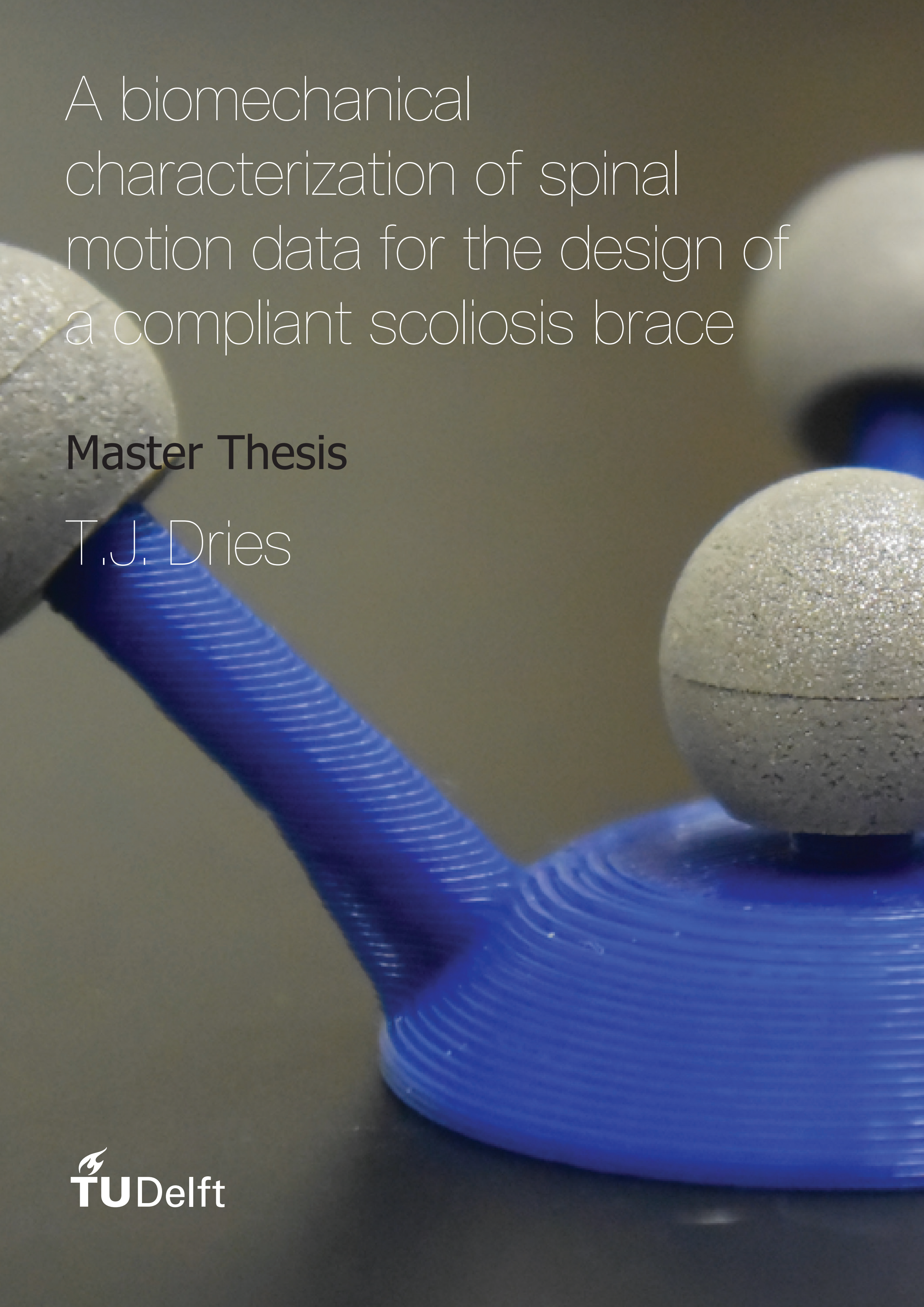
A biomechanical characterization of spinal motion data for the design of a compliant scoliosis brace



Master Thesis

T.J. Dries

TU Delft

A biomechanical characterization of spinal motion data for the design of a compliant scoliosis brace

Master Thesis

by

T.J. Dries

to obtain the degree of Master of Science
at the Delft University of Technology,
to be defended publicly on Tuesday May 22, 2018 at 1:30 PM.

Student number:	4091957	
Project duration:	September, 2017 – May, 2018	
Thesis committee:	Dr. ir. D. H. Plettenburg,	TU Delft, supervisor
	Prof. Dr. ir. J. L. Herder,	TU Delft
	Dr. ir. C. Ayas,	TU Delft
	Dr. ir. C. J. Kim,	Bucknell University

This thesis is confidential and cannot be made public until June 1, 2019.

Preface

This thesis is the final outcome of my study Biomedical Engineering at Delft University of Technology. I am very grateful for all the support I got throughout my entire studies from tutors, friends and family.
I would like to thank Charles for giving me the opportunity to work on this project, for his mentoring during the entire project, reviewing my work and for the trips I was able to join thanks to this project.
I would like to thank Dick for his support during the project and the constructive feedback.
I would like to thank Joost for joining me in this project. Thank you for the discussions we had about the project work, the productive contemplations after a hard day of work on the porch in Lewisburg, and for being my travel companion in the big American adventure.
I would like to thank Yanthe for always supporting me. You are the one calming me down, putting things into perspective and motivating me to keep on going when needed.
Finally, I would like to thank my parents and sister for supporting me in every way possible throughout all the years in Delft, Copenhagen and Lewisburg. Without you I would never have managed to become who I am.

T.J. Dries
Delft, May 2018

Contents

I

Thesis

1

Introduction

Adolescent Idiopathic Scoliosis (AIS) is a condition of the spine, often characterized by a three-dimensional spinal deformity. Treatment usually involves interventions like exercise, bracing, or if necessary, surgery. Often bracing is prescribed to stop curve progression so that surgery can be avoided. Traditional scoliosis braces are usually rigid devices, which displace the spine to the desired corrective position. With efficacies over 90%, these braces can be quite effective when worn enough. Unfortunately, the activities of daily living (ADL) for patients are reduced drastically when wearing a brace, and as a consequence compliance towards the braces is low. Since AIS develops in around 3% of all adolescents, of which approximately 10% has progressive curves that require treatment of some sort, the need for effective, comfortable bracing is high. To increase the ADL of patients, the current focus has been shifted towards designing a compliant scoliosis brace that can provide needed corrective forces and allow motion.

This thesis focusses on the evaluation of spinal motions through the design of a motion capture experiment. The goal of this work is to provide general knowledge about these spinal motions for clinicians, researchers and mechanism designers, such that they can make use of the provided analysis for the design of a new, compliant scoliosis brace. Parts of this analysis are implemented in a brace design quantification strategy, which can be used to facilitate such a brace design project. The key contribution of this master thesis is the characterization of spinal motions for specific vertebrae, to provide substantial kinematic data for the design of a compliant scoliosis brace.

1.1. Research Objectives

The research objectives of this thesis are defined as:

- To gather and analyze patient motion data of specific vertebrae during primary motions.
- To devise a structured strategy for the design of a brace design based on compliant shell mechanisms.

1.2. Thesis Outline

The main body of this thesis work consists of two parts, written as scientific papers. These papers focus on the two presented research objectives.

Before presenting these papers, first an introduction to the pathology is given and the main reason behind the project. This introduction is largely based on a literature review performed before commencing this graduation thesis. The focus of this literature review was to investigate and compare current brace designs and to investigate methods for spinal motions characterization. The literature review is depicted in Appendix A.

The first paper contains a thorough explanation on the set-up of a performed motion experiment with both scoliosis patients and healthy subjects. Used methods for gathering data and theories used in the analysis are explained. All relevant outcomes for both clinicians and mechanism designers from these motion studies are presented and discussed.

The second paper presents a function requirement quantification strategy that can be applied to design a compliant scoliosis brace. It presents collected information from different sources; outcomes of a validated brace are used in conjunction with outcomes from the motion studies in the first paper to translate the functional requirements into quantitative design specifications.

In the final part of this thesis recommendations for the future are given for both presented papers. In addition, a possible brace design is presented, showing a potential outcome of the current project as a whole.

2

Background

To understand the scope of this project, first an introduction to necessary anatomy is presented, after which the pathology itself is explained including the current treatment paradigm. This introduction serves as background knowledge and gives a more thorough understanding why the presented work contributes to the field of (scoliosis) research.

2.1. Anatomy

The spine is a complex, flexible structure consisting of 26 connected irregular vertebrae. Besides giving support to the entire trunk, connecting the skull with the pelvis, it also protects the spinal cord and serves as an attachment medium for the ribs of the ribcage. In early stages of life, the spine consists of 33 different vertebrae, of which nine eventually fuse to form two bony structures; the sacrum and the tiny coccyx. Figure 2.1 depicts the spinal column with all its vertebrae.

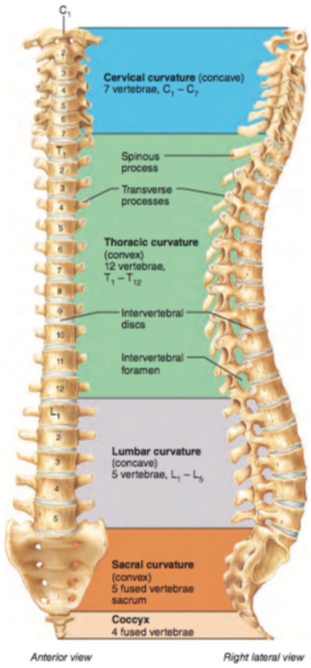


Figure 2.1: The spinal column, indication of the 5 spinal regions and the respective curvatures. Reproduced from [9]

Even though each vertebra differs in size, almost all vertebrae consist of the same general structure. The centrum is located at the centre of the vertebra, and is connected to a vertebral arch, which is protecting the spinal cord. Seven processes are connected to the vertebral arch, which form connection points for muscles and ligaments to move and stabilize the spinal column. The most present process is the spinous process, which is located posterior to the vertebrae. This spinous process is relatively easy to palpate underneath the skin. Figure 2.2 depicts the typical structural pattern of a vertebra.

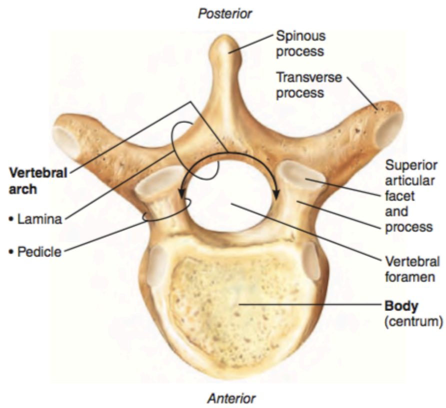


Figure 2.2: Typical vertebral structures, superior view of a thoracic vertebra. Only bone features are illustrated. Reproduced from [9]

Within the spine we can distinguish 5 different regions, from top to bottom: The cervical region (C1-C7), the thoracic region (T1-T12), and the lumbar region (L1-L5). The lumbar region is connected to the sacrum (S1-S5), which is connected to the pelvis. Inferior to the sacrum we have the tiny coccyx. These regions are depicted in Figure 2.1.

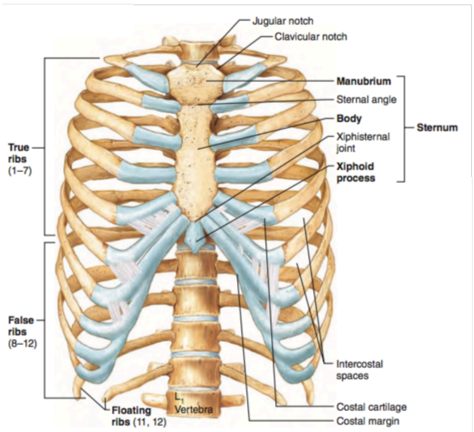


Figure 2.3: Skeleton of the thoracic cage, anterior view. Reproduced from [9]

Attached to the thoracic region, we find the ribs, which form the thoracic cage together with the thoracic vertebrae and the sternum. The thoracic cage is depicted in Figure 2.3. The rib cage consists of twelve pairs of ribs, which are all attached to the thoracic vertebrae posteriorly. Rib pairs 1-10 are attached to the posterior thoracic T1-T10 vertebrae, respectively, and have anterior attachments to the sternum (either directly or indirectly). Rib pairs 11 and 12 have no anterior attachments, and are therefore called floating ribs.

Each spinal region has its specific curvature when viewed laterally. These curvatures increase the flexibility of the spine, allowing it to function like a spring. [9]

2.2. Scoliosis

If an abnormal lateral curvature is present, this is referred to as scoliosis. Three different types of scoliosis can be distinguished: congenital scoliosis, neuromuscular scoliosis, and idiophatic scoliosis. [3] Scoliotic deformations can cause some serious health implications. On a physical level these implications might include severe deformations that may cause heart and lung problems. [1][23] Non-physical or psychological health implications might include struggling with self-image, an emotional pain. [3]

Two types of scoliosis curves can be distinguished; the S-shape and C-shape curve. As seen from the frontal or posterior side, these shapes indicate the typical spinal shapes we see in scoliosis. These two shapes are highlighted in Figure 2.4.

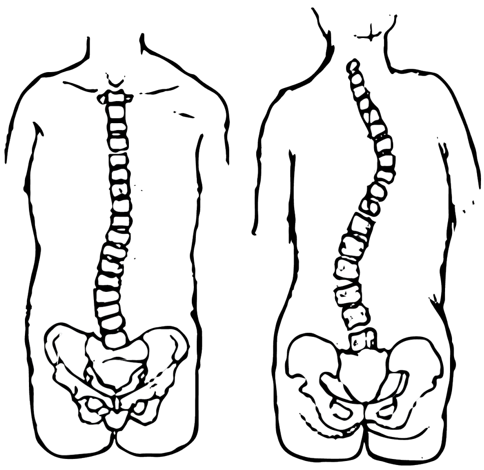


Figure 2.4: Different scoliosis curvatures. On the left a lumbar c-curve is shown. On the right a double major curve, or S-curve is shown. Edited image, original from [2]

Especially present during late childhood with unknown causes, Adolescent Idiopathic Scoliosis is the most common type of scoliosis, developing in around 3% of all adolescents. Approximately 10% of which needs to be treated, due to its progressing curves. [10] When the Cobb angle exceeds 10 degrees, the patient is diagnosed with Scoliosis.

Typically, the scoliotic curvature is characterized with the Cobb angle; the angle between the two most tilted vertebrae of the spine. [3] Figure 2.5 depicts a schematic overview of the Cobb angle measurement.

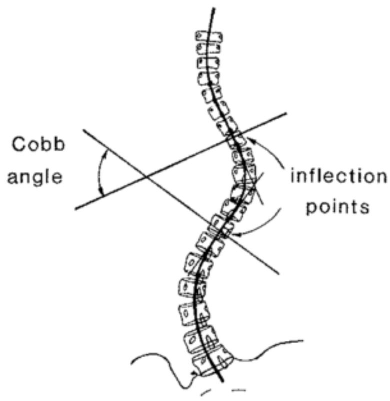


Figure 2.5: Measurement of the Cobb Angle, the angle between the two most tilted vertebrae. Reproduced from [20]

To overcome the possible problems scoliosis might bring, braces are prescribed when the Cobb angle measures between 25 and 40 degrees to prevent further progression of the spinal curvature. [15] If the Cobb angle exceeds 40 degrees usually surgery is performed, where the deformity is reduced by fusing the vertebrae together using metal rods and bolts, leading to a reduced spinal range of motion.[5]

2.3. Bracing

In-brace correction of the scoliotic spine is generally performed by applying external forces at the apex of the curve. These forces are applied by patient-specific scoliosis braces, which usually have to be worn up to 23 hours a day. [11] It has been shown that effective treatment depends on wear time [17][22], where effective treatment is often regarded a maximum curve progression ($< 5°$) at the end of growth. [14]

Typically, patients do not wear braces for the prescribed amount of time per day due to comfort and aesthetic reasons. Mostly rigid braces are used, which have a higher success rate compared to their flexible counterparts, but perform poorly in wearer compliance and comfort. Rigid braces also limit the patient's Range of Motion, which in its turn limits the patient's ability to perform several Activities of Daily Living (ADL). Furthermore, rigid braces are often bulky and patients have to wear oversized clothes to hide them. This can have psychological impacts on the patients and damage self-confidence.[12][16] Flexible braces perform better on these criteria, but show a lower success rate in terms of treatment. [13] Figure 2.6 shows three examples of brace designs.

(a) Boston Brace, reproduced from [8]

(b) TriAC Brace, reproduced from [21]

(c) SpineCor Brace, reproduced from [4]

Figure 2.6: Different types of braces

To maximize treatment efficiency, previous factors give rise to the development of a new hybrid brace solution. This brace solution aims to combine the effectiveness of rigid braces with the flexibility of the non-rigid braces, thus creating a brace which improves ADL while making sure the success rate is not decreased. Previous work by Nijssen and Ring showed possible brace designs using compliant shell mechanisms, creating a semi-rigid brace while simultaneously applying required correction forces. [13][16] Nijssen used a spatial mechanism approach to design scoliosis braces, allowing for the design of relatively thin shell elements to provide flexibility and transmit correction forces. Additionally, a currently unconventional two-fold force controlled strategy was used based on work by Nijenbanning [12], which made it very hard to validate the brace design, without taking into account the complex brace-tissue interaction. Furthermore, generalized motion data was not available, which is why the proof-of-concept brace design was focused on one specific subject.

Today, not much research has focused on spinal motion of scoliosis patients. Engsberg et al. measured the spinal range of motion pre- and post spinal fusion for the entire spine. [5] Skalli et al. used a similar experiment to investigate the contribution of the pelvis to the spinal motion before and after surgery, showing that most patients use the pelvis to compensate for reduced spinal motions both prior and after spinal fusion. [18] Solomito investigated the range of motion for three different spinal segments; the entire spine, the upper spine (C7 to T7) and the lower spine (T7 to S1), showing differences in motion profiles between healthy subjects and scoliosis patients. [19] More recently, Galvis et al. investigated the spinal mobility for different segments of the spine, showing an increased mobility for the scoliosis group compared to the control group. [6]

Since these researches have mostly been focusing on analysing spinal range of motion, the need for characterization of spinal motions from a mechanism design perspective remains for the design of a compliant scoliosis brace.

3

Spinal Motions

Paper: The spinal Range of Motion of Adolescent Scoliosis patients

This paper is to be submitted to Scoliosis and Spinal Disorders

CHARACTERIZATION OF SPINAL KINEMATICS IN ADOLESCENT SCOLIOSIS PATIENTS

Tim J. Dries
Department of BioMechanical Engineering
Delft University of Technology
The Netherlands

Charles J. Kim
Dep. of Mechanical Eng.
Bucknell University

ABSTRACT

Treatment possibilities for Adolescent Idiopathic Scoliosis (AIS) have been widely investigated over the last decades. Even though progress has been made in designing efficient treatment methods, not much is known about the spinal motions of AIS patients. In this work, spinal mobility is quantified and compared between scoliosis patients and healthy subjects using a primary motions experiment. In addition, bending strategies are evaluated between the groups and rotation axes are investigated. Together, this data can be applied when developing a new, compliant scoliosis brace.

BACKGROUND

Adolescent Idiopathic Scoliosis (AIS) is a three-dimensional deformity of the spine, which is characterized by a lateral curvature with a Cobb angle of more than 10 degrees and rotated vertebrae. [1] [2] Because spinal deformation can cause serious health implications, usually scoliosis braces are prescribed to stop the curvature from progressing. Current braces consist of rigid and flexible solutions. The rigid braces generally have a higher success rate, but perform poorly on comfort and the ability of the wearer to perform Activities of Daily Living (ADL). Flexible braces, on the other hand, generally perform better on these criteria but have been shown to be less effective in terms of treatment. [3] These facts give rise to the development of a new hybrid brace solution, which aims to combine the effectiveness of rigid braces with the flexibility of their flexible counterparts. Since ADL are highly influenced by the ability to move the spine, a better understanding of these spinal motions and the spinal mobility is required for the design of a semi-rigid, or compliant scoliosis brace. Observation of these topics could also give a better understanding of the effect of AIS on the mobility of the spine. Since not much research has been focused on the quantification of spinal mobility in conjunction with AIS, not much is known about the differences in spinal motion between scoliosis patients and healthy subjects. Only few studies have been performed with the aim of comparing the Range of Motions (RoM) of these two groups [4] [5], all of which used a very small group size or focused on differences between broad spinal regions. In this study, a method is proposed with which we are able to capture motion data for specific vertebrae. Using this method, the spinal mobility is measured during primary motion tasks; sagittal bend, lateral bend and axial rotation, or twist. The purpose of this experiment is to define specific contributions of different spinal sections to primary motion tasks and indicate possible differences in motion profiles between scoliosis patients and healthy subjects. It is hypothesized that the lower lumbar regions contribute a significant amount to the entire sagittal and lateral bend. Furthermore, spatial axes of rotation can be described for the captured vertebrae using screw theory, allowing for a rotational characterization of large part of the spine. Captured data from this experiment, such as the particular contributions to the total bend, rotation axis locations and possible generalization over groups can be put into practice for the design of a new, compliant scoliosis brace.

METHODS

Fifteen adolescent AIS patients between the age of 10 and 14 years old were included in this study. Recruitment of subjects took place during scoliosis clinic visits. The inclusion criteria consisted of a primary Cobb angle of at least 15 deg and no previous history of spinal surgery. The control group consisted of fifteen age- and gender matched subjects.

This study was approved by the Institutional Research Board at Bucknell University and by the Institutional Research Board of Geisinger Health Center. Written and informed consent and assent was obtained for all subjects ahead of each trial.

Motion trial design

Several researchers investigated spinal motions of scoliosis patients. The majority of this research focused on the ex-vivo calculation of Cobb angles or spinal range of motion before and after spinal fusion. Nijssen [3], Solomito [5] and Galvis et al. [4] investigated the relative motion ranges for different spinal segments. While Galvis et al. used an electromagnetic motion sensor system (Trakstar system, Ascension Technologies, Burlington, VT), this study uses an optoelectronic motion system, decreasing the possible influence of attached equipment on natural motions. The utilized system is a set-up consisting of 8 T10 series infra-red motion cameras (Vicon, Los Angeles, CA), acquiring data at 1000Hz (frames/sec).

Subjects in both groups were equipped with 6 retro-reflective marker tripods. The specific configuration of the tripods allows for gathering spatial motion data of one specific location, the attachment point on the spinous process. Each tripod consists of one retro-reflective marker attached to the center, with two retro-reflective markers connected at a 90 deg. angle with the centre marker at the intersection point. Tripods were constructed using fused deposition modelling, an example is depicted in Figure 1.

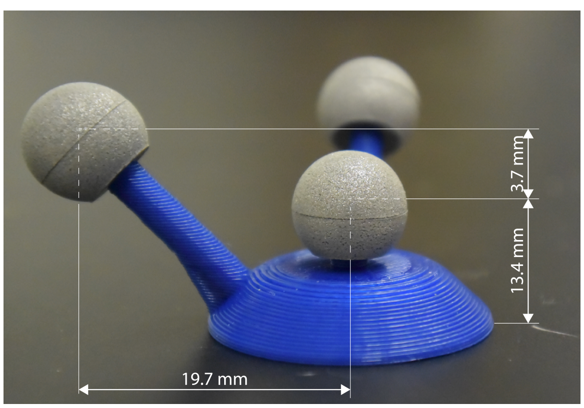


FIGURE 1: Marker tripod as used in this work

The tripods were placed on the T1, T4, T7, T10, L1 and L3 vertebrae. These locations were chosen due to the relative ease to locate (palpate) them in experimental setting and the potential to define different spinal regions from these locations. In addition, three markers were placed on the S1, and the left and right anterior superior iliac spine (LASIS and RASIS). The 3D-plane constructed by these markers serves as the base-frame for further analysis. The total marker set-up is depicted in Figure 2.

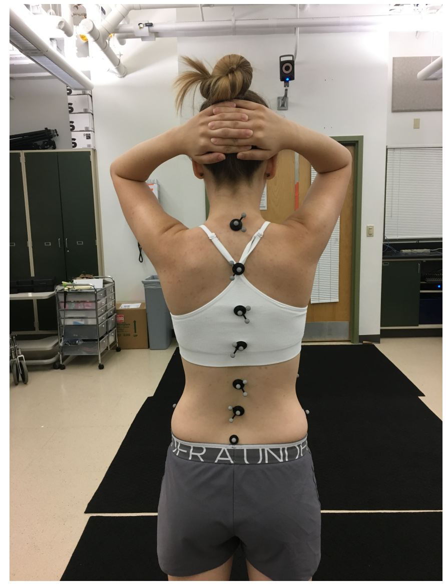


FIGURE 2: Marker set as used in this work

All subjects were instructed to perform 3 primary bends as far as comfortable; a sagittal bend, a lateral bend (to the left and right) and a coronal twist (to the left and right). For each bend, the bending tasks were demonstrated after which participants could practice before data collection started. In total, each subject was asked to perform the different motion tasks 15 times, which were captured in 5 consequent trials. Subjects were asked to keep their hands on the back of their head, as depicted in Figure 2, to preclude occlusion of markers. Demographic data of participants was collected at the time of participating in the motion trials.

Data processing

After data collection, data was analysed and processed in several steps. An overview of the data analysis is presented in Figure 3. Each analysis step will be explained in detail in a separate section.

Capture Data: Data was captured and refined using VICON Nexus software (Vicon, Los Angeles, CA) and possible gaps were filled using built-in gap-filling algorithms.

Filter: A custom-made MATLAB (Mathworks, Natick, MA) program was used to post-process the data. After capturing, data is filtered with a 4-th order Butterworth filter (cut-off frequency of 15 Hz) to remove possible noise from within the motion capture system and reduce skin motion artefact.

Force Rigid Bodies: Data is forced to be rigid to further reduce error build-up within the capturing system, ensuring exact marker set dimensions throughout the entire trial.

Frame Transformation: In this step, data is translated into the Pelvis frame, setting all motions relative to the Pelvis.
Frame Projection: After transformation, data is projected onto the planes-of-interest in order to isolate the in-plane motions for further analysis.
Screw Calculation: In this step, screw theory is used to calculate relative rotations, translations and the locations of spatial axes of rotation for each captured marker set. Screw theory describes any rigid body motion as a combination of a rotation about and a translation in the direction of the axis of rotation. [6] Using this theory, a motion can be subdivided into a primarily rotation, translation or a combination of the two. To find the screw locations (spatial axes of rotation) and corresponding angles of rotation about these screw vectors, a mathematical model is used, based on a model developed by Ring. [7]

In the end, data is clustered and plotted. The results are presented in the Results section.

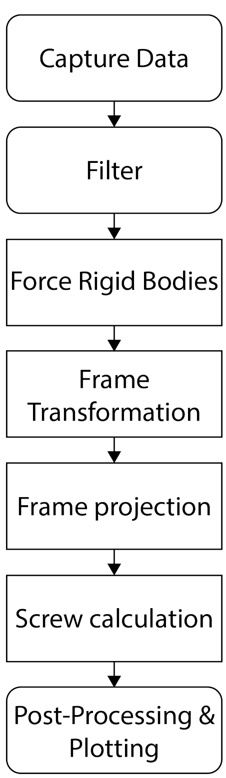


FIGURE 3: Process-tree of data analysis. Each box represents main function as utilized by Matlab program to evaluate captured motion data. Theory behind scripts in rectangular boxes is explained in further detail in the section below.

Force rigid bodies

While capturing the data during the trials, small error is introduced because of imaging errors within the VICON capturing system. These imaging errors are reduced by ensuring the dimensions of the pelvis frame are the same for each time frame, therefore forcing the captured Pelvis frame to be rigid.

Each marker set consists of three different markers A, B and C, where A is the central marker, B the top marker and C the bottom marker. For the pelvis, A_p is the single marker attached to the S1 vertebra, B_p is the marker attached to the right hip (RASIS) and C_p is the marker attached to the left hip (LASIS), where p denotes the relative marker set, in this case the Pelvis marker set. A simplified representation of the marker sets is shown in Figure 4.

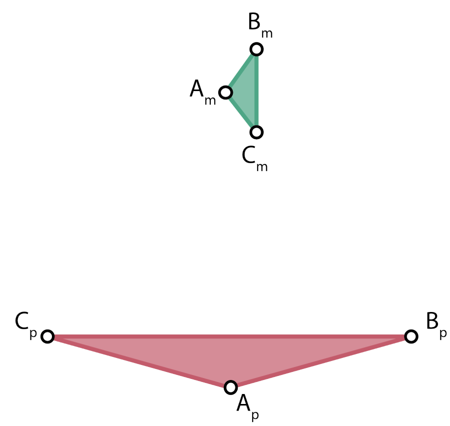


FIGURE 4: Simplified representation of basic marker sets. Marker set on pelvis (A_p on S1, B_p on RASIS,C_p LASIS) is shown, together with a possible configuration (A_m, B_m, C_m) of one of the marker tripods on T1, T4, T7, T10, L1 or L3 vertebrae.

To force the marker set to be rigid, the A_p marker is used as a base reference and the locations of the other two markers B_p and C_p are re-calculated, if necessary, for each time frame.

First, the magnitudes of the vectors between A_p, B_p and C_p are calculated and normalized for the first time frame ($t = 0$):

$$\hat{AB}_p(0) = \frac{(B_p(0) - A_p(0))}{|(B_p(0) - A_p(0))|} \tag{1}$$

$$\hat{AC}_p(0) = \frac{(C_p(0) - A_p(0))}{|(C_p(0) - A_p(0))|} \tag{2}$$

$$BC_p^{\hat{}}(0) = \frac{(C_p(0) - B_p(0))}{|(C_p(0) - B_p(0))|} \tag{3}$$

Next, the angle between these vectors is computed:

$$\theta_{BAC_p}(0) = \cos^{-1} \frac{A\hat{B}_p(0)^2 + A\hat{C}_p(0)^2 - B\hat{C}_p(0)^2}{2 * (A\hat{B}_p(0) \cdot A\hat{C}_p(0))} \tag{4}$$

The unit vectors for each time frame (i) are calculated similarly to Equation 1 and the orthogonal vector between them is computed by taking the cross product:

$$\vec{k(i)} = \vec{AB_p}(i) \times \vec{AC_p}(i) \tag{5}$$

The next step is using the Rodrigues Formula in order to calculate the new $A\hat{C}_p$ unit vector direction, by inserting the $A\hat{B}_p$ vector for each time frame and the angle between the vectors at the first time frame:

$$\begin{aligned} A\hat{C}_p{}'(i) = \; & A\hat{B}_p(i)\cos\theta_{BAC_p}(0) + (\vec{k} \times A\hat{B}_p(i))\sin\theta_{BAC_p}(0) \\ & + \vec{k}(\vec{k} \cdot A\hat{B}_p(i))(1 - \cos\theta_{BAC_p}(0)) \end{aligned} \tag{6}$$

Finally, the newly found unit vectors are multiplied with the magnitudes as found in Equation 1, 2 and 3 to produce the new (rigid) positions of the relative markers.

Translate marker data into Pelvis frame

To make sure that all the marker data is expressed in the pelvis frame, the pelvis frame is set as the global frame and all captured data is translated to this frame.

A vector between the origin and marker $A_p(0)$ on the Pelvis is calculated, where p denotes the marker location, (0) denotes the first time-frame and G denotes the global reference frame:

$$\vec{d} = {}^{G}A_p(0) - \begin{bmatrix} 0 \\ 0 \\ 0 \end{bmatrix} \tag{7}$$

This vector is subtracted from the captured data of all marker sets for each time frame (i), to shift the Pelvis onto the xy-plane:

$$\begin{aligned} {}^{1}A(i) &= {}^{G}A(i) - \vec{d} \\ {}^{1}B(i) &= {}^{G}B(i) - \vec{d} \\ {}^{1}C(i) &= {}^{G}C(i) - \vec{d} \end{aligned} \tag{8}$$

To transform the data, the angle between the global x- and z-components of ${}^{G}B_p$ is calculated using the four quadrant inverse tangent:

$$\phi_{B_p} = \tan^{-1}\left(\frac{{}^{1}B_p(1,3)}{{}^{1}B_p(1,1)}\right) \tag{9}$$

Using this angle the marker data is rotated about the y-axis into the xy-plane, by making use of the Euler rotation matrix:

$${}^{2}P = \begin{bmatrix} x' \\ y' \\ z' \end{bmatrix} = R_y(\phi_{B_p}) \cdot {}^{1}P = \begin{bmatrix} \cos(\phi) & 0 & -\sin(\phi) \\ 0 & 1 & 0 \\ \sin(\phi) & 0 & \cos(\phi) \end{bmatrix} \begin{bmatrix} x \\ y \\ z \end{bmatrix} \tag{10}$$

, where P represents the point in three-dimensional space:

$${}^{j}P = \begin{bmatrix} p(i,1) \\ p(i,2) \\ p(i,3) \end{bmatrix} \tag{11}$$

Consequently, data is rotated about the z-axis onto the x-axis. With the angle calculated:

$$\theta_{B_p} = \tan^{-1}\left(\frac{{}^{2}B_p(1,2)}{{}^{2}B_p(1,1)}\right) \tag{12}$$

And the rotation using Euler's rotation matrix about the z-axis:

$${}^{3}P = \begin{bmatrix} x'' \\ y'' \\ z'' \end{bmatrix} = R_z(\theta_{B_p}) \cdot {}^{2}P = \begin{bmatrix} \cos(\theta) & \sin(\theta) & 0 \\ -\sin(\theta) & \cos(\theta) & 0 \\ 0 & 0 & 1 \end{bmatrix} \begin{bmatrix} x' \\ y' \\ z' \end{bmatrix} \tag{13}$$

Finally, the data about is rotated about the x-axis into the xy-plane. Using the same method for the angle and Euler's rotation matrix:

$$\psi_{B_p} = \tan^{-1}\left(\frac{{}^{3}B_p(1,3)}{{}^{3}B_p(1,2)}\right) \tag{14}$$

And

$$^{P}P = \begin{bmatrix} x''' \\ y''' \\ z''' \end{bmatrix} = R_x(\psi_{B_p}) \cdot {}^{3}P = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\psi) & \sin(\psi) \\ 0 & -\sin(\psi) & \cos(\psi) \end{bmatrix} \begin{bmatrix} x'' \\ y'' \\ z'' \end{bmatrix} \quad (15)$$

Projection on plane of interest

After translating every data point into the Pelvis frame, the plane-of-interest is defined; the plane in which the locations of the screw rotational axes are to be calculated. For all three different motions, particular planes-of-interest can be distinguished. For the sagittal bend this is the sagittal plane, for the lateral bend the frontal plane and for the axial twist this is the transverse plane.

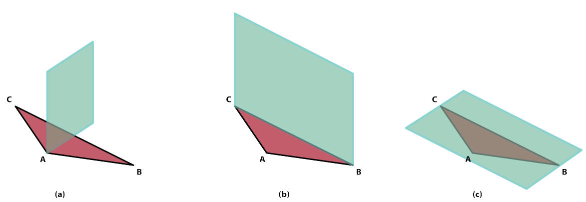


FIGURE 5: Planes of interest w.r.t. Pelvis frame for the three primary bends: a) Sagittal plane. b) Frontal plane. c) Transverse plane

As Figure 5 indicates, all the planes-of-interest are shown relative to the Pelvis. The orientations of the planes are calculated for each time frame by defining the normals to each of these planes:
First, the Z-normal to the Pelvis frame is calculated for each time-frame:

$$Z_{norm}(i) = ((A_p(i) - B_p(i)) \times (A_p(i) - C_p(i))) \quad (16)$$

Next, the Y-normal of the Pelvis is computed using the cross-product of the Z-normal from Equation 16 with the vector from C_p to B_p.

$$Y_{norm}(i) = (Z_{norm}(i) \times (B_p(i) - C_p(i))) \quad (17)$$

The X-normal is calculated in a similar manner, by taking the cross-product of the Z-normal from eq. 16 with the Y-normal from eq. 17:

$$X_{norm}(i) = (Z_{norm}(i) \times Y_{norm}(i)) \quad (18)$$

Using these normals, the marker data can be projected onto the relative planes-of-interest using vector projection:

$$\begin{aligned} &{}^{Y}A_m(i) = {}^{P}A_m(i) - \\ &\left(\left(\left({}^{P}A_m(i) - {}^{P}A_p(0)\right) \cdot Y_{norm}(i)\right) * Y_{norm}(i)\right) \end{aligned} \quad (19)$$

Calculate screws

A screw is defined by a direction vector $\vec{w}$, a point q on that vector, a pitch h and an angle of rotation γ. A screw without pitch depicts a pure rotation en its location represents a rotation axis. Screws are calculated in this work using the following strategy:

First, the direction unit vector of the screw is computed, by using the transformations of B and C. These transformations $\tilde{B}$ & $\tilde{C}$ are defined as:

$$^{Y}\tilde{B} = \frac{B_m(i) - A_m(i)}{|B_m(i) - A_m(i)|} - \frac{B_m(0) - A_m(0)}{|B_m(0) - A_m(0)|} \quad (20)$$

,

$$^{Y}\tilde{C} = \frac{C_m(i) - A_m(i)}{|C_m(i) - A_m(i)|} - \frac{C_m(0) - A_m(0)}{|C_m(0) - A_m(0)|} \quad (21)$$

The screw direction vector must be orthogonal to these vectors and so the cross-product is used to calculate vector, the axis of rotation $\vec{w}$:

$$\vec{w} = {}^{Y}\tilde{C} \times {}^{Y}\tilde{B} \quad (22)$$

Next, the angle of rotation is calculated by considering the movement of a point q about the rotation axis w. A graphic representation of this movement is shown in Figure 6.

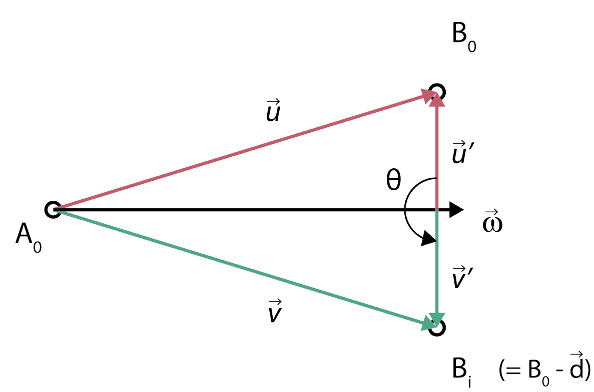


FIGURE 6: Graphic representation of movement of point about rotation axis $\vec{w}$ from B_0 to B_i

The normalized vector $\vec{u}$ form A_m to B_m of each marker set m is calculated for the first time-frame (0) and for all consequent time-frames (i):

$$\vec{u} = \frac{(B_m(0) - A_m(0))}{|B_m(0) - A_m(0)|} \tag{23}$$

$$\vec{v(i)} = \frac{(B_m(i) - A_m(i))}{|B_m(i) - A_m(i)|} \tag{24}$$

The perpendicular vectors from the rotation axis to $B_m(0)$ and to $B_m(i)$ are calculated:

$$\vec{u'} = \vec{AB_m}(0) - \vec{w}(i) \cdot \vec{AB_m}(0) \tag{25}$$

$$\vec{v'} = \vec{AB_m}(i) - \vec{w}(i) \cdot \vec{AB_m}(i) \tag{26}$$

Using the four quadrant inverse tangent, the angles of rotation relative to the first time-frame are found:

$$\gamma(i) = \tan^{-1} \frac{\left(\vec{u'} \times \vec{v'}\right) \cdot \vec{w}}{\vec{u'} \cdot \vec{v'}} \tag{27}$$

Now, the homogeneous transformation matrix T is constructed. This matrix consists of a displacement vector $\vec{d}$ and the Rotation matrix R, as defined from the Rodrigues formula:

$$R_{\hat{\omega}}(\gamma) = e^{\hat{\omega}\gamma} = I + \hat{\omega}\sin\gamma + \hat{\omega}^2(1 - \cos\gamma) \tag{28}$$

T is then constructed as follows:

$$T = \begin{bmatrix} R & \vec{d} \\ 0 & 1 \end{bmatrix} = \begin{bmatrix} e^{\hat{\omega}\gamma} & (I - e^{\hat{\omega}\gamma}) + h\gamma(\vec{\omega}) \\ 0 & 1 \end{bmatrix} \tag{29}$$

The pitch of the screw h is calculated, by using the relationship between the parallel component to the rotation axis w and the projection of d onto w:

$$h(i) = \frac{\vec{w}(i) \cdot \vec{d}}{\gamma(i)} \tag{30}$$

Lastly, the location of the screw $\vec{q}$ is found. $\vec{q}$ is defined as the point on the screw closest to the zero of the global frame and its derivation is shown in [7]:

$$q(i) = \frac{(\frac{N_d}{2} + (\vec{w}(i) \times N_d))}{2\tan(\frac{\gamma(i)}{2})} \tag{31}$$

, with N_d defined as the normal to $\vec{d}$:

$$N_d = \vec{d'} - (h(i) \cdot \gamma(i) \cdot \vec{w}(i)) \tag{32}$$

Clustering

When calculating screw rotation axis locations, results show varying axis positions including several outliers and axes that can be clustered or averaged. To calculate average screw locations, the calculated screws are clustered using DBSCAN method, a data-based density spatial clustering approach. [8] DBSCAN allows for finding clusters of data with only a minimum amount of input; eps, the minimum distance between two points and $minPts$, the minimum number of points that form a dense (clustered) region. [9] Using this method clusters are isolated and outliers can be eliminated. Figure 7 shows an example of clustering the generalized screw rotational axes of the L1 marker set, where the clustered screw axis is depicted in orange, the blue vectors show the screw axes that make up this cluster and the vectors in black indicate the outliers.

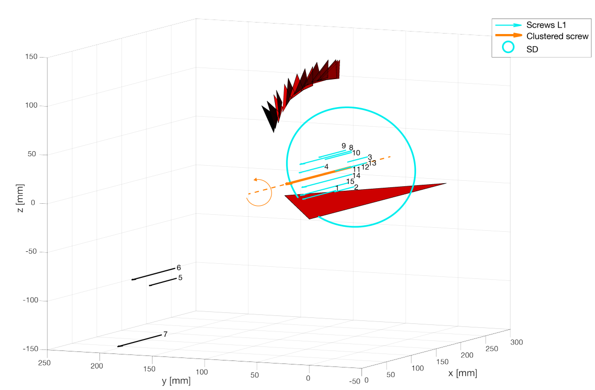


FIGURE 7: Averaged locations of screw rotational axes of L1 marker set for scoliosis group during sagittal bend, relative to pelvis. Clustered screw for all subjects is depicted in orange, with standard deviation from this clustered screw and screws that are clustered given in blue. Black vectors indicate outliers that are identified in the clustering algorithm and do not contribute to the clustered screw vector.

TABLE 1: Captured angles during three primary bends including confidence intervals for scoliosis group and control group.

		Rotation Angles [deg]					
Motion	**Group**	**T1**	**T4**	**T7**	**T10**	**L1**	**L3**
Sagittal Bend	**Scoliosis**	-	49.2 ± 5.2	49.5 ± 6.0	45.4 ± 4.6	36.1 ± 3.0	24.1 ± 5.0
	Control	-	45.9 ± 4.2	45.8 ± 5.2	44.9 ± 4.6	33.2 ± 3.3	20.8 ± 2.6
Lateral Bend	**Scoliosis**	55.5 ± 6.4	50.0 ± 4.4	40.3 ± 4.7	32.2 ± 3.0	13.8± 1.7	4.7 ± 0.8
	Control	74.2 ± 7.1	58.5 ± 5.4	52.0 ± 4.0	38.6 ± 4.0	21.0 ± 2.3	9.4 ± 1.7
Axial Twist	**Scoliosis**	-	44.8 ± 6.0	24.3 ± 3.6	28.5 ± 9.1	26.2 ± 4.6	21.1 ± 2.8
	Control	49.1 ± 5.2	25.0 ± 3.2	24.1 ± 4.4	32.3 ± 4.3	25.9 ± 9.6	

Motion Study Analysis

The motion study results are based on the three primary motions as performed by each subject. For each bend, the angles in the plane-of-interest (sagittal, frontal and transverse planes) are calculated and the relative contributions to the motion of each captured vertebra. In addition, average screw locations, describing the rotation axes of captured vertebrae are calculated and group means are plotted using custom made MATLAB programs. Rotation angles and screw rotational axes of captured vertebrae are calculated with relation to the pelvis for each subject motion trial and averaged out for each subject. For the sagittal bend and axial twist bending tasks, marker set T1 was disregarded, because of large inconsistencies and fluctuations in captured data, caused by marker occlusion or motion out of the camera field-of-view. Averaged subject data was gathered and combined for both groups. Statistical tests were used to evaluate possible differences in bend contributions between the scoliosis group and the control group. Data outliers were identified using statistical z-scores and where removed when the empirical rule ($> 3 * SD$) was met. Since group sizes were relatively small and data was not normally distributed, Wilcoxon-Mann-Whitney tests were used to determine statistical differences. [10] For the statistical tests, a significance level of $\alpha = 0.05$ was used, meaning two-tailed values $p < 0.05$ were considered statistically significant. All statistical analyses were performed using IBM SPSS Statistics, version 24 (IBM Corp., Armonk, N.Y., USA).

RESULTS

Subject Demographics

In total 30 subjects were used in this study; fifteen subjects in the study group and fifteen subjects in the control group. Subjects in the scoliosis group were diagnosed with scoliosis and undergoing treatment. Participants in the control group all had no diagnosis of scoliosis or other spinal disorders. All participants were female. The mean age, height and weight for both groups is depicted in Table 2.

TABLE 2: Mean demographics of participants in both groups

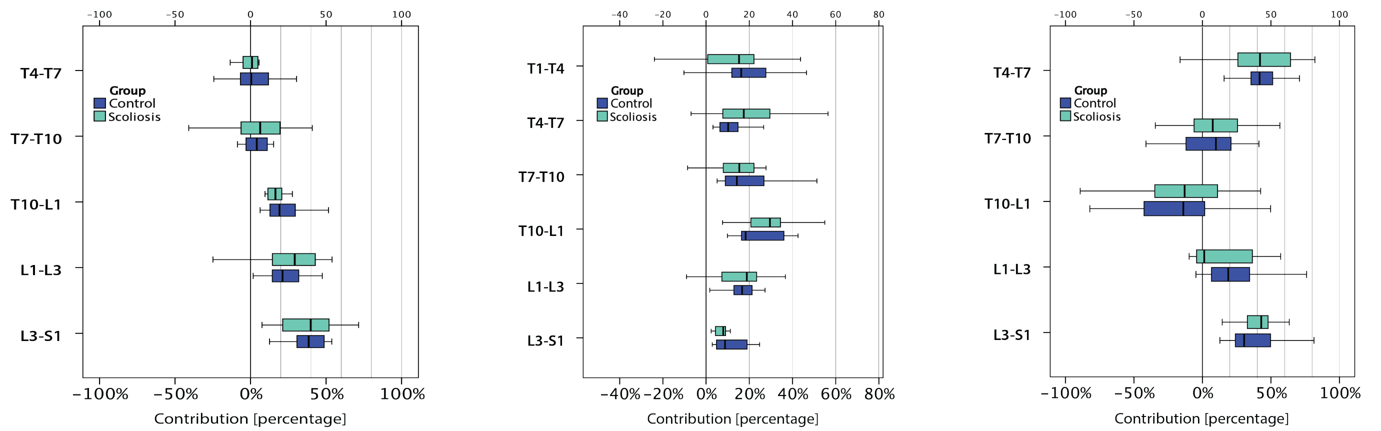
	Control Group	**Scoliosis Group**
Age [years]	12.0 ± 1.4	12.6 ± 1.9
Length [meters]	1.57 ± 0.07	1.58 ± 0.10
Weight [kg]	48.3 ± 12.7	49.1 ± 9.3

Bend Strategies

Rotation angle data for the three primary bends as collected for all 15 subjects of each group is depicted in Table 1. Results are calculated from trials where all data was recorded correctly.
Figure 8 depicts the relative contribution percentages of captured spinal segments for both the control group and the scoliosis group for the sagittal bend.

Similar bending strategies are apparent during sagittal bending for both groups. The scoliosis group shows a slightly larger contribution of the upper lumbar region (L1-L3) compared to the control group(Mdn =24.8%, Mdn = 21.8%). The low thoracic region of the control group (T10-L1) contributes more than the same region of the scoliosis group (Mdn =22.1%, Mdn =15.2%). On average, the lower lumbar region (L3-T10) contributes for 81.2% to the sagittal bend for the scoliosis group and for 87.3% for the control group. No significant differences are found when comparing the contributions of spinal sections to the sagittal bends between both groups, however both L1-L3 and T10-L1 sections show a small to medium-sized effect ($r = 0.14$ in both cases).

During lateral bending (Figure 8b), the scoliosis group shows significantly smaller rotation angles for both the lower



(a) Contributions of S1-T4 to sagittal bend. (b) Contributions of S1-T1 to lateral bend. (c) Contributions of S1-T4 to axial twist.

FIGURE 8: Comparison of averaged contributions of captured spinal segments of the scoliosis group to the averaged contributions of the same captured spinal segments of the control group during primary motion tasks.

lumbar L1 and L3 vertebra compared to the control group (p = 0.020, p = 0.019) and trends towards smaller rotation angles of the T1 and T7 vertebra (p = 0.06, p = 0.07).

The contribution of the T1-T4 section of the control group does not differ significantly from the scoliosis group (Mdn =17.6% ±15.5, Mdn =15.2% ±26.0, r = 0.20). A larger effect is seen for the T4-T7 section (Mdn =9.0% ±16.9, Mdn =17.4% ±23.9, r = 0.29). The T10-L1 section also shows (non-significant) differences in bend contributions to the lateral bend with a small to medium-sized effect (Mdn =19.0% ±13.3, Mdn =29.5% ±15.2, r = 0.24). On average, the lower lumbar region (T10-S1) contributes for 55.7% to the lateral bend in both groups.

Mobility in the axial twist tasks show different contributions of captured spinal segments to the total motion, as shown in Figure 8c. Both groups show a similar motion profile when performing axial twists, where we see the biggest contributions to the twist for the control group and scoliosis group from S1-L3 (39.2%, 40.8%) and from T7-T4 (43.0%, 38.8%). When comparing the groups, small differences are found between the control group and scoliosis group for T10-L1 (Mdn =-14.0% ±38.8, Mdn =-3.5% ±43.1, r = 0.18) and L1-L3 (Mdn =18.8% ±33.4, Mdn =8.0% ±30.5, r = 0.17), however these results are not significant.

Screw Locations

For each primary bend, screw locations for all subjects within each group were calculated. Figure 9 shows the screw locations of L1 vertebra for all subjects of the scoliosis group during the sagittal bend. The screw locations are shown for each 20 ms of the total bend, including standard deviations (horizontal lines).

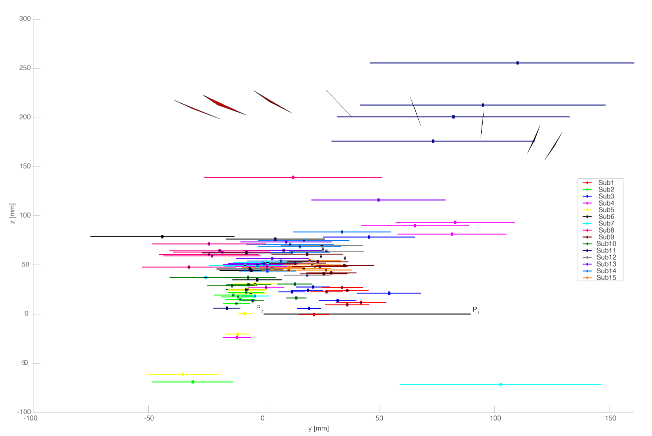
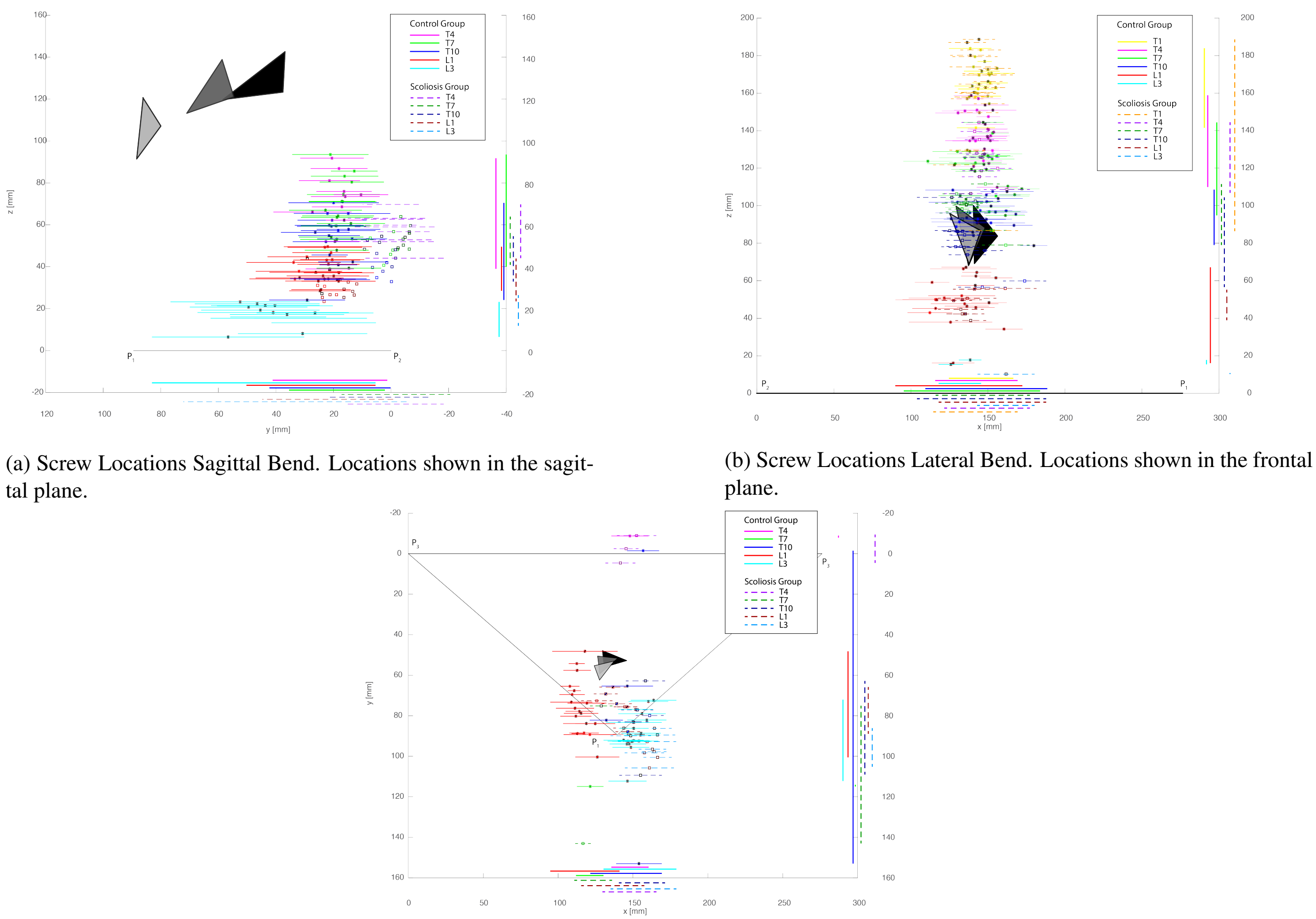


FIGURE 9: Screw Locations of L1 vertebra for 15 subjects of the scoliosis group. Each color represents a subject of the group. Standard deviations are indicated with horizontal lines. All screws were calculated relative to the pelvis frame, which is indicated in black. The motion of the actual marker set on L1 is shown by the red triangular shapes.

The simplified screw locations were clustered for both groups. Averaged locations for all three primary bends are shown in Figure 10.



(a) Screw Locations Sagittal Bend. Locations shown in the sagittal plane.

(b) Screw Locations Lateral Bend. Locations shown in the frontal plane.

(c) Screw Locations Axial Twist. Locations shown in the transverse plane.

FIGURE 10: Generalized screw locations for scoliosis group and control group of all marker sets during sagittal bend (Figure 10a), lateral bend (Figure 10b), and axial twist (Figure 10c). Locations are shown in planes-of-interest, including motions of T10 marker set. Solid lines represent control group, dashed lines represent scoliosis group. Relative spread is shown for each marker set of both groups on the horizontal and vertical axes.

For the sagittal bend, in general a larger spread of screw locations is seen for the control group, compared to the scoliosis group in z-direction. Additionally, the rotation axes of the control group are located more anteriorly, compared to those of the scoliosis group, except for the T4 marker set. The average location differences are depicted in Table 3. It is shown that on average, the screw locations of the control group are located posteriorly to those of the scoliosis group. These differences range from 3.55*mm* for L3 to 12.40*mm* for T4.

For the lateral bend, the averaged screw locations of the control group are located more distant from the pelvis, compared to the screw locations of the scoliosis group (proximal locations differ $5.72-21.67mm$). Differences in lateral positions are smaller, except for the L3 vertebra ($d=-25.94\pm 18.34mm$).

In the axial twist motion, screw locations of the scoliosis group are located more to the left lateral side of the body, compared to the locations of the control group ($-22.86-0.38mm$). For the y-direction, T4 and L1 are located more posteriorly for the control group than the scoliosis group ($d=40.06mm$, $4.25mm$), while T7, T10 and L3 are located more anteriorly

$(d = -59.9mm, 10.98mm, -4.39mm)$

DISCUSSION

The goal of this research was to analyse the spinal mobility of both scoliosis patients and healthy subjects, and to determine the relative screw motions of different vertebrae during three primary bends. It was hypothesized that both subject groups use their lower lumbar region most for sagittal bending and lateral bending, while the upper thoracic region of the spine is used more during axial twisting. This hypothesis was supported by the averaged motion data and bend contributions for both groups.

Marker Set-up

The marker set-up presented in this paper has benefits compared to existing methods. While most set-ups consist of single markers attached to bony landmarks [5], this paper introduces the use of marker tripods, which allows for calculation of rotation angles and screw rotation axes of the attachment points using screw theory. The presented method is highly reliable on retro-reflective markers, which have been placed manually by palpating the vertebrae. Since these markers are mainly attached to the skin, adipose and muscle tissue below the skin can cause the markers to move differently compared to the actual motion of the vertebra. This motion artefact was decreased by applying a Butterworth filter.

In addition, some markers might be disturbed by the movement of fabrics between the patient body and the marker tripods. These movements are most likely unrelated to the actual motion of the vertebrae underneath.

Data processing

In this research, subjects were asked to perform motion trials in a modified lab environment equipped with 8 infra-red motion cameras. Even though the pilot study was designed to capture all attached markers during the entire bend trajectories, markers have been occluded from the camera system because of motions out of the field of view. The authors have tried to eliminate these occlusions by monitoring the trials and eliminating possible error-some trials from the analysis. Additionally, data was refined using Vicon Nexus gap-filling algorithms, however these algorithms are not able to fill missing marker data at the beginning or end of trials. [11] This led to removal of trials, which might have influenced the power of the performed analysis.

Bend Contributions

The used methods in this paper present a method to analyse differences in bend contributions between the control group and scoliosis group. Relatively large deviations were found when calculating bend angles and contribution percentages. These deviations are possibly introduced by using the Euler method for rotation into the Pelvis frame. Efforts were made to remove these errors using filtering and by manually selecting only trials that did not contain these fluctuations. During lateral bending, significant differences in RoM were found for the lower lumbar vertebrae; scoliosis patients participating in this study showed significantly smaller rotation angles for L1 and L3 than the subjects in the control group. Research performed by Solomito [5] also showed larger lower spine (T7-S1) angles for the control group ($31^\circ \pm 5$), compared to the scoliosis group ($28^\circ \pm 8$). Similar research on spinal mobility performed by Galvis et al. [4] showed a higher mobility for the L1-L3 section, when comparing the lateral bend angles of scoliosis subjects with those of the control group. In this research, we saw a higher contribution to the lateral bend in this region for the control group. These differences may be explained by subject demographics; Galvis et al. included only AIS participants with right thoracic curves, while in this research scoliosis subjects with varying curvatures participated. Furthermore both Solomito and Galvis instructed participants to bend as far as possible, while in this research subjects were asked to bend as far as comfortable.

In-plane motions

Each of the bending tasks was designed and explained as a primary motion in one plane. During the analysis, considerable out-of-plane motions were observed, indicating a much more complex use of the spine when performing motions then expected. In this research the in-plane motion was isolated and analysed, as it is believed these motions are more relevant when using the data to design compliant scoliosis braces. Since it can be argued that spinal deformations are usually three dimensionally shaped, further research should investigate the spinal mobility and screw locations in 3D.

Screw Analysis

In this paper a new method is presented, which allows for analyzing screw axis locations for spinal motions. The results obtained by mean of this screw characterization from a primary bend experiment show a high variety of screw axis locations between subjects. As the first of its kind, it is not known what is causing this variance. It could be argued that different scoliotic characteristics could provide different results. As this study did not control for spinal curvature, curve severity or Risser grade, it is not clear what influence different scoliotic characteristics have on presented results. Even though other research showed that differences in Cobb angles were found to not influence the spinal

TABLE 3: Differences between screw locations of control group and scoliosis group for all captured marker sets during primary bends

		Control Group-Scoliosis Group					
		T1	**T4**	**T7**	**T10**	**L1**	**L3**
Sagittal Bend	y	-	-19.72 ± 13.95	21.78 ± 15.40	19.08 ± 13.50	4.58 ± 3.24	14.70 ± 10.39
	z	-	12.40 ± 8.77	11.06 ± 7.82	10.58 ± 7.48	9.71 ± 6.86	3.55 ± 2.51
Lateral Bend	x	1.42 ± 1.01	-0.19 ± 0.13	7.57 ± 5.35	14.49 ± 10.24	3.78 ± 2.67	-25.94 ± 18.34
	z	21.67 ± 15.33	20.20 ± 14.29	20.86 ± 14.75	16.23 ± 11.48	7.29 ± 5.16	5.72 ± 4.04
Axial Twist	x	-	-1.38 ± 0.98	-5.06 ± 3.58	0.38 ± 0.27	-22.86 ± 16.17	-8.45 ± 5.98
	y	-	40.06 ± 28.33	-59.87 ± 42.34	-10.98 ± 7.76	4.25 ± 3.01	-4.39 ± 3.10

mobility [12], it is likely that the spinal bio-mechanics are influenced by differences in scoliotic deformations, thus different screw trajectories are to be expected. Since the study was initially focused on determining differences between scoliosis patients and healthy subjects, screw data was averaged for the entire group using density based algorithms. The presented results give an indication how this theory can be used to analyse rotation axes of specific vertebra, future developments should consider patient specific spinal curvatures and control for curve severity and skeletal maturity to validate whether generalization of screw axes is possible.

CONCLUSION

When comparing the bending strategies of both healthy subjects and scoliosis patients for primary bends, no significant differences were found. When bending sagitally the lower lumbar region contributed up to 81.2% for the scoliosis group, while the same section provided 55.7% of the lateral bend. The thoracic region is used more when performing axial twists. During lateral bending tasks, healthy subjects showed a higher RoM, with larger contributions of L1 and L3 to the total bend. Screw locations differed widely across the two groups, especially during the sagittal bending and axial twisting tasks. The current study design did not allow for clear generalization of screw motion data. Further research should explore differences in screw motion trajectories for different types of scoliosis. Furthermore, research should be pursued to explore possible screw ranges in three dimensions and to translate these into design specifications for compliant scoliosis brace designs.

ACKNOWLEDGMENT

The authors would like to acknowledge NSF No. IIS - 1527133 as well as STW (HTSM-2012 12814: ShellMech) for the financial support of this project. The writers would also like to thank dr. Mark Seeley of Geisinger Health Center for his collaboration throughout this project.

REFERENCES

[1] Deacon, P., Flood, B. M., and Dickson, R. a., 1984. “Idiopathic scoliosis in three dimensions. A radiographic and morphometric analysis.”. *The Journal of bone and joint surgery. British volume,* ***66***(4), pp. 509–512.

[2] Weinstein, S. L., Dolan, L. A., Wright, J. G., and Dobbs, M. B., 2013. “Effects of bracing in adolescents with idiopathic scoliosis.”. *The New England journal of medicine,* ***369***(16), pp. 1512–21.

[3] Nijssen, J. P. A., 2017. “Design and Analysis of a Shell Mechanism Based Two-fold Force Controlled Scoliosis Brace”. Master’s thesis, Delft University of Technology.

[4] Galvis, S., Burton, D., Barnds, B., Anderson, J., Schwend, R., Price, N., Wilson, S., and Friis, E., 2016. “The effect of scoliotic deformity on spine kinematics in adolescents”. *Scoliosis and Spinal Disorders,* ***11***(1), p. 42.

[5] Solomito, M., 2011. “The Use of Motion Analysis Technology as an Alternative Means of Assessing Spinal Deformity in Patients with Adolescent Idiopathic Scoliosis”. Master’s thesis, Univeristy of Connecticut - Storrs.

[6] Ball, R. S., 1876. “The theory of screws: A study in the dynamics of a rigid body”. *Mathematische Annalen,* ***9***(4), pp. 541–553.

[7] Ring, J. B., Kim, C. J., Evans, J., and Beal, C., 2017. “A Passive Brace for the Treatment fo Scoliosis Utilizing Compliant Mechanisms”. Master’s thesis, Bucknell University, Lewisburg PA.

[8] Daszykowski, M., Walczak, B., and Massart, D. L., 2001. “Looking for natural patterns in data. Part 1. Density-based approach”. *Chemometrics and Intelligent Laboratory Sys-*

tems, ***56***(2), pp. 83–92.

[9] Ester, M., Kriegel, H. P., Sander, J., and Xu, X., 1996. "A Density-Based Algorithm for Discovering Clusters in Large Spatial Databases with Noise". *Proceedings of the 2nd International Conference on Knowledge Discovery and Data Mining*, pp. 226–231.

[10] Field, A., 2009. *Discovering Statistics Using SPSS.*, Vol. 58.

[11] Federolf, P. A., 2013. "A novel approach to solve the "missing marker problem" in marker-based motion analysis that exploits the segment coordination patterns in multi-limb motion". *PLoS ONE,* ***8***(10).

[12] Hresko, M. T., Mesiha, M., Richards, K., and Zurakowski, D., 2006. "A comparison of methods for measuring spinal motion in female patients with adolescent idiopathic scoliosis.". *Journal of pediatric orthopedics,* ***26***(6), pp. 758–63.

4

Quantification Strategy

Paper: Functional Requirement Quantification Strategy of a Compliant Scoliosis Brace

This paper is written as part of the design of a compliant scoliosis brace. After the actual fabrication, validation and testing of a concept brace, this paper will be submitted the ASME Mechanisms and Robotics Conference (MR).

FUNCTIONAL REQUIREMENT QUANTIFICATION STRATEGY OF A COMPLIANT SCOLIOSIS BRACE

Tim J. Dries
Dep. of BioMechanical Eng.
Delft University of Technology
The Netherlands

Joost R. Leemans
Dep. of Precision and Microsystems Eng.
Delft University of Technology
The Netherlands

Charles J. Kim
Dep. of Mechanical Eng.
Bucknell University
Lewisburg, PA. USA

ABSTRACT

3% of adolescents suffer from Scoliosis, a spinal deformation of which 10% needs to be treated. Bracing is a common treatment to prevent curve progression. Current scoliosis brace designs are usually rigid or flexible, which either score high in terms of success rate of treatment or in comfort and patient compliance. Attempts have been made to overcome these problems using a compliant brace design. Previous work has shown the ability of spatial mechanisms to combine correction with flexibility. In this work, the functional requirements that lead to correct treatment and allow motion are quantified by deducing validated brace characteristics using BraceSim and combining this with patient motion data. A strategy is proposed, reducing the complexity of the bio-mechanical problem into an isolated mechanical problem. This strategy provides a structure and serves as a structural process-tree to enhance the design of a new, compliant scoliosis brace.

INTRODUCTION

Adolescent Idiopathic Scoliosis (AIS) can be defined as a three-dimensional deformity of the spine, which is characterized by a lateral curvature with a Cobb angle of more than 10 degrees and rotated vertebrae. [1] [2] AIS develops in around 3% of all adolescents, of which approximately 10% has progressive curves that require treatment of some sort. [3]

Treatment possibilities for AIS include surgery, muscle stimulation and training and bracing, depending on the severity of the deformation. Severity of the deformation is measured by the Cobb angle, which is the angle between the two most tilted vertebrae of the spine. [4] When the Cobb angle exceeds 25 degrees, braces are prescribed to prevent further progression of the scoliotic deformation. [5] If the Cobb angle exceeds 40 degrees, usually surgery is performed in order to reduce the deformity by fusing vertebrae together using metal rods and bolts.

Among current state of bracing there are both rigid and flexible solutions. The rigid braces generally have a higher success rate, but limit the user in their Range of Motion (RoM), thus obstructing the wearer to perform their Activities of Daily Living (ADL). Flexible braces, on the other hand, generally perform better on these criteria but have been shown to generally have a lower success rate in terms of treatment. [6]

In the current project, the research objective is to design a scoliosis brace that corrects the spinal deformation, while allowing for motion during daily activities. Such a brace should be able to provide the high correction rates of rigid braces, while simultaneously using the non-rigidness of flexible braces to allow the patient to move and bend their torso.

Attempts have been made to develop semi-rigid or compliant braces. These promising brace designs show the unique abilities of shell mechanisms being compliant while simultaneously transmitting forces [6] [7].

Nijssen [6] first introduced a spatial mechanism approach to design scoliosis braces which allow motion and facilitate a force based correction utilizing compliant shell mechanisms. By introducing a new type synthesis approach along with an unconventional force based correction strategy the final result of Nijssen was difficult to validate [8], since the complex brace-tissue interaction was not taken into account [6]. Furthermore, no generalized motion data was available, which is why these proof-of-concept braces were focused on specific subjects.

The design strategy presented in this work, is based on the conventional displacement based control strategy while adding compliance using compliant shell elements. Compliant shell mechanisms are highly suitable for the application of human-assistive devices, such as the compliant scoliosis brace. Their monolithic lightweight thin-walled nature makes them ideal to clean, wear and conceal under clothes. Furthermore, the human spine does not have clear distinctive constraint and free motion directions, similar to compliant shell mechanisms.

The conventional displacement based control strategy improves the ability to validate a possible brace design, since the correction efficiency can be determined using BraceSim [9]. BraceSim is a validated software tool to simulate the effectiveness of a scoliosis brace prior to manufacture, and is developed by our collaborators from the École Polytechnique de Montréal.

The presented strategy reduces the complexity of a bio-mechanical compliant brace design problem into an isolated mechanical problem. Utilizing BraceSim, the main functional requirements of a compliant scoliosis brace are quantified into validatable kinematic design specifications. The kinematic design specifications include desired non-linear behaviour of compliant shell mechanisms. The proposed strategy contributes to the overall design process of designing a compliant scoliosis brace.

Case selection

The strategy presented in this paper, is explained on the basis of patient data provided by our collaborators from the École Polytechnique de Montréal. In this paper this analyzed patient is referred to as the correction study patient. The proposed strategy requires motion data of this patient, however the motion data of the correction study patient was not available. Therefore a motion study was performed on a different patient with similar scoliotic characteristics. This patient was selected by an experienced orthopaedic by comparing the X-rays of the braced patient (correction study patient) with the X-rays of all patients participating in a spinal motion characterization study performed by Dries [10]. From here on this patient is referred to as the motion study patient. Table 1 provides a comparison of the relevant data between the correction study patient and the motion study patient, this includes the spinal length and the Cobb angle.

TABLE 1. COMPARISON BETWEEN CORRECTION STUDY PATIENT AND THE MOTION STUDY PATIENT

	Length T4-S1	**Cobb angle**
Correction study patient	259.7 *mm*	25°
Motion study patient	302.5 *mm*	20°

Since the motion study patient and correction study patient have different body lengths, likely the motion characteristics differ between these subjects.

In this work, it is assumed that both patients have similar bend characteristics and that the captured spatial axes of rotation can be scaled by 0.86 to match the correction study patient's body length.

Since the Cobb angle differs by 20%, further research should be pursued to investigate motion characteristics for patients with different Scoliotic curvatures and Cobb angles, as discussed by Dries [10].

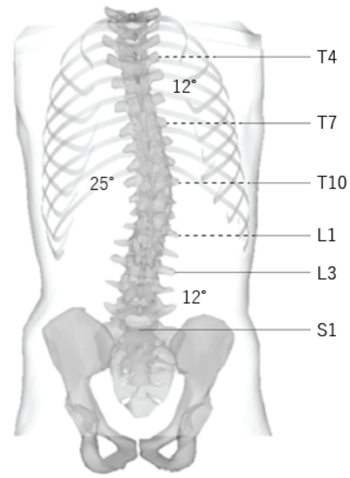


FIGURE 1. PATIENTS SPINE POSTERIOR VIEW

The correction study patient has the minimum Cobb angle for which braces are prescribed. [5] The spinal deformation of this patient is a so-called C-shaped curvature, having just one apex and consequently one focus area for correction. Patient information containing a finite element model (FEM) of the correction study patient's torso and spine was gathered using surface topography and bi-planar radio-graphs as described in [9]. A visualization of the spine, including the deformation angles, of the correction study patient is depicted in Figure 1.

QUANTIFICATION OUTLINE

This section presents the outline of the proposed strategy to quantify main functional requirements into validatable design specifications of a compliant scoliosis brace. The strategy steps are briefly introduced in this section and shown in Figure 2. Each strategy step will be explained in detail in a separate section and illustrated on the basis of the correction study patient.

Step I - Functional requirements: From the design objective follow the main qualitative functional requirements, which include the facilitation of correction and motion. The key component of the approach is to reduce the complex bio-mechanical functional requirements into isolated mechanical design specifications, by utilizing BraceSim.

Step II_A - Correction analysis: The correction strategy is based on the modification of a validated conventional scoliosis brace, designed by a certified orthotist using BraceSim. This validated conventional scoliosis brace is referred to as the benchmark brace. The analysis of the benchmark brace determines the correctional strategy.

Step II_B - Motion analysis: The motion characteristics determine the desired compliance of the to be designed brace.

Step III - Segment division: Based on correction and motion analysis, we determine which segment of the brace can be compliant and which segments need to be stiff.

Step IV - Isolated segment analysis: A modified benchmark brace is introduced that results in a similar correction as the original benchmark brace, obtained in BraseSim. In the modified benchmark brace the, to be designed, compliant segment is temporarily modelled as a stiff segment. The temporary stiff segment connects the permanent stiff segments. The temporary stiff segment is isolated to determine the required transmitted loads and deformation using BraceSim.

Step V - Design specifications: The design specifications follow from the temporary stiff segment and the motion characteristics. The stiff segment can be substituted by a compliant segment that can transmit the required loads but is compliant in desired directions.

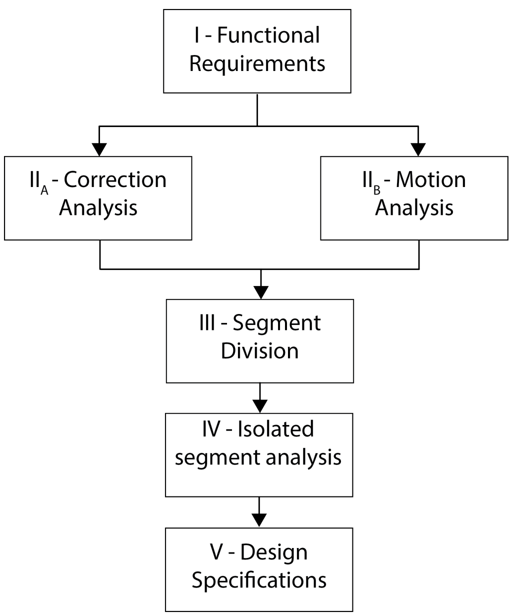


FIGURE 2. QUANTIFICATION STRATEGY DIAGRAM

I - FUNCTIONAL REQUIREMENTS

The main objective is to design a scoliosis brace that corrects the spinal deformations and allows for motion during activities of daily living.

From the design objective two seemingly contradicting main functional requirements can be extracted, given as:

- *Facilitate correction* to reduce the deformation of the spine.
- *Facilitate motion* necessary for activities of daily living.

Additional functional requirements are less crucial to present the introduced strategy and are therefore not addressed

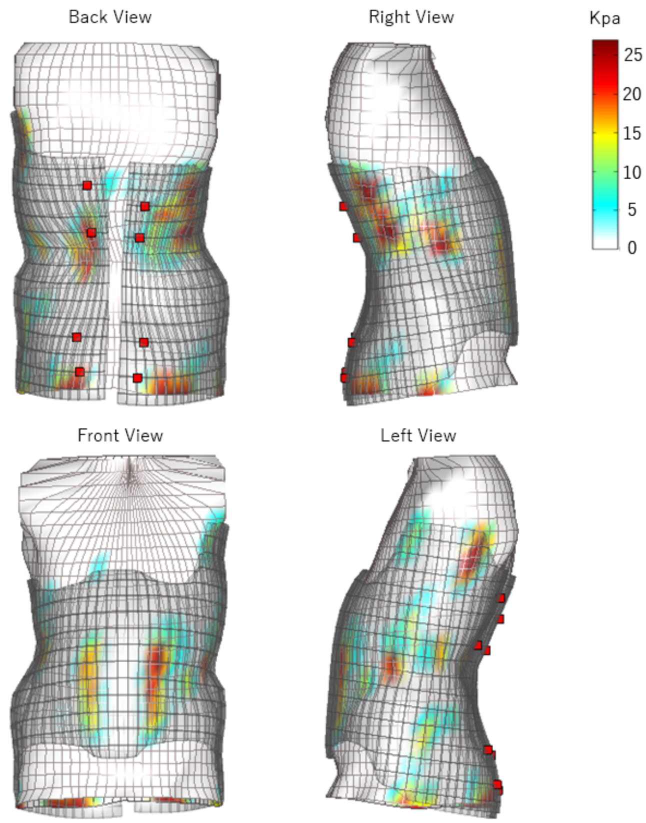


FIGURE 3. VALIDATED BRACE INCLUDING PRESSURES

in this paper. The main functional requirements are qualitative and should be translated into quantitative design specifications. The key contribution of this paper is the strategy that enables this quantification using the steps illustrated in Figure 2.

II$_A$ - CORRECTION ANALYSIS

The correction strategy should accomplish the functional requirement "facilitate correction". The correction strategy is based on a brace designed by a certified orthotist and validated using BraceSim, which is defined as the benchmark brace. The correction of the spinal deformation and the pressures resulting from the benchmark in-brace model serve as correction efficiency benchmark for the design specifications of a new compliant brace design.

The validated brace model and the in-brace results of the correction study patient are shown in Figures 3 and 4.

As can be observed by comparing Figures 1 and 4 the Cobb angle is reduced by 11 degrees under influence of the benchmark brace. We consider this as successful treatment, since usually brace treatment is regarded as being successful when curve progression is less than 5°. [11]

The pressure maps in Figures 3 and 4 show highest pressures

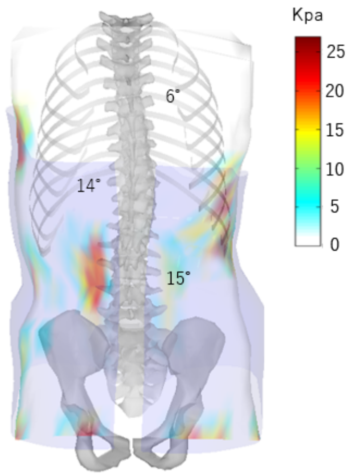


FIGURE 4. SIMULATED CORRECTION AND PRESSURE RESULTS OF BENCHMARK BRACE

of ±25 Kpa near the apex of the spinal curve, this is the region where most correction is needed. Other higher-pressure regions occur at the opposite side of this apex, on the left and right bottom part of the brace, and in the frontal region on the abdomen. The latter of which is regarded unnecessary and should be minimized as it provides a load orthogonal to the desired correction direction. The pressure maps show that transmitting correctional loads to the spine is done through predominantly hard-tissue regions in the torso. Loads applied on the primarily soft-tissue region are distributed via organs over larger regions of the spine. These larger load distributions result in less correction control of the spine. Furthermore the pressure on the soft-tissue region and inferior organs can be very uncomfortable and should therefore be minimized.

The primarily soft-tissue region is situated between vertebrae L5-T10 as shown in Figure 1.

II$_B$ - MOTION ANALYSIS

The motion characteristics define the desired compliance that should accomplish the functional requirement "facilitate motion".

In previous work by Dries [10], the contribution of different spinal regions to primary bends and the different loci about which particular vertebrae are rotating, were researched for both healthy subjects and scoliosis patients.

Dries showed that for both investigated groups (N=15), the largest portion of spinal deformation during basic motions (sagittal bending and lateral bending) is contributed by the region from vertebrae S1 to T10 (Scoliosis group 81.2%, Control group 87.3% for sagittal bends, Scoliosis group 55.7%, Control group 55.7% for lateral bends).

Contribution to bend tasks

Using the methods described by Dries [10], we are able to calculate the rotation angles and relative contribution of different spinal segments during primary bends, being sagittal bend tasks (flexion), lateral bend tasks and axial twists. Using these measurements, we identify the spinal segments where most motion is present during the primary bend tasks. The motion results of the Motion study patient have been scaled to the correction study patient and shown in table 2. During the sagittal bend, the lower regions of the spine (S1-T10) contribute to 82.4% of the total bend, while the higher regions (T10-T4) contribute for only 17.6%. For the lateral bend, these relative contributions are 73.4% for S1-T10 and 26.6% for T10-T4. For the axial twist motion, the lower region (S1-T10) provides 67.9% and the upper spinal region (T10-T4) contributes 32.1%.

TABLE 2. CONTRIBUTION TO PRIMARY BENDS FOR DIFFERENT SECTIONS OF THE SPINAL COLUMN. CONTRIBUTION IS GIVEN AS PERCENTAGE WITH STANDARD ERROR OF TOTAL BENDING TASK ANGLES.

Vertebra	Sagittal Bend	Lateral Bend	Axial Twist
T7-T4	4.6% ± 4.6%	9.1% ± 4.3%	30.8% ± 16.1%
T10-T7	13.0% ± 5.3%	17.5% ± 4.3%	1.3% ± 12.3%
L1-T10	30.1% ± 1.7%	42.4% ± 2.4%	44.5% ± 15.9%
L3-L1	19.6% ± 4.0%	22.3% ± 3.5%	-2.7% ± 7.1%
S1-L3	32.7% ± 5.7%	8.7% ± 1.0%	26.1% ± 2.3%

Location of screws for lower spinal region

For the spinal region where most motion occurs during primary bends (S1-T10), a spatial motion analysis is performed using a custom made Matlab program to indicate the relative locations about which the vertebrae are rotating. Screws are calculated in the absolute normal planes, neglecting the out-of-plane motion caused by natural motion and imaging errors. [10] The screws of the lower lumbar region (L5-T10) have been averaged and normalized to the pelvis. The screw locations of the T10 vertebra are depicted in Figure 5 for the three primary bending tasks. Since no pitch is involved, the visualized screws are rotation axes. This means the three bending motions in can be simplified to pure rotations about the depicted axes for this patient.

The exact locations including confidence intervals of the average screw origins in the pelvis frame are depicted in Table 3.

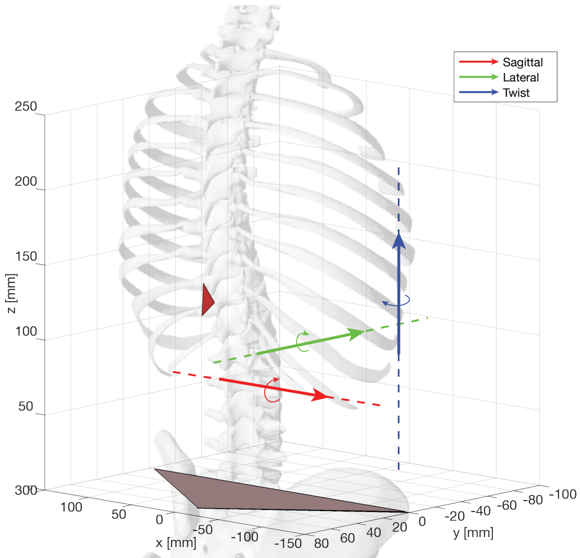


FIGURE 5. LOCATIONS OF GENERALIZED SCREW ROTATIONAL AXES DURING PRIMARY BEND TASKS FOR LOWER LUMBAR REGION (T10, RELATIVE TO PELVIS.)

TABLE 3. LOCATIONS OF AVERAGE SCREW ROTATIONAL AXES BETWEEN T10-S1 WITH RESPECT TO S1

	Sagittal Bend		**Lateral Bend**		**Twist**	
x	0	*mm*	-0.05 $\pm$ 9.8	*mm*	10.5 $\pm$ 30.0	*mm*
y	-14.2 $\pm$ 0	*mm*	0	*mm*	-81.4 $\pm$ 32.8	*mm*
z	63.6 $\pm$ 9.1	*mm*	97.6 $\pm$ 11.0	*mm*	0	*mm*

III - SEGMENT DIVISION

With the knowledge obtained by both the correction and the motion analysis the opportunity rises to reduce the complex biomechanical problem to an isolated spatial mechanism problem with quantified load and kinetic requirements. Both the ineffective correctional loads and the desired motion are situated in the soft-tissue region between vertebrae L5-T10. The regions that allow for effective correctional load transfer are the upper and lower hard-tissue regions (T10-T4 and region around Pelvis).

We propose a brace that applies loads in the hard-tissue regions and is compliant in the segment that spans the soft-tissue region.

The loads will be applied by an upper and lower correctional segment which are in contact with the predominantly hard-tissue

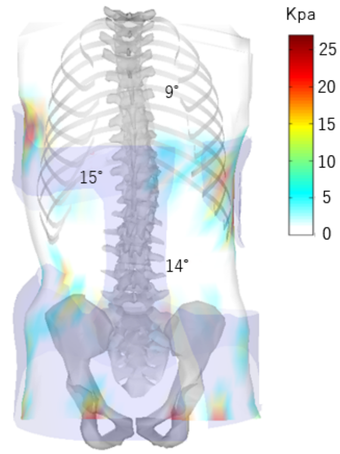


FIGURE 6. SIMULATED CORRECTION RESULTS OF POTENTIALLY COMPLIANT BRACE

regions. The two correctional segments are connected by a middle compliant segment that spans the predominantly soft-tissue regions without skin-contact. In the final design, the compliant segment will transmit the required loads while allowing motion during daily activities.

IV - ISOLATED SEGMENT ANALYSIS

The desired kinematic design specifications of the compliant segment are derived from a isolated segment of a modified benchmark brace. The in-brace correction profile and pressure results of the modified benchmark static brace are shown in Figures 6 and 7.

The modified static brace obtains similar correction as the original brace. Validation in BraceSim shows that the modified benchmark brace reduces the Cobb angle by $\pm$ 11 degrees. Figures 6 and 7 compared with Figures 3 and 4 show that the modified benchmark brace results in a similar reduction of the Cobb angle ($\pm$ 1 degree) and lower peak pressures are apparent on the skin of the patient.

The modified benchmark brace spans the compliance region with a temporary stiff segment, which is isolated as can be seen in Figure 8. The transmitted loads and the resulting deformation of the temporary stiff segment define the kinematic design specifications of the to be designed compliant segment.

Isolated Segment Simulation

A finite element model of the isolated temporary stiff segment is analyzed in ANSYS and post-processed in Matlab. We determine the loads transmitted by the temporary stiff segment spanning the compliance region. Any compliant mechanism that transmits these loads, while resulting in the same in-brace relative location of the upper and lower correction segment is a suit-

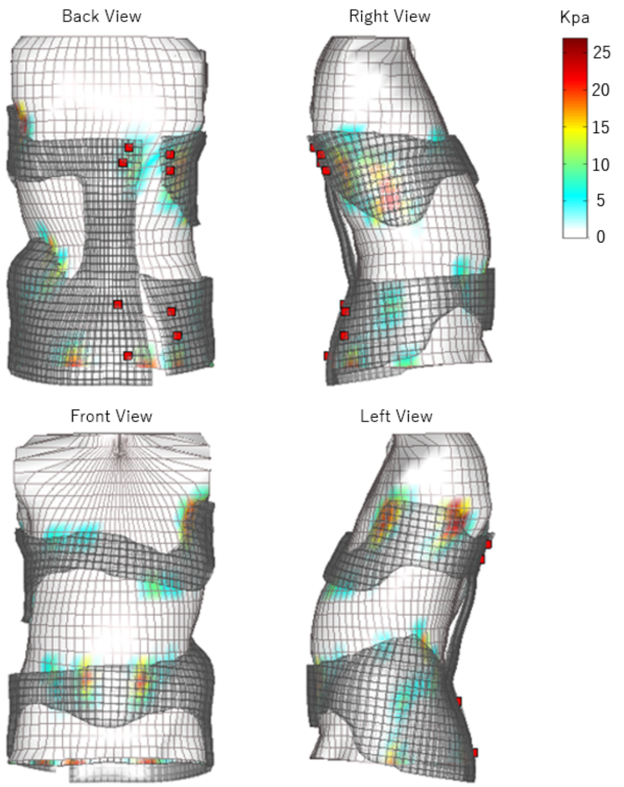


FIGURE 7. POTENTIALLY COMPLIANT BRACE INCLUDING PRESSURES

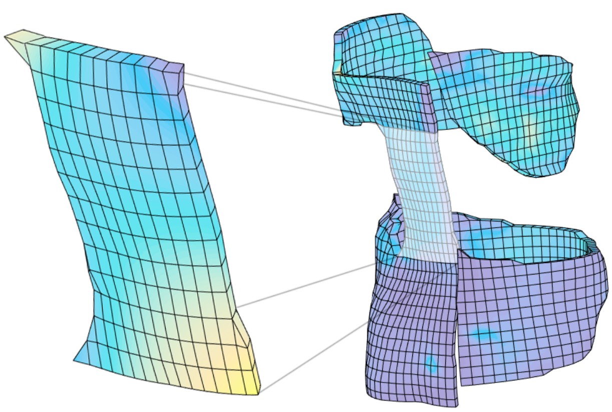


FIGURE 8. FINITE ELEMENT MODEL ISOLATED TEMPORARY STIFF SEGMENT

able substitution compliant mechanism. This includes a compliant mechanism with added desired compliance directions based on the motion study. The relative location is determined by subtracting the average node location of the lower edge from the upper edge in the in-brace configuration. The moments are determined around the average node location of the lower edge.

The transmitted loads and relative locations are given in table 4.

TABLE 4. KINETIC REQUIREMENTS

Axis	F	M	$\delta_{\mathbf{UE-LE}}$
x	3.44 N	0.02 Nm	$3.04 \cdot 10^{-02}$ m
y	0.51 N	0.55 Nm	$4.86 \cdot 10^{-04}$ m
z	3.31 N	0.12 Nm	$1.27 \cdot 10^{-01}$ m

V - DESIGN SPECIFICATION

The design specifications are based on the motion study and the temporary stiff segment FEM analysis results. The quantitative design specifications apply to the compliant segment that will substitute the temporary stiff segment of the modified benchmark brace. The temporary stiff segment FEM analysis determines the required loads and deformations. The motion study determined the desired compliance axes of the compliant segment. The complex bio-mechanical problem is reduced to an isolated mechanical problem with clear kinematic design specifications, which consists of average screw locations and kinematic requirements. These quantified results are shown in Tables 3, 4.

DISCUSSION

The goal of this research was to quantify the functional requirements for the design of a compliant scoliosis brace and translate them into design specifications. The strategy presented uses motion analysis of a scoliosis patient and correction analysis of an original non-compliant scoliosis brace to form these design specifications. A brace design based on the presented design specifications can be validated in terms of correction efficiency using BraceSim and compared with the original non-compliant brace design as presented in Figure 4.

The strategy presented in this paper is a new approach to convert qualitative functional requirements into quantitative design specification.

Previous research into compliant scoliosis bracesv utilizing compliant shell mechanisms did not account for brace-tissue interaction [6], where this strategy does using BraceSim results. We have analyzed the spinal motions of a scoliosis patient and shown what different vertebra contribute to the total motion, and about which twist axes these motions occur. This analysis was performed for the absolute body planes, eliminating out-of-plane motions which can be caused by natural motions or imaging errors.

Ideally the braced correction study patient should have been studied to allow for an exact design of a compliant brace, which can be validated using BraceSim and this patients X-rays and bodyscan. We reduce these differences by analyzing a patient with a similar Cobb angle and scoliotic progression as the correction study patient, without taking into account the angle of kyphosis or lordosis, and scaling the results to match the body length of the correction study patient. More research is needed to analyze possible differences in motion characteristics between scoliosis patients with similar Cobb angles and scoliotic progression.

The correctional loads needed for correcting the braced correction study patient have been analyzed using FEM analysis of a brace designed by an certified orthotist, and provide clear specifications for the design of compliant elements together with the compliance axes found in the motion studies. In future work it will be beneficial to model the edges of the temporary stiff segment as rigid, to ensure similar boundary conditions for the isolated temporary stiff element edges.

With the presented approach the required correctional loads are found in the patients non-deformed position. As a compliant brace design allows for motion, the required correctional loads in bent positions should be researched.

Currently, not many synthesis methodology are available in literature to generate concepts for compliant mechanism design. Using such a synthesis method one should be able to use the presented quantitative design specifications and translate them into spatial mechanisms allowing for compliant brace designs, which can be validated in terms of correction efficiency using BraceSim.

CONCLUSION

This paper introduced a strategy to reduce the complexity of a bio-mechanical problem into an isolated mechanical problem. The main qualitative functional requirements have been converted into quantitative design specifications for a spatial compliant mechanism. The explained strategy was proven feasible on the basis of an example patient. Using this method, generated concepts can be validated on correctional efficiency using BraceSim software. Further work should be focused on exploring and testing actual brace designs and the generalization of patient motion data.

ACKNOWLEDGMENT

The authors gratefully acknowledge NSF No. IIS - 1527133 as well as STW (HTSM-2012 12814: ShellMech) for their financial support of this project. The writers would also like to thank professor dr. Christiane Cauoette, Nikita Cobetto, PhD and prof. dr. Carl-Éric Aubin of the École Polytechnique de Montréal for collaborating and providing acces to BraceSim.

REFERENCES

[1] Deacon, P., Flood, B. M., and Dickson, R. a., 1984. "Idiopathic scoliosis in three dimensions. A radiographic and morphometric analysis.". *The Journal of bone and joint surgery. British volume,* ***66***(4), pp. 509–512.

[2] Weinstein, S. L., Dolan, L. A., Wright, J. G., and Dobbs, M. B., 2013. "Effects of bracing in adolescents with idiopathic scoliosis.". *The New England journal of medicine,* ***369***(16), pp. 1512–21.

[3] Nachemson, A. L., Lonstein, J. E., and Weinstein, S. L., 1982. "Report of the Prevalence and Natural History Committee of the Scoliosis Research Society.". In Denver: Scoliosis Research Society.

[4] Asher, M. A., and Burton, D. C., 2006. "Adolescent idiopathic scoliosis: natural history and long term treatment effects.". *Scoliosis,* ***1***(1), p. 2.

[5] Rigo, M., and Weiss, H.-R., 2008. "The Cheneau Concept of Bracing - Biomechanical Aspects". *The Conservative Scoliosis Treatment,* ***135***, pp. 303–319.

[6] Nijssen, J. P. A., 2017. "Design and Analysis of a Shell Mechanism Based Two-fold Force Controlled Scoliosis Brace". Master's thesis, Delft University of Technology.

[7] Ring, J. B., Kim, C. J., Evans, J., and Beal, C., 2017. "A Passive Brace for the Treatment of Scoliosis Utilizing Compliant Mechanisms". Master's thesis, Bucknell University, Lewisburg PA.

[8] Bulthuis, G. J., Veldhuizen, A. G., and Nijenbanning, G., 2008. "Clinical effect of continuous corrective force delivery in the non-operative treatment of idiopathic scoliosis: a prospective cohort study of the triac- brace". *European Spine Journal,* ***17***(2), pp. 231–239.

[9] Desbiens-Blais, F., Clin, J., Parent, S., Labelle, H., and Aubin, C.-E., 2012. "New brace design combining cad/cam and biomechanical simulation for the treatment of adolescent idiopathic scoliosis". *Clinical biomechanics,* ***27***(10), pp. 999–1005.

[10] Dries, T. J., 2018. "A Biomechanical Characterization of Spinal Motion Data for the Design of a Compliant Scoliosis Brace". Master's thesis, Delft University of Technology, Delft, The Netherlands.

[11] Richards, B. S., Bernstein, R. M., D'Amato, C. R., and Thompson, G. H., 2005. "Standardization of criteria for adolescent idiopathic scoliosis brace studies: SRS Committee on Bracing and Nonoperative Management.". *Spine,* ***30***(18), sep, pp. 2068–75; discussion 2076–7.

5

Conclusion

In this thesis, spinal motions have been analysed of both Scoliosis patients and healthy subjects. Characterizations of these motions form a necessary basis for the future design of a compliant scoliosis brace.

Chapter 2 focussed on the design and implementation of a motion trial experiment, which allowed for data collection of primary bends. A new retroreflective marker setup consisting of 6 marker tripods and 3 separate markers was designed to use in an optoelectronic motion capture system. A method was presented to post-process the collected data using screw theory, after which data could be analysed and compared between groups. Results from the chapter showed no significant differences between the two groups, but did show a general motion profile, as utilized by participants in the study for performing the primary bends. This motion profile showed a high contribution of the lower lumbar region of the spine during sagittal and lateral bends. The thoracic region contributed more during axial twists. Spatial axes of rotations were found for all subjects and differences between groups were presented. The ability to generalize these spatial rotational axes could not be approved, neither disproved using the experiment set-up. In general, data from the motion experiment can form a new basis for designing a compliant scoliosis brace.

Chapter 3 focussed on a quantification strategy for designing such a compliant scoliosis brace. This quantification strategy proposed a method to convert qualitative functional requirements into quantitative design specifications for spatial compliant mechanism design. These quantitative design specifications consist of captured motion data, which was part of the motion experiment from Chapter 3, together with kinetic data following from a correction analysis performed in BraceSim software. Using both kinematic data and correction simulations of a benchmark brace, a new brace design proposal was shown. In this new brace design the lower lumbar region of the spine is spanned by a specific segment, which can be replaced by a compliant element that meets the presented kinematic and kinetic requirements.

Together, the presented papers form new quantitative motion and design criteria that can be applied in the design of a compliant scoliosis brace.

6

Discussion

In this section, the presented work is discussed. Additionally, recommendations regarding the presented theory and results can be formed. In this section, both presented papers are discussed in chronological order including recommendations regarding the papers. Additionally, a possible design iteration is presented, showing the possible next steps in this project.

6.0.1. Chapter 3 Spinal Motions

In Chapter 3 spinal motions were analysed for both scoliosis patients and healthy subjects using primary motion experiments. In total, 15 Scoliosis patients and 15 healthy subjects participated in the experiment, which focused on characterizing spinal motion for specific spinal regions using a newly designed marker set-up. Results from these experiments showed a high contribution for the lower lumbar region in the sagittal and lateral bending task for both groups, which confirmed the hypothesis. The thoracic region contributed more during axial twists. Next to the contributions to the entire bend, spatial axes of rotation were calculated using screw theory. Efforts were made to generalize the results of these screw locations, however it remains unclear whether this generalization is grounded.

A possible explanation for these difficulties might be the varying types of scoliosis and scoliotic curvatures across participating subjects. Since no information on spinal curvatures and type of scoliosis was available for the subjects in the scoliosis group, a clear generalization was not possible. Furthermore the relatively small sample size might have influenced these outcomes as well.

The presented work in Chapter 3 contributes to the field of scoliosis research as it describes a new method to analyse spinal motions for specific vertebrae. Additionally, the analysis of spatial axes of rotations of specific vertebrae has not been performed before and may provide a better understanding of the motions of the human spine during primary bends as well as quantifiable input for mechanism designers for the design of compliant scoliosis braces.

It should be taken into account that the presented research focused primarily on in-plane motions. Since the spine is a three-dimensional system of interlinked vertebrae and scoliosis is usually a three-dimensional deformity, the presented results do not describe the entire motion of the spine or spinal segments.

Future work should therefore be devoted to several aspects:

- Investigate the possibility to generalize spatial screw axes by running a study design that controls for type of scoliosis, curve severity and possibly spinal maturity.
- Investigate whether out-of-plane motions contribute substantially to the primary bends.
- Perform experiments using the presented marker set-up for maximal bending tasks and compare outcomes with other research.

6.0.2. Chapter 4 Characterization Strategy

In Chapter 4, a new strategy is proposed to translate functional requirements into design specifications for the design of a compliant scoliosis brace. This strategy brings together spinal motion data as captured and analysed using methods in Chapter 3, with kinematic load requirements as derived from validated brace CAD models. The quantified results can be used to design a compliant shell element, which can be used in a scoliosis brace design.

The presented work focuses on the analysis of a patient-specific brace, combined with the motion results of a similar patient. Using the data a brace can be designed and validated using BraceSim software. This validation could prove the applicability of compliant shell mechanisms in scoliosis braces, increasing the RoM of patients, while making sure enough correction is provided.

The goal of this entire project is to design these compliant braces for a range of patients. The validation of a brace design following Chapter 4 will therefore be a proof-of-concept, after which a wider research into kinetic and kinematic requirements is preferred. Several recommendations are therefore:

- Use presented data to design a compliant brace concept, which can be validated using BraceSim and using pressure and motion tests.
- Design a generalized scoliosis brace using possible generalizations as stated in section 6.0.1, as the currently presented strategy focuses on patient-specific data

Design Iteration

Together with the work on the characterization of non-linear compliant mechanisms as presented by Leemans [7], new foundations have been developed to aid the design of such a brace. This new structure could, when applied correctly, be the starting point for a new brace design cycle, which will eventually lead to the actual production of a brace concept. This brace concept would then comply to the presented criteria in Chapter 4 and allow for computer simulations using BraceSim, as well as force and kinematic measurements for validation. As a first design iteration, it is recommended to design a brace concept allowing for a maximum of

two Degrees of Freedom, the sagittal and lateral direction.
As previous concepts were hard to validate, another recommendation is to use the conventional motion controlled correction strategy in the first concept. This concept can show the validity of the presented work and could thus become a proof-of-concept design. In conjunction with the presented design and collected data in Chapter 4, this might result in design ideas as shown in Figure 6.1.

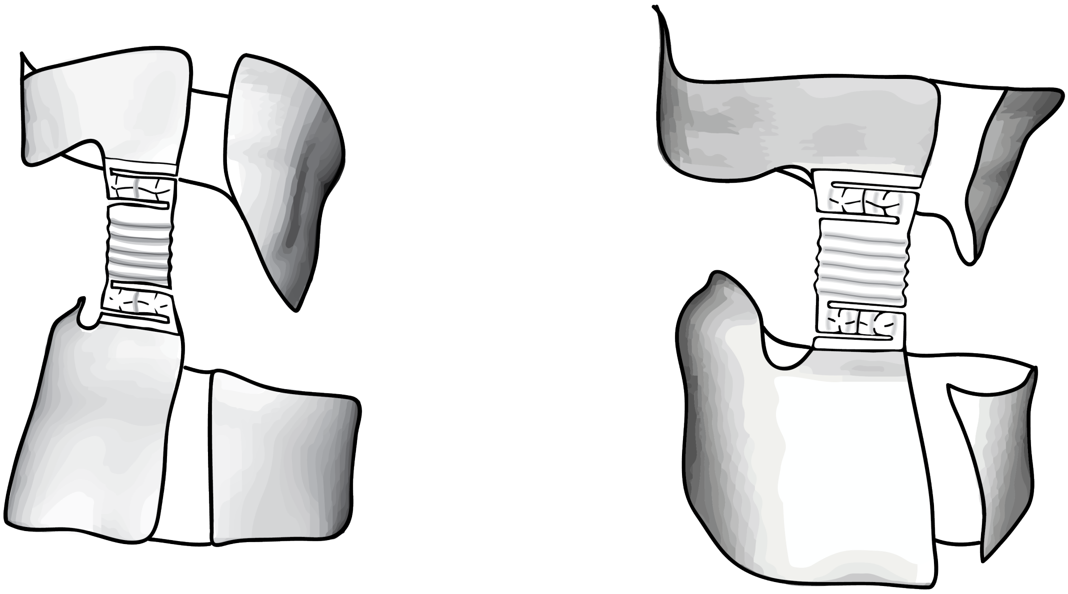


Figure 6.1: A possible 2DOF compliant scoliosis brace design showing the applicability of gathered data from Chapter 3 in conjunction with the strategy from Chapter 4 and the compliant shell mechanism characterization presented by Leemans [7]. Two patient optimized body shells are connected through a corrugated shell structure, allowing for bending in sagittal and lateral direction. The corrugated shell can be optimized for the corrective forces as found in Chapter 4 using methods described by Leemans. [7]

This design consists of a patient optimized shell, which provide the actual correction forces on the apex. Additionally, the 'bridge' part of the brace has been replaced by a corrugated shell mechanism, which allows the brace for rotations in sagittal and lateral directions. The corrugated shell can be optimized using the work presented by Leemans, with captured motion data (relative bend angles and contributions and screw locations) as captured in Chapter 3 as input.

Bibliography

[1] S Aaro and C Ohlund. Scoliosis and pumonary function. *Spine*, 9(2):220–2, 1984.

[2] Todd J. Albert, Moe R. Lim, Joon Y. Lee, and C. Regan. Adult scoliosis, 2016. URL https://musculoskeletalkey.com/adult-scoliosis-3/.

[3] Marc A Asher and Douglas C Burton. Adolescent idiopathic scoliosis: natural history and long term treatment effects. *Scoliosis*, 1(1):2, 2006. ISSN 1748-7161. doi: 10.1186/1748-7161-1-2.

[4] C Coillard, CH Rivard, and AB Circo. Effectiveness of the SpineCor brace based on the standardized criteria proposed by the S.R.S. for adolescent idiopathic scoliosis - up to date results. *Scoliosis*, 4(Suppl 2):O54, 2009. ISSN 1748-7161. doi: 10.1186/1748-7161-4-S2-O54. URL http://scoliosisjournal.biomedcentral.com/articles/10.1186/1748-7161-4-S2-O54.

[5] Jack R. Engsberg, Lawrence G. Lenke, Angela K. Reitenbach, Kevin W. Hollander, Keith H. Bridwell, and Kathy Blanke. Prospective Evaluation of Trunk Range of Motion in Adolescents With Idiopathic Scoliosis Undergoing Spinal Fusion Surgery. *Spine*, 27(12): 1346–1354, 2002. ISSN 0362-2436. doi: 10.1097/00007632-200206150-00018. URL http://content.wkhealth.com/linkback/openurl?sid=WKPTLP:landingpage{&}an=00007632-200206150-00018.

[6] Sarah Galvis, Douglas Burton, Brandon Barnds, John Anderson, Richard Schwend, Nigel Price, Sara Wilson, and Elizabeth Friis. The effect of scoliotic deformity on spine kinematics in adolescents. *Scoliosis and Spinal Disorders*, 11(1):42, 2016. ISSN 2397-1789. doi: 10.1186/s13013-016-0103-x. URL http://scoliosisjournal.biomedcentral.com/articles/10.1186/s13013-016-0103-x.

[7] Joost R Leemans. Characterization of non-linear compliant shell mechanisms: To enhance the design process of a compliant scoliosis brace. Master's thesis, Delft University of Technology, 2018.

[8] Mangino, L. All scoliosis braces are not the same. you be the judge... year = 2016, url = https://advance-physicaltherapy.com/2016/scoliosis-braces-created-equally-judge/.

[9] Elaine. N Marieb and Katja Hoehn. *Human Anatomy & Physiology (9th edition)*. Pearson, 9 edition, 2013. ISBN 978-0-321-74326-8.

[10] A L Nachemson, J E Lonstein, and S L Weinstein. Report of the Prevalence and Natural History Committee of the Scoliosis Research Society. In *Denver: Scoliosis Research Society*, 1982.

[11] Stefano Negrini, Silvia Minozzi, Josette Bettany-Saltikov, Nachiappan Chockalingam, Theodoros B. Grivas, Tomasz Kotwicki, Toru Maruyama, Michele Romano, and Fabio Zaina. Braces for idiopathic scoliosis in adolescents, jun 2015. ISSN 1469493X. URL http://doi.wiley.com/10.1002/14651858.CD006850.pub3.

[12] Gert Nijenbanning. *Scoliosis Redress: Design of a force controlled orthosis*. PhD thesis, Universiteit Twente, Enschede, 1998.

[13] Joep P A Nijssen. Design and Analysis of a Shell Mechanism Based Two-fold Force Controlled Scoliosis Brace. Master's thesis, Delft University of Technology, 2017.

[14] B Stephens Richards, Robert M. Bernstein, Charles R D'Amato, and George H. Thompson. Standardization of criteria for adolescent idiopathic scoliosis brace studies: SRS Committee on Bracing and Nonoperative Management. *Spine*, 30(18):2068–75; discussion 2076–7, sep 2005. ISSN 1528-1159. URL http://content.wkhealth.com/linkback/openurl?sid=WKPTLP:landingpage{&}an=00007632-200509150-00013http://www.ncbi.nlm.nih.gov/pubmed/16166897.

[15] Manuel Rigo and Hans-Rudolf Weiss. The Cheneau Concept of Bracing - Biomechanical Aspects. *The Conservative Scoliosis Treatment*, 135:303–319, 2008.

[16] J B Ring, Charles J Kim, Jeffrey Evans, and Craig Beal. A Passive Brace for the Treatment fo Scoliosis Utilizing Compliant Mechanisms. Master's thesis, Bucknell University, Lewisburg PA, 2017.

[17] D E Rowe, S M Bernstein, M F Riddick, F Adler, J B Emans, and D Gardner-Bonneau. A meta-analysis of the efficacy of non-operative treatments for idiopathic scoliosis. *J Bone Joint Surg Am*, 79(5):664–674, 1997. ISSN 0021-9355.

[18] Wafa Skalli, Reinhard D. Zeller, Lotfi Miladi, Gaëlle Bourcereau, Mélanie Savidan, François Lavaste, and Jean Dubousset. Importance of Pelvic Compensation in Posture and Motion After Posterior Spinal Fusion Using CD Instrumentation for Idiopathic Scoliosis. *Spine*, 31(12):E359–E366, 2006. ISSN 0362-2436. doi: 10.1097/01.brs.0000219402.01636.87. URL http://content.wkhealth.com/linkback/openurl?sid=WKPTLP:landingpage{&}an=00007632-200605200-00027.

[19] M.J. Solomito. The Use of Motion Analysis Technology as an Alternative Means of Assessing Spinal Deformity in Patients with Adolescent Idiopathic Scoliosis. Master's thesis, Univeristy of Connecticut - Storrs, 2011.

[20] Ian A F Stokes, Llynda C. Bigalow, and Morey S. Moreland. Three-dimensional spinal curvature in idiopathic scoliosis, 1987. ISSN 1554527X.

[21] A. G. Veldhuizen, J. Cheung, G. J. Bulthuis, and G. Nijenbanning. A new orthotic device in the non-operative treatment of idiopathic scoliosis. *Medical Engineering and Physics*, 24(3):209–218, 2002. ISSN 13504533. doi: 10.1016/S1350-4533(02)00008-5.

[22] Jeffrey W. Wiley, Jeffrey D. Thomson, Thomas M. Mitchell, Brian G. Smith, and John V. Banta. Effectiveness of the Boston brace in treatment of large curves in adolescent idiopathic scoliosis. *Spine*, 25(18):2326–2332, 2000. ISSN 03622436. doi: 10.1097/00007632-200009150-00010. URL https://journals.lww.com/spinejournal/Fulltext/2000/09150/Effectiveness{_}of{_}The{_}Boston{_}Brace{_}in{_}Treatment{_}of.10.aspx.

[23] R B Winter, W W Lovell, and J H Moe. Excessive thoracic lordosis and loss of pulmonary function in patients with idiopathic scoliosis. *Journal of Bone and Joint Surgery*, 57(7): 972–977, 1975.

II

Appendices

A

Literature Review

Literature Research

Tools to analyze scoliotic deformations and spinal movement in order to aid development of a new compliant scoliosis brace.

Tim Dries

December 28, 2017

Contents

1 Abstract

Adolescent Idiopathic Scoliosis can be defined by a three-dimensional deformity of the spine, which can be characterized by a lateral curvature with a Cobb angle of more than 10 degrees and rotated vertebrae. Bracing has been a common treatment for Scoliosis patients over the last decades. By applying forces on the spinal cord, the progression of the scoliotic deformation of the spine is inhibited or reversed. The currently used braces have been found to be quite efficient in terms of treatment, but often patients find them uncomfortable and too bulky to wear, especially because of the lack of compliance for bending and thus decreased Range of Motion. To design a new Scoliosis brace, knowledge about the currently available braces is needed, as well as knowledge about the spinal movement during primary motions to characterize the different axes of rotation between vertebrae. An overview of relevant available braces is presented to give insight in the effectiveness of these braces. Also, an overview of motion capture techniques is presented to help us design a motion capture experiment of these spinal movements in both healthy subjects and scoliosis patients. It is shown that a comparison of braces is very hard, due to different research approaches and the lack of a standardized research. Especially the patient compliance is heavily influenced by the measurement method, which is most often done using subjective techniques. Currently, most scoliosis motion research makes use of opto-electronic motion capture systems due to the accuracy of these systems. The found research does not present an analogous method how to capture motions of the spine, mostly due to different research goals in the literature.

2 Introduction

Scoliosis is typically defined as an abnormal curvature of the spine that can cause pain, and in severe cases cardio-pulmonary complications. Three different types of scoliosis can be distinguished: congenital scoliosis, neuromuscular scoliosis, and idiophatic scoliosis. [1] While the first two types of scoliosis have known causes, idiopathic scoliosis has no known cause. [2] This work is focusing on this last type of scoliosis.

Adolescent idiopathic scoliosis can be defined as a three-dimensional deformity of the spine, which is characterized by a lateral curvature with a Cobb angle of more than 10 degrees and rotated vertebrae. [3] [4] Adolescent idiopathic scoliosis (AIS) develops in around 3% of all adolescents, of which approximately 10% has progressive curves that require treatment of some sort. [5] Spinal deformation can cause some serious health implications. On a physical level these implications might include severe deformations that may cause heart and lung problems. [6][7] Non-physical or psychological health implications might include struggling with self-image, an emotional pain. [8]

Treatment possibilities for scoliosis include surgery, muscle stimulation & training and bracing, depending on the severity of the deformation. This deformation severity is usually measured using the Cobb-angle, which is the angle between the two most tilted vertebrae of the spine. [8] A visualisation of measuring the Cobb angle is shown in Figure 1.

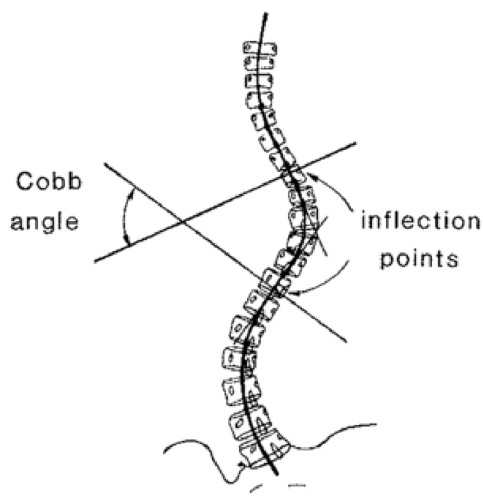


Figure 1: Cobb Angle Measurement, reproduced from [9]

Usually braces are prescribed when the Cobb angle measures between 25 and 40 degrees in order to prevent further progression of the spine deformation. [10] Using a brace, external corrective forces are applied to the trunk. [11] These forces are usually not expected to correct the deformation; the treatment focuses on inhibiting the progression of the spinal deformation rather than correcting the spinal curve. [12] Braces are only prescribed when the patients have growth remaining, since they have a greater change of curve progression. Usually this curve progression is indicated by the Risser sign; a grade of 0 or 1 indicates two more years of growth, while a grade of 5 indicates complete ossification of the apophysis, meaning little to no growth remaining. [13] If the Cobb Angle exceeds 40 degrees, usually surgery is performed in order to reduce the deformity by fusing vertebrae together using metal rods and bolts. The use of scoliosis braces is thought to be beneficial in terms of treatment, with Weinstein et al. showing in a multicenter study (BrAIST (Bracing in Adolescent Idiopathic Scoliosis Trial)) that the use of a scoliosis back brace prevents the need for scoliosis surgery in 72% of the cases.

[4] Commonly, scoliosis braces have to be worn for 16 to 23 hours a day, over the duration of 2 to 4 years in time. Different sorts of braces have been developed over the last decades, all of which have to comply to different criteria. These criteria are not only influencing the effectiveness of the brace in terms of treatment, they also influence the applicability of the brace and the comfort as experienced by the user. Research has shown that wearing a scoliosis brace can negatively influence the emotional and social well-being of a patient, which in its turn can lead to poor compliance towards the brace. [4] [14] [15] This reduced compliance towards the brace has been shown to be the most common cause of failure in treatment. [16] The reduced range of motion when wearing the brace and the bulkiness of the brace itself are said to be highly contributing factors to this (decreased) user compliance. [17]

2.1 Problem statement

Since the currently used scoliosis braces do not seem to perform sufficient enough in terms of either correction or patient compliance, there seems to be a need for the development of a new brace. This new brace will likely combine the performance in terms of correction of the rigid braces, with the flexibility of the flexible braces.

With this knowledge, we can form the following research question:

How to design a new scoliosis brace, which aims to combine the performance in terms of treatment of currently available rigid braces, with the increased ranges of motion as allowed by more flexible braces?

To answer this question, knowledge about the currently available braces is needed. Therefore one sub-question is stated as follows:

What can we learn from the current braces and how can we compare them?
Several secondary questions are formed as well:

- *How are current braces categorized?*
- *How are current braces validated?*

Since the new brace aims to allow for certain movements, knowledge about human motion is needed as well. Therefore the second sub-question is stated as:

How can we characterize spinal motions?
With the secondary questions:

- *What techniques exist to perform motion studies?*
- *What techniques are used in scoliosis research?*

Following these questions, the goal of this project is stated. The aim of this project is to develop a new scoliosis brace, which provides enough correction for effective treatment, but also allows the patient to move more freely. In this research, first different brace classifications and correction strategies will be elaborated on, in order to give us insight how we are able to compare different brace designs. Next, an overview of common scoliosis braces is given, showing the most relevant braces in the scope of this project. Also, an investigation on in-brace forces will be shown, indicating different methods to measure the correction forces. Since the user compliance is highly influenced by the activities of daily living of the patient, the other part of this study is focusing on capturing the spinal motion. Different types of motion studies will be highlighted and an overview of motion studies focusing on spinal kinematics is given. As a part of these studies, previous work within this project is highlighted in section 5.3.1. Next, information about marker set-up is given. In the end a conclusion and a discussion is presented.

3 Method

This literature report is carried out in order to get an overview of the important considerations, which should be taken into account when designing a scoliosis brace. It also serves as background knowledge for carrying out research with scoliosis patients and it points out interesting research considerations to be performed in the near future.

In this report, different literature searches have been used to find, review and include different scientific reports. A first search was performed in March 2^{nd}, 2017 in Scopus using search terms: "TITLE-ABS-KEY (scoliosis AND brace AND corrective)" resulting in 130 found articles. On March 16, 2017 additional searches were performed in Scopus using the terms: "TITLE-ABS-KEY ("orthoses" AND force AND passive)" and "TITLE-ABS-KEY (orthosis AND forces AND validation)" On May 9th, 2017 searches were performed using (kw: motion AND capture AND spine/spinal) on (TU Delft) WorldCat.org (all databases) resulting in 655 results.

Additional resources were found when referred to by consulted literature, suggested by researchers, experts or software like Mendeley or when recommended by different literature databases.

The found literature has been filtered, based on different criteria. Articles were excluded after a review of titles, keywords and abstracts because they were irrelevant to the chosen topic. Furthermore articles were excluded when only available in languages other than English or Dutch, or when they were only partially available through the TU Delft sources. Articles written before the year 2000 have been excluded, except for the articles about the background of scoliosis and first publications explaining a new type of scoliosis brace.

4 Scoliosis Braces

At this moment, varying braces are used in order to inhibit curve progression. All of these braces might have different properties, are used for treating different scoliosis cases and might consist of different materials when compared to each other. In order to understand the working principles of these braces, we have a look at the brace classification, validation, and patient compliance for several braces.

4.1 Brace classification

Because of the variety in braces, several attempts of classifying the braces into certain groups have been made. The most simple classification of braces is based on the anatomy of the human and the spinal region where the brace acts: cervical (C), thoracic (T), lumbar (L) and sacral (S). By using this type of classification, two main families have been used over the last 30 years: a) Cervical-Thoraco-Lumbo-Sacral Orthotics or CTLSO and b) Thoraco-Lumbo-Sacral Orthotics or TLSO.[18] In 2008 Negrini et al. [19] presented another, more generalized classification called BRACE MAP and in 2016 Grivas et al. [18] published the first step towards a revised classification system. Braces can also be classified by their rigidity, as shown by Ring [17] and Nijssen [20], or by their control strategy as shown by Nijenbanning. [12]

4.1.1 BRACE MAP classification

BRACE MAP is a classification made by Negrini et al. that has its origin by looking at the different properties of scoliosis braces. BRACE MAP stands for the following: Building, Rigidity, Anatomical classification, Construction of the Envelope, Mechanism of action, and Plane of action. [18] With these terms, each brace was classified assigning the beginning letters when the brace met the classificatory demands. (e.g. CpETAM3, Custom positioning, Elastic, TLS, Asymmetric, Movement principle and 3D correction was the characterized name of the SpineCor brace). [18] At the time, 13 braces were considered and all but two could be differentiated from each other following this system.

4.1.2 SOSORT Classification

Even though the BRACE MAP classification was the first of its kind, the authors agreed to redefine the system to make it more useful as a classification system. In 2014 the International Society for Scoliosis Orthopedic and Rehabiliation Treatment (SOSORT) asked the Brace Classification Study Group (BCSG) to evaluate the terminology used in the field of scoliosis treatment as well as the brace classification. This resulted in a first approach to brace classification and some 139 terms related to bracing. These definitions were presented by the BCSG in 2016, while the classification itself will be published in the near future. [18]

4.1.3 Rigidity

Another way of looking at different braces, is by looking at the material they consist of and in what way the patients are restricted in their movements when wearing the brace. We can distinguish rigid and flexible braces in this classification. Grivas et al. also highlighted rigidity as an orthotic classification, ranging from flexible, to semi rigid, to rigid, to high rigidity. [18] Rigidity is highly dependent on the material chosen to construct the material, since this influences the bendability of the entire brace. Examples of rigid braces are the Boston brace, Lyon brace and Sforzesco brace, while the SpineCor and TriAC braces can be classified as flexible braces. [20]

4.1.4 Control Strategy

Controlling or correcting a scoliotic spine is generally performed by correcting the spine with external forces at the apex of the curve. To compensate for this force, two counteracting forces are needed above and below the apex to reach equilibrium. For an S-shaped spinal cord, the top (counteracting) force is used on the top apex to stop the curve progression at that location. An example of this correction profile is shown in figure 2:

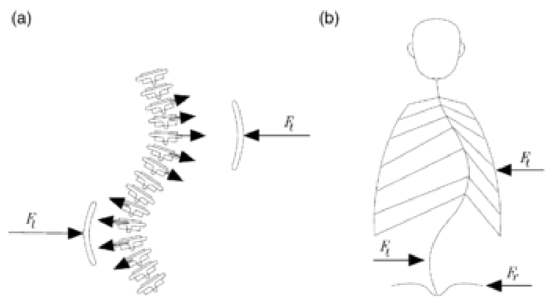
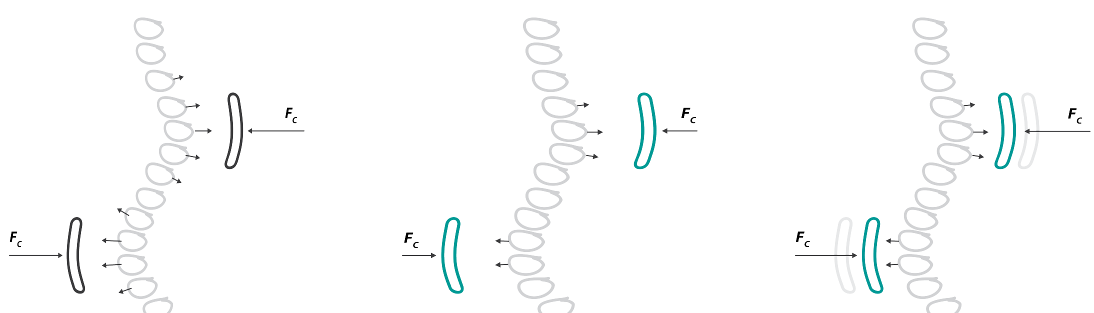


Figure 2: *(a) Counteracting progression forces acting on the scoliotic spine near the apex of the curves. (b) The progression forces together with a reaction force to obtain equilibrium in the orthosis.* Reproduced from [12]

As proposed by Nijenbanning [12], we can distinguish different control strategies when looking at different braces. Most of the currently designed rigid braces are so-called displacement controlled, meaning that as soon as the patient slightly changes his/her position within the brace the correctional forces decrease or even vanish and the intended correctional state is not reached. The flexible braces (SpineCor and TriAC) overcome this problem, since the patient is allowed to move, while the force exerting mechanism moves along with the patient. It is therefore that these braces are labelled to be force controlled; for each change in posture of the patient, the same forces are exerted on his/her body. Figure 3 shows a schematic overview of these two control strategies.



(a) Correction Profile (b) Displacement Control (c) Force Control

Figure 3: Correction strategies as proposed by Nijenbanning [12]. Figure 3a shows the correction profile, which consists of forces that correct the spine near the apex of the curves by counteracting the scoliotic progression forces. Figure 3b shows the behaviour of displacement controlled braces when the spine is slightly moved out of the initial correction position. In such a case, the correction forces decrease in magnitude. Figure 3c shows the force control strategy. This strategy makes sure that the correction forces will always have the same magnitude, since the mechanisms that exert these correction forces move along with the patient. Figures based on [12]

Table 1: Braces included in Zaina overview, adapted from [22] and edited by author.

[h]

Brace Name	Origin	Year	Developer	Rigidity	Material	Reference
Boston	USA	1972	J Hall, W Miller	Rigid	polyethelene	[23]
Charleston	USA	1979	F Reed, R Hooper	Rigid	polyethelene	[24]
Chênau and derivatives	France-Germany	1960	J Cheneau	Rigid	polyethelene	[25]
Dynamic Derotating	Greece	1982	N Vastatzidis	Rigid	polypropylene and aluminium	[26]
Lyon	France	1947	P Stagnara	Very Rigid	poylmetacrylate and radioluscent duralumin	[27]
Milwaukee	USA	1945	W Blount, A Schmidt	Rigid	polyethylene	[28]
PASB	Italy	1976	L Aulisa	Rigid	polyethylene	[29]
Providence	USA	1992	C d'Amato, B McCoy	Rigid	polyethylene	[30]
Rosenberger	USA	1983	R Rosenberger	Rigid	polyethlyene	[31]
Sforzesco	Italy	2004	S Negrini, M marchini	Very Rigid	copolyester $\pm$ radioluscent duralumin	[32]
SpineCor	Canada	1993	C Coillard, C Rivard	Elastic	Elastic tissue	[33]
TLI	The Netherlands	2012	P Van Loon	Rigid	polyethylene	[34], [35]
TriAC	The Netherlands	2002	GJ Nijenbanning, AG Veldhuizen	Low rigidity	Soft plastic and metallic connections	[36]
Wilmington	USA	1969	D MacEwen	Rigid	polyethylene	[37]

Although the force correction strategy seems to be beneficial over the displacement control strategy, it is not clear whether this strategy is more advantageous, since both the SpineCor and TriAC brace (which use this type of correction strategy) have been proven to be less efficient in terms of preventing curve progression. [21]

4.2 Types of braces

Lots of different braces have been developed and used over the last decades. Zaina et al. [22] performed a research in 2014 showing the, at that time most relevant state-of-the-art braces as used in North America and Europe. Their analysis includes the braces as shown in table 1. The table includes the names of the braces, the developers of the braces and the country of origin. It also gives an indication how rigid the braces are. This rigidity is based on the material used to build the brace, where low-flexible materials such as polyethylene make up rigid braces, while highly-flexible materials like elastic bands or tissues make up low rigid or even elastic braces. The table was extended with the year of publishing for each brace, indicating whether or not a brace has been introduced recently or has been on the market for quite some time already.

From this overview, the rigid Boston Brace and Chêneau Brace and the flexible TriAC and SpineCor braces seem to be most relevant in the scope of this project. Since the rigid Boston and Chêneau braces are the mostly prescribed braces around the world, we can learn a lot about the way they provide spinal correction. The flexibility of the TriAC and SpineCor braces can give us more insight on how to develop a compliant brace, which allows for movement of the patient.

4.2.1 Boston Brace

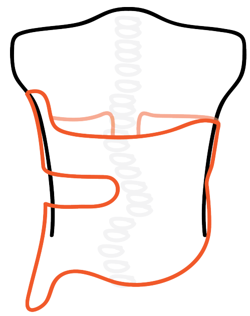


Figure 4: Schematic drawing of a possible configuration of the Boston brace (front view). A relief is shown opposite of the correction force. Also the opening at the posterior side is visible.

The Boston Brace (often referred to as TLSO brace) might be the most popular and well-known brace for treating scoliosis worldwide. It was developed by J Hall and W Miller in 1972 as a specific replacement of a Milwaukee brace. [23][22] The Boston brace is made out of copolymer plastic, where the thickness depends on the patient size. The inner shell is usually lined with aliplast, specifically at the waist to increase comfort. [38] The brace has to be worn for at least 16 hours a day. The Boston brace is said to be modular, meaning the base can be adjusted to treat several different scoliosis curves by adding and/or removing different extensions.

Usually three pads are added to the inner surface of the Boston brace to provide increased forces on the spinal curve. By putting them on the inside of the brace, they provide extra pressure on the body with the benefit of being easily adjustable. Using x-rays of the patient wearing the brace, the clinician is able to determine whether or not the pads are put on the correct location. From experience and mathematical models, it is found that the main pad pressure should be at the apex of the curve and below for nearly all deformities. Opposite of the pads, usually reliefs are made in the orthosis. These reliefs maximize the forces induced by the pads by assisting in derotation of the lumbar spine and reduction of the lumbar curvature. [38] A schematic drawing of the Boston brace is shown in Figure 4.

Validation
The Boston brace was validated multiple times, with a success rate up to 93% [4]. Weinstein et al. defined a treatment as being successful when the curve did not progress to 50 degrees or more when skeletal maturity was achieved. Gepstein et al. found a maximum treatment success of 81% in a comparative study, comparing the Boston brace with the Charleston brace. Treatment success was defined as $< 5°$ progression of curve deterioration from the start to the end of treatment, together with the absence for the need of corrective surgery. [39].

Patient Compliance
Next to the effectiveness of different braces, the multicenter BrAIST study also investigated the patient compliance towards the brace by conducting the Pediatric Quality of Life Inventory (PedsQL), ”a generic quality-of-life instrument used in studies of acute and chronic illness”. [4] Both of the patient groups indicated an average 82-point score on a scale from 0-100.
In their research, Gepstein et al. also objectively investigated the patient compliance towards the brace , and found that all their patients, or their parents, reported brace wear of at least 80% of the prescribed time. [39]

4.2.2 Chêneau Brace

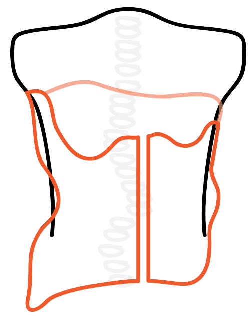


Figure 5: Schematic drawing of a possible configuration of the Chêneau brace (front view). The opening at the anterior side of the brace is clearly visible.

The Chêneau Brace was invented by J Chêneau in 1960. It was designed as a 3-dimensional correctional brace that opens on the anterior side. The brace is schematically shown in Figure 5. The Chêneau brace consists of both passive and active mechanisms. The passive mechanism is a 3-point system, which acts by hypercorrecting, elongating and unloading and consequently detoriating and bending the thorax. The active mechanism is a conjunction between the brace and the patient, since it is mainly influenced by 1) the vertebral growth, which acts as a corrective factor, 2) asymmetrically guided respiratory movements of the rib cage, 3) the repositioning of the spatial arrangement of the trunk muscles to provide physiological action, and 4) anti-gravitational effect. [22][40]

Validation
The Chêneau brace was proven to be quite efficient in treatment compared to other (a)symmetrical braces in several studies, with a success rate (stabilization of the curve or less than 6 degrees of curve progression) of 76% [41] or even 80% [42]. The latter was found during a prospective controlled trial, which had a quite homogeneous sample of patients that were at high risk for curve progression. [43]

Patient Compliance
Since the Chêneau brace is a rigid brace, it is interesting to look at the compliance of the patients wearing the device. Zaborowska-Sapeta et al. showed in their prospective study that the brace was "unwillingly accepted by adolescents because of aesthetic and functional reasons. The brace was considered, especially by girls, as an element making every day life activities difficult." [44] In 2006, Helfenstein et al. published an extensive research on patient compliance with the Chêneau brace, presenting an objective overall patient compliance of 67.5% (average compliance in terms of time actually wearing the device vs. the time the patient is prescribed to wear the device). The data was collected using temperature sensors placed inside the braces. They also showed that the subjective compliance of the patients was remarkably higher (94%). [45]

4.2.3 SpineCor Brace

The SpineCor was introduced in Montreal, Canada in 1993 by Christine Colliard and Charles Rivard. In contradiction to the other braces at that time, the SpineCor brace makes use of straps, resulting in a non-rigid brace that uses the patients motions to generate the actual correction forces. This force-controlled strategy as already highlighted in section 4.1.4, corrects the spine by influencing postural disorganization, muscular dysfunction, and unsynchronized spinal growth. [22] The brace consists of a rigid thermoplastic pelvic base, thigh and cross bands, a bolero and elastic corrective bands. [46] A

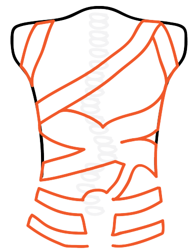


Figure 6: Schematic drawing of the SpineCor brace (front view). The configuration shows different elastic bands that are attached to the base-plate, which is located around the crotch.

schematic overview of the brace is shown in Figure 6. The SpineCor company also designed a newer version of the brace where the pelvic base, thigh and cross bands are replaced by a pelvic shorts. However, this newer brace design seems to be mainly focused on treating back pain in adults.

Validation
The SpineCor brace is designed to be a more tolerated and cosmetically accepted brace. It was proven to be effective for treating moderate curves, stabilizing or improving 64% of patients in a study respecting the SRS criteria, with only 18% of the patients reaching the surgery threshold. [47][22] Another, more recent study performed by Coillard, using a curve specific 'Corrective Movement© Principle' (CMP) showed even better results with 74.2% of 349 patients having received successful treatment, where this percentage represents the percentage of patients who have 5° or less curve progression. [33] Furthermore, the treatment was maintained after 2 years for 93.2% of 162 patients, showing this brace is quite effective for certain types of scoliosis. Szwed and Koban validated the brace as providing successful treatment in 76% of the cases, where success was defined as correction or stabilization of the scoliosis, with a maximum Cobb angle change of ±5°. [48]

Patient Compliance
The SpineCor system was designed to increase the patient compliance towards the brace in comparison to the rigid braces. It is less bulky, less heavy and presumably more comfortable to wear compared to the rigid braces available. It is more easy to 'hide' the brace underneath clothing, without requiring the patient to wear an over-sized t-shirt for example. One of the disadvantages of the brace is the fact that it complicates toileting, because of the pelvic base and the foundation structures that wrap around the pelvis and upper thighs [49], which leads to discomfort and decreased compliance towards the brace. Hasler et al. found a patient compliance score of 54% using an objective method with flexible temperature loggers. These loggers captured the brace wear time. The compliance score is then calculated by dividing the wearing hours with the prescribed 23 hours of brace wear. [50]

4.2.4 TriAC Brace

Similar to the SpineCor brace, the TriAC brace was also designed to improve cosmetic appearance and wearing comfort. It was developed by G.J. Nijenbanning and introduced together with dr. A.G. Veldhuizen in the Netherlands in 2001. The TriAC brace has its name derived from the three C's of Comfort, Control and Cosmesis, showing that these three principles are the main points of interest for this brace. [36] The brace is designed to apply continuous correction forces on the trunk, while allowing the patient to move freely despite wearing the brace. It consists of a thoracic part, including one of three force pads; a thoracic force pad. Next, it consists of a lumbar part, including the lumbar force pad and hip force pad. In between the thoracic and lumbar part, a flexible coupling is located at the opposite side of the thoracic force pad. A schematic overview of the TriAC brace is shown in Figure 7. The flexible coupling between the thoracic and lumbar part enables the patient to bend forward, backward and sideways, while making sure the corrective forces are maintained. [21]

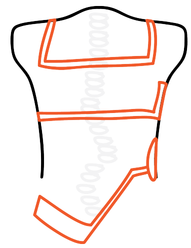


Figure 7: Schematic drawing of the TriAC brace (front view). The configuration shows the slender metal parts connecting a hip force pad with a lumbar force pad. Also a thoracic force pad is shown at the upper right side.

In contrast to the general 3-force-system correction strategy as shown in section 4.1.4, the TriAC brace also applies a forces in the frontal plane, as can be seen in figure 8. By doing so, the brace corrects the spine in the frontal plane similar to conventional braces, but in the sagittal plane it only acts on the thoracic region. Therefore, no pelvic tilt occurs, leading to improved flexibility without affecting the correction forces. [22]

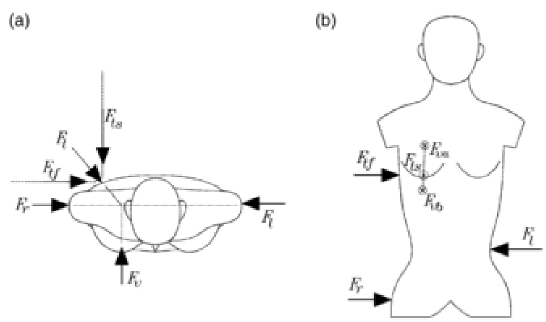


Figure 8: *(a) The 3-force system shown in the frontal plane, consisting of a* F_t *force close to the spine with a frontal* F_{tf} *and sagittal* F_{ts} *component. To obtain equilibrium the anterior force* F_v *is needed. (b) A possible location of this force* F_t *is near the breast, which is not preferred. Therefore two anterior forces are introduced, one above* F_{va} *and one below* F_{vb}*. The frontal component* F_{tf} *of the thoracic correction force forms a 3-force system together with* F_l *and* F_r*.* Reproduced from [36]

Validation

The TriAC brace has been validated by Roller and Nijenbanning. They found control or correction of the Idiophatic Scoliotic curves in 76% of the patients, with treatment success defined as a maximum curve progression of 5 degrees. They also state an improvement in the natural history of IS using the brace. [51] An important note is the fact that the brace is recommended to be used exclusively for the correction of lumbar curves.[52]

Patient Compliance

Patient compliance with the TriAC brace is found to be pretty high, due to its improved cosmetic appearance and comfort. Using the “Quality of Life Profile for Spine Deformities” survey, Zeh et al. reported high values for cosmetic effects (4.2 on a scale of 0-5, or 84%). Also the wearer time was indicated to be very high compared to conventional braces, which is also an important aspect in successful brace treatment. [52]

Table 2: Collected data on the four different brace designs

Brace Name	Year of research	Researcher	Success Rate	Patient Compliance
Boston	2013	Weinstein et al. [4]	93%*	82% [1]
	2002	Gepstein et al. [39]	81%**	80% [2]
Chêneau	2016	SY et al. [43]	80%**	
	2015	Maruyama et al. [41]	76%**	
	2006	Helfenstein et al. [45]		67.5% [3]
SpineCor	2008	Coillard et al. [33]	74.2%**	
	2012	Szwed et al. [48]	76%**	
	2010	Hasler et al. [50]		54% [4]
TriAC	2009	Roller, Nijenbanning [51]	76%**	
	2008	Zeh et al. [52]		84% [5]

* Successful treatment defined as a maximum curve of 50° when skeletal maturity is reached
** Successful treatment defined as a maximum curve progression of 5° at the end of treatment.
[1] Patient compliance measured subjectively using the Pediatric Quality of Life Inventory (PedsQL) questionnaire. Patients indicated an average 82-point score on a scale from 0-100.
[2] Patient compliance subjectively indicated by patients and their parents when asked how long the braces were worn each day. Score is a percentage of actually prescribed wearing time.
[3] Patient compliance objectively measured using temperature sensors placed inside the braces. Score is a percentage of actually prescribed wearing time.
[4] Patient compliance measured objectively using flexible temperature loggers. Score is a percentage of actually prescribed wearing time (% wearing hours/prescribed 23 h).
[5] Patient compliance subjectively measured using the "Quality of Life Profile for Spine Deformities" survey. Score indicates the average value for cosmetic considerations (4.2 on a scale from 0-5).

4.2.5 Brace Comparison

Even though we have collected the different success rates of the four presented scoliosis braces, this does not mean we can blindly compare these numbers with eachother, since they are all based on different studies, performed by different researchers. Not only does the definition of 'treatment success' change from research to research, also the length of the studies vary. Similarly we see a highly varying patient compliance rate for each of the presented braces. These percentages or scores do not only vary between braces, they are also very much depending on the kind of data collection as performed by the researcher in the particular study. Table 2 shows an overview of the braces, together with notes on what is actually investigated during the particular studies.

4.3 Brace Biomechanics

Not much is known about the biomechanics of different braces. As Rigo et al. showed in 2006, a part from the 'three point system' (as shown by Nijenbanning in Figure 2 (b), no other biomechanical model is really apparent or accepted in scoliosis bracing.[53] In 2008, Rigo and Weiss presented a novel way of looking at the different biomechanical aspects of a scoliosis brace by defining these for the Chêneau brace. [10] By looking at the different correction strategies provided by this brace, they defined a general framework with which clinicians will be quite capable of successfully designing and/or adjusting the brace for a specific patient. The analysis includes the derotation, deflection and sagittal normalization of the Chêneau brace. They argue that current scoliosis braces have not been designed or are not optimized for 3-dimensional correction. With the use of the above stated principles, the standards in brace correction could be improved, leading to an improvement of overall correction effect in 3 dimensions.

4.4 Brace Forces

Next to the correction strategies and the brace biomechanics, the forces that actually provide the correction to the spine are very interesting to know and understand. Currently, braces are usually designed or modified by experienced clinicians, makings small adjustments to a basic brace system in order to get an efficient patient-specific scoliosis brace. Even though this method works out well in practice, not much is known about the actual forces that are exerted by the brace onto the torso. Knowing what forces are needed for efficient design helps us in understanding how to design an effective scoliosis brace. The locations and magnitudes of these forces could then be translated into criteria for the design of a new scoliosis brace.

4.4.1 BraceSim ®

Since each patient has a specific scoliosis pattern, it makes sense to design braces specifically for each patient. The BraceSim® software is specially designed for simulating and evaluating different brace designs on a particular patients body. After gathering a 3D scan of the patient, clinicians are able to design a patient specific brace, which is then evaluated in the BraceSim® software. BraceSim® makes use of a biomechanical finite element model of the trunk, together with a CAD model of the patients spine. After designing the brace digitally, the user is able to set different strap forces (and inherently brace forces), after which the simulation is run for these specific strap forces. These simulations and the biomechanical finite element model were first designed by Aubin [54] and are still used today. A simulation result of the BraceSim® software is shown in Figure 9, where the BraceSim® software shows the 'original' spinal configuration, together with a simulated pressure mapping of the designed brace and its simulated treatment result over time.

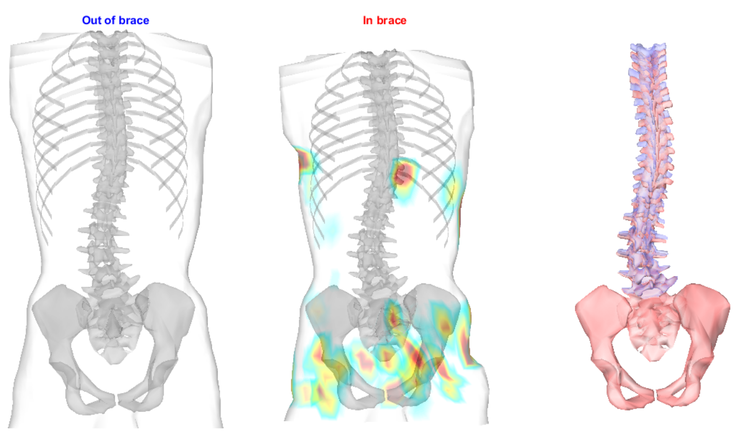


Figure 9: (a) Original state of scoliotic spine. (b) In brace simulation of the spine and the resulting pressures as caused by the particular brace design on the patients body. (c) Simulation result of the spinal configuration, showing the original state in red and the final configuration in blue.

Based on Nijenbanning [12], Nijssen and Ring [20] researched a pressure profile on a scoliosis patient with a thoracic curve of 20 degrees and a lumbar curve of 34 degrees in collaboration with Laboratoire d'Innovations CAO and Genie Orthopedique at the Ecole Polytechnique de Montreal. Using the BraceSim® software, a correctional profile was determined for this specific case study, where treatment was simulated using a general Boston brace design. The pressure profile and the forces can be seen in Figure 10.

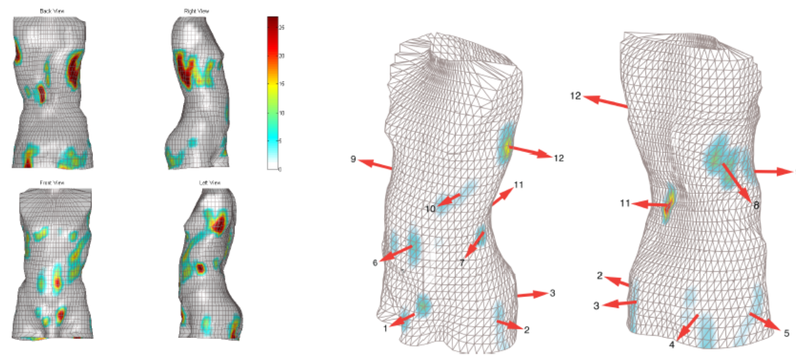


Figure 10: (a) Pressure profile resulting from a Boston type brace design simulation using BraceSim® software(b) Derived corrective forces from the pressure profile locations, reproduced from [20]

The exact data of the derived forces can be seen in Table 3.

Table 3: Forces derived from the pressure profile generated using BraceSim®. Reproduced from [17]

			Force Components (N)	
Force #	**Force Magnitude (N)**	x (front-back)	y (lateral)	z (vertical)
1	31	31	1.2	1.8
2	23	-4.9	-22	-2.5
3	31	24	-18	-0.1
4	32	30	10	0.4
5	36	23	27	-1.6
6	44	-44	-1.2	0.0
7	40	-36	-18	-1.7
8	76	49	55	16
9	38	-8.1	37	-2.3
10	22	-21	-4.8	-1.1
11	39	38	0.0	-9.4
12	50	-0.3	-50	-4.2

From the derived forces, Nijssen and Ring found only the forces 12, 9 and 8 to be actual corrective forces (in order of importance), the remaining forces are necessary to obtain equilibrium. Using this simulation results, researchers and clinicians are able to more specifically design a brace specifically for a patient.

4.4.2 Force measurement instruments

Another way of determining the forces needed for efficient treatment could be found by investigating for instance the Boston brace. We can try to measure the forces of the brace, which is specifically designed and/or adjusted for a patient, in between the brace and the body. By doing so, we can directly generate a pressure profile similar as shown in the section above, without being dependent on the simulation software. Naturally, this feels like a more objective measure to determine the correction forces induced by scoliosis braces, however we have to take into account that not all braces are adjusted in perfection to the patient, furthermore the correction forces may very well decrease over time when the scoliotic spine is corrected.

Van den Hout et al. used the electronic PEDAR measuring device (Novel, Munich, Germany) to measure the direct forces exerted by the pads in a Boston brace in 16 patients. The PEDAR device is designed as an in-shoe measuring system, which records static and dynamic pressure contribution under the planar surface of the foot. With the use of this device, they were quite able to measure the different forces applied by several Boston braces on different patients. The results show a significant larger mean corrective force through the lumbar brace pad in comparison to the corrective force through the lumbar pad. Furthermore, they found significant differences in the exerted forces when changes in posture occurred. [55] The latter indicates the earlier stated problems with these displacement controlled orthoses (see Section 4.1.4).

Romano et al. used a pressure distribution system, consisting of a sensor made out of a rectangular plastic sheet of 7.62 cm wide, 20.32 cm long and 0.1016 mm thick. [56] This device measures changes in resistance by having a conductive paste interposed between two insulating sheets. As they indicate, several parameters can influence the measurements, namely:

1. the magnitude of the forces to be measured;
2. the temperature during the measurements;
3. the time taken to perform the measurements;
4. the interval between two consecutive measurements; and
5. the sensor wear and tear

Using this system, after delicately calibrating to minimize the biasing factors as shown above, average pressure values were found at different locations of the Milwaukee brace during rest and during different exercises.

In 2012, Chou et al. presented their research on using finite element (FE) method modelling to determine pad positions in a boston brace for enhancing the corrective effect of a Boston brace. With the use of an advanced FE model, they simulated the pressures as applied by the brace when using different strap forces. To validate their model, they used an X-sensor pad (XSENSOR Technology Corporation, Florida, USA) to measure the contact pressure at both the lumbar and thoracic pads. [57] *Unfortunately the exact method of using this pressure-measuring system is not presented in this research.*

Measuring the forces inside a brace is not a very common research topic, however some researchers already investigated these forces in the 1970s. Galante et al. used specially designed dynamometers, which make use of strain gauges to measure the strain from which the applied forces can be calculated. [58] Even though it is interesting to see how this research was performed, the outcomes are of less importance today; the research focused on the mandibular and occipital supports of the Milwaukee brace. The braces used nowadays do not contain this 'neck support system' (see Figure 11) and consequently we can not use the measuring set-up used by these authors.

Giesberts, Sluiter and Hekman are currently using home-made inductive pressure sensors to measure the pressure at two distinct locations on the baby foot in between the baby foot and a cast. With the use of this data they try to figure out the stress-relaxation rate in the treatment of clubfoot, with which they will give recommendations in the treatment (Ponseti method) of this particular disease. [59][60] The sensors used are pretty small with a diameter of 10 mm and a thickness of alsmost 2 mm. Forces are measured periodically, extending the battery life time to a total of 12 days. With the use

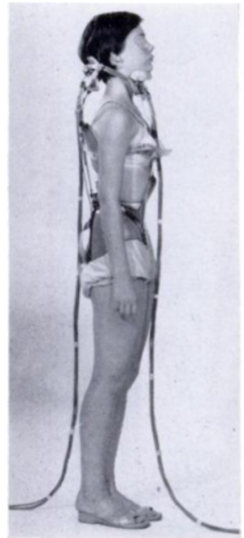


Figure 11: A patient wearing the Milwaukee brace, instrumented with dynamometers at the mandibular supports. Reproduced from [58]

of a temperature sensor, data is corrected for the influence of temperature. The sensor is shown in Figure 12.

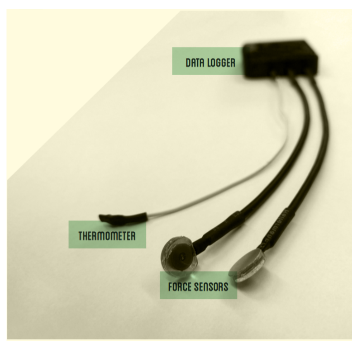


Figure 12: The sensors used by Giesberts et al. for measuring pressure in a cast for treating clubfoot (Ponseti method), reproduced from [61]

At this time, this measurement set-up is working properly; the sensors work, battery life is sufficient and no significant drift or hysteresis is observed. By using Cutinova as material in between the sensors and the skin, pressure marks were averted. Since this is still an on-going research, only preliminary results have been presented. [61] The sensors used by Giesberts, Sluiter and Hekman could also be used to measure the pressure on the skin as induced by a scoliosis brace at different locations. Similar to the research on clubfoot, it would be interesting to see how the pressure profile develops over time and how this pressure profile relates to spinal correction. Since the sensors used are home-made, no official results or data about maximum allowed force is available. An update from August 2016 showed a maximum recorded force close to 30N in a research on habitual toe walkers, further research using these sensors will be needed to deduce exact properties of the sensors.

5 Motion capture studies

5.1 Activities of Daily Living

As stated earlier, user compliance is of high importance for a brace to be effective for treatment or not. This compliance is highly correlated to the Activities of Daily Living (ADL) of the patient; the ADL indicate how well a person is able to perform basic household activities. Since most braces reduce the ability to move around freely, it is obvious that the perceived ADL is reduced as well. If a patient is aware of this reduced ADL, his or her compliance towards the brace decreases most likely and ultimately so the brace becomes less effective (since treatment success is highly correlated with wearing time of the brace, as shown in [4] [62] [63].

In order to translate the ADL to requirements and criteria they need to be quantified. Bible [64] quantified the motions of ADL using a decomposition of linear combinations of three primary motions; sagitall bending, lateral bending and twisting. The planes of the body used to describe the primary motions can be seen in Figure 13.

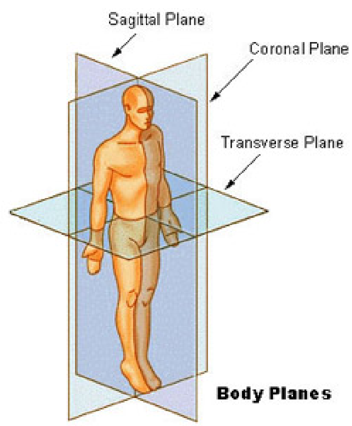


Figure 13: Planes of the body used to describe the primary motions, reproduced from [64]

Lateral bending and twisting can be described as bending in the coronal plane and the transverse plane, respectively. Sagittal bending is defined as the bending motion in the sagittal plane, and is often referred to as flexion/extension. The results for 15 ADL can be found in Table 4. What we see in table 4 is that the primary motions are of quite high importance in the total Range of Motions in daily life, which is mostly visible during simple motion tasks such as sit-to-stand motions or walking up or down the stairs.

Efforts have been made to determine whether or not the Activities of Daily Living (ADL) are reduced significantly when wearing different types of scoliosis braces. [33][63]. These studies have given rise to the design of a new type of brace, namely a brace with reduced rigidity, creating a more flexible design which ultimately improves the ADLs of the patient and so increases the patient compliance towards the brace. As shown earlier on in this research (Section 4.2.3 and 4.2.4), some more flexible braces have been on the market for a while now. These braces however did not meet the high correction rates compared to the Boston brace. It is believed that more rigid braces, which allow the patient to perform certain movements (which makes these braces flexible to some extent), allow for a better recovery and will increase the patient compliance towards the brace.

With the use of motion capture studies we can get a more thorough understanding of the movements of idiopathic adolescent scoliosis patients. Differences in motion and posture with or without brace,

Table 4: ADL and percentage of full range of the primary motions necessary to complete them, reproduced from [64]

	Average Percentage of Full Active ROM		
ADL	Flexion/Extension	Lateral Bending	Axial Rotation
Stand to sit	37	20	12
Backing Car	10	16	18
Reading	4	6	6
Feeding	5	8	9
Socks	22	19	14
Shoes	20	20	16
Sit to stand	39	14	10
Washing Hands	12	15	12
Make-up	7	11	8
Squatting	52	31	18
Bending	59	29	18
Walking	11	19	19
Up stairs	13	22	20
Down stairs	11	21	18
Maximum	59	31	20
Average	22	18	14

motion profiles of (patients with) different types of scoliosis and a range of patient dimensions can be found by cleverly using and designing motion capture studies.

5.2 Types of motion capture studies

Different systems can be used to do motion capture studies. Since this study is focusing on scoliosis, we focus on motion capture studies for the back and/or torso. Several techniques have been developed to allow for capturing human motions in different disciplines, namely:

- Goniometric measurements
- Accelerometers
- Film digitization
- Optoelectronic motion analysis

These motion capturing methods all have their specific advantages and disadvantages, which will be discussed below.

5.2.1 Goniometric measurements

Goniometric measurements are generally used to measure the range of motion (ROM) of different joints of the body. A goniometer is placed on or over the particular joint, which allows the user to read the angle of rotation. With these measurements of the joint angles, the ROM can be or estimated. Most commonly a two-arm goniometer is used, because of its simplicity and relatively low cost.[65] The simplicity of using a goniometer also leads to some disadvantages of the technique. For example the starting position, centre of rotation of the joint, limb axes and horizontal positions can only be estimated visually (the latter using a spirit level embedded in the tool).[66] Another disadvantage is the action plane of goniometers; it is mainly used to detect movement in one plane, while spinal movements are often 3 dimensional. Another disadvantage is the visibility of the goniometer itself; when applying a goniometer to small joints, it might become very hard to read the rotation angles. The latter in combination with the limited Degrees Of Freedom (DOF) of the device will limit the use of these devices for spinal measurements, especially when one is interested in between-vertebrae movements. Clinicians generally only use goniometers for the spine to measure the total Range of Motion, as shown in Figure 14.

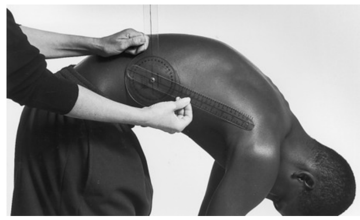


Figure 14: Goniometer alignment at end ROM of lumbar flexion, reproduced from [67]

5.2.2 Accelerometers

With the use of accelerometers, several researchers have been able to measure physical activity, since these accelerometers measure body movements in terms of acceleration. Accelerometers can be either uniaxial, biaxial or triaxial depending on their sensitivity to 1,2, or 3 orthogonal planes, respectively.[68] Next to the linear acceleration, accelerometers can also be used to measure the tilt angle relative to gravity. Examples of such can be found in smartphones, continuously measuring the orientation of the device. Due to the cost effectiveness and size of accelerometers, multiple sensors can be placed on the spine. The tilt measurement allows researchers to successfully measure the Ranges of motion of the spine as shown by Luinge and Veltink in [69] and [70] and by Alqhtani et al. [71]. Using 6 triaxial accelerometers, the Alqhtani et al. measured the ROM of 5 adjacent spinal segments by attaching the sensors to the T1, T4, T8, T12, L3 and S1 spinous processes. With the use of these accelerometers, they present reliable data with very small errors, showing that this technique is suited well for spinal motion analysis. Additionally, they give insight into the specific contributions of different spinal regions to the primary motions. [71]

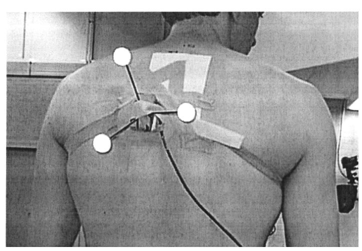


Figure 15: IMU as placed on the back of the trunk, at the T4/T5 level just left of the spine. Three reflective markers were placed on the IMU housing in order to measure the reference orientation of the accelerometer using a VICON system as a reference, reproduced from [70].

Since accelerometers are on its own only capable of measuring linear acceleration, measurement devices are often equipped with both accelerometers in combination with gyroscopes and magnetometers. These devices, inertial measurement units (IMUs), use the linear acceleration components as collected by the accelerometers together with the rotational rate captured by the gyroscope and the heading data collected by the magnetometer are able to detect changes in roll, pitch and yaw. By doing so, IMUs are capable of accurately describing the orientation of affixed bodies. IMUs come in different sizes and shapes, with quite accurate measurements for small, relatively cheap devices. [72] One of the disadvantages of IMUs is the dependency on external devices to deliver power and/or collect the data. The wiring to these devices could potentially introduce error to the measurements. Wireless devices are available, but come in bigger size and with increased weight.

Luinge and Veltink used an IMU sensor together with used an optoelectronic motion system, in order to validate the tilt angle measurements as received from the IMU. Their setup is shown in Figure 15.

5.2.3 Film digitization

Film digitization, or cinematography, makes use of the video recording of an actual movement. By applying a certain frame rate, several distinct images are created, which altogether show the movement sequence. From these images the key data can be indicated or extracted, such as the joints or other anatomical landmarks of interest. By doing so for every created still image (one every millisecond for instance), the movement data can be extracted and plotted against time. [73] The process of film digitization is time-consuming and inaccuracies are easily introduced, especially with low-resolution recordings. However, since image recognition software proves to be more and more accurate, developments and improvements in this area could very well be a reality in the coming decades.

Using digitization, research has shown to be able to analyze the spinal motion in several occasions. Okawa et al. used a videofluorscopy technique to measure the motion of the entire lumbar region of the spine, comparing healthy subjects with patients with chronic low back pain and degenerative spondylolisthesis. [74] Davis et al. used a similar technique together with a device that assists the patient to bend using a prescribed arc motion platform to measure the Ranges of Motion of patients and compare these to traditional methods using radiographs. [75]

5.2.4 Optoelectronic motion analysis

By using optoelectronic motion systems, we make use of markers affixed to bony landmarks or other locations of interest on the (human) body. Usually these optoelectronic systems use one of the two different types of markers; retro-reflective markers that reflect infrared light back to the motion cameras or active markers (LED diodes) that produce light which is tracked by the motion cameras. [76] The data is recorded using specialized software on a computer, showing the trajectories of the separate markers in 3-dimensional images and recordings. The associated software calculates the orientation of each marker in space and this information is usually saved into a .csv-file, which allows researchers to post-process the data in programming software like MATLAB by MathWorks®. Well-known advanced systems of brands like VICON or Optitrack allow for accurate measurements using multiple cameras in both big and smaller spaces. As shown by Merriaux et al. static experiments in a VICON set-up show a very small absolute positioning error of 0.15 mm with an even lower variability of 0.015 mm. [77] They also show hat the accuracy drops in dynamic experiments, due to several reasons, which include marker size and the speed of movements. It is therefore important to properly tune the sampling rate of the system and to carefully choose the marker size when designing the protocol for a new motion study.

In 2004, Cerveri et al. designed a specific marker set to measure only the vertebra in the lumbar region of the spine. [78] They placed tiny, 2mm diameter markers on the spinous processes of the lumbar region (T11 through S1), as well as their transverse processes, as shown in Figure 16. Using this marker set, they were able to describe the general motion of the spine. However, they had large difficulties in determining the motions between the vertebrae due to skin movement artifact. The movement of the skin has increasing influence on the captured data when the markers are getting smaller and smaller.

Figure 16: Marker set as used by Cerveri, reproduced from [79]. Original source: [78]

Next to the accuracy, another advantage of optoelectronic motion systems is the ability to track lots of markers simultaneously at a very high sample frequency, creating a digital data set which moves (almost) exactly similar to the actual subject. When using these systems one has to take into account the possible occlusion of markers; the cameras should be carefully positioned and possibly more cameras are needed to track every marker over the entire experiment time. Furthermore the cameras and software are very expensive to buy, with prices of $10.000 per camera or even more. Another disadvantage of the system is the dependence on a controlled environment; camera set-up and calibration is of high importance and any reflective attributes in the environment have to be blocked in order to collect the data most accurately.

5.3 Motion analysis in Scoliosis research

Motion capture studies in conjunction with scoliosis have been performed by several researchers.

In 2000, Masso and Gorton designed a marker set to measure the static posture of scoliosis patients. [80] The marker set they used did not show any spinal motion, since markers on the trunk were placed only on the right and left anterior superior iliac spine, the left and right ninth rib and on the first sacral vertebra (Figure 17). This research shows that optoelectronic motion systems can also play a role in static measurements, such as postural analysis or local asymmetries of scoliosis patients.

Figure 17: Marker set as used by Masso and Gorton, reproduced from [79]

In 2002, Engsberg et al. measured the spinal range of motion pre- and post spinal fusion, using an optoelectronic motion analysis system. [81] They combined a widely used clinical gait marker set with an additional marker set along the spine (Figure 18, to evaluate the global motion of the spine as well as the motion of specific regions of the spine. [79] The patients were asked to perform a series of bending exercises; spinal flexion and extension, lateral bending to both sides and axial rotation of the spine. Each patient was tested two months prior the spinal fusion and then again one year and two years after the spinal fusion. After evaluation and exclusion of motion data of 6 patients due to marker obstruction, the results showed significantly reduced lateral motion and spine flexion. For the non-fused segments no significant change was noticeable, showing that the spinal fusion only leads to reduction in motion in the actually fused regions.

Figure 18: Marker set as used by Engsberg, reproduced from [79]

In 2003, Frigo et al. defined specific marker places on bony landmarks (Figure 19. With the use of a kinematic computation model, previously used in multifactorial gait analysis, they were able to calculate the angles of kyphosis, lordosis, and coronal plane angles to represent Cobb angle measurements, as well as shoulder tilt during gait of healthy subjects. [82] They found that the spine angles in a normal population showed a deviation of 5 deg. Even though their research did not include scoliosis patients, it provides a good base for measuring the movement of the different segments of the spine and shoulder girdle in young females.

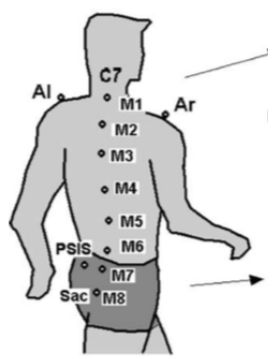


Figure 19: Marker set as used by Frigo et al., reproduced from [82]

In 2006, Skalli et al. used a similar marker configuration as the one used by Engsberg et al. [81], to measure the contribution of the pelvis to the spinal motion before and after spinal fusion. In this research, the marker set was used in a dynamic analysis, which included flexion and extension, lateral bending and axial rotation. [83] [79] In addition to Engsberg, Skalli also showed significant differences in the spine range of motion pre and post spinal fusion, but he also found significant differences between the healthy control group and the patient group. Additionally, they found that most patients use the pelvis to compensate for this reduced spinal motion both prior and after spinal fusion.

In 2008, Wong and Wong presented a smart garment that has been developed as a portable and user-friendly trunk posture monitoring system. With the use of this garment, shown in Figure 20 information of the trunk posture can be collected and feedback can instantly be provided to the user if needed. With the use of this relatively cheap system, consisting mainly of accelerometers, a quite accurate result was found.[84]

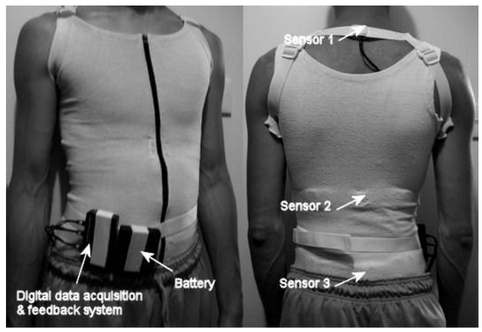


Figure 20: A patient wearing the smart garment, reproduced from [84]

In 2012, Shaw et al. presented a way of using the iPhone for measuring the Cobb angle.[85] With the use of the accelerometers and inclinometers in mobile phones, clinical measurement applications, such as Cobb angle measurement are possible. Their research shows the applicability of measuring these angles directly from a radio-graph using an iPhone. They show that these measurements are equivalent to using a manual protractor (a goniometric tool) , with reduction of measurement time of about 15%.

Currently, the most common way of investigating children with AIS and to see the progress of their spinal deformities is through the use of repeated radiography. This type of radio-graphic imaging might bring some unwanted consequences induced by the ionizing radiation, as Solomito argues. [79] In 2011, Solomito presented his research, proving the ability to measure the sagittal vertical alignment, coronal vertical alignment, the proximal angle, distal angle, the angle of lordosis and the angle of kyphoses by using 10 retro-reflective markers. He placed these markers on the right and left clavicles close to the sternal notch, on the spinous process of C7, T7 and L3, on the 10^{th} rib on the right and left side, superior and medial of the floating ribs, as shown in Figure 21. By adding force plates he says it is also possible to determine both spinal asymmetries in the frontal and transverse plane and the balance of the spine for each subject. Furthermore he uses Euler rotation sequences to determine the kinematic profiles of the spine. Solomito also investigated the range of motion for three different spinal segments; the entire spine, the upper spine (C7 to T7) and the lower spine (T7 to S1). With these measurements he notes differences in motion profiles between the healthy test group and the scoliotic patient group.

Figure 21: Marker set as used by Solomito, reproduced from [79]

In 2016, Galvis et al. investigated the influence of AIS on the spinal kinematics. [86] Position data at the manubrium, T1, T3, T6, T10, L1, L3 and S1 was collected using electromagnetic 6DOF motion sensors of the Trakstar system (Ascension Technologies, Burlington, VT). Each participant was asked to perform a maximum flexion and extension bend, left and right lateral bend and left and right anterior-lateral flexion bend at a self-selected speed. From the collected position data, angles between spinal regions was calculated. Galvis et al. showed that measurements using electromagnetic motion sensors gives accurate results and allows for comparison between healthy subjects and scoliosis patients. Although expected to see a reduced mobility in the scoliosis group, their results show a general increased mobility for this group compared to the healthy subjects.

In addition to the configurations performed by different researchers as shown above, advanced systems to externally investigate spinal deformities are also available. One of such systems is the DIERS Formetric 4D system, which removes the need of palpating by taking ex-vivo non-marker measurements using a projector and a camera, as shown in Figure 22.

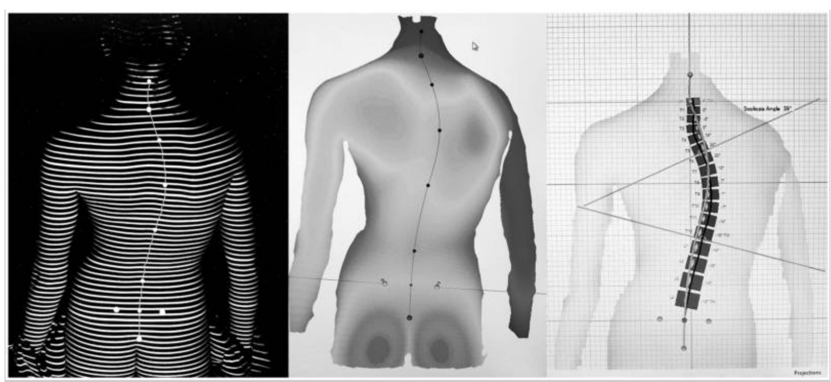


Figure 22: Topographic analysis and software output from the Formetric 4D device, reproduced from [87]

This system has shown to be accurate in estimating Cobb angles, with predictions correlating strongly to measurements from radiographs. [87]

Simi® Reality Motion Systems developed several motion analysis systems. In their Aktisys 3D Spine Analysis Protocol,they show that commonly 5 markers are needed for spinal examinations; one on each shoulder, one on each superior iliac process and one on the C7 vertebra. With the use of this set-up

shoulder rotation and obliquity, scoliotic tendencies, deviation of the spine from the vertical, pelvic rotation and pelvic obiliquity can be analyzed. For a segmental analysis, markers are placed on the lumbar, thoracic and cervical vertebrae (C4, T1 and L1) together with two markers on the PSISs of the pelvis. With the use of the segmental analysis the relative movement of each of the spinal segments whilst in motion can be determined. [88]
Simi® also developed a Motion 3D System, with which researchers can accurately measure the intricacies of the spine's movement. The system makes use of markers placed on the spinal processes of each vertebra, providing a detailed description of the dynamics of the vertebrae and their relative movement to one another. [88]

5.3.1 Kinematic Mapping by Nijssen and Ring

Nijssen used motion capture systems to analyze the kinematic mapping of the human torso, based on the method Ring [89] presented to describe motion between rigid rings around the body as a screw motion. [20] Motion capture markers were placed on specific places on the spine on the S1,L4,L1,T8 and T4 vertebrae. For each marker, two additional corresponding markers were placed at the same height on the sides to create a triangular shape in order to approximate a rigid body, as can be seen in Figure 23.

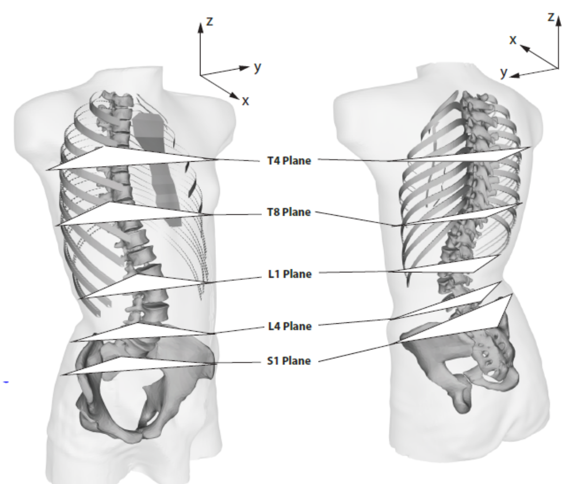


Figure 23: Approximated rigid body planes on the human torso, reproduced from [20]

The placement of these markers on the spine was chosen to have an even distribution along the spine. Since the S1 vertebra is fused with the pelvis, the S1 marker acts as a constraint together with the markers on the pelvis. For each triangular shape, the assumption of rigid bodies was made, and the motion was determined using screw motions for one specific subject. With the results from this motion study, Nijssen and Ring evaluated the three primary motions that decompose the ADL (see section 5.1)[64]. For each one of these primary motions, they show the resulting average screw locations and screw direction between the triangular planes, as can be seen in Figure 24.

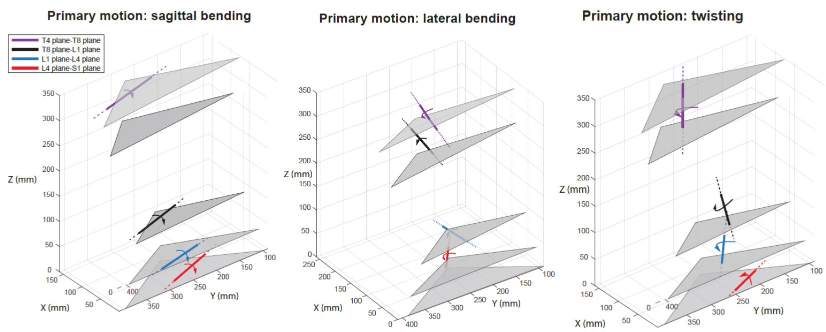


Figure 24: The resulting average rotation axes in 3D space, reproduced from [20]

.

The magnitudes of each angle of rotation around the screw axis are shown in Table 5.

Table 5: Measured rotations occurring between the triangular segments for the tree primary motions. The last column gives the motion in percentages relative to the largest occurring rotations per primary motion. Reproduced from [20]

Motion	**vertebrae**	**maximum angle**	**percentage**
Sagittal (S)	S1-L4	27.58°	84%
	L4-L1	32.42°	99 %
	L1-T8	32.69°	100 %
	T8-T4	10.75°	33 %
Twist (T)	S1-L4	9.32°	18%
	L4-L1	32.17°	63 %
	L1-T8	48.32°	95 %
	T8-T4	50.92°	100 %
Lateral (L)	S1-L4	7.57°	17%
	L4-L1	35.36°	78 %
	L1-T8	45.33°	100 %
	T8-T4	8.69°	19 %

The planes were assumed to be rigid, however the planes stretch over time as can be seen in Table 6.

Nijssen [20] concludes that between L4 and the T8 vertebrae the most lateral bending occurs, between T11 and S1 the most sagittal bending occurs and between the L1 and T4 the most twisting occurs. However, these conclusions are drawn while assuming that: the three motion markers form a rigid body, the motion of the human torso can be described by only four segments and the small pitches of the screw motions can be neglected. Without a pitch a screw motion can be represented by the rotation axis only.

Table 6: Stretching of the planes. Reproduced from [20]

Motion	Plane	Back (mm)	Left (mm)	Right (mm)
Sagittal (S)	S1-L4	16.56	14.6	4.46
	L4-L1	31.6	10.64	12.18
	L1-T8	24.3	7.87	8.33
	T8-T4	7.14	10.78	6.07
Twist (T)	S1-L4	2.55	5.88	3.31
	L4-L1	5.03	16.65	23.31
	L1-T8	9.99	37.48	20.07
	T8-T4	2.92	17.83	8.84
Lateral (L)	S1-L4	2.87	8.11	6.82
	L4-L1	8.97	26.8	54.03
	L1-T8	7.88	53.86	49.8
	T8-T4	4.28	10.84	9.13

5.4 Marker set-up

After having decided where to put the tracking markers and what locations need to be captured, the type of marker configuration needs to be chosen, together with a decision how to attach the markers to the body. A practical set-up is required to easily track the movements of different patients, without having to take long preparation time.

In 2010 Tranberg presented his thesis work on the influence of soft-tissue artefacts on analyses of body motions based on optical markers. He showed some skin motion artefacts influencing the recorded data and suggested 'further comparative studies with simultaneous use of skeletal and superficial skin markers to explore this issue further'. [90]

As shown earlier in section 5.2.4, Cerveri et al. also had difficulties measuring the motions between the vertebrae due to skin movement artefacts. [78]

Rast et al. concluded that adding markers and using a point cloud algorithm did not improve the between-day reliability of the recordings of trunk movement; their hypothesis of using redundant number of markers and an optimization algorithm will even out errors due to marker placement and soft tissue artefacts was proven to be not true. [91]

5.4.1 Rigid marker set-up

Markers can be adhered directly on the patient's skin using adhesive glue, but we can also use rigid marker set-ups to reduce the soft tissue artefact. In their study on different technical marker sets for measuring 3D pelvic motion during gait, Bruno and Jarden used a rigid thermoplastic plate with four non-collinear retro-reflective spherical markers fixed to the surface of this plate, which was mounted on the lateral aspect of each thigh using the Superwrap™ (Fabrifoam Inc., Exton, PA, USA). [92] Their set-up and the use of this plate is shown in Figure 25.

Figure 25: Marker setup used by Bruno and Jarden including four non-collinear retro-reflective markers mounted on a rigid thermoplastic plate, attached to the body using the Superwrap™ (Fabrifoam Inc., Exton, PA, USA), reproduced from [92]

In 2015, Lee and Jung evaluated marker placement for hand movements objectively using a static hand mock-up with different marker attachment methods. They show a significant difference between different marker sets for dynamic evaluation. This is why they recommend to use three markers per segment or a cluster marker, because they are less affected by skin movements. [93] Such a cluster marker can also be seen in Figure 25, where four different markers are clustered on a rigid thermoplastic plate, which in its turn is attached to the body using a elastic foam band.

De Baets et al. (2013) used clusters of 3 or 4 markers to acquire scapular rotation movements after stroke. These markers were placed on tripods or cuffs, depending on the locations on the body. [94]

Schmid et al. (2016) combined two marker sets (The Plug-in Gait full body and the IfB marker set) to evaluate and quantify the spinal kinematics of AIS patients during gait.[95] With the use of these marker sets they were able to accurately describe the biomechanical behavior of the spine during gait. The combined marker sets are shown in Figure 26.

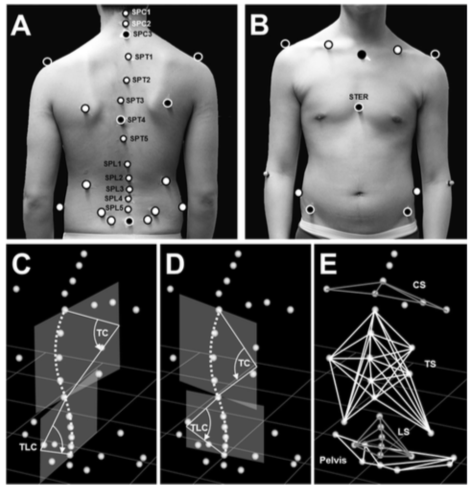


Figure 26: **(A,B)** photograph of the combined Plug-in Gait full body and IfB marker sets. White circles with black filling represent markers used for both models, black circles with white filling represent markers used for the IfB model, and markers with black circles where only used for the Plug-in Gait full body model. (SPC1-3: spinous processes of C3, C5 and C7; SPT1-5: spinous processes of T3, T5, T7, T9 and T11; SPL1-5: spinous processes of L1–L5). **(C,D)** Visual representation/definition of the thoracolumbar/lumbar and thoracic curvature angles in the sagittal plane (C) and frontal plane (D). **(E)** Visual representation of the rigid pelvis, lumbar, thoracic and cervical segments. Figure reproduced from [95]

The data derived using a VICON 12-camera system was used to quantify several kinematic aspects of the spine. Using the markers placed over the spinous processes for example, the researchers determined the spinal curvature. For instance, markers SPT1-SPT5 and SPT5-SPL5 (where SP = Spinous Process, T = thoracic and L = lumbar) defined the sagittal thoracic and lumbar curves, respectively. In the frontal plane the toracic and thoracolumbar/lumbar curves were obtained using markers SPT1-SPT5 and SPT5-SPL4. [95]

6 Conclusion

This literature overview gives insight into the work performed in the field of adolescent idiopathic scoliosis braces and kinematic characterization/motion studies of adolescent subjects diagnosed with a scoliotic deformation of their spine.

It has been shown that the common treatment for scoliosis is through the use of a certain type of scoliosis brace. These scoliosis braces have been developed and evaluated over the last decades, with different correction profiles and materials leading to varying results in terms of inhibition of curve progression and patient compliance towards the brace. Different types of brace categorizations can be used when talking about scoliosis. For this project a combination of the control strategy and rigidity categorizations seems to be the most relevant. The two mostly used rigid scoliosis braces have been presented together with two rather recently developed flexible braces. It has been shown that a comparison of one brace to another is not easily established, since braces were validated using different types of research and no standardized method is found. Furthermore, the patient compliance towards the brace is shown to vary a lot from brace to brace, with values ranging from 84% for the TriAC brace, to 54% for the SpineCor brace.
Next to the available braces, the biomechanical aspects of scoliosis braces have been highlighted, with different techniques to measure the forces as exerted by the braces in between the brace and the patient's skin. It has been shown that several techniques exist, like the BraceSim software which is used to simulate the (needed) forces on the human body. Not only the brace forces have an influence on the overall functioning on the brace, also the patient's compliance towards the brace is of importance. This compliance is highly influenced by the comfort experienced by the patient while wearing the brace. The activities of daily living (ADL) are an important indication of the abilities of the patient while wearing the brace and the compliance of the patient towards this brace. The ADL can be quantified using a decomposition of linear combinations of the three primary motions; sagittal bending, lateral bending and coronal twisting.
To measure the ADL on different subjects, motion capture studies can be performed. These different types of motion capture studies aim to characterize the spinal kinematics during for instance the three primary motions as described above. Motion capture techniques can be divided into different categories; goniometric measurements, accelerometers, film digitization, optoelectronic motion analysis and videofluoroscopy. From these techniques, the optoelectronic motion analysis is the mostly researched and used method in scoliosis research due to its relative accuracy and ability to detect small motion changes. Most generally, researchers attach retro-reflective markers to bony landmarks or points of interest, which are tracked by infra-red light emitting, high resolution motion cameras.
It has been shown that no analogous method is used in available literature; due to different research goals, each researcher has used varying marker set-ups and analysis methodologies. Subsequently it is not possible to compare the results of these analyses on a one-to-one base; especially when investigating the spinal movements, where the slightest changes in (marker) set-up could result in highly different results. Therefore we can draw the conclusion that the currently available literature does not offer an analogous method to determine spinal ranges of motion in (scoliosis) patients or healthy subjects; pre-eminently it provides us insight in the different methodologies to perform these kind of researches.

In order to design a new scoliosis brace, which aims to combine the performance in terms of treatment of currently used rigid braces with the increased ranges of motion as allowed by more flexible braces, it is clear that we can learn a lot by looking at the currently available braces. The new brace should use the rigidity of the Boston brace at the locations where correction is needed. To determine at which locations this new brace design should bend, a motion analysis focusing on the motions of different spinal segments is needed. To perform such a motion analysis, the preferred option would be to use an optoelectronic system, since this seems to be the most common method in scoliosis research because of its accuracy.

7 Discussion

The goal of this study was to give recommendations on how to design a new type of scoliosis brace, combining the performance of rigid braces with the flexibility of non-rigid braces.

For this study, several search terms have been used to find articles about scoliosis braces and motion studies focusing on the spine. With the use of these search terms, the same articles should be found and so, a repeated literature search should lead to a similar outcome. Next to the articles found directly in the literature search, a lot of articles have been found when referred to by other articles, or when recommended by search engines or software. It is not known whether the same articles would be recommended to other users of these platforms when consulting the found literature.

Scoliosis is a patient-specific disease, meaning each spinal curvature is unique and changes from patient to patient. Therefore the treatment is also highly patient-specific and so scoliosis braces should be specifically designed for the patient. Especially when comparing scoliosis braces in terms of treatment success, this might bias the results, since some types of scoliosis are easier to treat than others. Furthermore each of the scoliosis braces is adjusted to the patient and so no scoliosis brace is exactly similar.

When measuring the patient compliance towards the brace, this was mainly calculated using the hours of brace wear of the patient. Even though this number is an indication of patient compliance, it does not take other aspects into account that might influence the patient compliance, such as the cosmetic value. On the contrary, when quantifying the patient compliance objectively, the hours of brace wear seems to be the only available measurement. Furthermore, since objective patient compliance measurements are only available for a few brace designs in literature, it is very hard to compare these compliance values for different braces with each other.

Although brace forces are very interesting from an engineering perspective, all braces are currently designed by clinicians based on their experience. Generally the straps are used to increase the exerted forces on the body, together with modified pressure pads on the inner side of the brace. This is inherently the reason why not much literature on brace forces is available.

It is shown that motion capture experiments are capable of measuring patient-specific motions quite accurately. Since all presented motion analyses in research have had different goals, it is quite hard to compare the results from these experiments. Different camera set-ups, marker set-ups and marker locations might have a big influence on the captured results. Also the available resources for each researcher is a contributing factor to the differences in the presented research projects. When developing and validating a (new) measurement method, this has to be taken into account.

Patient demographics have not been taken into account in this research, which is mainly focusing on scoliosis patients in the United States of America. Since not all research presented has been performed in the US, or using US subjects, results might be biased or hard to compare.

This study was performed to give an overview of the produced braces and the performed motion analysis studies in scoliosis research. Most of the used literature was written in the last two decades, while some of the referred papers date from over 40 years ago. Even though efforts have been made to validate these old sources, changes in research methodologies over the years could have a big influence when these old projects would be performed today.

8 References

[1] N. H. Miller, "Cause and natural history of adolescent idiopathic scoliosis," 1999.

[2] S. L. Weinstein, L. A. Dolan, J. C. Y. Cheng, A. Danielsson, and J. A. Morcuende, "Adolescent idiopathic scoliosis," Lancet, vol. 371, pp. 1527–1537, 2008.

[3] P. Deacon, B. M. Flood, and R. a. Dickson, "Idiopathic scoliosis in three dimensions. A radiographic and morphometric analysis.," The Journal of bone and joint surgery. British volume, vol. 66, no. 4, pp. 509–512, 1984.

[4] S. L. Weinstein, L. A. Dolan, J. G. Wright, and M. B. Dobbs, "Effects of bracing in adolescents with idiopathic scoliosis.," The New England journal of medicine, vol. 369, no. 16, pp. 1512–21, 2013.

[5] A. L. Nachemson, J. E. Lonstein, and S. L. Weinstein, "Report of the Prevalence and Natural History Committee of the Scoliosis Research Society.," in Denver: Scoliosis Research Society, 1982.

[6] S. Aaro and C. Ohlund, "Scoliosis and pumonary function," Spine, vol. 9, no. 2, pp. 220–2, 1984.

[7] R. B. Winter, W. W. Lovell, and J. H. Moe, "Excessive thoracic lordosis and loss of pulmonary function in patients with idiopathic scoliosis," Journal of Bone and Joint Surgery, vol. 57, no. 7, pp. 972–977, 1975.

[8] M. A. Asher and D. C. Burton, "Adolescent idiopathic scoliosis: natural history and long term treatment effects.," Scoliosis, vol. 1, no. 1, p. 2, 2006.

[9] I. A. F. Stokes, L. C. Bigalow, and M. S. Moreland, "Three-dimensional spinal curvature in idiopathic scoliosis," 1987.

[10] M. Rigo and H.-R. Weiss, "The Cheneau Concept of Bracing - Biomechanical Aspects," The Conservative Scoliosis Treatment, vol. 135, pp. 303–319, 2008.

[11] S. Negrini, S. Minozzi, J. Bettany-Saltikov, N. Chockalingam, T. B. Grivas, T. Kotwicki, T. Maruyama, M. Romano, and F. Zaina, "Braces for idiopathic scoliosis in adolescents," jun 2015.

[12] G. Nijenbanning, Scoliosis Redress: Design of a force controlled orthosis. PhD thesis, Universiteit Twente, Enschede, 1998.

[13] J. H. Hacquebord and S. S. Leopold, "In brief: The risser classification: A classic tool for the clinician treating adolescent idiopathic scoliosis," Clinical Orthopaedics and Related Research, vol. 470, no. 8, pp. 2335–2338, 2012.

[14] S. L. Chan, K. M. Cheung, K. D. Luk, K. W. Wong, and M. S. Wong, "A correlation study between in-brace correction, compliance to spinal orthosis and health-related quality of life of patients with Adolescent Idiopathic Scoliosis," Scoliosis, vol. 9, no. 1, p. 1, 2014.

[15] L. Rivett, A. Rothberg, A. Stewart, and R. Berkowitz, "The relationship between quality of life and compliance to a brace protocol in adolescents with idiopathic scoliosis: a comparative study," BMC Musculoskeletal Disorders, vol. 10, no. 1, p. 5, 2009.

[16] C. Nnadi and J. Fairbank, "Scoliosis: A review," Paediatrics and Child Health (United Kingdom), vol. 24, no. 5, pp. 197–203, 2014.

[17] J. B. Ring, C. J. Kim, J. Evans, and C. Beal, A PASSIVE BRACE FOR THE TREATMENT OF SCOLIOSIS UTILI PhD thesis, Bucknell University, Lewisburg PA, 2017.

[18] T. B. Grivas, J. C. de Mauroy, G. Wood, M. Rigo, M. T. Hresko, T. Kotwicki, and S. Negrini, "Brace Classification Study Group (BCSG): part one–definitions and atlas," Scoliosis and Spinal Disorders, vol. 11, p. 43, dec 2016.

[19] S. Negrini, F. Zaina, and S. Atanasio, "BRACE MAP, a proposal for a new classification of braces," Studies in Health Technology and Informatics, vol. 140, pp. 299–302, 2008.

[20] J. P. A. Nijssen, Design and Analysis of a Shell Mechanism Based Two-fold Force Controlled Scoliosis Brace. PhD thesis, Delft University of Technology, 2017.

[21] G. J. Bulthuis, A. G. Veldhuizen, and G. Nijenbanning, "Clinical effect of continuous corrective force delivery in the non-operative treatment of idiopathic scoliosis: A prospective cohort study of the triac-brace," European Spine Journal, vol. 17, no. 2, pp. 231–239, 2008.

[22] F. Zaina, J. C. De Mauroy, T. Grivas, M. T. Hresko, T. Kotwizki, T. Maruyama, N. Price, M. Rigo, L. Stikeleather, J. Wynne, and S. Negrini, "Bracing for scoliosis in 2014: State of the art," European Journal of Physical and Rehabilitation Medicine, vol. 50, no. 1, pp. 93–110, 2014.

[23] H. G. Watts, J. E. Hall, and W. Stanish, "The Boston brace system for the treatment of low thoracic and lumbar scoliosis by the use of a girdle without superstructure," Clin Orthop Relat Res., vol. 126, pp. 87–92, 1977.

[24] C. T. Price, D. S. Scott, F. E. Reed, and M. F. Riddick, "Nighttime bracing for adolescent idiopathic scoliosis with the Charleston bending brace. Preliminary report.," Spine, vol. 15, no. 12, pp. 1294–9, 1990.

[25] S. Negrini, T. M. Hresko, J. P. O'Brien, and N. Price, "Recommendations for research studies on treatment of idiopathic scoliosis: Consensus 2014 between SOSORT and SRS non-operative management committee.," Scoliosis, vol. 10, no. 1, p. 8, 2015.

[26] T. B. Grivas, A. Bountis, I. Vrasami, and N. V. Bardakos, "Brace technology thematic series: the dynamic derotation brace.," Scoliosis, vol. 5, p. 20, 2010.

[27] D. Mauroy, J. C. D. Mauroy, C. Lecante, and F. Barral, "" brace technology " thematic series-the lyon approach to the conservative treatment of scoliosis," Scoliosis, vol. 4, no. March, 2011.

[28] J. E. Lonstein and R. B. Winter, "The Milwaukee brace for the treatment of adolescent idiopathic scoliosis. A review of one thousand and twenty patients.," The Journal of bone and joint surgery. American volume, vol. 76, no. 8, pp. 1207–21, 1994.

[29] A. G. Aulisa, G. Mastantuoni, M. Laineri, F. Falciglia, M. Giordano, E. Marzetti, and V. Guzzanti, "Brace technology thematic series: the progressive action short brace (PASB).," Scoliosis, vol. 7, p. 21, 2012.

[30] C. R. D'Amato, S. Griggs, and B. McCoy, "Nighttime bracing with the Providence brace in adolescent girls with idiopathic scoliosis.," Spine, vol. 26, no. 18, pp. 2006–2012, 2001.

[31] T. Gavin, W. H. Bunch, and V. M. Dvonch, "The Rosenberger Scoliosis Orthosis," JACPOC, vol. 21, no. 3, p. 35, 1986.

[32] S. Negrini, G. Marchini, and F. Tessadri, "Brace technology thematic series - The Sforzesco and Sibilla braces, and the SPoRT (Symmetric, Patient oriented, Rigid, Three-dimensional, active) concept," Scoliosis, vol. 6, no. May, p. 8, 2011.

[33] C. Coillard, A. Circo, and C. H. Rivard, "A new concept for the non-invasive treatment of Adolescent Idiopathic Scoliosis: the Corrective Movement principle integrated in the SpineCor System," Disabil Rehabil Assist Technol, vol. 3, no. 3, pp. 112–119, 2008.

[34] P. J. van Loon and R. van Erve, "The Development of TLI (Thoracolumbar Lordotic Intervention) as an Effective Bracing Concept for the Postural Spinal Problems - A Review," Spine, vol. 4, no. 2, 2015.

[35] P. J. van Loon, M. Roukens, J. D. Kuit, and F. B. Thunnissen, "A new brace treatment similar for adolescent scoliosis and kyphosis based on restoration of thoracolumbar lordosis. Radiological and subjective clinical results after at least one year of treatment.," Scoliosis, vol. 7, no. 1, p. 19, 2012.

[36] A. G. Veldhuizen, J. Cheung, G. J. Bulthuis, and G. Nijenbanning, "A new orthotic device in the non-operative treatment of idiopathic scoliosis," Medical Engineering and Physics, vol. 24, no. 3, pp. 209–218, 2002.

[37] G. S. Bassett, W. P. Bunnell, and G. D. MacEwen, "Treatment of idiopathic scoliosis with the wilmington brace. results in patients with a twenty to thirty-nine-degree curve," Journal of Bone & Joint Surgery, vol. 4, no. 68, pp. 602–605, 1986.

[38] J. H. Wynne, "The Boston Brace System Philisophy, Biomechanics, Design & Fit," in The Conservative Scoliosis Treatment (T. B. Grivas, ed.), pp. 370–384, IOS Press, 2008.

[39] R. Gepstein, Y. Leitner, E. Zohar, I. Angel, S. Shabat, I. Pekarsky, T. Friesem, Y. Folman, A. Katz, and B. Fredman, "Effectiveness of the Charleston bending brace in the treatment of single-curve idiopathic scoliosis.," Journal of pediatric orthopedics, vol. 22, no. 1, pp. 84–87, 2002.

[40] T. Kotwicki and J. Cheneau, "Biomechanical action of a corrective brace on thoracic idiopathic scoliosis: Cheneau 2000 orthosis," Disabil Rehabil Assist Technol, vol. 3, no. 3, pp. 146–153, 2008.

[41] T. Maruyama, Y. Kobayashi, M. M, and N. Y, "Effectiveness of brace treatment for adolescent idiopathic scoliosis," Scoliosis, vol. 10, p. S12, 2015.

[42] H. R. Weiss, "History of soft brace treatment in patients with scoliosis: a critical appraisal," Hard Tissue, vol. 2, p. 35, 2013.

[43] N. Sy, M. Borysov, M. Moramarco, X. F. Nan, and H.-R. Weiss, "Bracing Scoliosis - State of the Art (Mini-Review)," Current Pediatric Reviews, vol. 12, pp. 36–42, 2016.

[44] K. Zaborowska-Sapeta, I. M. Kowalski, T. Kotwicki, H. Protasiewicz-Fałdowska, and W. Kiebzak, "Effectiveness of Ch??neau brace treatment for idiopathic scoliosis: Prospective study in 79 patients followed to skeletal maturity," Scoliosis, vol. 6, no. 1, pp. 1–5, 2011.

[45] A. Helfenstein, M. Lankes, K. Ohlert, D. Varoga, H.-J. Hahne, H. W. Ulrich, and J. Hassenpflug, "The objective determination of compliance in treatment of adolescent idiopathic scoliosis with spinal orthoses.," Spine, vol. 31, no. 3, pp. 339–344, 2006.

[46] The SpineCorporation Limited, "The spinecor® dynamic corrective brace," 2013. [Online] http://www.spinecor.com/ForProfessionals/SpineCorDynamicCorrectiveBrace.aspx.

[47] C. Coillard, A. B. Circo, and C. H. Rivard, "Effectiveness of the SpineCor brace based on the standardized criteria proposed by the S.R.S. for adolescent idiopathic scoliosis - up to date results," in 6th International Conference on Conservative Management of Spinal Deformities (J. C. de Mauroy, ed.), (Lyon), p. saa15, 2009.

[48] A. Szwed and M. Ko??ban, "Results of SpineCor dynamic bracing for idiopathic scoliosis," in Studies in Health Technology and Informatics, vol. 176, pp. 379–382, 2012.

[49] M. Wong, J. Cheng, T. Lam, B. Nq, S. Sin, S. Lee-Shum, D. Chow, and S. Tam, "The Effect of Rigid vs Flexible Spinal Orthosis on the Clinical Efficacy and acceptance of the Patients with AIS," Spine, vol. 33, no. 12, pp. 1360–1365, 2008.

[50] C. C. Hasler, S. Wietlisbach, and P. Büchler, "Objective compliance of adolescent girls with idiopathic scoliosis in a dynamic SpineCor brace," Journal of Children's Orthopaedics, vol. 4, no. 3, pp. 211–218, 2010.

[51] M. Roller and G. Nijenbanning, "Results of a Prospective Cohort Study of the TriaC Brace and its Clinical Effects," Orthopädie-Technik Quarterly, vol. III, 2009.

[52] A. Zeh, M. Planert, S. Klima, W. Hein, and D. Wohlrab, "The flexible Triac???-Brace for conservative treatment of idiopathic scoliosis. An alternative treatment option?," Acta Orthopaedica Belgica, vol. 74, no. 4, pp. 512–521, 2008.

[53] M. Rigo, S. Negrini, H. R. Weiss, T. B. Grivas, T. Maruyama, T. Kotwicki, and Sosort, "'SOSORT consensus paper on brace action: TLSO biomechanics of correction (investigating the rationale for force vector selection)'," Scoliosis, vol. 1, p. 11, dec 2006.

[54] C. E. Aubin, J. Dansereau, J. A. De Guise, and H. Labelle, "[A study of biomechanical coupling between spine and rib cage in the treatment by orthosis of scoliosis]," Ann Chir, vol. 50, no. 8, pp. 641–650, 1996.

[55] J. Van den Hout, L. Van Rhijn, R. Van den Munckhof, and A. Van Ooy, "Interface corrective force measurements in Boston brace treatment," European Spine Journal, vol. 11, no. 4, pp. 332–335, 2002.

[56] M. Romano, R. Carabalona, S. Petrilli, P. Sibilla, and S. Negrini, "Forces exerted during exercises by patients with adolescent idiopathic scoliosis wearing fiberglass braces.," Scoliosis, vol. 1, p. 12, 2006.

[57] W. K. Chou, C. L. Liu, Y. C. Liao, F. H. Cheng, Z. C. Zhong, and C. S. Chen, "Using finite element method to determine pad positions in a boston brace for enhancing corrective effect on scoliotic spine: A preliminary analysis," Journal of Medical and Biological Engineering, vol. 32, no. 1, pp. 29–35, 2012.

[58] J. Galante, A. Shultz, R. L. Dewald, and R. D. Ray, "Forces Acting in the Milwaukee Brace on Patients Undergoing Treatment for Idiopathic Scoliosis," Journal of Bone and Joint Surgery, vol. 52, no. 3, pp. 498–506, 1970.

[59] R. B. Giesberts, E. E. G. Hekman, P. G. M. Maathuis, and G. J. Verkerke, "Quantifying the Ponseti method," Journal of the Mechanical Behavior of Biomedical Materials, vol. 66, no. November 2016, pp. 45–49, 2017.

[60] R. B. Giesberts, V. Sluiter, and E. Hekman, "Adaptation rate in the treatment of clubfoot.." [Online] https://www.researchgate.net/project/Adaptation-rate-in-the-treatment-of-clubfoot, 2017.

[61] R. B. Giesberts, "Clubfoot - towards a rapid treatment of clubfoot," 2017. Poster presented at BME2017, Jan 26–27, University of Twente, Twente, Netherlands, [Online] https://www.utwente.nl/en/et/clubfoot/.

[62] F. Landauer, C. Wimmer, and H. Behensky, "Estimating the final outcome of brace treatment for idiopathic thoracic scoliosis at 6-month follow-up," Pediatric Rehabilitation, vol. 6, pp. 201–207, jul 2003.

[63] E. Lou, J. V. Raso, D. L. Hill, J. K. Mahood, and M. J. Moreau, "Correlation between quantity and quality of orthosis wear and treatment outcomes in adolescent idiopathic scoliosis.," Prosthetics and orthotics international, vol. 28, no. 1, pp. 49–54, 2004.

[64] J. E. Bible, D. Biswas, C. P. Miller, P. G. Whang, and J. N. Grauer, "Normal functional range of motion of the lumbar spine during 15 activities of daily living," Clinical Spine Surgery, vol. 23, no. 2, pp. 106–112, 2010.

[65] R. D. Lea and J. J. Gerhardt, "Range-of-motion measurements," Joint Surg am., vol. 77, no. 5, pp. 784–798, 1995.

[66] M. Yazdifar, M. R. Yazdifar, J. Mahmud, I. Esata, and M. Chizaria, "Evaluating the hip range of motion using the goniometer and video tracking methods," Procedia Engineering, vol. 68, pp. 77–82, 2013.

[67] Muscoskeletal Key, "Measurement of range of motion of the thoracic and lumbar spine," 2016. [Online] https://musculoskeletalkey.com/measurement-of-range-of-motion-of-the-thoracic-and-lumbar-spine/.

[68] N. Gebruers, C. Vanroy, S. Truijen, S. Engelborghs, and P. P. De Dyen, "Monitoring of physical activity after stroke: a systematic review of accelerometry-based measures.," Arch Phys Med Rehabil, vol. 91, pp. 288–97, 2010.

[69] H. J. Luinge and P. H. Veltink, "Measuring orientation of human body segments using miniature gyroscopes and accelerometers," Medical and Biological Engineering and Computing, vol. 43, no. 2, pp. 273–282, 2005.

[70] H. J. Luinge and P. H. Veltink, "Inclination measurement of human movement using a 3-D accelerometer with autocalibration.," IEEE transactions on neural systems and rehabilitation engineering : a publication of the IEEE Engineering in Medicine and Biology Society, vol. 12, no. 1, pp. 112–121, 2004.

[71] R. S. Alqhtani, M. D. Jones, P. S. Theobald, and J. M. Williams, "Reliability of an accelerometer-based system for quantifying multiregional spinal range of motion," Journal of Manipulative and Physiological Therapeutics, vol. 38, no. 4, pp. 275–281, 2015.

[72] Adafruit, "Adafruit precision nxp 9-dof breakout board," 2017. [Online] https://www.adafruit.com/product/3463.

[73] D. A. Winter, Biomechanics and Motor Control of Human Movement. Hoboken, New Jersey: John Wiley and Sons, 2005.

[74] A. Okawa, K. Shinomiya, H. Komori, T. Muneta, Y. Arai, and O. Nakai, "Dynamic motion study of the whole lumbar spine by videofluoroscopy.," Spine, vol. 23, no. 16, pp. 1743–9, 1998.

[75] R. J. Davis, D. C. Lee, C. Wade, and B. Cheng, "Measurement Performance of a Computer Assisted Vertebral Motion Analysis System.," Int J Spine Surg, vol. 9, p. 36, 2015.

[76] D. B. Chaffin, G. B. J. Andersson, and B. J. Martin, Occupational Biomechanics 4th edition. New Jersey: John Wiely and Sons, 2006.

[77] P. Merriaux, Y. Dupuis, R. Boutteau, P. Vasseur, and X. Savatier, "A study of vicon system positioning performance," Sensors (Switzerland), vol. 17, no. 7, 2017.

[78] P. Cerveri, A. Pedotti, and G. Ferrigno, "Non-invasive approach towards the in vivo estimation of 3D inter-vertebral movements: Methods and preliminary results," Medical Engineering and Physics, vol. 26, no. 10, pp. 841–853, 2004.

[79] M. Solomito, The Use of Motion Analysis Technology as an Alternative Means of Assessing Spinal Deformity in Pati PhD thesis, Univeristy of Connecticut - Storrs, 2011.

[80] P. D. Masso and G. E. Gorton, "Quantifying Changes in Standing Body Segment Alignment Following Spinal Instrumentation and Fusion in Idiopathic Scoliosis Using an Optoelectronic Measurement System," Spine, vol. 25, no. 4, pp. 457–462, 2000.

[81] J. R. Engsberg, L. G. Lenke, A. K. Reitenbach, K. W. Hollander, K. H. Bridwell, and K. Blanke, "Prospective Evaluation of Trunk Range of Motion in Adolescents With Idiopathic Scoliosis Undergoing Spinal Fusion Surgery," Spine, vol. 27, no. 12, pp. 1346–1354, 2002.

[82] C. Frigo, R. Carabalona, M. Dalla Mura, and S. Negrini, "The upper body segmental movements during walking by young females," Clinical Biomechanics, vol. 18, no. 5, pp. 419–425, 2003.

[83] W. Skalli, R. D. Zeller, L. Miladi, G. Bourcereau, M. Savidan, F. Lavaste, and J. Dubousset, "Importance of Pelvic Compensation in Posture and Motion After Posterior Spinal Fusion Using CD Instrumentation for Idiopathic Scoliosis," Spine, vol. 31, no. 12, pp. E359–E366, 2006.

[84] W. Y. Wong and M. S. Wong, "Smart garment for trunk posture monitoring: A preliminary study," Scoliosis, vol. 3, no. 7, 2008.

[85] M. Shaw, C. J. Adam, M. T. Izatt, P. Licing, and G. N. Askin, "Use of the iphone for cobb angle measurement in scoliosis," Scoliosis, vol. 21, no. 6, pp. 1062–1068, 2012.

[86] S. Galvis, D. Burton, B. Barnds, J. Anderson, R. Schwend, N. Price, S. Wilson, and E. Friis, "The effect of scoliotic deformity on spine kinematics in adolescents," Scoliosis and Spinal Disorders, vol. 11, no. 1, p. 42, 2016.

[87] J. M. Frerich, K. Hertzler, P. Knott, and S. Mardjetko, "Comparison of Radiographic and Surface Topography Measurements in Adolescents with Idiopathic Scoliosis," The Open Orthopaedics Journal, vol. 6, no. 1, pp. 261–265, 2012.

[88] Simi Reality Motion Systems GmbH, "Simi concept for clinical motion analysis and applied biomechanics - catalogue," 2014. [Online] http://www.simi.com/fileadmin/user_upload/Dokumente/Downloads/Simi_Clinical_and_Sports_Catalogue.pdf.

[89] J. B. Ring and C. Kim, "A Passive Brace to Improve Activities of Daily Living Utilizing Compliant Parallel Mechanisms," in Volume 5A: 40th Mechanisms and Robotics Conference, p. V05AT07A015, ASME, aug 2016.

[90] R. Tranberg, Analysis of body motions based on optical markers applications. PhD thesis, University of Gothenburg, 2010.

[91] F. M. Rast, E. S. Graf, A. Meichtry, J. Kool, and C. M. Bauer, "Between-day reliability of three-dimensional motion analysis of the trunk: A comparison of marker based protocols," Journal of Biomechanics, vol. 49, no. 5, pp. 807–811, 2016.

[92] P. Bruno and J. Barden, "Comparison of two alternative technical marker sets for measuring 3D pelvic motion during gait," Journal of Biomechanics, vol. 48, no. 14, pp. 3876–3882, 2015.

[93] K.-S. Lee and M.-C. Jung, "Quantitative comparison of marker attachment methods for hand motion analysis," International Journal of Occupational Safety and Ergonomics, vol. 21, no. 1, pp. 30–38, 2015.

[94] L. De Baets, S. Van Deun, K. Desloovere, and E. Jaspers, "Dynamic scapular movement analysis: Is it feasible and reliable in stroke patients during arm elevation?," PLoS ONE, vol. 8, no. 11, 2013.

[95] S. Schmid, D. Studer, C. C. Hasler, J. Romkes, W. R. Taylor, S. Lorenzetti, and R. Brunner, "Quantifying spinal gait kinematics using an enhanced optical motion capture approach in adolescent idiopathic scoliosis," Gait and Posture, vol. 44, pp. 231–237, 2016.