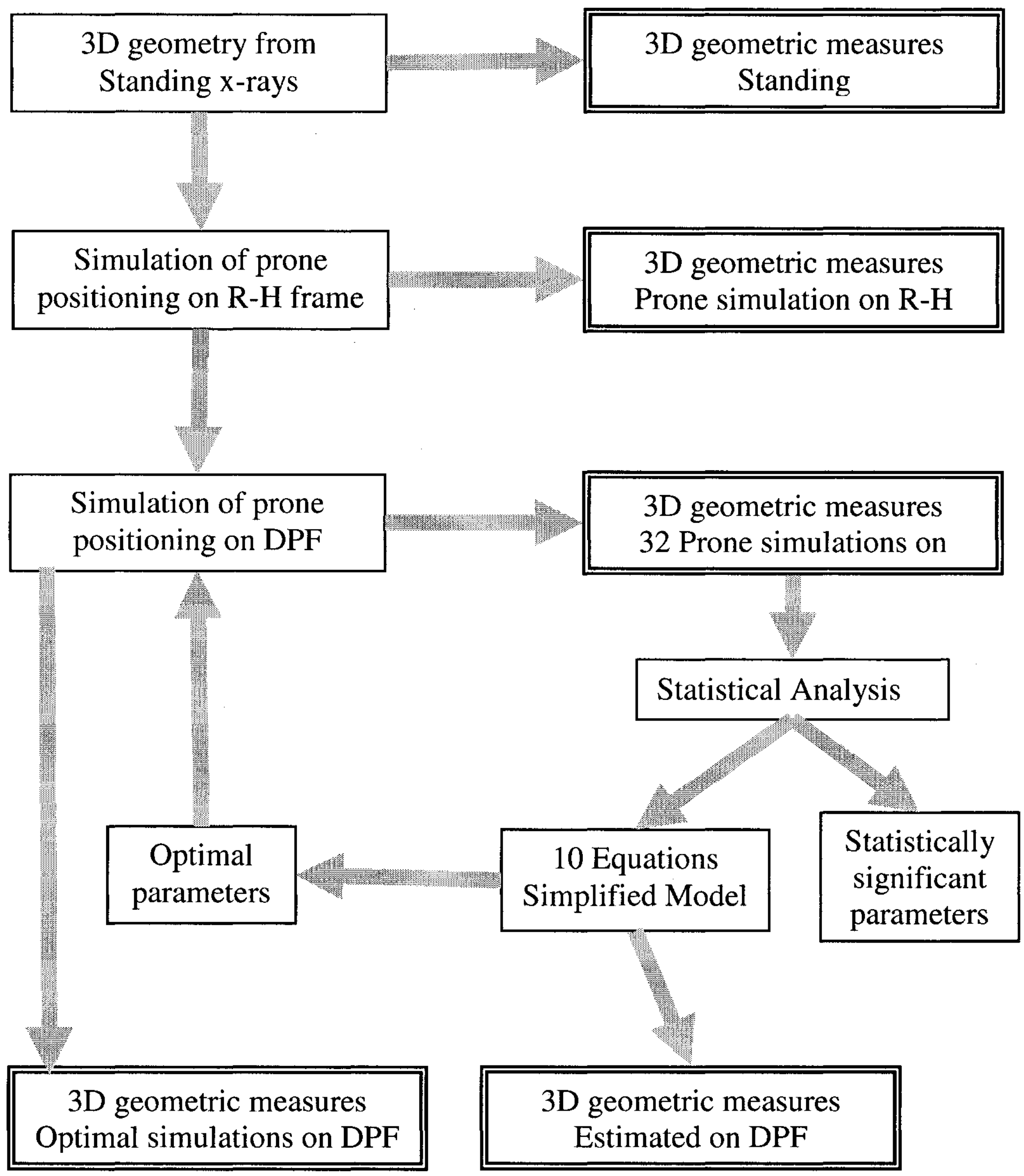

Figure 5.2: Article 3 Figure 2: Flow chart summarising the materials and methods These were used in order to optimise patient positioning on a dynamic positioning frame (DPF) and compare that to the position on a standard Relton-Hall (R-H) type frame. The results of the 3-D geometry are highlighted.

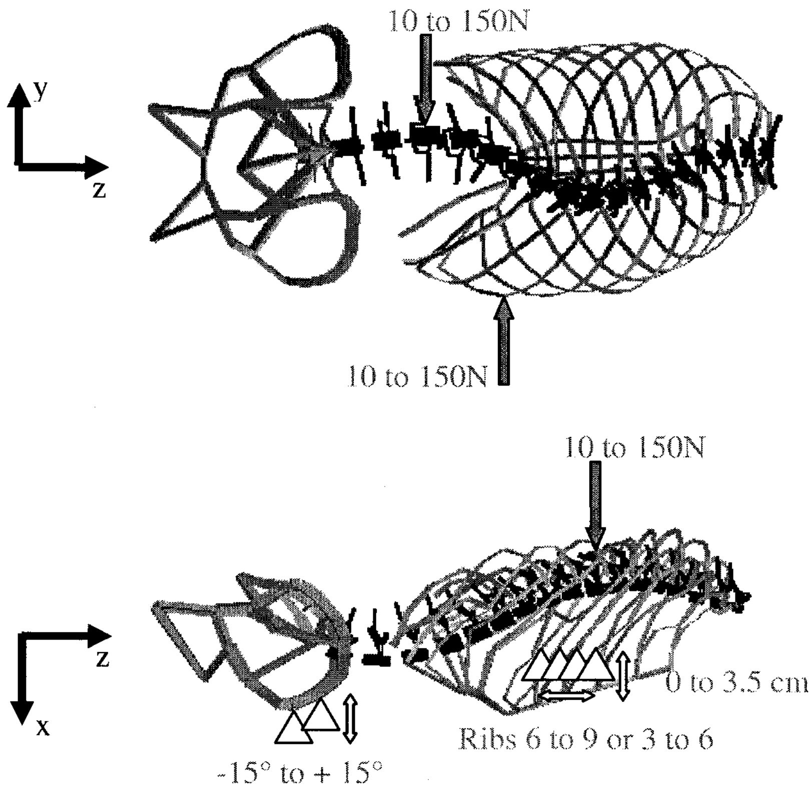

Figure 5.3: Article 3 Figure 3: AP and lateral view of a finite element model of a scoliotic patient.

Various parameters were adjusted including pelvic inclination (-15° to ±15°), chest cushion location (under ribs 6 to 9 or 3 to 6), chest cushion height (0 to 3.5cm), rib hump force, lateral thoracic force, and lumbar force (all 10N to 150N).

Table 5.1 Article 3 Table 1: Geometric measures for the standing and initial prone simulation.

Ideal correction and desired ideal range. Minimum and maximum simulated geometric measures from the 32 simulations.

	Decompensation (mm)	Balance (mm)	Main Thoracic Cobb (°)	Lumbar Cobb (°)	Thoracic AVT (mm)	Lumbar AVT (mm)	AVR (°)	Kyphosis (°)	Lordosis (°)	Rib Hump (°)
Standing	-5.0	-43.0	62	-42	-46.4	8.5	-40	45	-37	-5.0
Prone	-10.2	-23.1	47	-34	-32.3	9.2	-40	38	-32	-4.4
Ideal correction	0.0	0.0	31	-21	-23.2	4.2	-20	25	-50	-2.5
Minimum range	-2.5	-21.5	31	-21	-23.2	4.2	-20	10	-40	-2.5
Maximum range	-4.0	-34.4	50	-33	-37.1	6.8	-32	40	-60	-4.0
Maximum simulated	4.2	113.9	55	-28	-3.3	15.0	4	64	-10	31.1
Minimum simulated	-41.9	-75.2	36	-37	-34.5	4.6	-46	30	-42	-7.9

Table 5.2: Article 3 Table 2: Significant model parameters and their importance for each geometric measure

	Decompensation (mm)	Balance (mm)	Main Thoracic Cobb (°)	Lumbar Cobb (°)	Thoracic AVT (mm)	Lumbar AVT (mm)	AVR (°)	Kyphosis (°)	Lordosis (°)	Rib Hump (°)
Pelvice inclination			3			3	2	2	1	
Chest Cushion Location		1	6					1	2	2
Chest Cushion Height		3	5			4		3	4	
Rib Hump Force		2	2	2	3		1	4	3	1
Lateral Thoracic Force			1		1	1	3		5	3
Lateral Lumbar Force	1		4	1	2	2				

Table 5.3: Article 3 Table 3: Optimised geometric measures and corresponding model parameters.

	Suggested Model Parameters						Resultant Geometric Measure									
Geometric measure optimised	Pelvice inclination (°)	Chest cushion location (1st rib)	Chest cushion height (cm)	Rib Hump Force (N)	Lateral Thoracic Force (N)	Lateral Lumbar Force (N)	Decompensation (mm)	Balance (mm)	Main Thoracic Cobb (°)	Lumbar Cobb (°)	Thoracic AVT (mm)	Lumbar AVT (mm)	AVR (°)	Kyphosis (°)	Lordosis (°)	Rib Hump (°)
Decompensation	-12	3	0.0	112	150	10	0	-3	40	-31	-21	15	-19	35	-21	12
Balance	2	4	0.0	81	80	80	-18	0	45	-32	-22	10	-23	38	-28	7
Main Thoracic Cobb	-15	6	3.5	10	150	10	-11	-36	35	-31	-10	14	-41	41	-25	-1
Lumbar Cobb	-15	6	0.0	10	139	150	-21	10	37	-29	-11	11	-40	39	-21	-3
Thoracic AVT	-4	4	0.0	81	68	78	-17	-2	45	-32	-23	9	-25	37	-26	6
Lumbar AVT	15	4	0.0	18	10	150	-21	-29	49	-29	-30	4	-39	41	-37	-1
AVR	9	4	0.0	90	81	80	-19	-3	45	-32	-21	10	-20	38	-30	10
Kyphosis	-15	3	0.0	150	10	150	-29	-13	49	-32	-26	6	-14	29	-22	15
Lordosis	15	3	3.5	10	10	150	-20	-73	50	-31	-30	4	-41	46	-43	0
RibHump	8	4	0.0	14	58	75	-16	-2	45	-31	-24	9	-39	43	-32	-3
Equal weighting	12	4	0	68	97	10	-13	-3	43	-33	-20	12	-24	40	-32	7

Table 5.4: Article 3 Table 4: Calculated difference from linear regression predicted geometric measures and the finite element model simulated geometric measures.

		Decompensation (mm)	Balance (mm)	Main Thoracic Cobb (°)	Lumbar Cobb (°)	Thoracic AVT (mm)	Lumbar AVT (mm)	AVR (°)	Kyphosis (°)	Lordosis (°)	Rib Hump (°)
Training set n=32	Average	2.5	5.4	0.4	0.8	1.2	0.1	1.5	1.2	0.4	1.9
	Standard deviation	1.7	3.8	0.3	0.5	0.9	0.1	1.1	1.1	0.2	1.6
Test Set n=11	Average	2.6	7.8	0.8	0.6	2.0	0.1	2.8	1.7	1.3	2.3
	Standard deviation	2.3	4.8	0.5	0.5	1.8	0.1	1.3	1.5	0.6	1.7

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CHAPTER SIX: DISCUSSION

The aim of this project was to design, test and model a dynamic positioning frame (DPF) to improve the position of scoliotic patients during surgery. Personalised finite element modeling of the scoliotic spine was used in order to better understand the biomechanics of patient positioning and to provide a tool to help recommend adjustments of the DPF components. Two hypotheses were presented. First, the utilization of a DPF can improve the correction of the spinal deformity and the overall trunk geometry compared to the Relton-Hall frame used in the conventional approach. Second, the osseo-ligamentous structures of the patient, the effect of gravity, anaesthesia and the surgical position can be modeled, and such computer models can be used to recommend the placement of the positioning frame components on a patient for correction of the spinal deformities.

To begin, a DPF that could be used in the operating room had to be designed and fabricated. The DPF is novel in that the cushions can be displaced in three directions and can also rotate. By being able to independently modify the position of the cushions, particularly in the x direction, the balance, kyphosis, rib hump and Cobb angles of the patient can be modified. External correction cushions provide a novel way of freeing up the hands of the resident or surgeon's assistant and can provide constant corrective force to the trunk deformities. This study was the first in which active correction was attempted to provide even greater correction of the deformity.

The initial clinical study (n=11) showed that the patient's external trunk geometry was modified when lying on the DPF. Others have found that the prone positioning alters the scoliotic deformity (Delorme et al. 2000, Behairy Y et al. 2000, Mac-Thiong et al. 2002, Polly et al. 1998). Our study compared the patient's external trunk geometry while standing, lying prone on the Relton-Hall frame and lying prone on the DPF. On the DPF the patient's torsional deformity significantly improved as delta hump decreased from 8.3° to 3.7° compared to an insignificant correction of 8.3° to 7.6° on the Relton-Hall frame. Kyphosis height, originally at 8 cm, significantly flattened on the Relton-Hall frame (5 cm) but was 6 cm on the DPF which was not a significant reduction. The Jackson table, similar to the Relton-Hall frame, has also shown a significant decrease in kyphosis from 34° to 28° due to prone positioning (Marsicano et al. 1998). The DPF could provide an advantage over the Jackson and Relton-Hall frames since it can be adjusted to provide less of a reduction in kyphosis. In addition, spine height significantly increased on the Relton-Hall frame and then again significantly more on the DPF. This lengthening is likely due to a combination of the straightening of both sagittal and frontal curves. Since the sagittal curve straightens less on the DPF, but the overall straightening (spine height) is significantly more, this straightening can be attributed to further correction in the frontal plane.

During the second clinical study on a smaller group of patients (n=3) in the operating room, spine geometry was extracted from x-rays while the patients were lying on the DPF. Due to the cassette size, 28 cm x 36 cm (11x14 inch), the entire spine could not be imaged and therefore kyphosis and lordosis could not be measured. However, the thoracic and lumbar Cobb angles could be measured and the average correction of both was 40% prior to instrumentation. Comparing this to the study of Delorme et al. where 37% correction was observed there appears to be similar correction (Delorme et al. 2000). The final correction of the three patients averaged 68% and 75% for the thoracic and lumbar cures respectively while Delorme et al. found 57% for both curves. A bias must be disclosed that could be partially responsible for this large overall correction. These first three patients were selected as they were "uncomplicated" AIS. There is also the fact that the study is not blind and the surgeons were obviously aware that they were using the DPF. The full potential of the DPF was not tested in these trials as the external correction cushions were not utilised. It is anticipated that even more correction will be observed on the intra-operative x-rays once the correction cushions are in place. The results of these preliminary cases are promising and show the feasibility of using the DPF. A larger clinical trial is underway to determine if there is a significant improvement in the amount of intra-operative and post-operative correction achieved with the DPF.

In addition to measuring the external trunk geometry or the internal spine geometry, pressures were measured using a force sensing array (FSA) pressure mapping device (Vista Medical, Winnipeg) during both clinical studies. During the first clinical study the pressures were higher on the DPF compared to the Relton-Hall frame, yet peak pressures of 300 mmHg were recorded on both systems. The higher pressures on the DPF with external forces are not a completely negative result as they show that the applied corrective forces are being transferred through the patient. However, the pressures became a major concern in this study and the pressure results were used to help redesign the cushions in order to make the pressures equivalent to, or less than, the Relton-Hall system to ensure patient safety.

Prior to this study the pressure at the patient cushion interface had not been measured on scoliotic patients in the prone position. During surgery, pressures as high as 260 mmHg have been recorded (Neander et al. 1991). Not only is the pressure a factor, but surgeries lasting more then 2.5 hours pose a greater risk of pressure sores (Hoshowsky et al. 1994), making scoliosis patients susceptible. The design of the DPF incorporated Tempur foam which was found to significantly reduce the pressures but not sufficiently to prevent the occurrence of pressure sores (Defloor et al. 2000). The research into pressure measurements intra-operatively is limited but much work has been done in the wheelchair population. It is difficult to compare the magnitude of the peak pressures during surgery to those in a wheelchair since the positioning during the surgery is only a temporary position and it is not repeated day after day. The anatomy in contact with the cushions is also different, however the order of magnitude of the pressures is worth noting. In the wheelchair population those who developed pressure sores had significantly higher peak pressures averaging 115 $\pm$ 45 mmHg and the location of the pressure sore coincided with the location of the peak pressure measured on an FSA (Brienza et al. 2001). Although we did not observe any pressure sores, this is consistent with our three clinical cases in the operating room where reddening of the skin also coincided with peak pressures on the FSA. The pressures on the DPF were higher during surgery compared to the day before. This could be due to the anaesthesia and the difference in pressure requires further investigation. There is still insufficient evidence to determine a pressure threshold at which patients are at risk of developing pressure sores or neurological complications when positioned prone on the operating table. Future designs of positioning systems need to be more aware of the complications that can arise due to improper positioning and take this into account from the initial concepts.

After using the DPF certain improvements or modifications to the design are noted. For the purpose of this study the DPF was designed to accommodate paediatric scoliosis patients. Future designs should be able to adjust for different spine surgeries and accommodate adults including the increasing number of obese patients in the North American population. Stevens et al. (1996) recommend the Jackson table when positioning obese patients to allow for decompression of the viscera and decrease blood loss. A commercial prototype of the DPF, suitable for adult spine cases, should be rated to a maximum of 250 kg. Before moving to a commercial prototype, modifications must be made to the corrective cushions as those designed for this project must be gas sterilised and are not completely radiotransparent. Ideally the corrective cushions would be reusable and could be autoclaved. However, from a practical and hygienic point of view, the base could be autoclaved and a pre-packaged, single use, sterile cushion could be snapped on. Improvements can also be made to the base cushions so that they are low profile to facilitate patient transfer.

As required in the initial design criteria, the DPF is sufficiently radiotransparent to provide exposure of the spine and partial ribcage during the standard AP and lateral x-rays. However, part of the cushion support is visible on the edge of the x-rays. Fluoroscopy is sometimes used instead of an AP or lateral x-ray or when an *en face* x-ray of the vertebra pedicle canal is desired for pedicle screw insertion or tool placement. Based on pedicle orientation and inclination in scoliotic patients (Parent et al. 2004), and apical vertebral rotation (Beuerlein et al. 2003) the C-arm could be required to rotate up to 45° from the AP position when taking an *en face* x-ray. If a fluoroscopy is to be used, ideally, the DPF should be completely radiotransparent. To provide the greatest freedom in taking images during surgery the fluoroscopy and/or the cassette should be able to be positioned anywhere from head to toe. These modifications to the materials are possible if using carbon fibre, Delrin®, Nylon® or other types of plastics however it comes at an increased cost and difficulty in manufacturing.

In addition to the above mentioned design and clinical trials there was a biomechanical modeling component. The modeling component serves to better understand the biomechanics involved in patient positioning. Since the same patient can not be operated on two different systems, finite element modeling serves to virtually test the effect of the Relton-Hall system versus the DPF on the same patient.

The positioning model developed in this study was presented in Chapter 3. First the material properties of the model were personalised to the supine bending x-rays. This was accomplished by simulating the supine position with gravity, the interaction of the patient's trunk with the table, and then finally the lateral displacement of T1. This provides an improvement over existing models where an admitted limitation was that the side bending was simplified by only constraining L5 and the lateral displacement of T1 (Petit 2002). Though the bending simulation used in this study better incorporates gravity and the interaction with the table, it is still limited as imposing a fixed displacement to T1 is a rigid constraint and does not take into account the active muscle forces required to bend.

The simulation of gravity, by not only applying forces in the posterior anterior direction but, by also applying a traction type force in the cranial direction to compensate for the absence of gravity acting down on the spine in compression is a novel technique. The model was validated against x-rays taken intra-operatively before instrumentation. The positioning model developed in this study can be used to provide more information to existing models that simulate the surgical procedure. It has been noted that existing finite element models are limited since they have neglected to take into account patient positioning and have only simulated the instrumentation (Stokes et al. 1999).

This model is an improvement over existing models but still poses some limitations. First, as mentioned in the first article, for the anaesthesia factor to be a useful tool it must be able to be predicted from pre-surgical data. There is perhaps a relationship to age, patient weight, race, dose, type of anaesthesia and other factors that may make one patient more relaxed compared to another. Another, limitation is that during surgery, when the prone AP x-ray is taken, the ligaments, muscles and some facets have been resected. In a recent model by Lafage et al. (2004), they simulate the facet resection but do not say what percentage of correction this contributes to. As mentioned earlier it is believed that prone positioning is responsible for about 25% correction and that an additional 10 to 15% is due to the anaesthesia and the opening of the patient. Due to ethical reasons we are not able to take multiple images to provide clinical validation of the contribution of the patient prone, prone with anaesthesia and prone with anaesthesia and surgical resection.

The prone positioning model was used, in Chapter 5, to test the sensitivity of the model, determine significant parameters and finally determine optimal cushion placement. The results of the model can be used to suggest design modifications to the DPF or help prescribe its use. Every model parameter had a significant effect on at least five of the geometric measures. It is not surprising that the lateral thoracic force improved the main thoracic Cobb angle. It is interesting to note that chest cushion location significantly altered the balance, kyphosis, lordosis and rib hump. It was shown in the first clinical study (Chapter 4) that chest cushion height had a significant effect on kyphosis height but chest cushion location was originally considered less important and was therefore not modified. Since chest cushion location had a significant effect on kyphosis and balance, care should be taken by the clinician when positioning the chest cushions. Simultaneous optimisation of all ten geometric measures resulted in an overall correction of 75% simulating the DPF compared to 55% simulating the standard prone position.

In order to test the different parameters of the model an experimental design was simulated. The concepts of experimental design have been widely used in engineering and are often used to save on costs when destructive experiments are required in developing an optimal product. The major "cost" when running finite element simulations is the time required for the simulation to converge. The experimental design offers an efficient way of optimising the material properties and cushion placement. In addition, being able to extract the regression coefficients of the finite element model and use this as a simplified mathematical model has great potential to reduce computational time. The error of the simplified mathematical model compared to the finite element model is under 2° for the Cobb angles. To the author's knowledge, the results of an experimental design and have not been used with the goal of replacing a finite element model. It should be noted that for non linear finite element models a Box-Benkhen type design at three modalities should be used, while for linear models a Box-Hunter experimental design at two modalities should suffice.

Although the finite element model is a useful tool to better understand the biomechanics of patient positioning there are certain limitations of the model and this study. One of the major limitations is that the muscles were not simulated. A simplified model without muscles was used because the muscles are not active during surgery. However, the muscles are still indirectly taken into account as the segmental stiffness of the spine was adjusted based on the bending tests which incorporated the muscles. Because the muscles were not present the direct effect of anaesthesia on the muscles was not simulated. Another limitation is that the model was only used to simulate the immediate effect of positioning and not the outcome of the entire surgery after instrumentation. By simulating optimal positioning followed by instrumentation one could see if the forces required to correct the deformity are in fact lower then when the patient is positioned on the Relton-Hall frame. In order to see the immediate effects of the positioning and the instrumentation the gravity and positioning components of this model would have to be combined with an instrumentation model such as the one of Lafage et al. (2004) or Aubin et al. (2003). The long term biomechanical effects of improved patient positioning and overall surgical effects need to be studied. In order to truly see the long term effects the growth would also have to be incorporated into the biomechanical model (Villemure et al. 2002). In addition, the model is simplified in that it does not represent the true volumetric geometry of the vertebra. If a true volumetric model was used, other clinical measures could be extracted from the model such as the effect of positioning on intervertebral disk space which would be useful when planning a discectomy procedure.

It should also be noted that the positioning itself is simplified and could be improved in future versions of a positioning model. At present, boundary conditions are used to represent the cushions. In order to make this more realistic they could be replaced with contact elements which would provide a more realistic stiffness similar to the actual cushions and would allow for the patient to separate from the surface of the cushion. Another improvement would be to add the soft tissues and skin to completely simulate the patient's trunk. The abdomen has already been added to a model simulating brace biomechanics (Perie et al. 2003). The additional benefit, and added accuracy of such modifications, would have to be analysed in future studies. Unfortunately, such improvements to the model would come at a cost of increased computational time. As mentioned in the third article, computational time is already a concern and it is relatively long at around 20 minutes. The objective of this study was not to optimise computational time but in future studies improvements could be made the existing finite element model to reduce computational time. To further reduce computational time a flexible multi-body model (like ADAMS) could be used. This has already been done for instrumentation simulations were solutions converged in around 1 minute (Aubin et al. 2003).

One final limitation is that the simulation of the DPF has not been clinically validated. The finite element model used in these simulations has been validated in a recent study of brace biomechanics with twelve patients (Perie et al. 2003). The position of the patients was also validated on the Relton-Hall frame for two patients (Duke et al. 2005). In simulating the DPF an efficient method of studying and optimising 32 different configurations of cushion placement was presented. It would be unrealistic to subject a patient to all 32 different positions and validate the model for each configuration. Ideally, the precise cushion placement should be noted during surgery and a retrospective study can be performed to clinically validate the model. Comparisons were not made between the model and the three patients operated on the DPF for a number of reasons. First, the external correction cushions were not used and only the base cushions were in place. The positioning of the base cushions were at the surgeon's discretion and were potentially not optimised. Finally, the 3D intra-operative reconstruction of the patients was not available since the necessary calibration object was not present as it is undergoing design improvements. Ideally, a prospective study can be performed where optimal simulation and cushion placement is predetermined and prescribed during the surgery. The intra-operative geometry of these patients can then be compared to the model output and to other patients operated in the traditional manner.

Overall the positioning model verified the second hypothesis in that the osseo-ligamentous structures of the patient, the effect of gravity, anaesthesia and the surgical position were modeled, and the model was used to recommend the adjustment of the DPF components for spinal deformities. Indirectly, since the simulation of the DPF improved the overall correction of the spinal deformity compared to the simulated Relton-Hall frame, the model also served to test the first hypothesis.

CHAPTER SEVEN: CONCLUSIONS

The aim of this thesis was to design, fabricate, test and model a dynamic positioning frame (DPF) with the first hypothesis that this DPF can improve the correction of the spinal deformity and the overall trunk geometry compared to the Relton-Hall frame used in the conventional approach. The computer modeling aspect of this project was necessary to test the second hypothesis which was that the osseo-ligamentous structures of the patient, the effect of gravity, anaesthesia and the surgical position can be modeled, and such computer models can be used to recommend the location and magnitude of displacement to apply on a patient for correction of the spinal deformities.

The above hypotheses were met by attaining specific goals set out by four smaller objectives. The conclusions to these objectives will be presented below.

In order to satisfy the first objective, which was to refine a finite element model such that it is able to simulate the influence of patient position and the effects of anaesthesia while incorporating patient personalised material properties, an original approach to modeling the effect of gravity and anaesthesia was presented in Chapter 3. Personalisation of the material properties by means of the bending test is not a novel idea but the method of using a Box-Benkhen experimental design to simultaneously satisfy and optimise the left and right bending x-rays is a new approach. Bending simulations were also more realistic, compared to previous work, as they took into account the supine position, gravity and the interaction of the patients back with the examining table as well as the T1 displacement. The Relton-Hall cushions were simulated as boundary conditions under the patient's pelvis and rib cage. Gravity was simulated as a distributed load applied to each vertebra in the anterior posterior direction but also as a "traction type" load in the cranial direction to remove the compressive gravity forces naturally present when standing. An anaesthesia factor was applied to the model to decrease the stiffness of the soft tissues. Results showed significant correction of the thoracic and lumbar curves due to positioning. This correction was increased, and the simulations better correlated to intra-operative x-rays, when patient personalised material properties and anaesthesia was taken into account. For the first patient, simulated correction was from 62° to 47° and 43° to 31° for the thoracic and lumbar curves respectively. For the second patient, thoracic correction was 70° to 55° while the lumbar corrections was 40° to 32°.

The second objective was to set out the design criteria and to incorporate these specifications into the design improvement and fabrication of a prototype of the DPF that could be used in the operating room. Once the design criteria were established they were attributed specific weightings. Then, a virtual prototype was designed in CAD software and this was used to produce the technical drawings necessary for fabrication. The DPF was fabricated with cushions which contoured to the sides of the patient, could be translated in three principle directions and one degree of rotation to adjust to different sized patients. External correction cushions were also designed and fabricated.

Once the DPF was designed, this led to the third objective which was the clinical testing on a group of scoliotic patients first, in a clinical setting and then, in the operating room. The results of the first clinical study show that there is an overall improvement in the trunk geometry when patients are positioned on the DPF. In particular, significant improvements were found in the spine height which increased from 41.9 cm to 47.6 cm and the delta hump which decreased from 8.3° to 3.7°. The kyphosis height, originally at 8 cm, significantly flattened on the Relton-Hall frame (5 cm) but was 6 cm on the DPF which was not a significant reduction. Unfortunately, higher pressures were noted on the DPF and modifications were made to the cushions before proceeding to the clinical study in the operating room.

By using the DPF in the operating room the feasibility of this system was demonstrated. The first patient's thoracic and lumbar curves corrected by 38% and 36% intra-operatively and 73% and 50% post operatively. The second patient's thoracic and lumbar curves corrected by 42% and 36% intra-operatively and 74% and 84% post operatively. Finally the third patient had a 48% correction of the lumbar curve intra-operatively while the post-op curves corrected to 57% and 90% for the thoracic and lumbar curves. The pressures measured on the DPF were higher for all three patients when in the operating room under anaesthesia. Unfortunately, at the present time there are insufficient cases to determine if there are significant differences in these cases compared to patients operated on a Relton-Hall frame. However, using the DPF in the operating room provided a great opportunity to get feedback from the surgeons and to suggest modification, for future generation designs.

Because it is impossible to operate on the same patient using two or more different positioning systems, and to further understand the biomechanical potential of the DPF, computer modeling was utilised for the final section of this thesis. The fourth and final objective was to modify the model created in objective 1 to simulate the DPF and test various adjustment parameters and recommend placement of the positioning system components prior to, and during surgery based on the results of the computer simulations. Modifications to the model were made such that six parameters of the DPF were tested while ten clinical geometric measures were analysed. This model illustrated what parameters had a significant effect and optimal cushion placement was determined. Using the equations, which resulted from the statistical analysis of Box-Hunter experimental design, a simplified mathematical model of the complex finite element model was created. Using this novel methodology, and the resulting equations, there is great potential for surgical simulators that could quickly analyse different input parameters. With the modeling we were able to show that there is improved overall correction of the scoliotic deformity when the DPF is simulated (75%) compared to simulations of the Relton-Hall system (55%). Looking specifically at the thoracic and lumbar Cobb angles they were 62° and 42° for the standing geometry. These corrected to 47° and 34° when simulated on the Relton-hall frame and further corrected to 43° and 33° when positioned on the DPF. This improvement in overall geometry not only addresses the second hypothesis, which was directly related to modeling, but can also be translated to virtually satisfy the first hypothesis which addresses the improved correction of scoliotic deformity when comparing the DPF and the Relton-Hall system.

The culmination of the above described work helped answer the questions addressed in the hypothesis but also led to more questions and interesting areas of research. There are many recommendations for the design, clinical and modeling aspects of this work.

The following are recommendations to the design and further clinical studies used to validate the system.

- The clinical trial of the DPF in the operating room must continue so that it can be determined if significant differences are found between the DPF and the Relton-Hall system. Ethics approval has been obtained to do a paired comparison of the x-rays of 30 patients on the DPF with 30 patients on the Relton-Hall frame;
- Future modification to the design of the DPF should take into account reduced pressures, offer complete radiotransparence, adjust to other spine surgeries, accommodate obese adult patients, and improve patient transfer in addition to the existing design criteria;
- Measures of the pressures during standard scoliotic cases on the Relton-Hall system would be useful to help quantify the existing pressure levels intra-operatively and could establish a safe threshold of maximum acceptable pressure.

The following recommendations pertain more specifically to the modeling section of this work.

- Since it has been clearly shown that the positioning of the patient has a biomechanical effect on the scoliotic deformity, it is recommended that any models that simulate the surgical procedure should also include patient positioning simulation;

- A better understanding of the biomechanics of lateral side bending is required. A retrospective clinical study of the bending x-rays is recommended where not only the Cobb angles are measured but also the displacement of T1. The displacement of T1 may be an indication of patient effort and therefore how valid the bending test is as a predictor of spine flexibility. Once the biomechanics of bending are better understood this can be reliably incorporated into the computer models;
- The simulation of the patient position can be made more realistic by simulating the patient cushion interface with contact elements and adding soft tissues, including muscles, to the model;
- The analysis of the geometric measures from the model should be automated and the vertebral levels form which to measure the Cobb need to be selected so that the pre-simulation and post-simulation curves are analysed between the same end vertebrae as presently done in the clinical setting;
- Further investigations into experimental designs and statistical analysis are recommended to potentially create a simplified mathematical model of the current complex finite element model for use in a surgical simulator.

Designing a positioning frame that can reduce the patient's scoliotic deformity compared to the Relton-Hall frame is a novel concept. The DPF was able to correct the trunk deformity and its feasibility was demonstrated in the operating room. The finite element model (Chapter 3) which simulated the patient positioning, gravity and anaesthesia, should be used with existing instrumentation models in order to obtain the correction that is observed intra-operatively. By merging the DPF with the finite element model the positioning of the patient can be optimised and specific cushion adjustments can be prescribed. And finally, the innovative methods developed using experimental design can be used to create simplified models, representing the finite element models, which could be used for surgical simulators. In the future, the positioning of the patient will be approached from a different perspective knowing that it can have a profound effect on the biomechanics of the patient.

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ANNEX 1

DETAILS OF THE STEPS REQUIRED TO SIMULATE AND OPTIMISE CUSHION PLACEMENT

As mentioned in the article presented in Chapter 5, one limitation that is preventing the finite element model from becoming a useful clinical tool is the length of time and different programs involved in the model construction and data analysis. This annex is provided to review the steps required to simulate and optimise cushion placement using the finite element model.

To create the finite element ANSYS model stereo radiographs must be manually digitized. From that point, a number of in-house programs are used to create the finite elements model nodal points and initial material properties. At present, the digitisation and model creation can be done in approximately six to eight hours. These steps were established and validated from the previous work of others (Aubin et al 1995, Delorme et al 2003).

Once the base model is created a number of steps are first completed to determine the patient personalised material properties and they are then repeated to optimise the cushion placement as shown in the flow chart in Figure A1.1. The detailed procedure for optimising the cushion placement is described below.

Statistica is first used to determine the experimental design, for the case presented here with 6 manipulated variables there are 32 runs. Once the different tests are determined, 32 specific finite element model programs are created, to apply the boundary conditions and external forces. In Ansys, a sub program containing the geometry and material properties is first called, then one of the 32 specific programs is called and finally a program to extract the final nodes is run. These 3 programs are repeated over and over again 32 times. These simulations can be run consecutively from a single input file and 32 different output files will be written. Creating the 32 specific files takes around a days work and running the simulations can be done overnight.

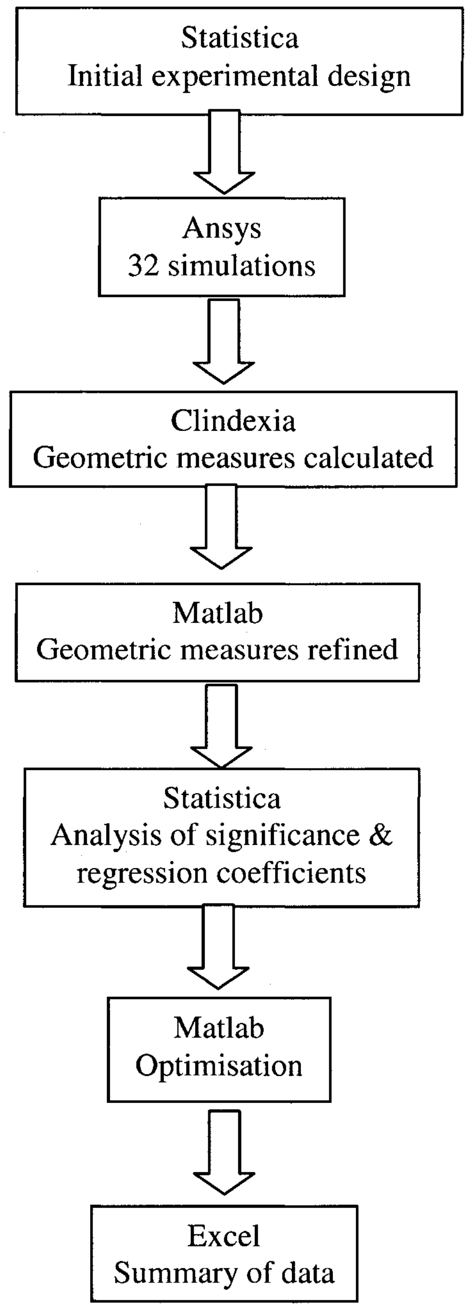

Figure A1.1 : Flow chart representing the steps required to find the optimal cushion placement.

The output files can then be read by Clindexia (an in-house program) where three other in house programs (angle.exe, colonne.exe and cotes.exe) are called to determine various clinically relevant geometric measurements and write a *.cot, *.cob and *.col output files containing geometric information for the ribcage, Cobb angles and spine respectively. Additional geometric measures, not directly provided by Clindexia, are extracted with the help of Matlab programs to provide more geometric measures specific to the patient. For example, with Matlab, the Cobb angles are always measured with the same end vertebra. This is done because clinically the surgeon compared his before and after Cobbs at the same vertebral levels. This option is presently not available in Clindexia since angle.exe calculates the Cobbs based on the inflection points which can change as the shape of the spine changes. In order to truly compare apples to apples the vertebral level should be kept constant. If not, the vertebral levels need to at least be noted.

The geometric measures are then transferred back to Statistica and added to the original experimental design as the dependent or responding variables. The results of the 32 simulations are shown in Table A1.1. Statistical analysis is performed on each of the geometric measures as a function of the model parameters. Paretto charts are analysed to see what model parameters have a significant effect. Figure A1.2 shows a sample Paretto chart for the main thoracic Cobb. The regression coefficients are also extracted using Statistica (Table A1.2). Once the regression coefficients are obtained there are 10 equations, corresponding to the geometric measures, with 22 coefficients, corresponding to the 6 model parameters with 15 interactions and one error coefficient.

The equations are entered into Matlab where the cost function, presented in Equation 5.2.1, is optimised using the fmincon function. The fmincon function finds the minimum for a constrained number of variables (MathWorks Natick, Massachusetts, USA). Optimisation is performed with the desired geometric measure appropriate weighting and the corresponding model parameters are calculated.

The equations are also entered into Excel where calculations can be easily made to determine the resulting geometric measures as a function of different model parameters. For example if a surgeon would like to try a specific cushion placement the resulting geometric measures will be displayed. It should also be mentioned that Excel is used during intermediate steps to keep tract of the data.

Ideally, the results of the Ansys simulations could be entered into one software program where the entire statistical analyse and optimisation is performed in one step and displayed with a graphical user interface for the surgeon to enter his desired geometric measures or DPF parameters.

Table A1.1:Results of the 32 simulations

	Imput Parameters						Output parameters									
	Pelvice inclination	Ribs	Rib height	Rib Hump Force	Lateral Thoracic Force	Lateral Lumbar Force	Decompensation (mm)	Balance (mm)	Main Thoracic Cobb (°)	Lumbar Cobb (°)	Thoracic AVT (mm)	Lumbar AVT (mm)	AVR (°)	Kyphosis (°)	Lordosis (°)	Rib Hump (°)
1	-15	3	0	10	10	10	-10	-26	48	-34	-31	10	-39	33	-25	-4
2	15	3	0	10	10	150	-21	-48	49	-31	-30	5	-41	39	-40	-3
3	-15	6	0	10	10	150	-23	15	48	-30	-28	6	-39	37	-23	-5
4	15	6	0	10	10	10	-7	53	50	-31	-34	9	-42	53	-32	-8
5	-15	3	3.5	10	10	150	-23	-61	49	-32	-29	5	-40	39	-29	-3
6	15	3	3.5	10	10	10	-9	-75	48	-33	-30	9	-41	47	-42	-3
7	-15	6	3.5	10	10	10	-10	5	49	-33	-32	10	-40	48	-25	-6
8	15	6	3.5	10	10	150	-16	45	52	-28	-35	5	-45	64	-34	-7
9	-15	3	0	150	10	150	-31	-12	49	-32	-24	5	-18	30	-21	11
10	15	3	0	150	10	10	-20	-30	50	-35	-28	9	-8	32	-35	18
11	-15	6	0	150	10	10	-15	84	48	-32	-27	10	-21	42	-14	5
12	15	6	0	150	10	150	-33	114	54	-33	-28	5	-8	52	-20	6
13	-15	3	3.5	150	10	10	-22	-45	49	-36	-25	9	-16	36	-25	13
14	15	3	3.5	150	10	150	-36	-58	52	-34	-24	5	0	36	-39	26
15	-15	6	3.5	150	10	150	-34	75	51	-32	-24	5	-9	49	-13	10
16	15	6	3.5	150	10	10	-25	93	55	-37	-30	9	-7	60	-21	6
17	-15	3	0	10	150	150	-27	8	39	-31	-9	10	-38	48	-19	-3
18	15	3	0	10	150	10	-19	-38	37	-33	-7	14	-38	41	-36	2
19	-15	6	0	10	150	10	-1	1	36	-30	-17	15	-43	35	-22	-6
20	15	6	0	10	150	150	-27	15	37	-29	-7	10	-38	46	-32	-1
21	-15	3	3.5	10	150	10	-18	-53	37	-34	-6	15	-42	43	-26	-2
22	15	3	3.5	10	150	150	-34	-70	39	-32	-3	10	-40	48	-40	4
23	-15	6	3.5	10	150	150	-30	-20	36	-30	-4	11	-46	45	-24	-7
24	15	6	3.5	10	150	10	-15	0	38	-33	-11	14	-33	54	-34	3
25	-15	3	0	150	150	10	4	-6	41	-32	-25	15	-11	32	-20	16
26	15	3	0	150	150	150	-17	-22	42	-30	-19	10	2	31	-33	29
27	-15	6	0	150	150	150	-15	91	41	-29	-18	11	-5	40	-10	17
28	15	6	0	150	150	10	-22	114	44	-37	-14	14	1	50	-19	13
29	-15	3	3.5	150	150	150	-20	-36	42	-31	-16	11	-6	36	-22	24
30	15	3	3.5	150	150	10	-6	-55	42	-33	-20	14	4	35	-37	31
31	-15	6	3.5	150	150	10	-9	17	40	-31	-17	13	-6	39	-14	20
32	15	6	3.5	150	150	150	-42	92	47	-37	-11	9	1	58	-21	8

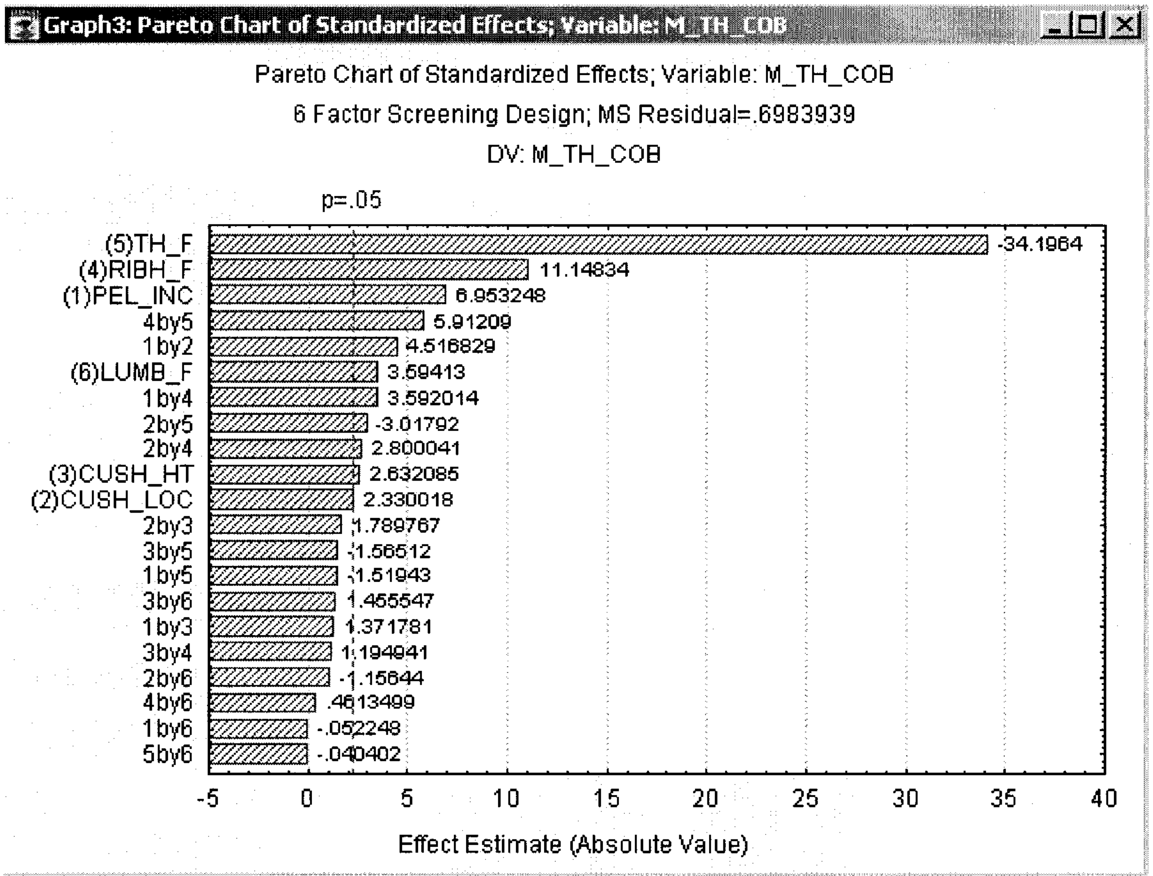

Figure A1.2: Paretto chart for the main thoracic Cobb.

All model parameters that extend beyond the p=0.05 line are considered to have a statistically significant effect on the main thoracic Cobb.

Table A1.1: Regression coefficients for the equations.

Representing the 10 geometric parameters. k=constant, a=pelvic inclination, b=chest cushion location, c=chest cushion height, d=rib hump force, e=lateral thoracic force, f=lateral lumbar force.

	Decompensation (mm)	Balance (mm)	Main Thoracic Cobb (°)	Lumbar Cobb (°)	Thoracic AVT (mm)	Lumbar AVT (mm)	AVR (°)	Kyphosis (°)	Lordosis (°)	Rib Hump (°)
k	-20.095	-98.218	48.299	-37.630	-26.455	8.747	-42.487	30.365	-35.818	-7.455
a	0.262	-2.545	-0.101	0.144	0.029	-0.036	0.083	-0.236	-0.650	0.465
b	2.898	21.716	0.208	1.034	-1.656	0.135	-0.188	2.457	1.161	-0.181
c	1.047	-11.880	-0.336	0.010	0.299	0.088	0.499	0.828	-1.677	1.747
d	0.021	-0.365	-0.012	0.017	-0.003	0.004	0.174	-0.117	-0.040	0.182
e	0.022	0.300	-0.064	0.020	0.121	0.039	0.009	0.078	0.039	0.034
f	-0.069	-0.091	0.011	0.024	-0.004	-0.034	0.018	-0.005	0.001	0.039
ab	-0.050	0.589	0.030	-0.023	-0.012	0.001	-0.016	0.124	0.060	-0.081
ac	0.032	0.103	0.008	0.002	-0.029	0.001	0.003	0.022	0.006	-0.009
ad	-0.002	0.002	0.001	-0.001	0.000	0.000	0.002	0.000	0.000	0.000
ae	-0.002	-0.001	0.000	0.000	0.001	0.000	0.001	-0.001	0.000	0.000
af	0.001	-0.002	0.000	0.000	0.000	0.000	-0.001	0.000	0.000	0.000
bc	-0.150	1.191	0.101	-0.004	-0.097	-0.029	-0.038	0.342	0.211	-0.229
bd	-0.023	0.140	0.004	-0.007	0.010	-0.001	0.001	0.024	0.021	-0.017
be	-0.011	-0.076	-0.004	-0.004	0.007	-0.001	0.002	-0.017	-0.005	0.000
bf	-0.002	0.008	-0.002	0.000	0.003	0.001	-0.003	0.000	0.000	-0.005
cd	-0.006	-0.010	0.001	-0.001	0.000	0.000	0.010	-0.004	0.001	0.004
ce	-0.009	-0.029	-0.002	-0.001	0.007	0.000	-0.004	-0.007	-0.002	0.000
cf	-0.004	0.018	0.002	-0.002	0.001	0.000	-0.005	0.002	-0.001	-0.006
de	0.001	0.000	0.000	0.000	-0.001	0.000	0.000	0.000	0.000	0.000
df	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000
ef	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000	0.000