

École polytechnique de Louvain

Quantification of Intervertebral Efforts and Muscle Forces through Multibody Modeling and Electromyographic Measurements: a Scoliosis Case Study

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Abstract

Scoliosis is a deformity of the spine in the three planes, with the primary deformation occurring in the frontal plane. The principal objective of this master's thesis was to identify which vertebral segments of a scoliotic spine were subjected to the most mechanical stress. This information could serve as a valuable tool to reduce the reliance on surgeon intuition in the planning of scoliosis surgery when it is indicated. In order to achieve this, it was necessary to compute the efforts in each intervertebral disc and back muscle forces. This was done in the case of a static back extension, the Sorensen posture, based on the hybrid method developed and applied on healthy subjects by Hinnekens et al. [1]. Accordingly, the current study adapted the hybrid approach to render it applicable to the context of scoliosis. Furthermore, the adapted method was applied to a single healthy subject with similar anthropometric data to validate the adaptation and to highlight the differences between a subject with scoliosis and without. The intervertebral efforts and the muscle forces were successfully quantified for the lumbar spine and the associated back muscles. The results demonstrated several similarities with the existing literature, which was promising. Consequently, this study offers motivation to pursue the quantification of such biological markers, with the aim of providing assistance in the medical field and generalizing the results to a larger population of scoliotic patients.

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Acronyms

A

A3 Method for which the angles between each vertebra was computed with coordinates of the vertebra taken on an A3-paper. [64](#), [65](#), [67](#), [68](#), [70](#), [71](#)

A4 Method for which the angles between each vertebra was computed with coordinates of the vertebra taken on an A4-paper. [10](#), [64](#), [65](#), [67](#), [68](#), [70](#), [71](#)

AF Annulus Fibrosus. [16](#)

AIS Adolescent Idiopathic Scoliosis. [6](#), [14](#), [20–22](#), [77–80](#), [82–84](#)

ATS Anterior Thorascopic Surgery. [22](#)

AVR Axial Vertebral Rotation. [42](#), [43](#), [85](#), [86](#)

B

BW Body Weight. [7](#), [9](#), [10](#), [37](#), [46–48](#), [50](#), [52](#), [53](#), [57](#), [58](#), [70](#), [71](#), [80](#), [83](#), [87](#)

E

EMG Electromyography. [7](#), [9](#), [16](#), [23](#), [24](#), [26](#), [29](#), [31](#), [32](#), [34](#), [35](#), [39](#), [45](#), [47–50](#), [59](#), [73](#), [78–80](#), [84](#)

G

GIMP Method for which the angles between each vertebra was computed with coordinates of the vertebra measured through the open source software GIMP [2]. [64](#), [65](#), [67](#), [68](#), [70](#), [71](#)

H

Hybrid approach Hybrid method developped by Hinnekens [1]. [24](#), [26](#)

I

ILpL Iliocostalis Lumborum pars Lumborum. 32, 33, 39, 58

ILpT Iliocostalis Lumborum pars Thoracis. 32, 33, 39, 54, 56, 58, 60, 62

IVD Intervertebral Disc. 6–10, 15, 16, 21, 40, 44, 50–52, 54, 68–71, 74–76

L

LD Latissimus Dorsi. 7, 8, 16, 27, 32–34, 39, 41, 45, 47, 49, 54, 57–60, 63, 78, 80, 82

LIC Loaded Isometric Contraction. 9, 10, 24, 29, 34, 37, 47, 52, 57, 58, 82

LP Lumbar Paravertebral. 7, 8, 18, 32, 34, 39, 47, 49, 50, 54, 55, 58–61, 79, 84

LTpL Longissimus Thoracis pars Lumborum. 32, 33, 39

LTpT Longissimus Thoracis pars Thoracis. 32, 33, 39, 54, 56, 60, 62

M

Manual Method describing the measurement of the angles between each vertebra directly on the radio. 8, 64–68, 70, 71

MBS Multibody System. 7, 37, 39, 40, 44, 60, 64, 65, 74, 82

MIS Minimally Invasive Surgery. 22

MRI Magnetic resonance imaging. 21

MVC Maximum Voluntary Contraction. 9, 24, 29, 34, 37, 46, 47, 77, 78, 85

N

NP Nucleus Pulposus. 6, 16, 21

P

PMIS Posterior Minimally Invasive Surgery. 22

Q

QL Quadratus Lumborum. 7–9, 18, 32–34, 39, 47, 49, 54, 55, 58–61, 79

R

RMS Root Mean Square. [34](#), [47](#), [49](#), [50](#), [79](#), [84](#)

T

TP Thoracic Paravertebral. [7](#), [8](#), [32](#), [34](#), [39](#), [47](#), [49](#), [54](#), [56](#), [58–60](#), [62](#), [83](#), [84](#)

V

VBT Vertebral Body Tethering. [22](#)

Chapter 1

Introduction

1.1 Context

Scoliosis, a deviation of the spine in the frontal plane with a curvature exceeding 10 degrees, has been studied for years, since Hippocrates [3]. However, the pathology still raises many unknowns, such as surgery planning when surgical correction is necessary. The most common type of scoliosis is the adolescent idiopathic scoliosis (AIS), more present in girls than boys, which has no known cause and affects the population over the age of ten [4]. To better understand the issue just introduced, one needs to know some anatomical basics.

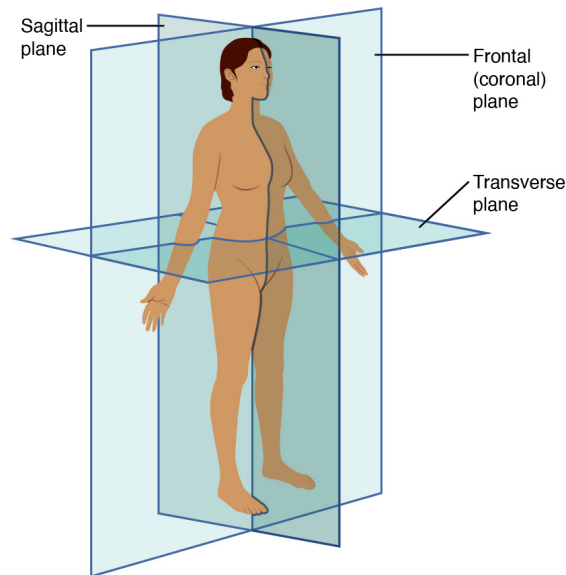


Figure 1.1: Anatomical planes. Adapted from [5]

When describing human anatomy, the following conventions apply: the frontal or coronal plane is the plane that divides the body into anterior and posterior parts. The sagittal plane divides the body longitudinally into right and left parts, while the transverse plane divides it into upper and lower parts (Fig. 1.1).

It is also important to note the terminology used to describe anatomical directions. The direction towards the midline of the body is called the medial direction, and its opposite is the lateral direction. In contrast, the direction towards the trunk is the proximal direction, and its opposite is called the distal direction. The next important anatomical topics, presented hereafter, concern main components of the spine, back muscles, and more about scoliosis.

1.1.1 The spine: vertebrae and intervertebral discs

The spine is mainly composed of vertebrae and intervertebral discs (IVDs). The vertebrae are bone components. Their shape differ depending on their location in the spine, but they all have a spinous process and two transverse processes, which serve as attachment sites for muscles (Fig. 1.2). Among the vertebrae, the fused coccygeal and sacral ones form the coccyx and the sacrum. Above them, vertebrae can move relative to each other: 5 are lumbar (*L1* to *L5*), 12 are thoracic (*T1* to *T12*) and 7 are cervical (*C1* to *C7*) (Fig. 1.3). In the sagittal plane, a healthy spine shows four natural curves called the cervical lordosis, the thoracic kyphosis, the lumbar lordosis and the sacral curve. In the frontal and transverse planes in normal conditions, the spine shows little to no angle between consecutive vertebrae.

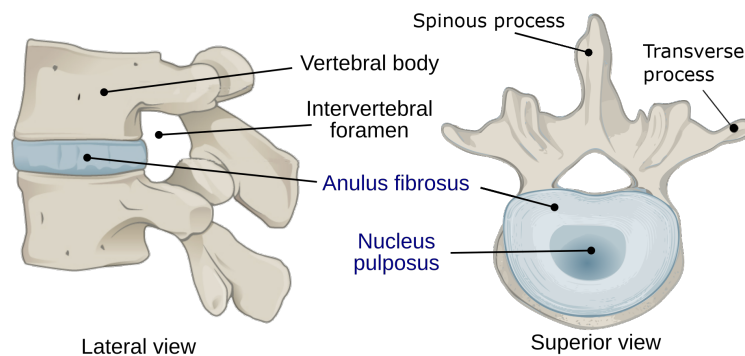


Figure 1.2: The intervertebral disc and the vertebra. Adapted from [6].

The intervertebral discs, located between the vertebrae, are essential components of the spine. They serve several functions, including allowing movement

between vertebrae and absorbing shocks. They enable the spine to remain strong yet flexible. To accomplish such a task, **IVDs** are composed of three fibrocartilaginous structures: an inner structure called the nucleus pulposus (**NP**), an outer annulus fibrosus (**AF**), and outer cartilaginous endplates that anchor each disc to its corresponding vertebrae (Fig. 1.2) [7].

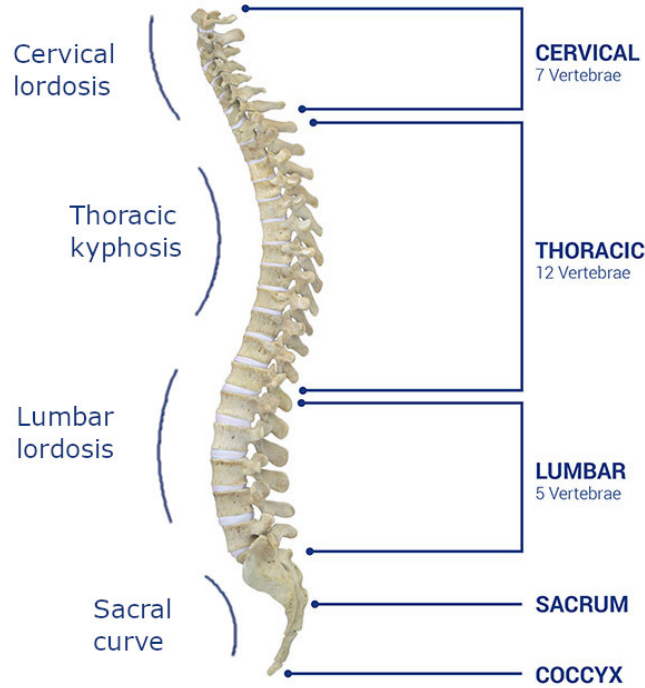
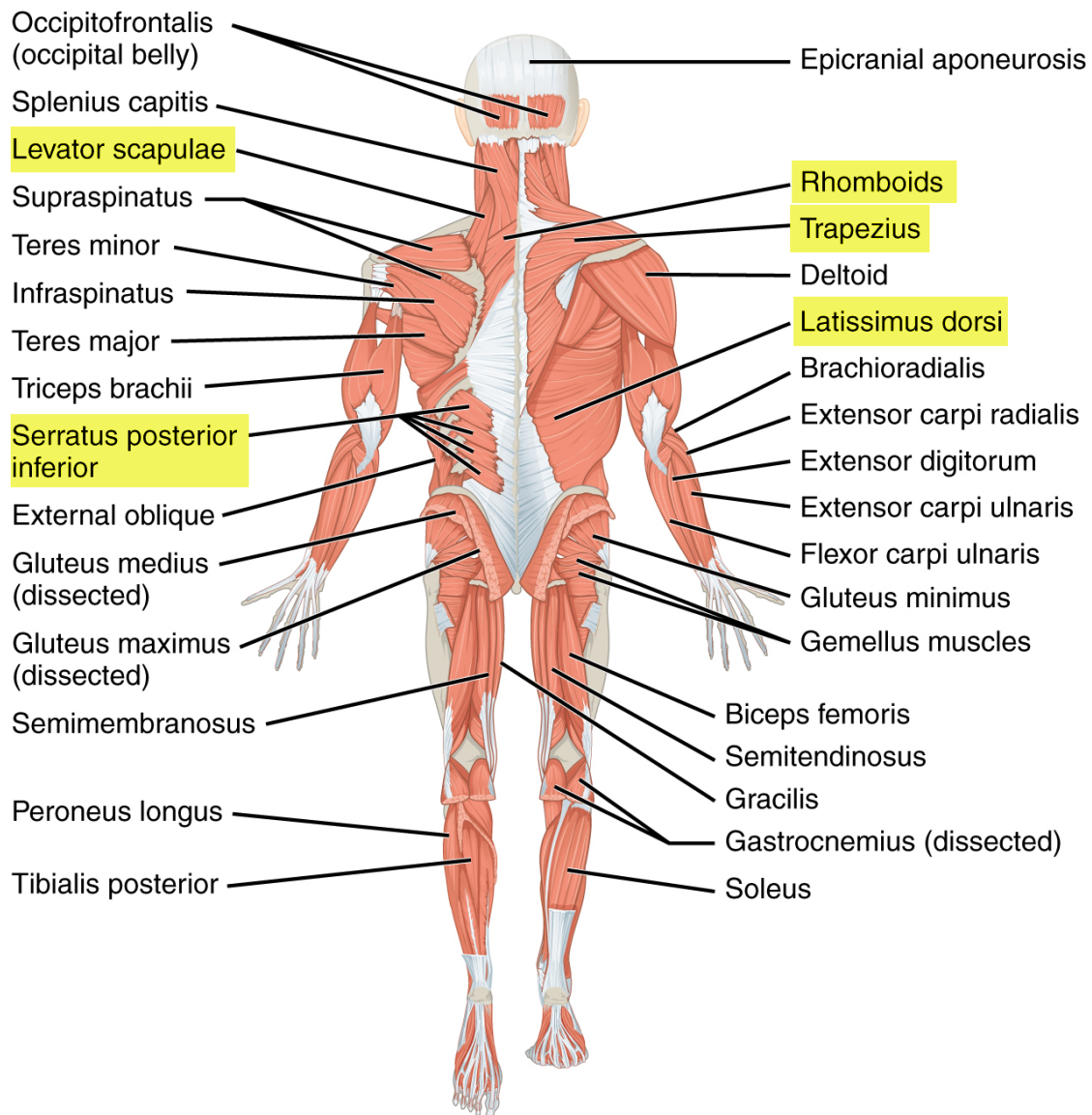


Figure 1.3: Normal curvatures of the spine. Adapted from [8].

1.1.2 Back muscles

The muscles of the back are divided into two groups [9]: superficial and deep back muscles. The superficial group consists of five muscles: the trapezius, the latissimus dorsi (**LD**), the levator scapulae, the rhomboids and the serratus posterior (Fig. 1.4).

The deep back muscles consist of three subgroups: the more superficial ones, the deep ones and the deepest ones. As explained in Section 2.2, the experimental data utilized in the method consists of the activity of the muscles surrounding the spine, collected using electrodes. Electromyography (**EMG**) is the name of the technique used to measure the intensity of the activity of muscles through electrodes. It is a passive measure of the electrical potential produced during



Major muscles of the body.
 Right side: superficial; left side: deep (posterior view)
 Figure 1.4: Superficial back muscles. Adapted from [10].

muscle contraction. Surface electrodes, which are used for measurements in this study, can only measure the signal of superficial muscles. This is why only the most superficial layer of deep muscles will be depicted here. This one comprises the splenius acting on the neck, and the erector spinae acting on the spine. The erector spinae can be further divided into three muscles: the iliocostalis laterally, the longissimus which is intermediate, and the spinalis medially. Each of these three muscles can be further divided into three muscles, depending on the region of vertebrae they are connected to. The iliocostalis includes a cervicis, a thoracis and a lumborum part. The longissimus has a capitis, a cervicis and a thoracis part. Finally, the spinalis has a capitis, a cervicis and a thoracis part (Fig. 1.5). All muscles of the erector spinae act as spinal extensors when contracted bilaterally, and bend the spine in unilateral contraction.

Even if only superficial muscles could be detected by surface electrodes, one exception remains. The quadratus lumborum ([QL](#)), which is considered to be a deep back muscle, can be measured through surface electrodes with the anatomical expertise of a physiotherapist. The fibers of the [QL](#) can be detected where the surface over it is very thin. This zone, called the superior lumbar triangle, is defined by the lower line of the rib cage, the iliac crest, and the lumbar paravertebral muscle ([LP](#), see section 2.2) [1].

Every muscle is constituted of several fascicles. These are detailed in Section 2.3.

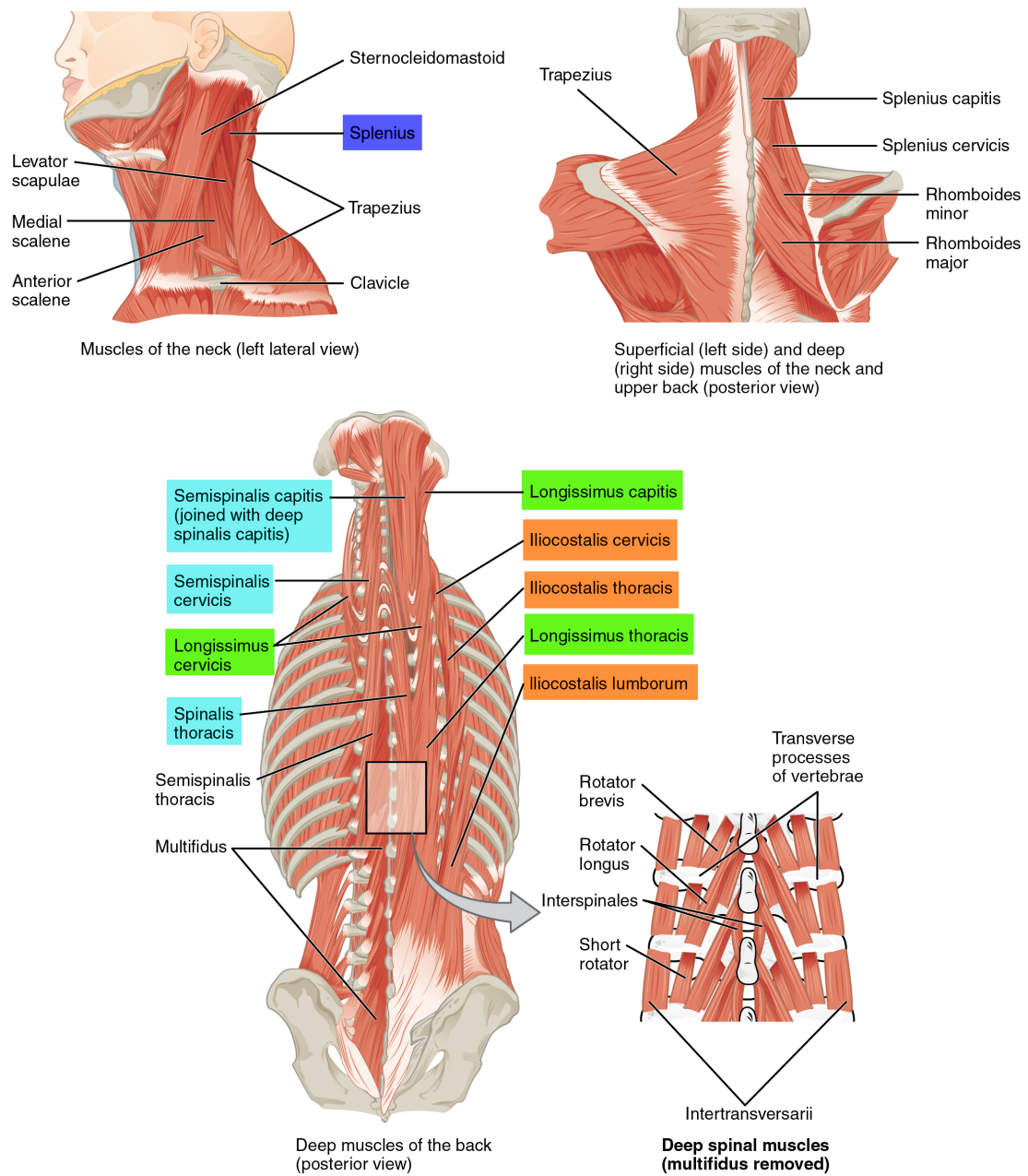


Figure 1.5: Deep back muscles. Adapted from [11]. The three parts of the spinalis are in light blue. The three parts of the longissimus are in green. The three parts of the iliocostalis are in orange.

1.1.3 Scoliosis

Up to now, only the anatomy of a healthy spine associated with normal back muscles has been described. However, spinal disorders can occur, such as scoliosis. As the most common form of scoliosis is AIS, this study focuses on AIS.

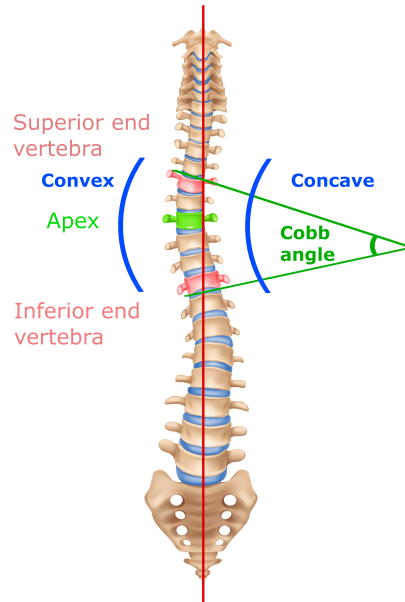


Figure 1.6: Anterior view of a scoliotic spine. Adapted from [12]. The vertebrae colored in red correspond to the upper and lower end vertebrae, whereas the one colored in green is the apex.

Scoliosis is a deformation of the spine in the three planes. The greatest deformation occurs however in the frontal plane. This pathology naturally leads to a concave side of the spine where body elements are compressed, and a convex side where body elements are stretched (Fig. 1.6). The vertebra that is the furthest from the midline of the body is the apex, whereas the vertebrae with the greatest tilt at the ends of the curve are the superior and inferior end vertebrae. When lines are drawn in the frontal plane parallel to the upper endplate of the superior end vertebra and to the lower endplate of the inferior end vertebra, the angle between these two lines defines the Cobb angle. The severity of the scoliosis is measured by this angle.

Scoliosis leads to many alterations. In addition to daily pain, patients may also experience slipping discs, respiratory insufficiency, or other pathological conditions. For example, when compressed asymmetrically, the IVDs see their NP migrate in the opposite direction of the compression. This phenomenon, called nuclear migration, is predictable in most cases and is then referred to as the dynamic disc model, but presents unpredictable patterns when degeneration is present within the IVDs. In the case of scoliosis, NP migration away from the concavity is observed (Fig. 1.7) [13].

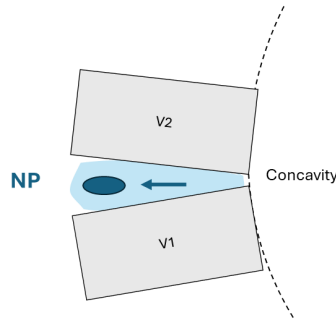


Figure 1.7: Schematic of the NP migration. The two grey bodies V1 and V2 represent two consecutive vertebrae, the light blue shape represents one IVD and the dark blue shape is the NP of the IVD.

Scoliosis also has an impact on back muscles. However, this impact remains unpredictable. Differences of muscle volume are more frequent at the apex. They are slightly more frequently larger on the concave side than on the convex side. However, skinfold thickness shows to be always greater on the concave side at the apex [14]. In addition, N. Gaudreault et al. [15] suggested that muscle activity shows a significant difference from one side to the other at the lower level of the spine, but no great difference at the apex. The study suggested a compensatory strategy at the lower level of the spine. Finally, based on magnetic resonance imaging (MRI) of females with AIS, paraspinal muscles on the concave side, due to prolonged compression, showed reduced muscle activity and an altered muscle composition [16].

Thanks to physiotherapy and bracing, surgery can be avoided most of the time. However, surgery is recommended for adolescents with a Cobb angle greater than 45°-50°. This concerns about 10% of AIS patients [4]. Currently, the majority of

scoliosis surgeries consist of fusing some of the vertebrae together, called spinal fusion (Fig. 1.8). This naturally leads to a reduction of the spine mobility. The higher the number of fused vertebrae, the more important this reduction. However, if not enough vertebrae are fused, the scoliosis may recur and the surgery will fail. Therefore, the decision regarding the extent of spinal fusion is important and must be made with as much certainty as possible.

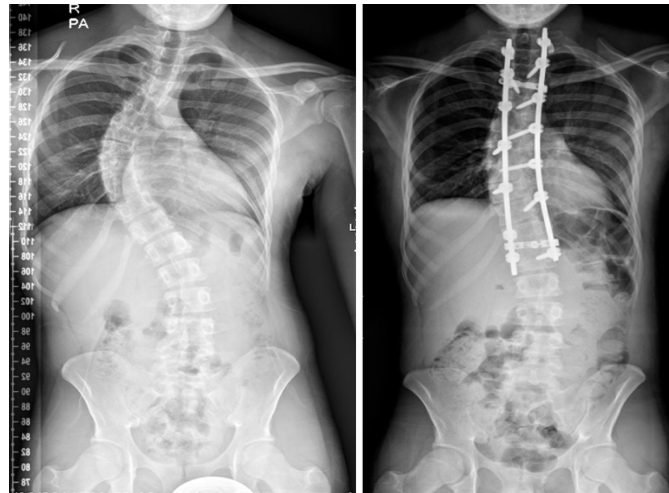


Figure 1.8: Anterior-posterior X-ray of a scoliotic spine before and after spinal fusion due to an AIS [4].

1.2 State of the art

Despite the fact that approximately 29,000 AIS surgeries are conducted annually in the United States [17], surgical strategies remain disparate among surgeons. This discrepancy reflects a lack of certainty in scoliosis surgery planning, as highlighted by the survey of Benedikt Schlager et al [18]. Open surgery remains the main approach, as minimally invasive surgery (MIS) is relatively new. To date, MIS comprises three techniques: anterior thoroscopic surgery (ATS), posterior minimally invasive surgery (PMIS), and vertebral body tethering (VBT). However, ATS is associated with a steep learning curve, high complication rates, and implant failures. PMIS also has a steep learning curve and a longer surgical time. VBT is promising, but still has high complication and re-operation rates [19].

Despite ongoing developments in scoliosis surgery, all techniques still involve spinal fusion. In spinal fusion surgery, the selection of the optimal fusion levels is

crucial. However, to date, no standardized selection method exists [20], [21]. To address this, it is imperative to provide surgeons with objective decision-making tools, such as intervertebral efforts [22].

Intervertebral efforts are important biological markers, and thus, this work is not the first one to use them. However, these quantities are not so easily found when it comes to pathology groups such as scoliosis subjects. Indeed, the most common quantities to find for this pathology group remain **EMG** muscle activity results [23], [24]. In addition, in order to provide these efforts, one needs to dispose of correct spine geometry and spine kinematics. Only then muscle forces can be computed. Abedrabbo brought a high focus to the spine geometry and kinematics in the case of scoliosis and was able to find intervertebral efforts, but muscle forces results remained limited [25].

There are several possibilities for evaluating these quantities: purely experimental approaches, purely mathematical approaches, or a combination of both. As stated by Hinnekens [1], mathematical approaches based on optimization processes must address the redundancy problem, arising from the over-actuation of the human body. Consequently, results of these approaches often drift from reality. However, these processes are based on a generic model. Therefore, the scope of these approaches is less limited than experiments.

Experiments require volunteers, and the more there are, the more time-consuming the experiments become, which is a strong limitation for the scope of the experimental studies. Moreover, experimental approaches are able to provide only quantities that can be measured, which are only indirect measurements of the desired quantities. In addition, muscles forces are even less frequent in the literature as not every method allows their computation. Therefore, a hybrid method is more suitable for this study.

Hybrid methods exhibit the advantages of closer-to-reality and non-measurable results. However, they also inherit the disadvantages of the noisy nature of the data and of a limited scope due to the time-consuming experiments. Nevertheless, they allow for the avoidance of the redundancy problem associated with mathematical approaches, and therefore appear to be more advantageous than limiting. The hybrid method of Hinnekens [1] combined **EMG** as experimental data with a scaled musculoskeletal model. This approach used the concept of an **EMG**-based distribution of a global equivalent force, which is detailed in Section 2.1. The results demonstrated promising and realistic outcomes.

The [Hybrid approach](#), selected in this work, was developed by Hinnekens and applied on 19 healthy subjects, but not yet to a pathology group. However, as this method employs a multibody model, the latter, though generic for healthy patients with only a scaling process, must be adapted to each scoliosis patient due to the variety of curves in scoliosis pathology. Consequently, this study applies the method to a single scoliosis subject and one healthy one with close anthropometric data for comparison.

1.3 Objectives

The aim of the study is to identify the vertebral segments of a scoliotic spine that are subjected to the most mechanical stress. In order to achieve this, it is necessary to compute the efforts in each intervertebral disc and back muscle forces. In the long term, these quantities, objective biological markers for the preservation of spinal integrity, could be used as a tool in spinal surgery planning to reduce the role of surgeon intuition in this phase.

In the present work, data acquisition is achieved through the hybrid approach proposed by Hinnekens et al. [1] on two subjects: a healthy one and a scoliosis one. The proposed approach combines non-invasive experimental measurements with multibody models. It allows to make the most of the experimental data, while also providing the quantification of intervertebral efforts and muscle forces that cannot be measured experimentally. However, the method described by Hinnekens is intended to be applied to healthy patients. Therefore, in the following work, it is adapted to scoliosis patients in several aspects, and applied to one scoliosis subject. The adapted method is also applied to one healthy patient with similar anthropometric data, in order to validate the adaptation and to highlight the differences between a subject with scoliosis and without.

1.4 Outline of the work

First, the method to acquire the desired data is thoroughly detailed, starting with an explanation of the concept of a global equivalent force underlying the method. Then, the remainder of the method explanation is divided according to its two main steps. On the one hand, the experimental step involving the use of [EMG](#) is carried out for two sets of tests: [Maximum Voluntary Contraction \(MVC\)](#) tests and [Loaded Isometric Contraction \(LIC\)](#) tests. On the other hand, the modeling

phase is described, which involves the use of an adapted musculoskeletal model to compute the inverse dynamics of the case under study. Then, the results of the experiments are presented, along with those from the simulation of the model. Finally, a sensitivity analysis is carried out to observe the impact of the geometry on the results, as well as a comparison between the adapted and non-adapted method of Hinnekens, after which results are discussed and a conclusion is drawn.

Chapter 2

Methods

The method used is the [Hybrid approach](#) presented by Hinnekens et al. [1], consisting of [EMG](#) measurements used as input for the multibody model. As the method described by Hinnekens is intended to be applied to healthy patients, it is adapted to scoliosis patients in the following section.

The intervertebral efforts of this study are computed for the Sorensen posture, which consists of an isometric back extension (Section 2.2). It was selected among others as it is associated with a high level of contraction of back muscles, which highlights the efforts in the spine. Furthermore, it is performed without the use of weight machines, which would have made it more complex to achieve an accurate modeling. A static configuration was chosen because it allows the speed and length dependencies of the force produced by each muscle to be neglected, therefore not involving the Hill model in the muscle. In fact, these dependencies greatly complicate the modeling algorithm, and do not have a major impact on the distribution of the intervertebral efforts in the present case of a static configuration.

First, the Hinnekens method is introduced. Secondly, the experimental measurements are described. Finally, the modeling procedure is outlined.

2.1 The hybrid approach

The hybrid approach of Hinnekens is defined as an [EMG](#)-based deterministic distribution of a global equivalent force into each muscle fascicle of the model. Consequently, an important concept lies behind the procedure: when multiple muscles of the body are contracted in order to produce a movement or maintain a position, a global equivalent force can be defined for the configuration at hand as a result. As suggested in [1], this global unknown force, noted F_{equiv} , will be

spread into each force produced by each fascicle of each muscle. This concept is formalized in the following equation:

$$F_{fa,i} = w_{fa,i} * w_m * F_{equiv} \quad (2.1)$$

where w_m is the proportion of the global equivalent force produced by the muscle m , $w_{fa,i}$ is the proportion of the force produced in the muscle m by the i^{th} fascicle of that muscle, and $F_{fa,i}$ is the force produced by the i^{th} fascicle of the muscle m . The factors $w_{fa,i}$ were computed by Hinnekens [1], using the maximum isometric forces $F_{max,i}$ of the different fascicles found in the OpenSim model [26], using the following equation for one muscle:

$$w_{fa,i} = \frac{F_{max,i}}{\sum_i F_{max,i}} \quad (2.2)$$

In order to take into account the asymmetry caused by the spinal deformity, it is very important to note that the subscript m stands for the identification of a muscle, where the left and right muscles are considered as distinct, contrary to [1]. Therefore, if one studies the latissimus dorsi muscle, two muscles are counted for the right side ($m_1 = \text{LDR}$) and the left side ($m_2 = \text{LDL}$), respectively. Following this reasoning, the weights used in Hinnekens method were adapted here to take into account this left and right distinction, using w_{LDR} and w_{LDL} for the right and left LD instead of w_{LD} that is used in [1].

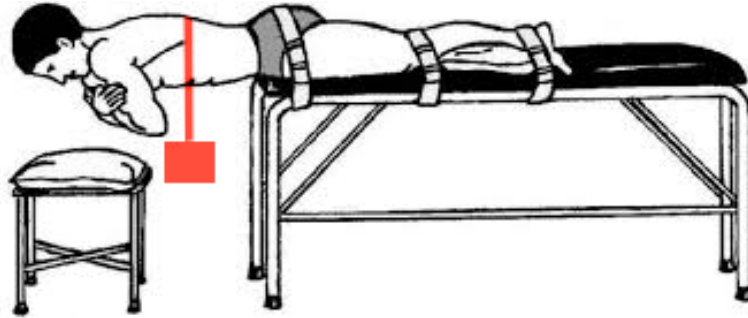


Figure 2.1: Schematic of the Sorensen posture. The added load is represented in red, whereas the exercise at body weight is performed without this red load. Adapted from [27].

It is possible to be more accurate by taking into account the so-called muscle tone and the load rate of each muscle. In fact, following the load a muscle has

to carry, and therefore the force it has to produce, it is suggested to establish a linear regression between the load carried and the level of muscle contraction. In the absence of added load, the level of contraction that exists is the muscle tone w_{F0} , while the increase in the level of activation for an increase in the load applied is the load rate of the muscle w_{Fr} . This splits the weight of one muscle w_m into these two contributions. Equation 2.1 is then transformed into its more refined version, where the global equivalent force is split into its contribution at body weight $F_{equiv,0}$ and its contribution due to the added load $F_{equiv,r}$ (Fig. 2.1), resulting in $F_{equiv} = F_{equiv,0} + F_{equiv,r}$:

$$F_{fa,i} = w_{fa,i} * w_{F0,m} * F_{equiv,0} + w_{fa,i} * w_{Fr,m} * F_{equiv,r} \quad (2.3)$$

Because they represent the proportion of the force that is produced by the muscle, the weights w_{F0} and w_{Fr} are found following this rule: if the linear regression of the muscle activity of muscle m as a function of the load has the form $y_m = a_m x + b_m$ (Fig. 2.2),

$$w_{F0,m} = \frac{\text{intercept}_m}{\sum_i \text{intercept}_i} = \frac{b_m}{\sum_i b_i} \quad (2.4)$$

And

$$w_{Fr,m} = \frac{\text{increment}_m}{\sum_i \text{increment}_i} = \frac{a_i}{\sum_i a_i} \quad (2.5)$$

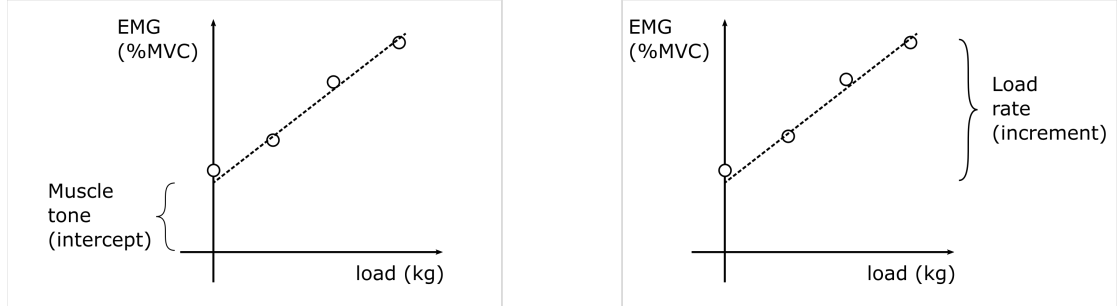


Figure 2.2: Schematic of the linear regression of muscle activity as a function of the load carried used to find the muscle tone and the load rate. Adapted from [1].

Firstly, experiments are carried out to determine the intensity of the activity of each measured muscle, with and without added load, in order to determine w_{F0} and w_{Fr} of each muscle. This procedure is detailed in Section 2.2.

Secondly, the weights found experimentally are introduced into the multibody equations to perform inverse equilibrium and find the intervertebral efforts. This algorithm is presented in Section 2.3.

Fig. 2.3 summarizes the different steps of the method.

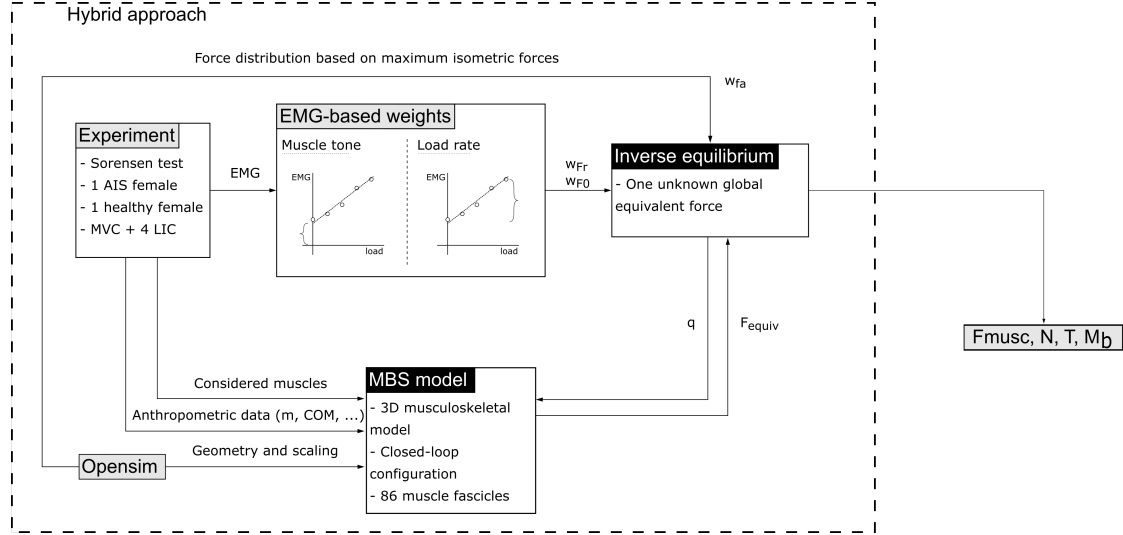


Figure 2.3: Overview of the method proposed by Hinneken. Adapted from [1].

2.2 Experimental electromyographic measurements

EMG provides the electrical activity of the measured muscle over a period of time. However, the signals obtained are not directly comparable with results from other studies. Indeed, the intensity of the signal depends on the conditions of the measurement, which can be altered due to skinfold thickness, skin moisture and many other parameters. For this reason, the signals must be normalized. One common method for normalizing the signals is to use the **MVC**, which involves having the subjects contract their muscles to the maximum possible for a specified period of time. On the other hand, as the objective of the experiment is to determine the muscle tone $w_{F0,m}$ and the load rate $w_{Fr,m}$ acquired through linear regression of the muscle activity compared to the load carried, a series of static tests with different values of added load must be performed, **LIC** tests.

First, the setting of the experiment is detailed. Then, the **MVC** tests are presented. Finally, the **LIC** tests are explained.

2.2.1 Experimental setting

Subjects

The two participants were female, with similar age and body mass. Additionally, they were half-sisters and engaged in the same physical activity for the same period of time, track and field. Table 2.1 displays their anthropometric data. Subject 1 is healthy while Subject 2 is scoliotic. They were first given an explanation of the experiment. They were informed that they could withdraw at anytime from the experiment without justification. For Subject 2, recent X-rays (dating March 2023) were also provided (Fig. 2.4).

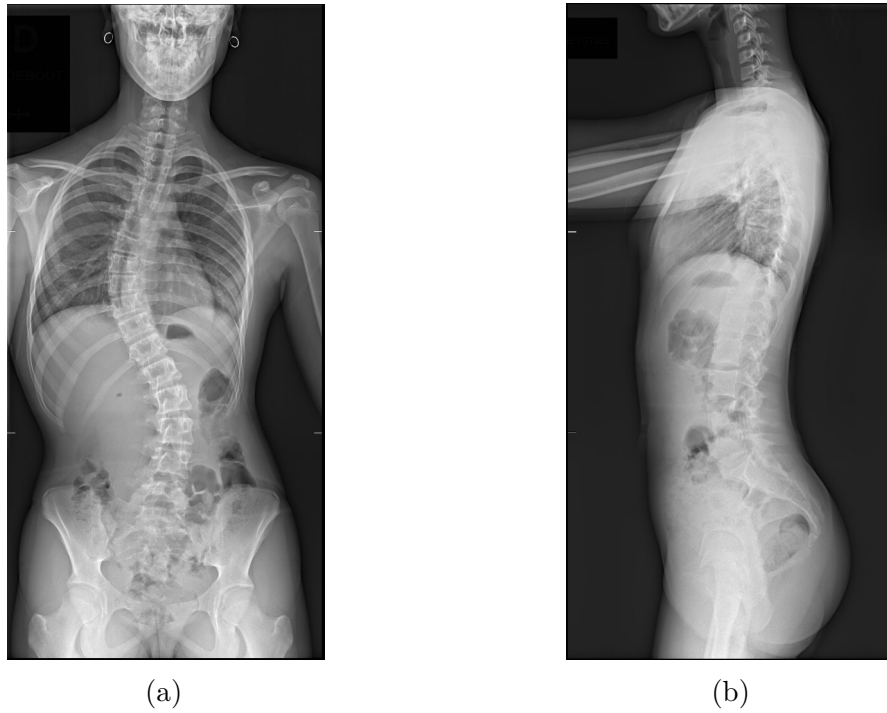


Figure 2.4: X-rays of subject 2 presenting a scoliosis. (a): anterior-posterior view. (b): sagittal view.

Sorensen test

The Sorensen test, described first in 1964 by Hansen and made popular by Biering-Sorensen in 1984, is commonly used to predict low back pain in males [27]. The task

	Subject 1 - Healthy	Subject 2 - Scoliosis
Sex	Female	Female
Age	14	13
Mass [kg]	50	46
Height [cm]	158	165
Medical history	None	Scoliosis
Dominant side	Right	Right
Inferior limb lengths [cm]	76R - 76L	83R - 83L
Frontal balance of the pelvis	Yes	Yes
Gibbosity	No : 5° lumbar right side	Yes : 9° lumbar left side, 10° thoracic right side
Thoracic depth (at T7 level) [cm]	14.3	14.8
Thoracic width (at T7 level) [cm]	23.5	23.1

Table 2.1: Anthropometric data of the subjects.

consist in maintaining an unsupported upper body position for a certain amount of time while lying horizontally with the lower limbs secured to the examination table (Fig. 2.5). The original test was modified in this study. The subjects were required to hold the position for a specified duration, whereas the original test was to be continued until exhaustion.

Materials

The method used here was non-invasive, which means that [EMG](#) only used surface electrodes and not intramuscular ones. In fact, this method is preferred because intramuscular electrodes as well as being invasive, require some expertise or even a formal certification. On the other hand, surface electrodes placement procedure only requires skin preparation to reduce the skin contact impedance. However, this type of electrodes requires a precise knowledge of the anatomy and the recording quality can be poor for high body mass index population. Nevertheless, the patients of the present study are not considered to be part of this population and a physiotherapist took the responsibility of locating the electrodes on the required muscles. In addition, conventional surface electrodes allow the subject to move freely, which is essential for the experiment [\[28\]](#).

The same electrode system as Hinnekens was used for the two subjects (Table 2.2). Indeed, the same electrode size and spacing allow a better comparison as they are crucial measurement parameters. This is because the depth and the area of

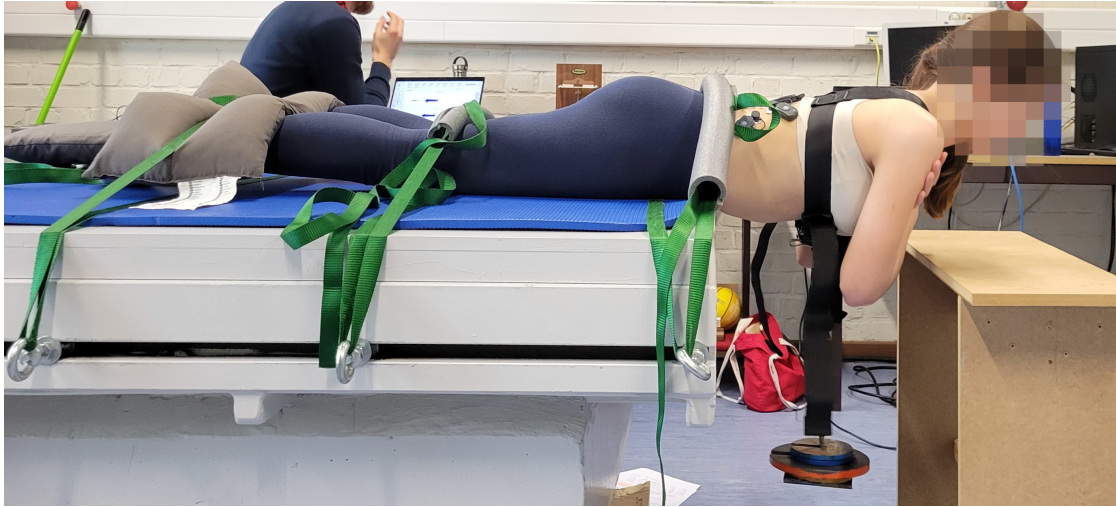
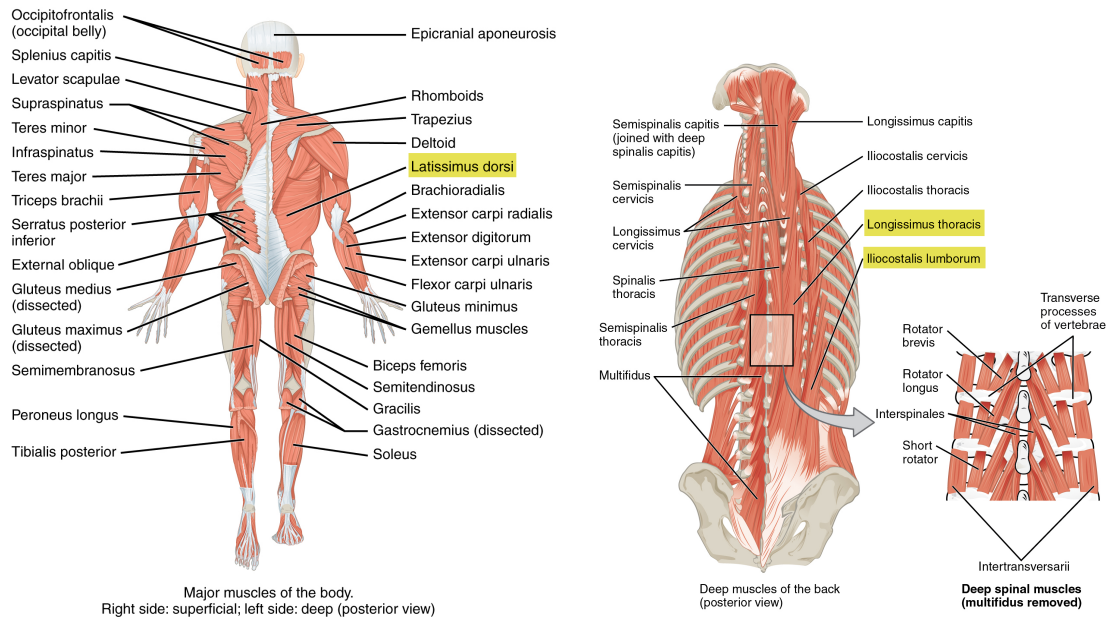


Figure 2.5: Sorensen test posture. The green slacks serve to secure the lower limbs of the participant to the examination table, whereas the harness located on the thorax serves to suspend loads.

measurement increases with the size of the electrode and the distance between them.

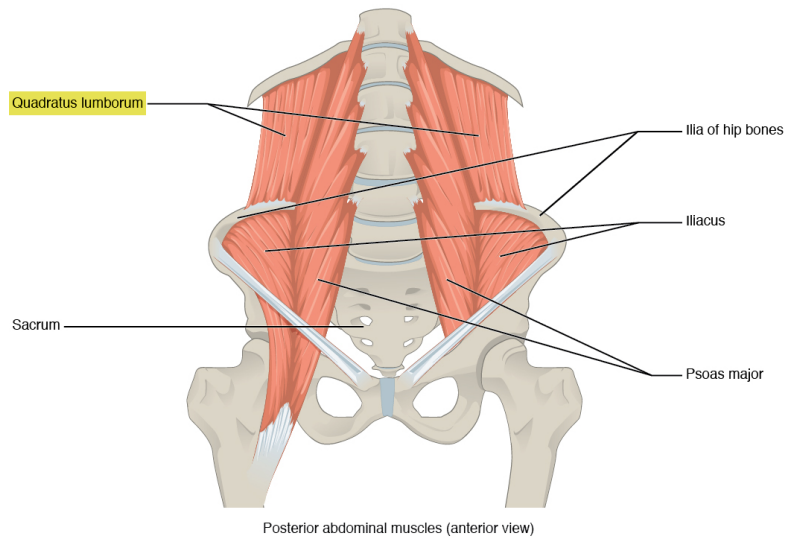
For the purposes of this study, only back muscles that were active during the Sorensen test and could be measured through surface electrodes were considered. These are the iliocostalis lumborum pars thoracis and pars lumborum (**ILpT** and **ILpL**), which are active in extension and bending of the spine, as well as the longissimus thoracis pars thoracis and pars lumborum (**LTpT** and **LTpL**), which are also active in the aforementioned movements. The latissimus dorsi in the thoracolumbar region (**LD**) is active in the medial rotation of the humerus, the adduction and extension of the arm at the shoulder joint, and also in the hyperextension of the back. The last recorded muscle is the quadratus lumborum (**QL**), which is active in the stabilization of the lumbar area and in pelvic lateral tilting (Fig. 2.6). Knowing the anatomical locations of these muscles is of great importance for electrodes placement in the **EMG** experiment.

ILpT, **ILpL**, **LTpT** and **LTpL** can be classified in two muscle groups depending on their location: the lumbar paravertebral (**LP**) including the **LTpL** and **ILpL**, and the thoracic paravertebral (**TP**) including the **LTpT** and **ILpT**. This classification simplifies the analyses of the measurement. Consequently, in the studied muscles, the **LP** and **QL** muscles act mostly in the lumbar region, and the **TP** and **LD** muscles act mostly in the thoracic region.



(a) Superficial back muscles. **Latissimus Dorsi (LD)** is highlighted. Adapted from [10].

(b) Deep back muscles. **ILpT, ILpL, LTpT and LTpL** are highlighted. Adapted from [11].



(c) Posterior abdominal muscles. **Quadratus Lumborum (QL)** is highlighted. Adapted from [29].

Figure 2.6: Back muscles.

Model	BTS FREEEMG 100
Electrodes type	conventional surface electrodes as pairs integrated into a single device
Mother electrode dimensions (LxWxH)	41.4mm x 24.8mm x 14mm
Satellite electrode dimensions (DxH)	16mm x 12mm
Inter-electrodes distance	20mm
Transmission frequency	2.4GHz
Input impedance	100M Ω
Resolution	16 bit
Sample rate	1kHz
Sensitivity	1 μ V
Accuracy	$\pm 2\%$
Operating temperature	0-40°C

Table 2.2: Specifications of the [EMG](#) device.

As Hinnekens logically observed for the Sorensen test, the abdominal muscles did not show a high level of contraction. For this reason, their action has been disregarded for this study.

A total of eight pairs of electrodes were placed in the following order, with R and L standing for the right and left part of the muscle group: [LPR](#), [LPL](#), [TPR](#), [TPL](#), [QLR](#), [QLL](#), [LDR](#), [LDL](#) (Fig. 2.7). The electrodes were placed on the muscle belly in the direction of the muscle fibers with the help of a physiotherapist.

The software BTS EMG Analyser was used to record and save [EMG](#) data. Regarding the force measurement for the [MVC](#) tests, a force sensor was used. A force sensor (Student Strain Gage Data Acquisition Device) was used as well to visualize the force produced in real time. It was placed on a harness (Chest'Air, Petzl) worn by the subjects, and attached to the ground for the [MVC](#) trials. The harness was also used to apply the different loads for the [LIC](#) trials.

After measurements, the raw [EMG](#) signals were rectified, filtered and smoothed with a low-pass 5th order Butterworth filter and a cut-off frequency of 10Hz, to follow the same procedure as in Hinnekens et al. [1]. After signal treatment, in order to dispose of comparable signals, the root-mean-square ([RMS](#)) value of every filtered signal over 15 seconds of the [LIC](#) tests were normalized with the [RMS](#) value over 6 seconds of the [MVC](#) tests, giving the following formulas for [EMG](#) amplitude



Figure 2.7: Electrodes positioning.

computation:

$$EMG\ amplitude = \frac{RMS\ of\ filtered\ signal\ of\ LIC\ test\ over\ 15s}{RMS\ of\ filtered\ signal\ of\ MVC\ test\ over\ 6s} \quad (2.6)$$

Preprocess

Before starting the experiment, relevant anthropometric data were measured to have information about frontal balance of the pelvis, gibbosity and difference in leg sizes. Moreover, depth and width of the thorax at the level of T7 were measured for both patients. The maximum allowed gibbosity to be considered healthy was 5° , and it is important to note that the healthy patient showed a gibbosity of 5° in the lumbar region towards the right side (Table 2.1).

After body segment measurements, the subjects proceeded to a general warm-up of 5 minutes of cycling, followed by two exercises of specific warm-up: the bridge and the short lever arm Sorensen posture on a Swiss ball (Fig. 2.8).

After warm-up, the skin was cleaned with alcohol and degreased with ether to lower the skin impedance for EMG measurements. The eight pairs of electrodes were then positioned following muscles location with a 20 mm spacing between the



(a) Bridge warm-up



(b) Short lever arm Sorensen warm-up

Figure 2.8: Specific warm-up exercises.

two electrodes of a same pair (Fig. 2.7).

Finally, the subject was ready to get installed for the Sorensen test. To ensure stability, ankles, knees and pelvis were fixed to the frame with slacks (Fig. 2.5).

2.2.2 Maximum voluntary contraction tests

The subject had to maintain the position for 6 seconds. MVC trials had to be repeated at least three times with one minute rest in between in order to have two best trials from which to take the mean.

2.2.3 Loaded isometric contraction tests

For the LIC tests, the participants each performed a series of 4 static contractions holding different loads attached to the harness. The loads were corresponding to 0, 5, 10 and 15 percent of the participant's body weight (BW), and they were applied in a random order to prevent the learning effect (Table 2.3). At the end of the experiment, the participants were asked to evaluate the difficulty and the pain felt during the tests on a scale of one to ten, ten being the most difficult or the most painful.

	Load 1		Load 2		Load 3		Load 4		Subject Answers	
	[%]	[kg]	[%]	[kg]	[%]	[kg]	[%]	[kg]	Difficulty/10	Pain/10
Healthy	0	0	10	5	5	2.5	15	7.5	2	1
Scoliosis	10	4.6	0	0	15	6.9	5	2.3	4-5	5

Table 2.3: Loads carried by the participants and answers to the felt difficulty and pain.

2.3 Multibody modeling

A multibody system (MBS) is defined as an assembly of several rigid or flexible bodies, connected by joints. The bodies are subjected to external and internal forces and torques. As soon as the number of bodies in a multibody system increases, the complexity of determining the equations of movement increases rapidly. Consequently, the use of numerical MBS modeling tools can be of significant assistance

in determining the aforementioned equations. For this study, the model used was the three dimensional model developed by Hinnekens [1] with the ROBOTRAN software [30], comprising 31 rigid bodies:

- The pelvis
- The sacrum
- The 5 lumbar vertebrae
- The 12 thoracic vertebrae
- The 2 twelfth ribs
- The 2 eleventh ribs
- The 2 arms, 2 forearms and 2 hands
- The head (the cervical vertebrae were assumed to be rigid and part of the head body)
- The vertical load applied to the subject through a harness at the level of the ninth thoracic vertebra.

The ROBOTRAN model of Hinnekens is based on a fully articulated thoracolumbar spine and ribcage model of a male aged 25 with a height of 175 cm and a weight of 78 kg [26], which was available in the OpenSim Software [31]. In order to fit the different subjects, the models are scaled using factors:

- $SF_{m,i} = \frac{m_{subject,i}}{78}$ with $m_{subject,i}$ the mass of the i^{th} subject, to scale the subject mass
- $SF_{h,i} = \frac{h_{subject,i}}{1.75}$ with $h_{subject,i}$ the height of the i^{th} subject, to scale the center of mass locations and the body dimensions

In this model, it is important to note that the only vertebrae allowed to move relative to one another are the S1 and L5 vertebrae, considered as a hinge representing the movement of the spine relative to the pelvis in the Sorensen posture. Therefore, the only joint that is non forced-driven is the S1/L5 joint. Consequently, only muscle fascicles acting over this joint were modeled, meaning the ones having insertion points on each side of it (Fig. 2.9 and Table 2.4).

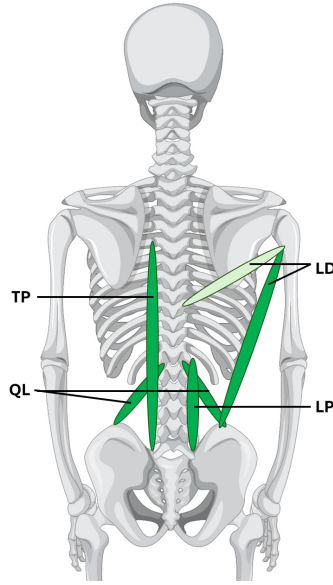


Figure 2.9: Scheme of the fascicles of the four muscle groups. Adapted from [32]. The light green LD fascicle is not modeled in the MBS, but its activity is measured during EMG. See Section 2.3.3.

Muscle	Fascicle	Number of fascicles (insertion points)
LP	ILpL	4 (L1→L4)
	LTpL	5 (L1→L5)
TP	ILpT	8 (R5→R12)
	LTpT	6 (T7→T12)
		6 (R7→R12)
QL	QL-p from I1	1 (I1-L3)
	QL-p from I2	3 (L4→L2)
	QL-p from I3	3 (L1→L3)
	QL-a from I2	1 (T12)
		1 (R12 ¹)
	QL-a from I3	1 (T12)
		3 (R12 ^{1,2,3})
LD	LD from I1	1 (R11)

Table 2.4: Muscle fascicles included in the model of Hinnekens [1]. Adapted from [1]. QL-p and QL-a stand for the posterior and anterior fiber of the QL muscle respectively. For the insertion points, L stands for lumbar vertebra, T for thoracic vertebra, and R for rib. I1, I2 and I3 are different insertion points on the pelvis. Each subscript of R12 refers to a different insertion point on this rib.

2.3.1 Formalism

In the context of [MBS](#), generalized coordinates q are the variables used to describe the position and orientation of each body. They can be independent (q_u), dependent (q_v) or forced-driven (q_d). The dependent variables can be expressed as a function of the independent ones, such as $q_v = q_v(u)$, whereas the forced-driven variables are known functions of the time, such as $q_d = q_d(t)$.

In this model, the only independent coordinate $q_{u,hinge}$ is the rotation of the lowermost [IVD](#), between S1/L5. However, this rotation is known, but the global equivalent force described in Section 2.1 is the only unknown of the system. Therefore, the resolution of the system necessitates the application of a numerical method called inverse equilibrium with exchange of variable. In this approach, the exchange of the only independent coordinate $q_{u,hinge}$ is assigned to the global equivalent force F_{equiv} .

The general equation of motion of a [MBS](#) subjected to constraint is :

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{c}(\mathbf{q}, \dot{\mathbf{q}}, \mathbf{frc}, \mathbf{trq}) = \mathbf{Q}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{J}^T(\mathbf{q})\lambda \quad (2.7)$$

$$\mathbf{h}(\mathbf{q}) = \mathbf{0} \quad (2.8)$$

Where:

- $\mathbf{M}(\mathbf{q})$ is the symmetric positive-definite mass matrix;
- $\mathbf{c}(\mathbf{q}, \dot{\mathbf{q}})$ contains the centripetal, Coriolis and gyroscopic terms as well as the external forces $\mathbf{frc}(\mathbf{q}, \dot{\mathbf{q}})$ and torques $\mathbf{trq}(\mathbf{q}, \dot{\mathbf{q}})$;
- $\mathbf{Q}(\mathbf{q}, \dot{\mathbf{q}})$ are the efforts within the joints;
- $\mathbf{h}(\mathbf{q})$ is the column-vector of constraints;
- $\mathbf{J}(\mathbf{q}) \triangleq \frac{\partial \mathbf{h}}{\partial \mathbf{q}^T}$ is the constraint Jacobian matrix.

Its reduced form in case of a static equilibrium (see [\[1\]](#)) becomes:

$$\mathbf{c}_r(\mathbf{q}_u^*, \mathbf{q}_d, \mathbf{0}, \mathbf{frc}, \mathbf{trq}) - \mathbf{Q}_r(\mathbf{q}, \mathbf{0}) = \mathbf{0} \quad (2.9)$$

with $q_u^* = q_{u,hinge}$.

And when applying the exchange of variable $\{\mathbf{q}_{u,hinge}\} \leftrightarrow \{F_{equiv}\}$, the final equation is:

$$\mathbf{c}_r(\mathbf{F}_{equiv}, \mathbf{q}_d, \mathbf{0}, \mathbf{frc}, \mathbf{trq}) = \mathbf{Q}_r(\mathbf{q}, \mathbf{0}) \quad (2.10)$$

2.3.2 Adaptation of the model

The model was adapted for this study in order to eliminate the symmetry assumption of right and left muscles and to incorporate joints between each vertebra to insert the geometry of the scoliosis in the three planes. Furthermore, the version of the model of [1] including the LD muscle was utilized as explained in section 3.1.

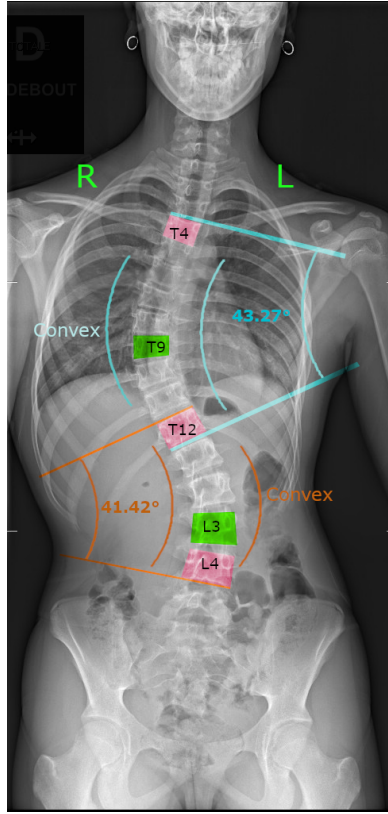


Figure 2.10: Anterior-posterior view of the spine of Subject 2. Superior and inferior end vertebrae are highlighted in red, apical vertebrae are highlighted in green, Cobb angles are represented. R and L stand for the right and left sides of the subject. Blue reports to the thoracic curve and orange to the lumbar curve. Convex sides of the two curves are shown as well.

The joints used to represent the curvatures of the scoliosis in the three planes for Subject 2 were set as forced-driven coordinates and their value did not change throughout the simulation. To that end, measurements were taken on the X-rays (Fig. 2.10). Regarding the rotations in the sagittal plane, the coordinates of the anterior corners of the vertebral bodies were measured (Fig. 2.11, left side), and a simple algorithm based on basic geometry and trigonometry allowed to compute consecutive rotations. Regarding the frontal plane, the coordinates of the middle point of the endplates of each vertebral body were used (Fig. 2.11, right side). This is because these points had to be computed for the rotations in the transverse planes. Indeed, the axial vertebral rotation (AVR) was computed based on the Stoke method [33]. This method uses the distance between the midline of the vertebral body to the two pedicles in order to determine the AVR, noted γ here (Fig. 2.11, right side). Its expression is depicted hereafter.

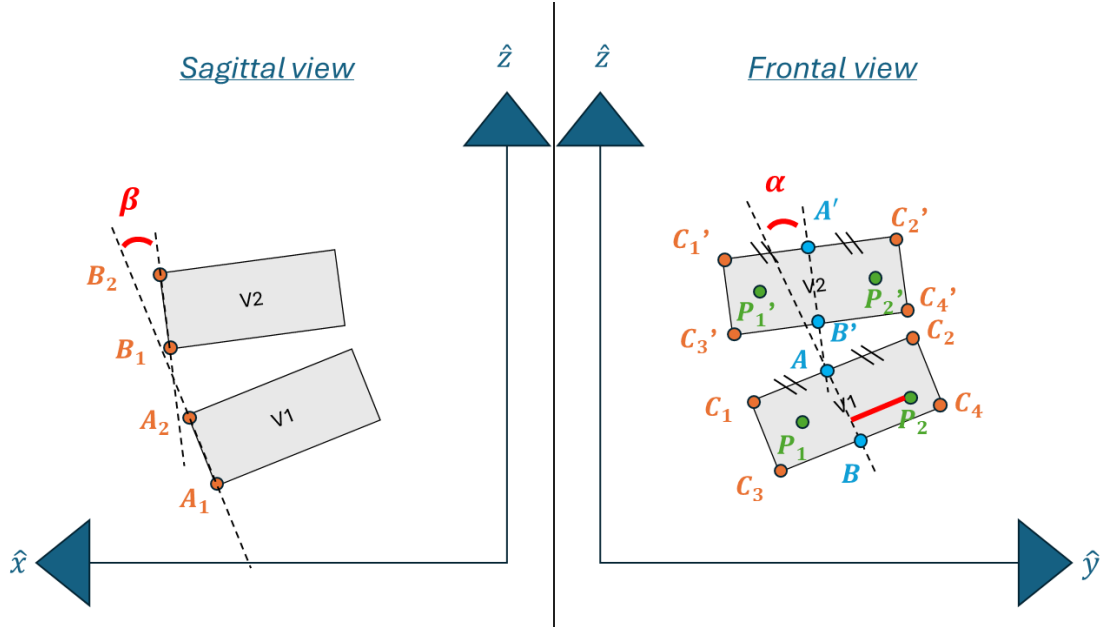


Figure 2.11: Scheme of the points measured to compute the angles between each vertebra in the three planes. Orange points are the coordinates of corners of vertebral bodies. Blue points are the coordinates of points at the intersection of the vertebral endplates and the midline of the vertebral bodies. Green points are the coordinates of the pedicles of the vertebra. The thick red line shows the distance between the midline of the vertebral body and a pedicle. Frontal view based on [33].

Computation of α :

$$\alpha = \arctan\left(\frac{A'_y - B'_y}{A'_x - B'_x}\right) - \arctan\left(\frac{A_y - B_y}{A_x - B_x}\right)$$

Computation of β :

$$\beta = \arctan\left(\frac{B_{2,y} - B_{1,y}}{B_{2,x} - B_{1,x}}\right) - \arctan\left(\frac{A_{2,y} - A_{1,y}}{A_{2,x} - A_{1,x}}\right)$$

Computation of γ :

Let m and n be the distance between the midline of the vertebral body and the pedicles P_1 and P_2 respectively. According to Stokes:

$$m = \left| \frac{(A_y - B_y)P_{1,x} + (A_x - B_x)P_{1,y} + B_x(B_y - A_y) + B_y(A_x - B_x)}{\sqrt{(A_y - B_y)^2 + (A_x - B_x)^2}} \right|$$

$$n = \left| \frac{(A_y - B_y)P_{2,x} + (A_x - B_x)P_{2,y} + B_x(B_y - A_y) + B_y(A_x - B_x)}{\sqrt{(A_y - B_y)^2 + (A_x - B_x)^2}} \right|$$

Then, if γ_1 is the [AVR](#) of the vertebra 1 and γ_2 is the [AVR](#) of the vertebra 2, the relative [AVR](#) of the vertebra 2 compared to the vertebra 1, γ , can be computed as follows:

$$\gamma_1 = \arctan\left(0.5 \frac{m - n}{m + n} \left(\frac{w}{d}\right)\right)$$

$$\gamma_2 = \arctan\left(0.5 \frac{m' - n'}{m' + n'} \left(\frac{w'}{d'}\right)\right)$$

$$\gamma = \gamma_2 - \gamma_1$$

Where $\frac{w}{d}$ are the width-to-depth ratio for the vertebrae from L4 to T4 given by Stokes et al [34]. Consequently, the assumption was made that the [AVR](#) of the vertebrae from S1 to L3 and from T3 to T1 was null. Even though this assumption seems reasonable for T3 to T1, as vertebra L3 is the apex of the lumbar curve, it suggests that it shows an [AVR](#) that is non-negligible. However, as the [AVR](#) of L3 could not be computed with this method, it was set to 0 anyway.

2.3.3 Post-processing

As the unknown of the [MBS](#) equations is the equivalent force F_{equiv} , the latter split into its two contributions $F_{equiv,0}$ and $F_{equiv,r}$, this is what gets out of the [MBS](#) algorithm. However, what is of interest are the intervertebral efforts as well as the muscle forces. The convention for the positive signs of the efforts are illustrated in Figure 2.12. In the musculoskeletal model, [IVDs](#) are modeled as joints. Therefore, the forces acting in the [IVDs](#) consists of joint forces which are computed during the algorithm. From there, the intervertebral efforts can be deduced using the [IVDs](#) transverse area for results in pressure [MPa] or using the subject's mass for results in force per kg [N/kg].

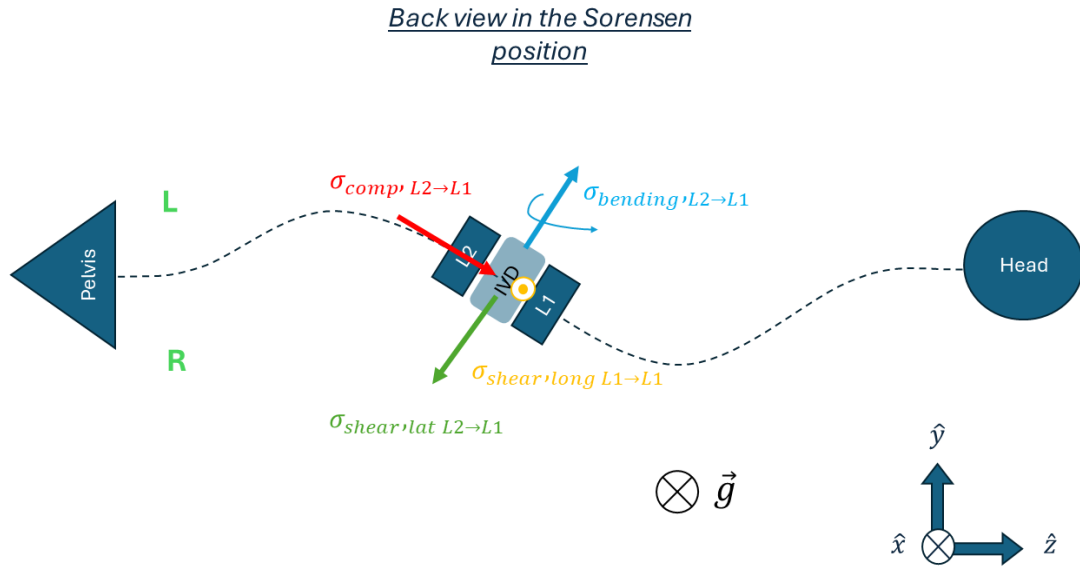


Figure 2.12: Intervertebral resultant efforts with arrows indicating the positive value direction. Efforts are from vertebra L2 on vertebra L1.

However, finding the muscle forces is not as direct and requires one additional computing step. Indeed, the term corresponding to the force developed by one muscle has already been implicitly introduced in Equation 2.3. As a reminder, $w_{F0,m}$ and $w_{Fr,m}$ are the proportion of the equivalent force produced by muscle m in the absence of additional load and for an increase in load applied, respectively. As a result, the force produced by muscle m in the absence of additional load is:

$$F_{musc,0} = w_{F0,m} * F_{equiv,0} \quad (2.11)$$

And the total muscle force produced when an additional load is applied is:

$$F_{musc,r} = F_{musc,0} + w_{Fr,m} * F_{equiv,r} \quad (2.12)$$

Knowing the proportion of the force developed by every fascicle of each muscle, it is also possible to retrieve the force developed by each fascicle:

$$F_{fa,i} = w_{fa,i} * F_{musc,r} \quad (2.13)$$

In the present model, it is important to remember that the thoracic spine is considered rigid. Consequently, only efforts acting in the lumbar spine were computed. Moreover, because only the fascicles having insertion points at the level of the S1/L5 joint were modelled, only these fascicle forces could be computed. For the **LD** muscle, this comprises only one of its fascicle. However, during the experiment, the electrodes recorded the activity of the entire muscle at the position of the electrodes. That is why it is important to note the difference between **EMG** signal intensity and F_{musc} intensity, as they do not represent the same number of fascicles for the **LD** muscle.

Chapter 3

Results

First, results of the experiment are exposed in order to validate the measurements used afterwards in the model, but also to already show some interesting facts. Then, results from the simulation of the model are presented. After some research, it was found that the model used by Hinnekens [26] was no longer available, and that another full-body model with a fully-articulated thoraco-lumbar spine scaled for every age between 6 to 18 years old was available, a model of Schmid et al. [35]. This new model displayed different muscle properties such as the maximum isometric forces. Consequently, the weights associated with the force contribution of each fascicle for this new model are different. Therefore, the results were computed once more using the weights based on the new model, and these new results are presented at last.

3.1 Experimental electromyographic measurements

3.1.1 Maximum voluntary contraction tests

The mean forces developed by the subjects during the MVC tests measured thanks to the force sensor attached at the harness are presented in Table 3.1.

	S1 - Healthy	S2 - Scoliosis
Mean MVC force [N]	191.73	220.39
Mean MVC force [% of BW]	39.09%	48.84%

Table 3.1: Mean force values developed by each subject during the MVC tests given in Newtons and in percentage of body weight. The average is computed from the two best trials.

Both the mean force in [N] and the mean force relative to the subject's body weight are higher for the subject with scoliosis. Subject 2 produces approximately 10% more of its BW.

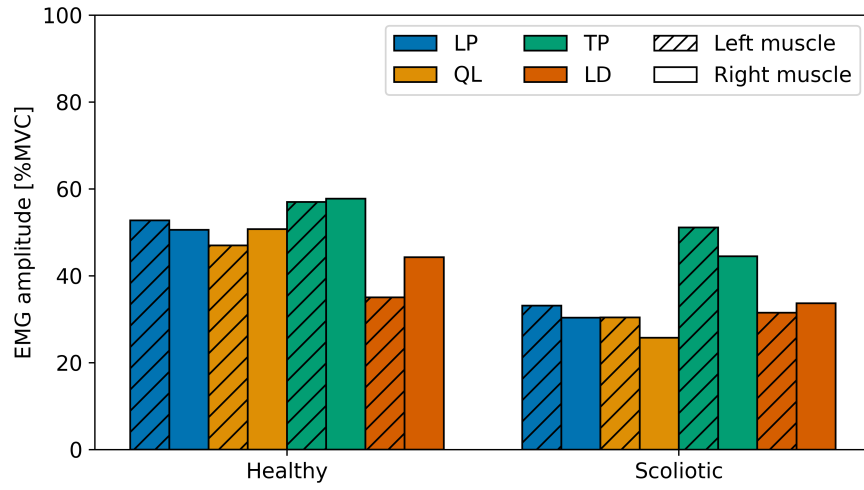
3.1.2 Loaded isometric contraction tests

As a reminder, the EMG amplitude is defined as the ratio of the RMS value of the filtered LIC signal over 15s over the RMS value of the filtered MVC signal over 6s. The proportion of the EMG amplitude of each muscle is represented in Fig. 3.1.

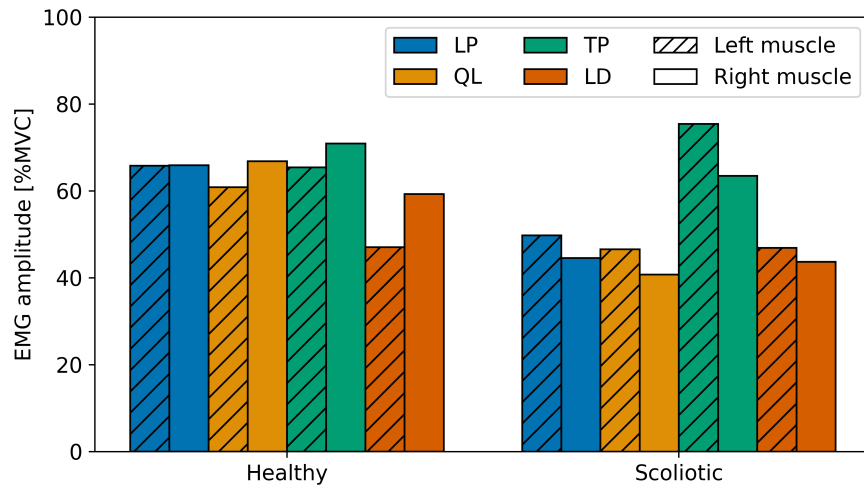
After signal treatment, the weights of equation 2.3 could be computed using equations 2.4 and 2.5. As the muscle tones are obtained based on the RMS signals, they depict the same profiles than Fig. 3.1.

In general, the subject affected by scoliosis contracted all her muscles to a lower intensity relative to the maximum possible, with an average over the muscles of 14% of lower relative contraction for no added load (Fig. 3.1a). However, for an increment in load applied, the increment in activation was significantly higher in the case of scoliosis. Indeed, for an increase in load applied of 15% of BW (Fig. 3.1b), the average increase in EMG amplitude of each muscle was 28% for the healthy subject and 47% for the scoliosis subject. Regarding the level of activation of every muscle, it was relatively the same for each muscle for the healthy subject, with the LD being slightly less activated. However, for the scoliosis subject, it was the level of activation of the TP that was significantly higher than for her other muscles, even more regarding the left side.

Moreover, in the healthy case, muscle tones between right and left muscles were quite similar, except for the LD which showed a more pronounced asymmetry: the left LD had a muscle tone 21.24% lower than its right side. However, Subject 2 presented a more pronounced asymmetry between right and left muscles for every muscle group. The right LP muscle tone was 13.12% less than its left side. The right QL muscle tone was 13.86% less than its left side. The right TP muscle tone was 17.02% less than its left side, and the left LD muscle tone was 14.50% less than its right side. In average, the importance of the asymmetry for the healthy subject was around 9.09%, whereas the one of the scoliosis subject was around 14.63%.



(a) Static trial with 0% of added BW load



(b) Static trial with 15% of added BW load

Figure 3.1: Proportion of the EMG amplitude of each muscle for the two subjects. Fig. 3.1a corresponds to the static trial with 0% of added BW load, and Fig. 3.1b corresponds to the static trial with 15% of added BW load.

For later comparison with results of the literature, two more quantities were displayed hereafter:

1. The linear regression that was used to compute the muscle tones and the load rates (Fig. 3.2);
2. A ratio of the **RMS** value of the non-normalized **EMG** signals of the convex side over the concave side [23], called the **RMS-EMG** ratio:

$$rms \text{ EMG ratio} = \frac{rms \text{ EMG of convexity}}{rms \text{ EMG of concavity}} \quad (3.1)$$

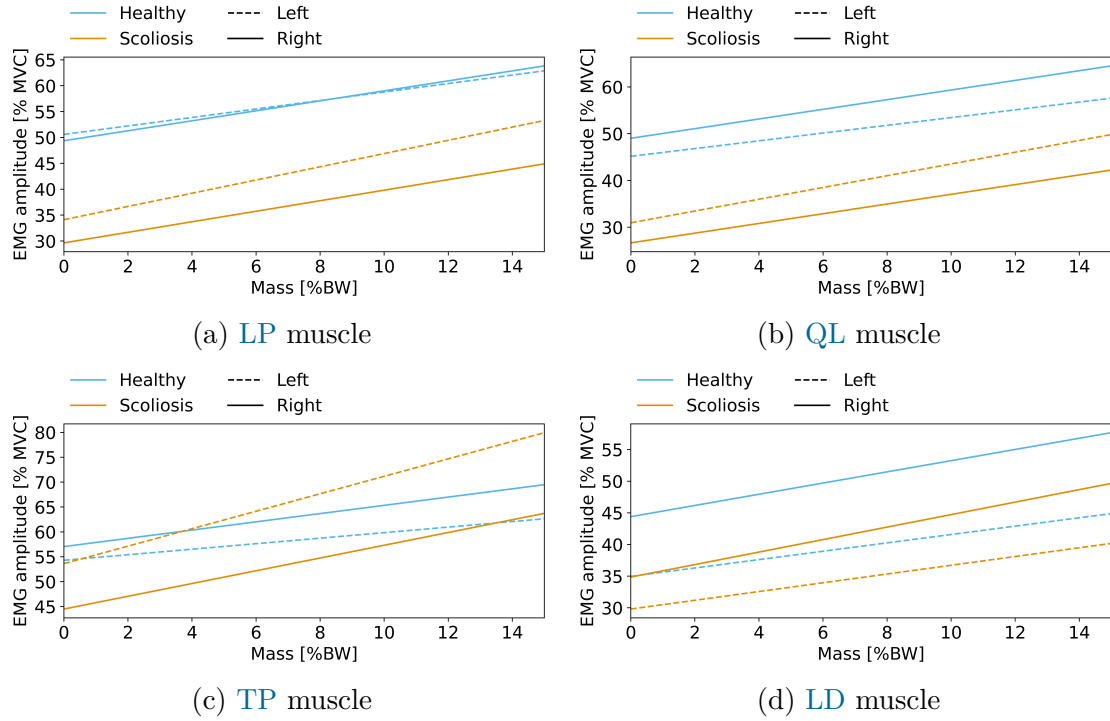


Figure 3.2: Linear regression of the **EMG** amplitude vs load carried relationship for the four muscle groups and the two subjects.

Fig. 3.2 showed consistently higher slopes for the scoliotic case compared to the healthy one, but lower values of ordinates at origin. In addition, the lines of the healthy subject demonstrated closer values of slopes and ordinate at origins between left and right than the lines of the scoliotic subject, except regarding the ordinate at origins of the **LD** which are closer for the scoliotic case.

The **RMS-EMG** ratio was computed at the apical level of the major curve, i.e. the lumbar curve, for the paraspinal muscles. Indeed, the signals at the apex were available as the apex is L3 for the lumbar curve, and the **LP** was measured at that level. The value of the **RMS-EMG** ratio for the lumbar curve was 1.74.

3.2 Quantification of intervertebral efforts and muscle forces

3.2.1 Intervertebral efforts

The intervertebral efforts, i.e. the compressive and shear forces in [N/kg] and the bending torque in [Nm/kg], for the two subjects for each lumbar **IVDs** are illustrated in Figs. 3.3 and 3.4, respectively.

It can be observed that, in general, the efforts for every load were always greater at every level for the scoliosis patient. Table 3.2 shows the impact of the scoliosis on the efforts at each lumbar level. One exception to the latest statement was observed: the longitudinal shear force at the L4/L3 **IVD** was slightly smaller for the scoliosis patient only for 0% **BW** load. Table 3.2 also revealed that for the L5/L4 segment, the geometry of the scoliosis increased all efforts to more or less the same extent, but mostly the bending torque. One additional noteworthy observation was the lateral shear force at the L2/L1 **IVD**, which was greater in magnitude for the scoliosis subject but in an opposite direction.

IVD level	L5/L4		L4/L3		L3/L2		L2/L1	
Percentage of BW load	0%	15%	0%	15%	0%	15%	0%	15%
Compressive force	+19	+39	+20	+39	+24	+44	+16	+32
Longitudinal shear force	+19	+36	-1	+5	-	-	-	-
Bending torque	+27	+44	-	-	+2	-	+9	+54

Table 3.2: Increase (%) of the efforts of the **IVDs** of the scoliotic subject relative to the healthy subject for 0% and 15% of added **BW** load. The lateral effort is not shown as the healthy subject's lateral shear stress is almost 0 N/kg. Some values are missing because one of the subjects displayed very small value at these levels making the comparison in % not valid.

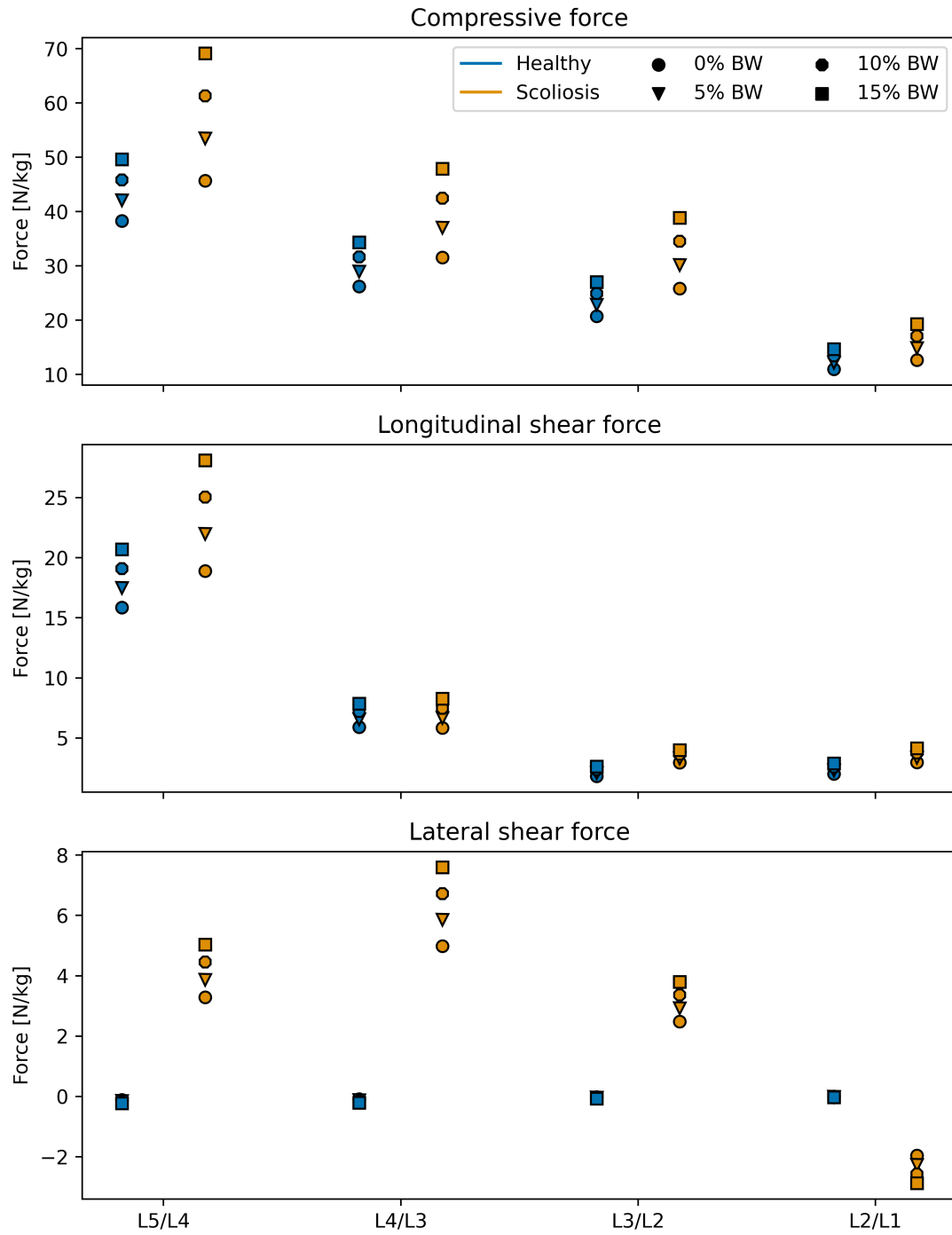


Figure 3.3: Normalized compressive, longitudinal shear and lateral shear forces in [N/kg] in the lumbar IVDs for the two subjects.

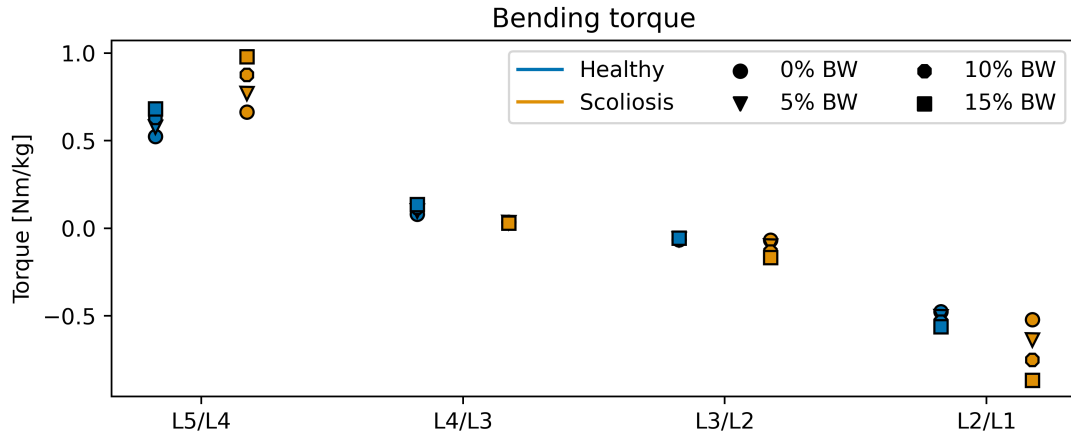


Figure 3.4: Normalized bending torques in [Nm/kg] in the lumbar IVDs for the two subjects.

Secondly, an increase in load was consistently accompanied by an increase in effort, except for the bending torque at the intermediate IVD levels, and with a greater impact on the scoliosis patient. Table 3.3 compares the impact of an increasing load on the efforts at every lumbar IVD for the two subjects. This impact was indeed almost always greater for the scoliosis subject, with the two exceptions being for the longitudinal shear stress at the L3/L2 and L2/L1 IVDs that were slightly greater for the healthy subject. It is also interesting to remark that the increase in effort for an increase of load applied was relatively homogeneous over the different lumbar segments. Finally, increasing the added load to 10% and 15% of BW increased the compressive force at the apex by 35% and 52% compared to the unloaded case, respectively. This is useful for further literature comparison.

Thirdly, the lateral shear force showed the greatest contrast between the two subjects, with the healthy subject showing minimal lateral shear force in comparison to the scoliosis subject.

Finally, for the compressive and longitudinal forces as well as for the bending torque, the greatest difference between the two subjects was observed at the L5/L4 level. However, for the lateral shear force, the greatest difference between the two subjects was observed at the L4/L3 level.

In order to facilitate comparison with existing literature, equivalent stresses have been calculated for both subjects in the LIC with 10% of added BW load (Table 3.4). To obtain these aforementioned equivalent stresses, it was necessary to

	L5/L4	L4/L3	L3/L2	L2/L1
Compressive force				
Healthy	9.0%	9.4%	9.2%	10.2%
Scoliosis	14.8%	14.9%	14.6%	15.1%
Longitudinal shear force				
Healthy	9.3%	10.1%	13.2%	12.8%
Scoliosis	14.1%	13.3%	10.6%	11.7%
Lateral shear force				
Healthy	-	-	-	-
Scoliosis	15.3%	15.1%	15.2%	13.7%
Bending torque				
Healthy	9.2%	18.8%	-	5.6%
Scoliosis	13.9%	-	35.2%	18.5%

Table 3.3: Average increase (%) in the efforts for 5% additional [BW](#) load. Some values are missing because one of the subjects displayed very small value at these levels making the comparison in % not valid.

	L5/L4	L4/L3	L3/L2	L2/L1
Compressive stress				
Healthy	1.70	1.17	0.92	0.50
Scoliosis	2.09	1.45	1.18	0.58
Longitudinal shear stress				
Healthy	0.71	0.27	0.09	0.10
Scoliosis	0.85	0.25	0.12	0.13
Lateral shear stress				
Healthy	-0.01	-0.01	0.00	0.00
Scoliosis	0.15	0.23	0.11	-0.09
Bending torque				
Healthy	2.25	0.42	-0.21	-1.91
Scoliosis	2.87	0.10	-0.44	-2.47

Table 3.4: Maximum equivalent stresses [MPa] for both subjects for 10% added [BW](#) load for the Sorensen test.

know the value of the area of the IVD. Since the subjects were aged 14 and 13, the area of the IVDs was set to 13.5 [cm²] [36]. These results highlighted that the most important forces in magnitude were the compressive forces. The bending torque was also important in magnitude at the two extreme lumbar levels.

3.2.2 Muscle fascicles forces

To compute the force in every modeled fascicle of every muscle group, Equation 2.3 was used once more. It should be noted that the LP and QL were considered as acting in the lumbar region (Fig. 3.5), whereas the TP and LD acted in the thoracic region (Figs. 3.6 and 3.7). As the fascicles of the TP are numerous, they have been divided in two figures, with the fascicles that are part of the ILpT muscle (Fig. 3.6a) and the ones of the LTpT muscle (Fig. 3.6b).

The initial observation is that the profiles of the distribution of force among the fascicles were similar for the two subjects, with different magnitudes (Figs. 3.5 and 3.6).

Secondly, when considering all muscle groups, it is interesting to look at Table 3.5, which is a direct reflection of the muscle tone associated with each muscle. However, it is very crucial to note that this table accounts for the entire LD, including all its fascicles, even those not directly influencing the S1/L5 joint. Consequently, its amplitude is not negligible. However, the only fascicle acting on the S1/L5 joint is illustrated in Fig. 3.7 and accounts for approximately 7% of the total LD muscle force. For a more detailed understanding, refer to the discussions chapter (Chap. 6).

On the one hand, Table 3.5 illustrates that, for both subjects, the muscle contributing the most to the equivalent force produced was the TP, while the LD contributed the least for the healthy subject and the least contribution came from the QL for the scoliosis subject. On the other hand, the scoliosis subject relied more on its thoracic muscles to produce force, whereas the healthy subject had a more homogeneous dependency over the muscles. Finally, the table demonstrated opposite trends between the right and left muscles for the QL and the TP. For the healthy subject, the QLR and TPR contributed more to the force than their left counterparts. Conversely, the scoliosis subject exhibited the opposite pattern. However, both subjects relied more on the right side of their LD and on the left side of their LP, although not at same extent of disproportionality.

Thirdly, when comparing fascicles between them, the fascicles contributing the most to the equivalent force from all fascicles of all muscle groups, were the fascicles

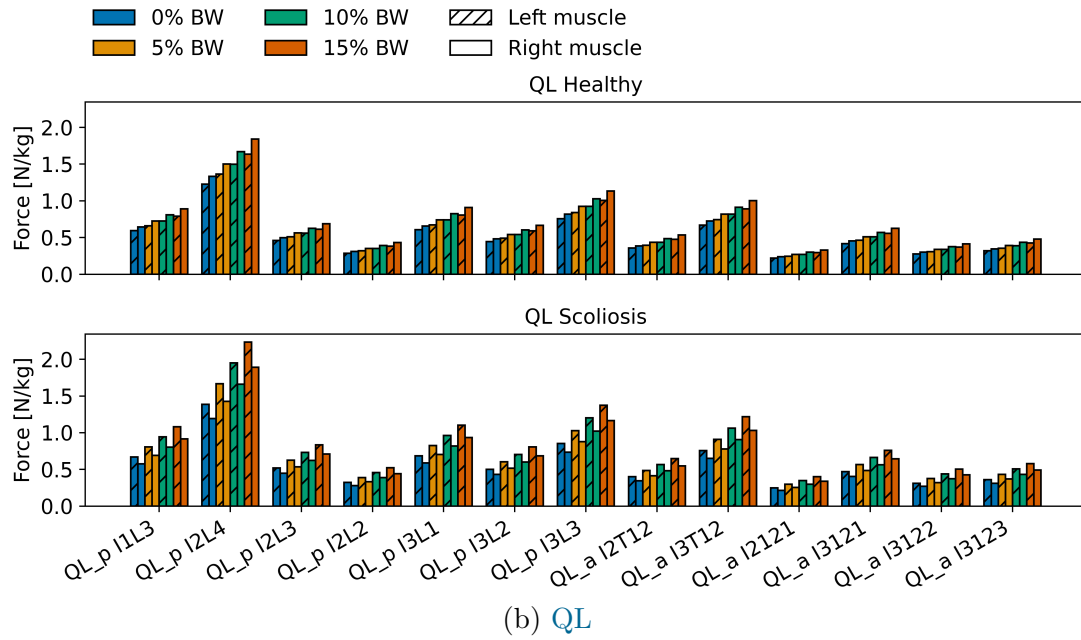
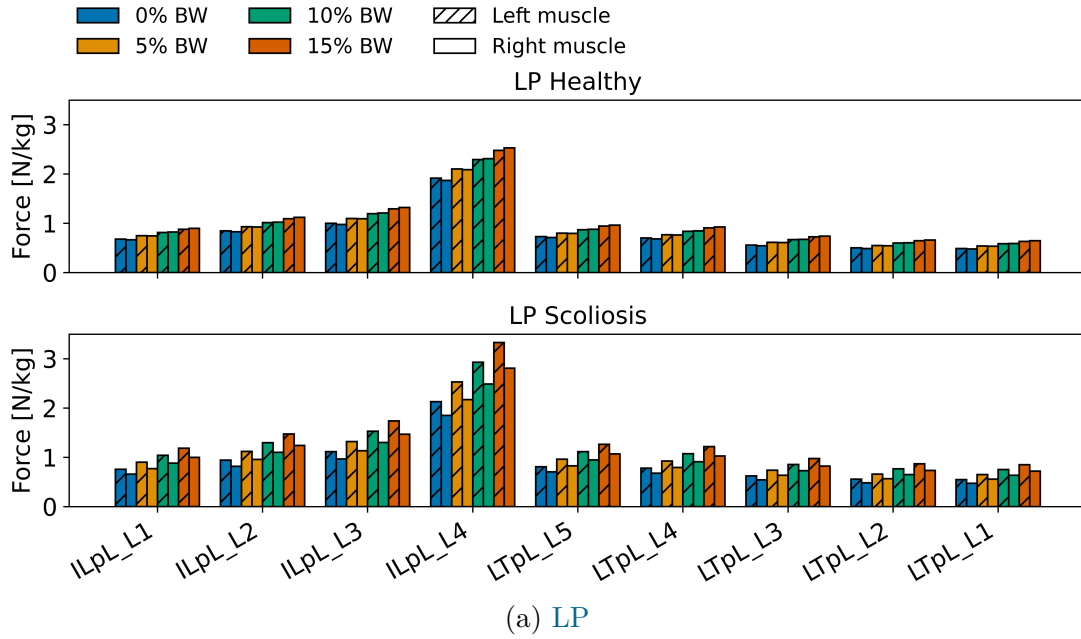
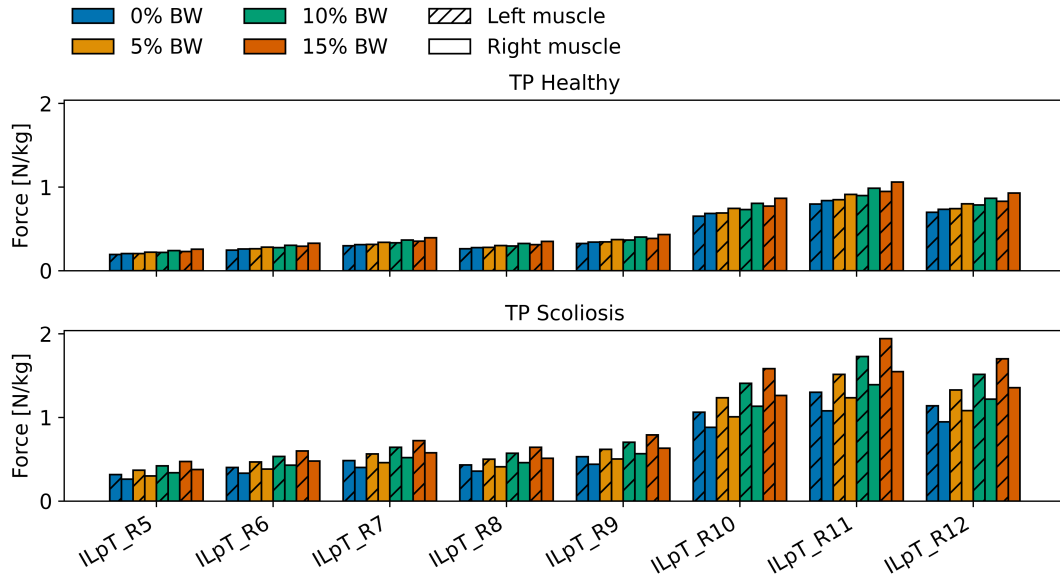
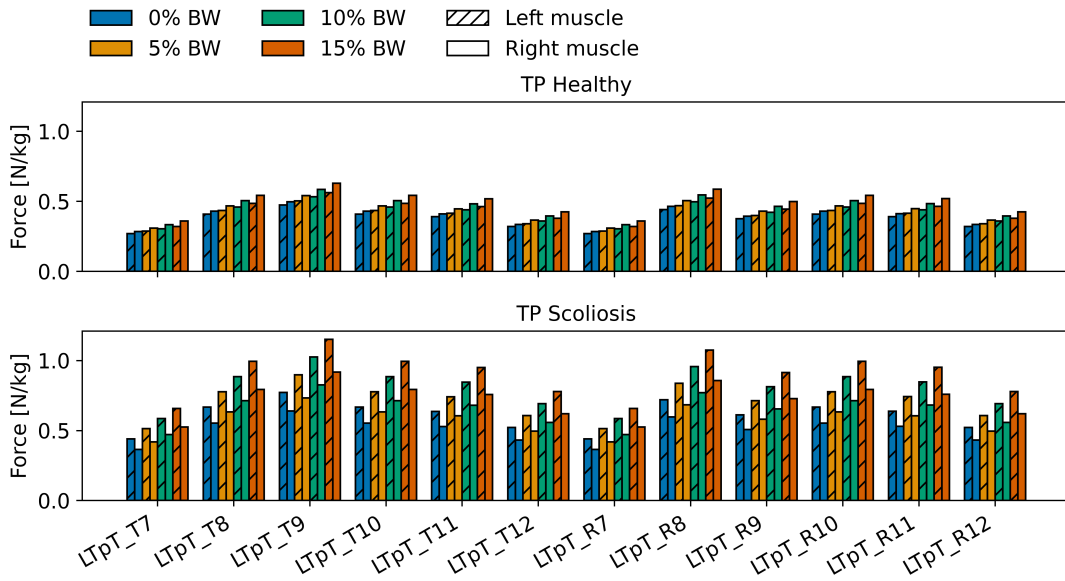


Figure 3.5: Forces in the fascicles of the muscles LP and QL for the two subjects for the different static tests, the two muscles acting in the lumbar region.



(a) TP - ILpT fascicles



(b) TP - LTpT fascicles

Figure 3.6: Forces in the fascicles of the muscle TP for the two subjects for the different static tests, one of the two muscles acting in the thoracic region.

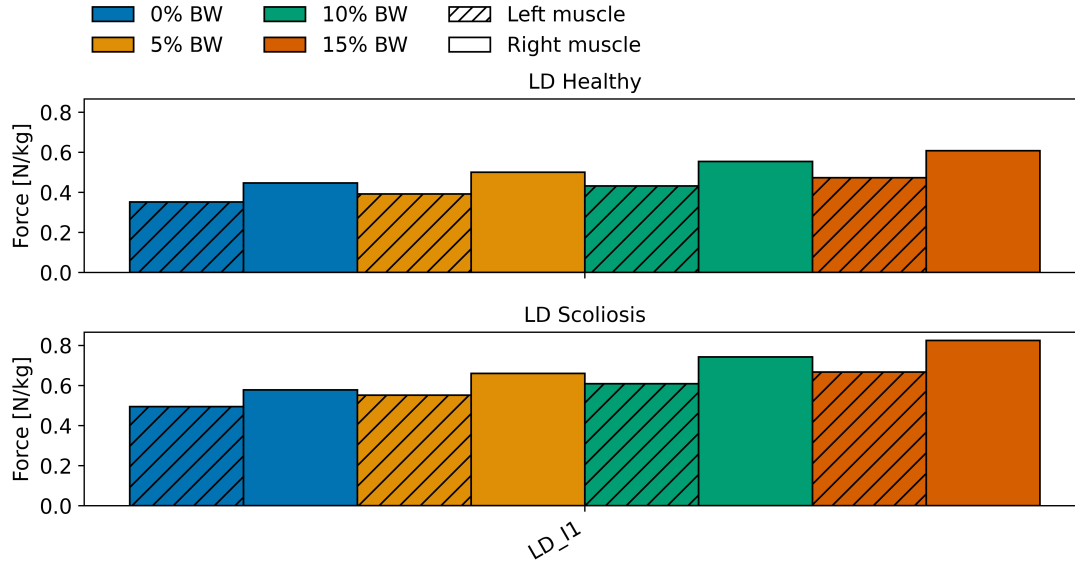


Figure 3.7: Forces in the fascicles of the muscle LD for the two subjects for the different static tests, the second muscle acting in the thoracic region.

		F_{musc} (N) (% of F_{equiv})		
		R	L	R+L
LP	Healthy	362 (12.82%)	371 (13.14%)	733 (25.96%)
	Scoliosis	330 (10.43%)	380 (12.02%)	710 (22.45%)
QL	Healthy	360 (12.75%)	331 (11.72%)	691 (24.47%)
	Scoliosis	297 (9.39%)	344 (10.88%)	641 (20.27%)
TP	Healthy	419 (14.84%)	398 (14.10%)	817 (28.94%)
	Scoliosis	495 (15.65%)	597 (18.88%)	1092 (34.53%)
LD	Healthy	326 (11.54%)	256 (9.07%)	582 (20.61%)
	Scoliosis	388 (12.27%)	332 (10.50%)	720 (22.77%)

Table 3.5: Equivalent force distribution among the four muscle groups for the LIC with 0% of added BW load given in [N] and in % of the equivalent force. R and L stands for the right and left muscle of the muscle group, respectively. R+L sums up the right and left parts of the muscle group.

of the **ILpL** inserting at the level of the L4 vertebra. In addition, the fascicles of the **QL** inserting at this same L4 vertebra were also the ones producing the most force relative to the other **QL** fascicles. For the **TP**, the fascicles producing the most force were those of the **ILpT** inserting around the tenth, eleventh, and twelfth ribs, with the ones inserting on the eleventh ribs showing the highest force for both subjects.

Furthermore, a comparison of the right and left fascicles revealed that the asymmetry observed at the fascicle level followed a similar pattern to that observed at the muscle level (Table 3.6), but provided a further analysis. The **LP** demonstrated a greater force production on the left side in the case of the healthy subject for the **LICs** with 0% and 5% added **BW** load, but the trend started reversing between 5 and 10% added **BW** load, as the **LIC** tests with 10% and 15% added **BW** load showed greater force among the right fascicles of the **LP** for the healthy subject. Regarding the scoliosis subject, it was observed that the asymmetry increased with the charge. However, as the Hinnekens method used a linear regression over the entire domain, variation of the slopes in function of the load could not be observed. A linear regression by parts over the distinct segments of the domain could reveal new results by highlighting the presence of a critical load. Consequently, a linear regression by parts had been conducted (Fig. 3.8) for rapid investigations. Yet, it was not employed into the model, as the complete method would have to be revised to account for varying load rates and muscle tones across each segment of the domain.

Subject	LIC trial	LP	QL	TP	LD
Healthy	0% BW	-2.42	+8.53	+5.07	+26.97
	15% BW	+2.15	+12.55	+11.99	+28.72
Scoliosis	0% BW	-13.12	-13.86	-17.02	+16.96
	15% BW	-15.72	-15.24	-20.29	+23.61

Table 3.6: Mean percentage variation of the mean right fascicles force relative to the mean left fascicles force for **LIC** with 0% and 15% of added **BW** load.

Fig. 3.8 showed variations in the slopes for different loads. Indeed, the healthy subject consistently showed an increase in the slope of the linear regression for every muscle around 10% of added **BW**. However, the scoliosis subject, yet more unpredictable, seemed to show an increase in the slope earlier on than the healthy subject.

Finally, Table 3.6 allowed to observe that the degree of the asymmetry increased as the load increased, for both subjects.

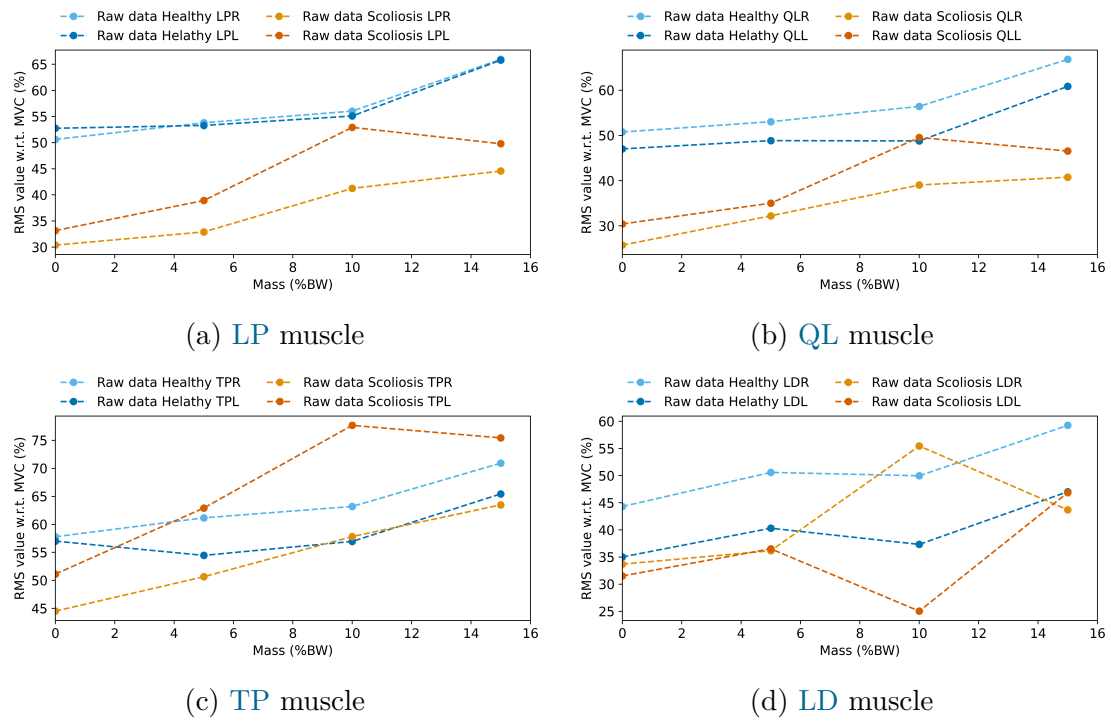
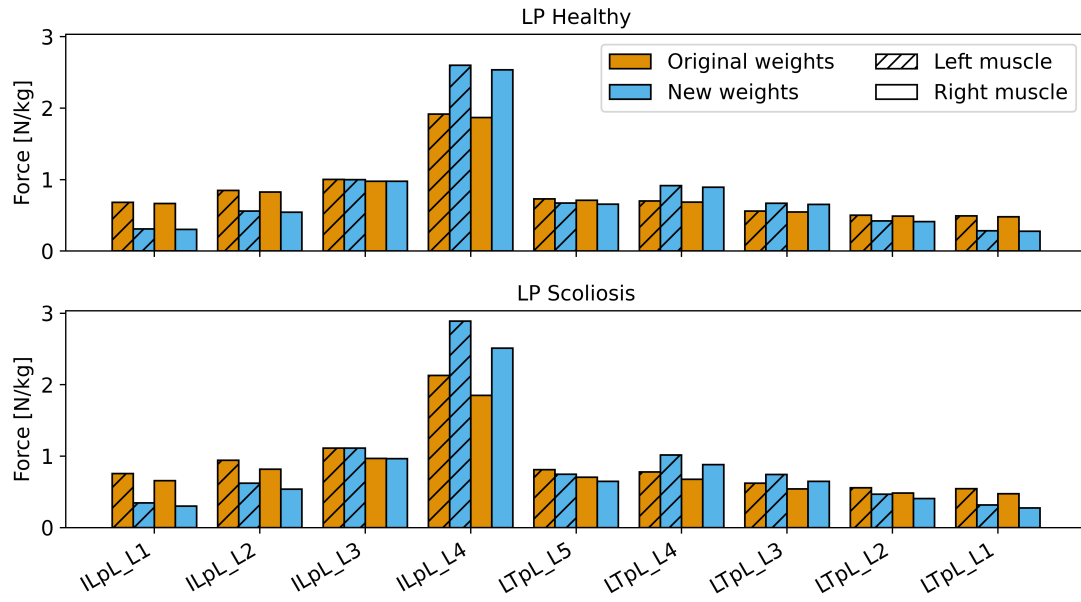


Figure 3.8: Linear regressions by parts of the EMG amplitude vs load carried relationship for the four muscle groups and the two subjects.

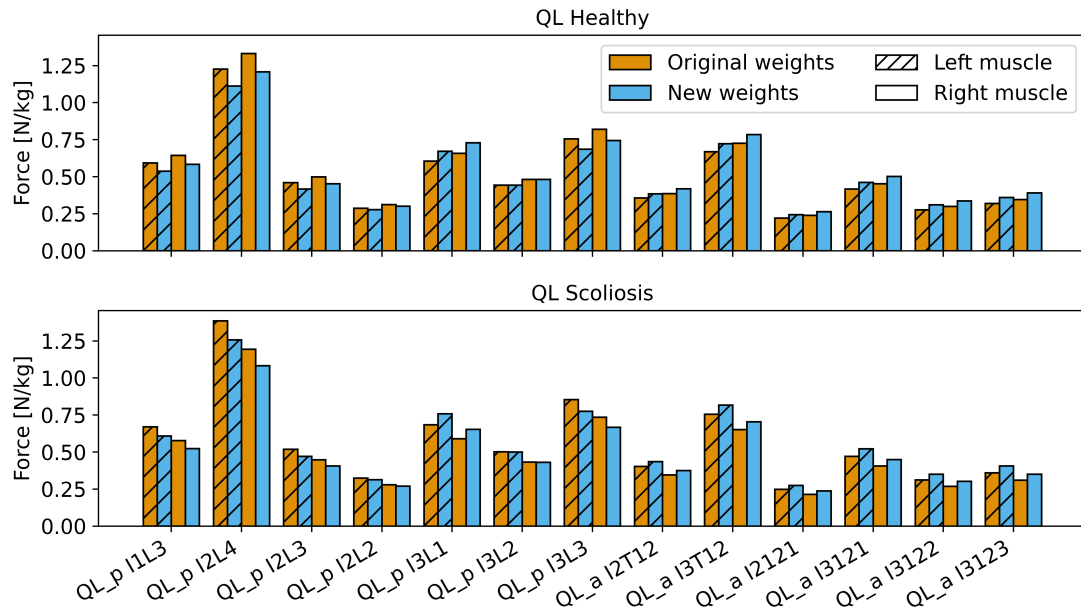
3.2.3 Comparison with a new OpenSim model

For the new model, the only changing parameters were the weights w_{fa} of the fascicles, for both subjects. Therefore, the impact on the intervertebral efforts remained minimal, whereas the greatest contrast laid within the muscle fascicle forces. These forces have been computed for the static trial at body weight and are displayed in Figs. 3.9, 3.10 and 3.11.

Upon comparison of the two OpenSim models, the first observation was a similar distribution of the forces amongst the fascicles of the two models for every muscle, except for the **LTpT** fascicles of the **TP** muscle which depicted a slightly different distribution. After a more in-depth comparison, the **LP** muscle showed a more marked disproportion between the weakest and strongest fascicles for the new model. The **QL** muscle showed a very similar force distribution. The **ILpT** fascicles of the **TP** muscle followed the same trend as the **LP** muscle, accentuating the disproportions in the new model. Finally, for the **LD**, the fascicle that is modeled in the **MBS** was given more weight in the new OpenSim model.

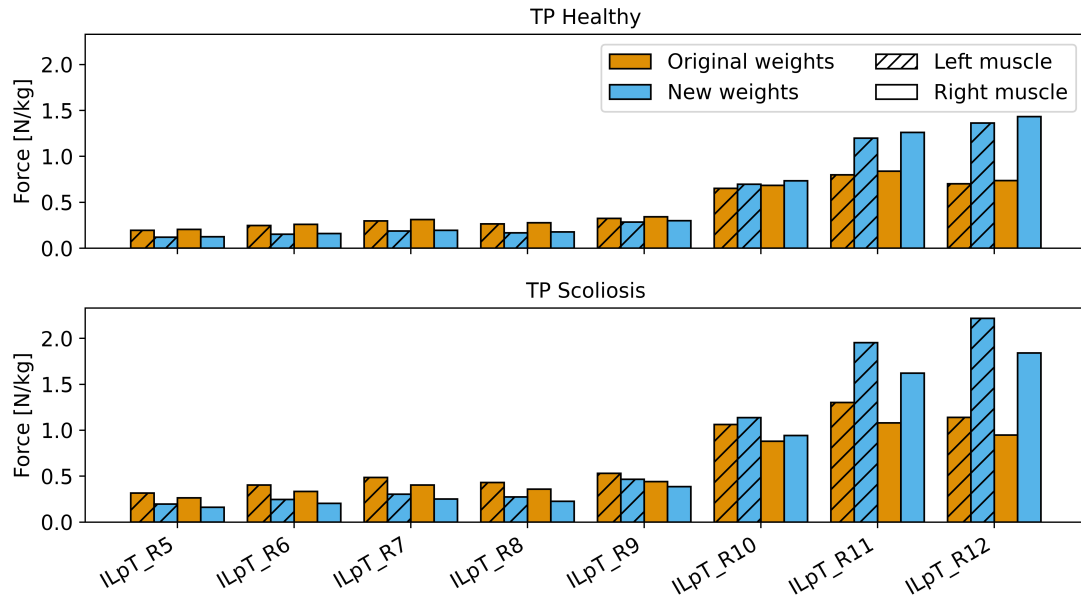


(a) LP

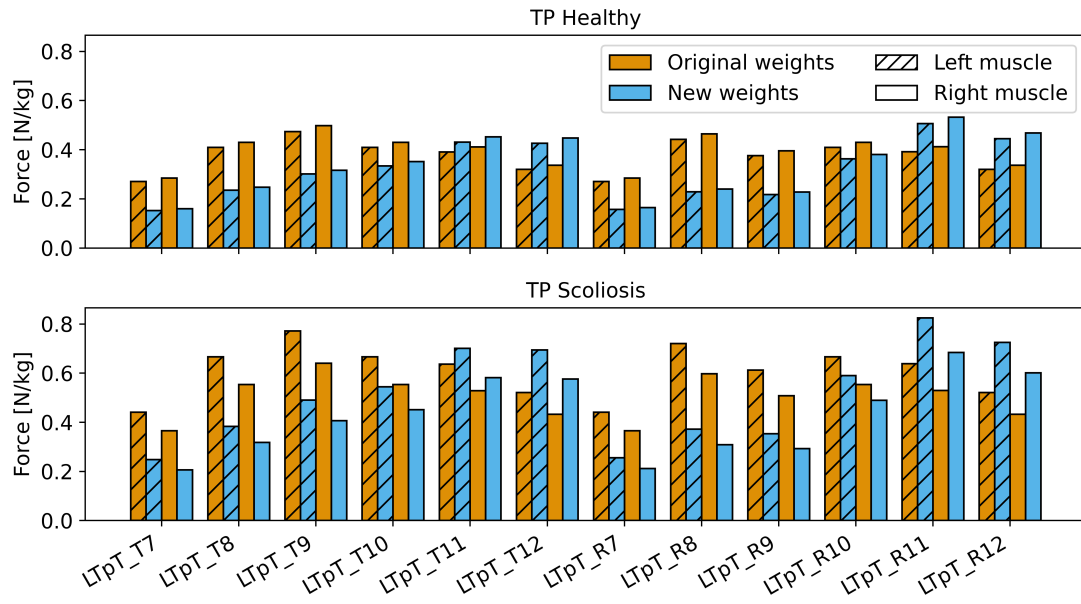


(b) QL

Figure 3.9: Forces in the fascicles of the muscles LP and QL for the two subjects for the two models for the static test at body weight.



(a) TP - ILpT fascicles



(b) TP - LTpT fascicles

Figure 3.10: Forces in the fascicles of the muscle TP for the two subjects for the two models for the static test at body weight.

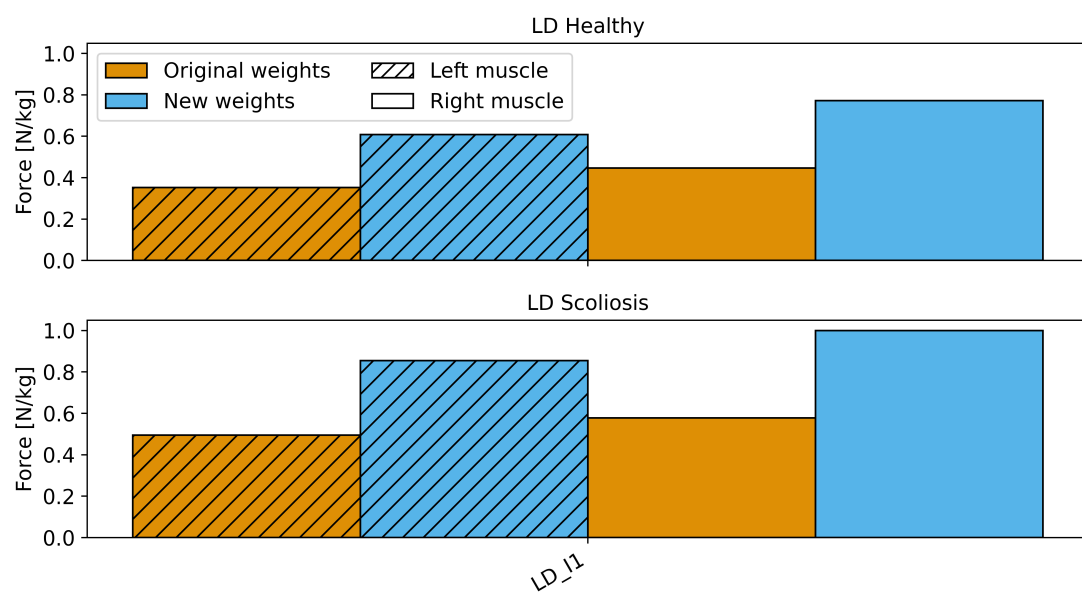


Figure 3.11: Forces in the fascicles of the muscle **LD** for the two subjects for the two models for the static test at body weight.

Chapter 4

Sensitivity Analysis

4.1 Impact of the geometry on the intervertebral efforts

4.1.1 Description of the different geometries

The geometry of the model is defined by the angles between each of the vertebrae inserted in the [MBS](#) model in the three planes for the scoliosis subject. Therefore, this section will only focus on the second subject. These angles have been computed in four different ways. First, based on the X-rays, the angles in the frontal and sagittal planes were simply measured directly on the radio. However, this method, called here the [Manual](#) method, could not provide the angles in the transverse plane which require the knowledge of coordinates in the frontal plane, as explained in subsection [2.3.2](#). The three other methods used the measurements of the coordinates of points as described in subsection [2.3.2](#). These coordinates were measured in three ways:

- On printed X-ray on A4 sized paper, this is the method called [A4](#);
- On printed X-ray on A3 sized paper, this is the method called [A3](#);
- Using the open source software GIMP [\[2\]](#), a software to manipulate images that allows to find coordinates of clicked points in pixels. This is the method called [GIMP](#).

4.1.2 Comparison of the different inputs

The variation in angles from one vertebra to the next in the three planes are illustrated in [Fig. 4.1](#) for the four methods. This figure showed a close profile for

the four methods with non-negligible variations. From these graphs, the [A4](#) and [A3](#) methods seemed to be quite close, whereas the [GIMP](#) method seemed to be less precise due to its higher and opposite variations. For the [Manual](#) method, the angles in the transverse plane of method [A4](#) were used in the model.

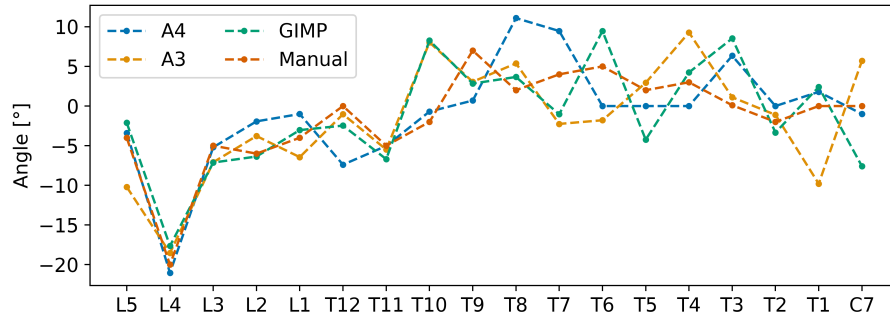
Fig. 4.1 only shows the variation in angles from one vertebra to the next, therefore the global rotation in the frontal plane of each vertebra can be visualized in Fig. 4.2, taken from the three dimension visualizer of the multibody model in ROBOTRAN [30]. This visualization suggests that the [A4](#) method seemed to be the closest one to reality. In addition, it can be visualized that the [A3](#) method exaggerated the distance to the midline of the body for the thoracic apical vertebra and deviates for the lumbar apical vertebra, whereas the [Manual](#) method seemed to diminish every rotation globally but kept the same apical vertebrae. Finally, the [GIMP](#) method deviated for the upper apical vertebra T9, and seemed to consider L2 as the apical vertebra instead of L3.

Upon observation of the different geometries, it was confirmed that the apical vertebrae were the ones with the least frontal rotation (Fig. 4.1b), but the most axial rotation (Fig. 4.1c). Furthermore, the upper and lower end vertebrae were the ones with the most frontal rotation.

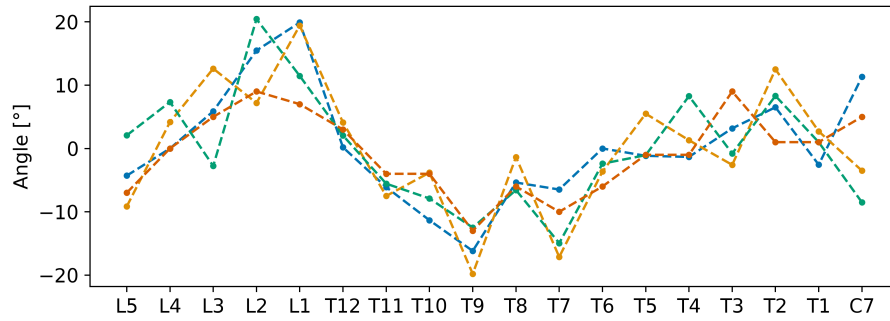
Knowing the Cobb angles of the scoliosis subject, a validation of the different geometries for the frontal plane could be made (Table 4.1). Because the [A4](#) method was the one closer to reality, this method was the one selected for the final [MBS](#) model and every result presented before were obtain for the [A4](#) geometry. Therefore, this method is taken as the reference and other methods will be compared relative to it.

		Real values	Manual	A4	A3	GIMP
Thoracic Cobb angle	[°]	43.27	36.0	44.79	49.03	43.44
	(%)	(100)	(83.20)	(103.51)	(113.31)	(100.39)
Lumbar Cobb angle	[°]	41.42	31.17	41.43	43.32	31.17
	(%)	(100)	(75.25)	(100.02)	(104.59)	(75.25)
Mean value	(%)	(100)	(79.23)	(101.77)	(108.95)	(87.82)

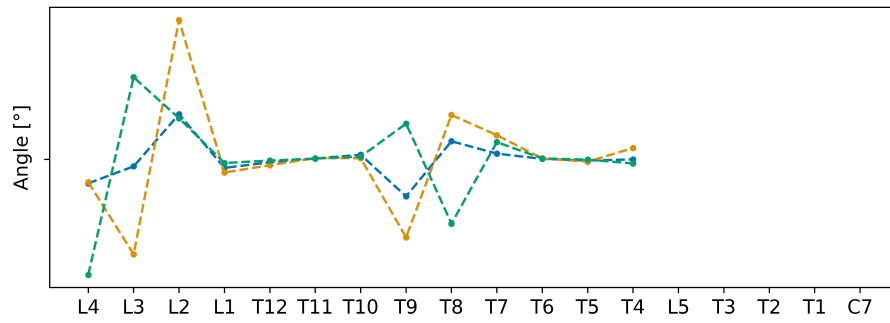
Table 4.1: Comparison of the Cobb angles computed for the different geometries in [°]. The values between brackets show the percentage relative to the real value of the angle. To overview the veracity of the values over the two Cobb angles (thoracic and lumbar), the mean of two angles in percentage of the real values has been computed in the last row.



(a) Sagittal plane angles

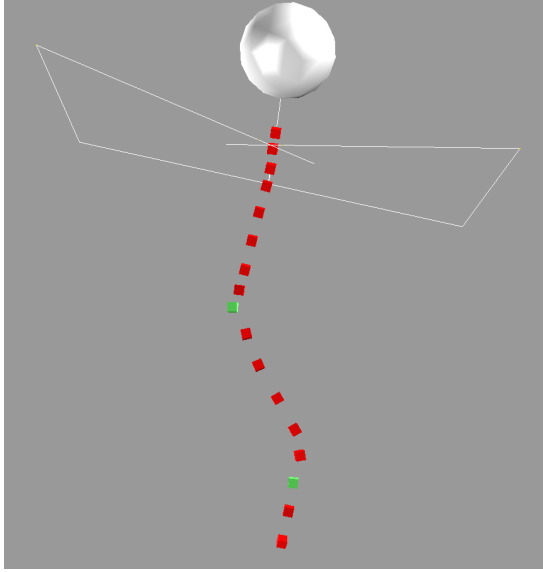


(b) Frontal plane angles

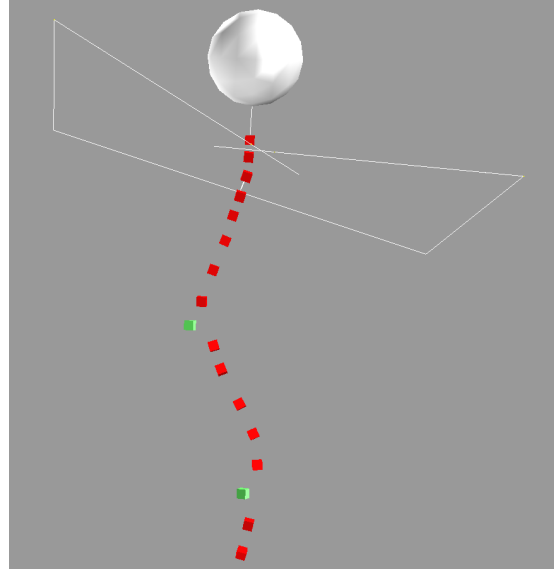


(c) Transverse plane angles

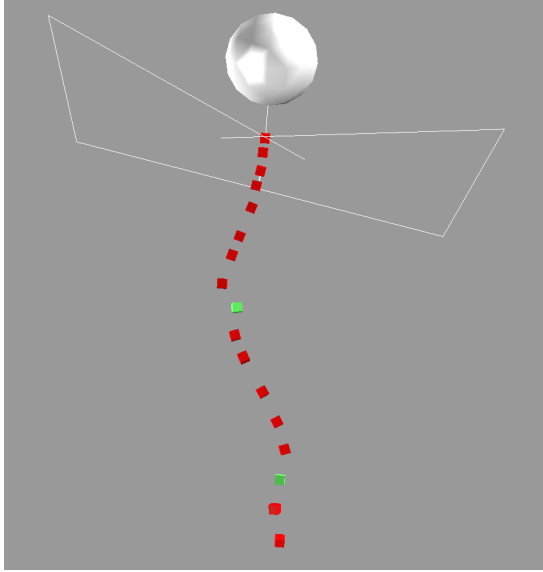
Figure 4.1: Comparison of the different geometries. The angles between successive vertebrae are presented for the different geometries in the three planes. The [Manual](#) method could not compute the angles in the transverse plane and was assigned the angles of the A4 method for that plane.



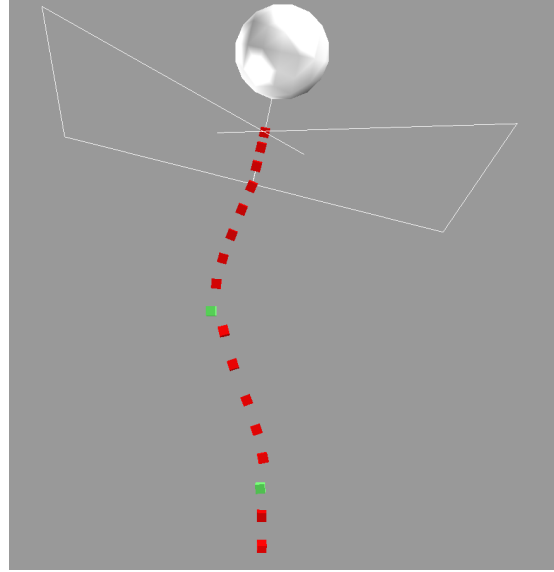
(a) Frontal geometry of the [A4](#) method.



(b) Frontal geometry of the [A3](#) method.



(c) Frontal geometry of the [GIMP](#) method.



(d) Frontal geometry of the [Manual](#) method.

Figure 4.2: Comparison of the different geometries in the frontal plane (anterior view). Image taken from the three dimension visualizer of the multibody model in ROBOTRAN [30]. The upper sphere represents the head, the thin white lines represent the arms, the red cubes represent the successive vertebrae with the two green ones being the two apical vertebrae, L3 and T9.

Table 4.2 shows the variations in angles of the different geometries relative to the A4 geometry, only for the lumbar regions as only there the efforts were computed. According to this table, the Manual method is the closest one to the A4 method. However, this method was the one that was the furthest away regarding Cobb angles.

	A3		GIMP		Manual	
Difference with A4	Mean	Max	Mean	Max	Mean	Max
Frontal angle [°]	4.75	8.29	6.27	8.65	4.30	12.90
Sagittal angle [°]	4.14	6.80	3.00	4.93	2.71	7.39
Transverse angle [°]	6.94	20.50	6.94	19.93	0	0

Table 4.2: Difference relative to the A4 method for the angles of the three planes for each geometry in the lumbar region.

4.1.3 Comparison of the intervertebral efforts

Figs. 4.3 and 4.4 illustrate the compressive and shear forces in [N/kg] and the bending torque in [Nm/kg], respectively.

Firstly, the lateral shear force was the effort that was most affected by the change in geometry. For all efforts except the lateral shear force, the general trend over the vertebrae was followed by the four geometries. The compressive force increased towards the pelvis as well as the longitudinal shear force, and the bending torque was maximal at the extreme lumbar levels. However, the lateral shear force suggested different scenarios. The A4 and Manual geometry showed lateral shear force profiles as curves with their maximum at the L4/L3 IVD, and they both considered L3 as the lumbar apical vertebra. The GIMP geometry had its maximum lateral shear force at the L3/L2 IVD, while it considered L2 as the lumbar apical vertebra. Finally, the A3 geometry showed a lateral shear force increasing towards the pelvis. This shows that the lateral shear forces tended to be maximal around the apical vertebra.

Secondly, regarding the compressive force, the A3 geometry seemed to cause more compression. The GIMP geometry caused similar compressive forces than the reference geometry, while the Manual geometry laid between the A3 and the reference geometry. The longitudinal shear force showed the same dependency on geometry than the compressive force for the levels L5/L4 and L4/L3. However, although the GIMP geometry remained the most similar to the reference in terms of longitudinal shear, the A3 and Manual geometries showed a change of trend. The A3 geometry caused less longitudinal shear for the L3/L2 and L2/L1 IVDs, and

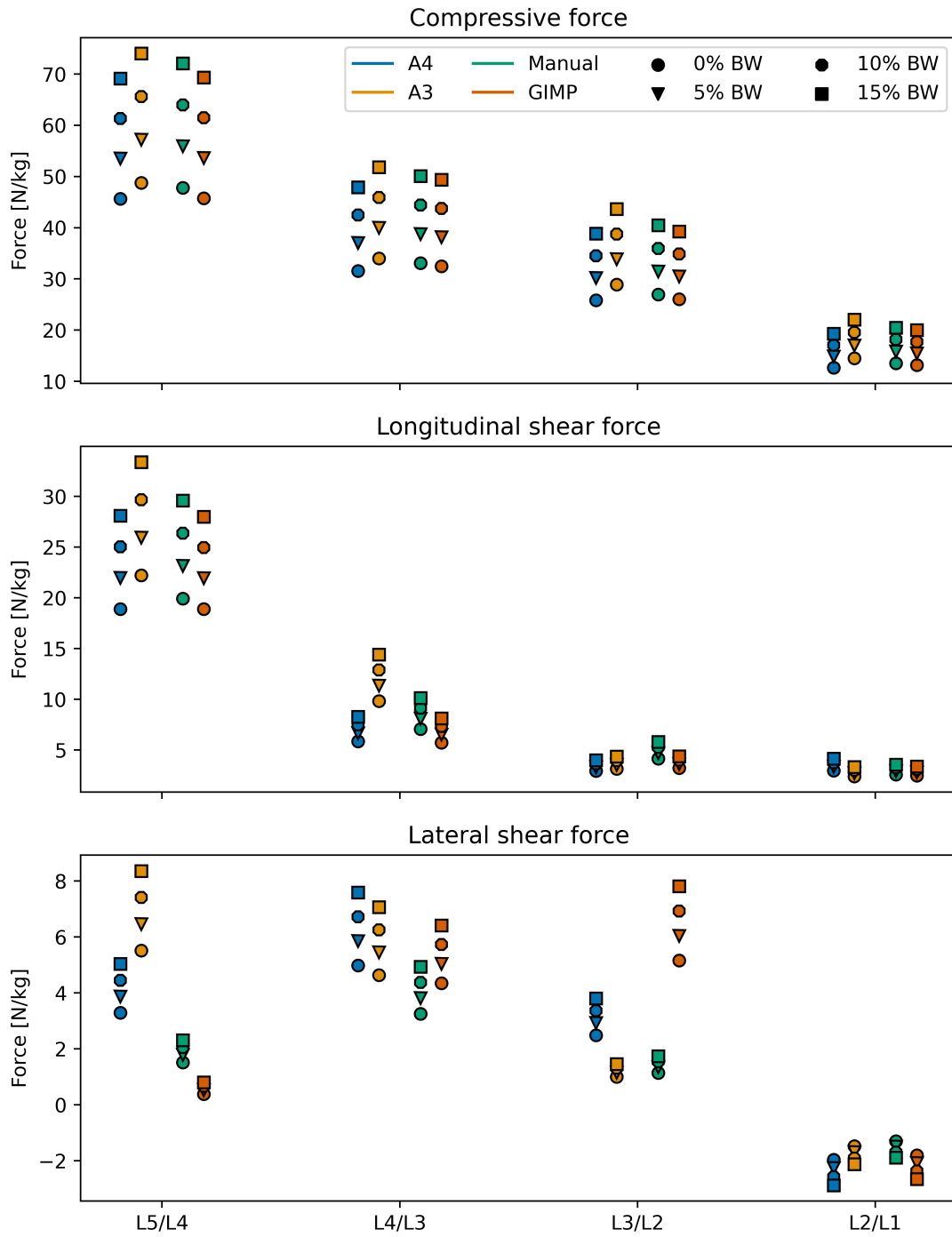


Figure 4.3: Normalized compressive, longitudinal shear and lateral shear forces in [N/kg] in the lumbar IVDs for the different geometries.

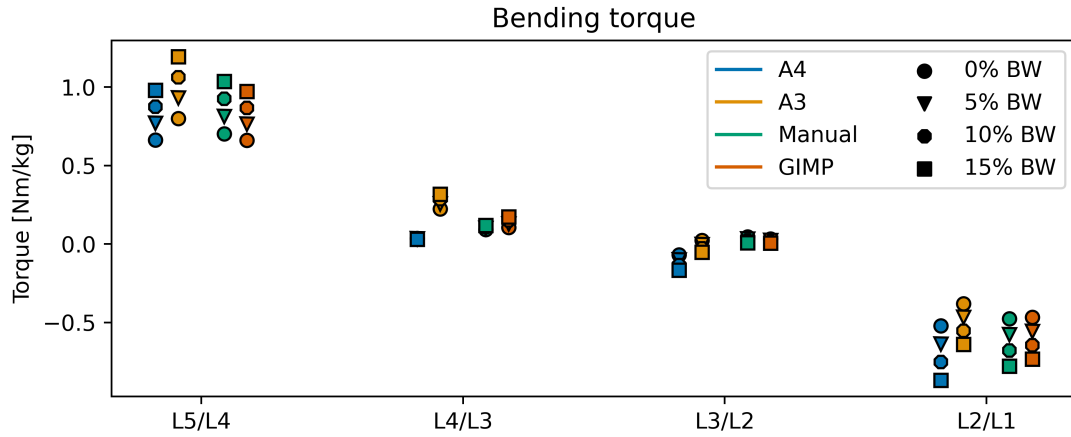


Figure 4.4: Normalized bending torques in [Nm/kg] in the lumbar IVDs for the different geometries.

the [Manual](#) method showed more longitudinal shear for the L3/L2 level but less for the L2/L1 level and remained higher in effort than the [A3](#) geometry. The bending torque showed similar magnitudes for the [Manual](#) and [GIMP](#) geometry compared to the reference, and the [A3](#) geometry seemed to shift the bending torques towards more positive values by keeping a similar profile.

To objectively quantify the impact of the geometry on the intervertebral efforts, Table 4.3 shows the relative increase in every effort for every IVD level relative to the [A4](#) method for 0% of added BW load.

Table 4.3 clearly showed a relationship between the lateral shear and the frontal angle. Indeed, the [Manual](#) method showed, in the lumbar region, smaller frontal angles than the [A4](#) method and the lateral shear of the [Manual](#) method were always smaller than the ones of the [A4](#) method. Furthermore, the [GIMP](#) method was associated with higher variations both over and under-estimating frontal angles compared to the reference, and the efforts followed the same variations compared to the reference. However, for the other efforts and other angles, no such direct correlations could be drawn. Indeed, by comparing Tables 4.2 and 4.3, the relationship between the geometry and the resulting efforts was likely to be complex, depending on multiple parameters at once. For example, the compressive forces were probably impacted by the sagittal angles, but also by the transverse ones, and it was hard to determine which one had the most impact. Therefore, this seems to indicate that the computation of the internal efforts for scoliosis patients needs to be based on a subject-specific model, including the right geometry, which can be a

	L5/L4	L4/L3	L3/L2	L2/L1	Mean
Compressive forces					
A3	+6.8%	+7.8%	+12.0%	+14.4%	+10.2%
GIMP	+0.1%	+2.9%	+1.0%	+3.9%	+2.0%
Manual	+4.6%	+4.8%	+4.4%	+6.8%	+5.2%
Longitudinal shear					
A3	+17.6%	+68.0%	+6.5%	-19.5%	+18.1%
GIMP	-0.1%	-2.0%	+9.3%	-16.6%	-2.3%
Manual	+5.4%	+21.0%	+40.5%	-12.9%	+13.4%
Lateral shear					
A3	+67.8%	-6.9%	-59.8%	-24.7%	-5.9%
GIMP	-88.3%	-12.7%	+107.5%	-7.9%	-0.4%
Manual	-53.8%	-34.8%	-54.0%	-33.9%	-44.1%
Bending torque					
A3	+20.4%	+535.7%	-132.9%	-27.2%	+99.0%
GIMP	-0.5%	+198.9%	-147.1%	-10.5%	+10.2%
Manual	+5.7%	+159.9%	-166.0%	-8.8%	-2.3%

Table 4.3: Increase in % of every effort at every IVD level relative to the A4 method for 0% of added BW load.

tedious task.

Finally, every geometry showed the same impact due to the load in the same proportion at every level for every effort (Figs. 4.3 and 4.4).

4.1.4 Conclusion

First of all, the higher the values of the frontal angles, the more important the lateral shear efforts, with the maximum lateral shear stress being seemingly located at the apical vertebra. However, for the other efforts, their relationship with geometry was complex and needed subject-specificity. Nevertheless, the lateral shear was the effort that was the most impacted by the geometry, as the other efforts followed each time the same profiles, with varying magnitudes.

Chapter 5

Validation of the Computation of the Weights $w_{F0,m}$ and $w_{Fr,m}$ through the Adapted Method

As a reminder, the original hybrid approach of Hinnekens, on which this study is based, was developed for healthy subjects, assuming a symmetry between right and left muscles. In this context, Hinnekens computed the weights of the contribution of each muscle to the global equivalent force, $w_{F0,m}$ and $w_{Fr,m}$ in Equation 2.3, using a single linear regression for both the right and left parts of the muscles. In this linear regression, the data points were obtained as a mean between the right and left parts.

As the present study aimed to obtain results for a scoliosis subject, the method of Hinnekens had to be adapted to let go of this symmetry assumption. To that end, distinct linear regressions were computed for each right and left muscles. The algorithm of the present study consequently executed twice as many linear regressions per subject as the algorithm used in Hinnekens' work. This resulted in two times more weights $w_{F0,m}$ and $w_{Fr,m}$ being associated with each subject, as explained in Section 2.1.

It is therefore of interest to validate the adaptation effectuated. In the present chapter, the name original method is attributed to the one developed by Hinnekens for healthy subjects, assuming symmetry between the right and left muscles. In contrast, the name adapted method is attributed to the one used in the present study, taking into account the asymmetry between the right and left muscles. As the adapted method is also applied to a healthy subject in the present study, and as the original method is intended to be applied to healthy subjects, the original method was applied to the healthy subject of the present study, and these two scenarios are compared hereafter. If the adaptation is appropriate, the results of

the original and of the adapted methods should be close. Nevertheless, as even the healthy subject exhibits some degree of gibbosity, some asymmetry should be revealed by the adapted method and disregarded by the original one.

5.1 Muscle tone

Before comparing the outcome of the modeling of the two methods, the inputs of the modeling can already be compared. These inputs were retrieved from the experimental [EMG](#) measurements. The signals are already processed in different ways to obtain the muscle tone w_{F0} and the load rate w_{Fr} . Fig. [5.1](#) shows the muscle tone computed for the adapted method and for the original one.

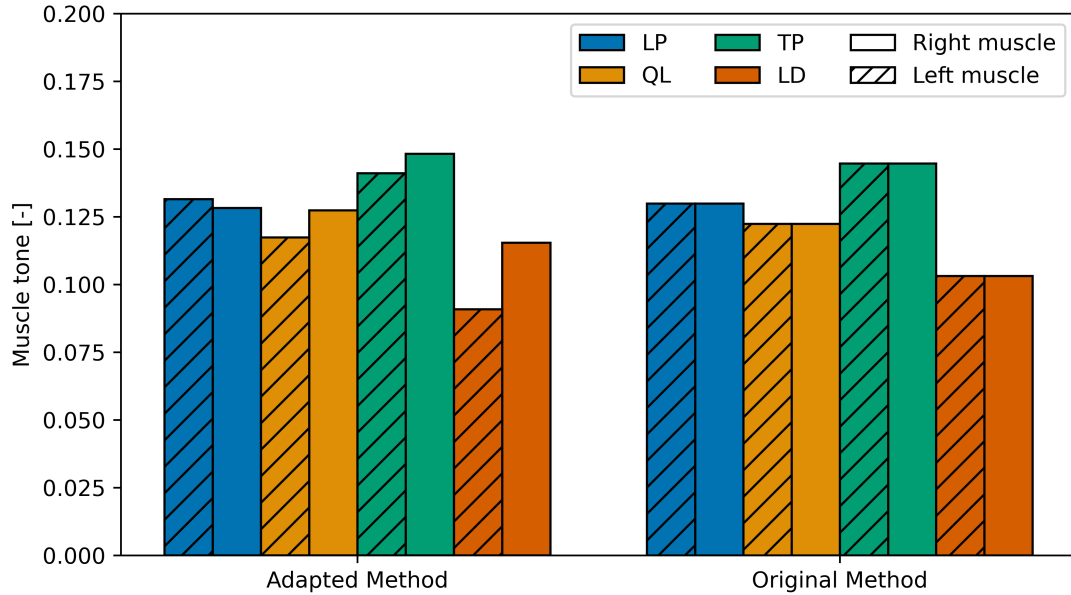


Figure 5.1: Muscle tone for the original and the adapted methods for the healthy subject.

Upon observation of the muscle tones computed for the two methods, no significant variations were seen. Indeed, despite some weak asymmetries shown by the adapted method, the contribution to the force by the different muscles followed a very similar distribution for the two cases. It is interesting to note that the weight given in the case of the original method seemed to be approximately the mean of the right and left weights of one muscle in the adapted method. As these quantities

are used as input into the [MBS](#) model, no significant variations should be present in the results of the modeling of the two scenarios, i.e. in the intervertebral efforts.

5.2 Intervertebral efforts

Figs. [5.2](#) and [5.3](#) illustrate the compressive and shear forces in [N/kg] and the bending torque in [Nm/kg], respectively. As expected, no significant variations in the efforts of the two methods were observed. Upon closer examination, the compressive force, the longitudinal shear force, and the bending torque demonstrated no variation at all across the different [IVD](#) levels. However, the lateral shear stress was evidently null for the original method due to the symmetry assumption, but showed some asymmetry with the adapted method. These results were entirely logical, as the only additional feature of the adapted method compared to the original one was that it took into account asymmetries in the frontal plane, where only the lateral shear force was affected by it. The compressive force, longitudinal force, and bending torque were not affected by asymmetries in the frontal plane.

5.3 Conclusion

The comparison of the two methods yielded the expected results, i.e. similar muscle tones and efforts with only some light asymmetry revealed by the adapted method, consequently leading to some weak lateral shear. Therefore, the results were the expected ones and the adaptation of the new method used in the present study was validated.

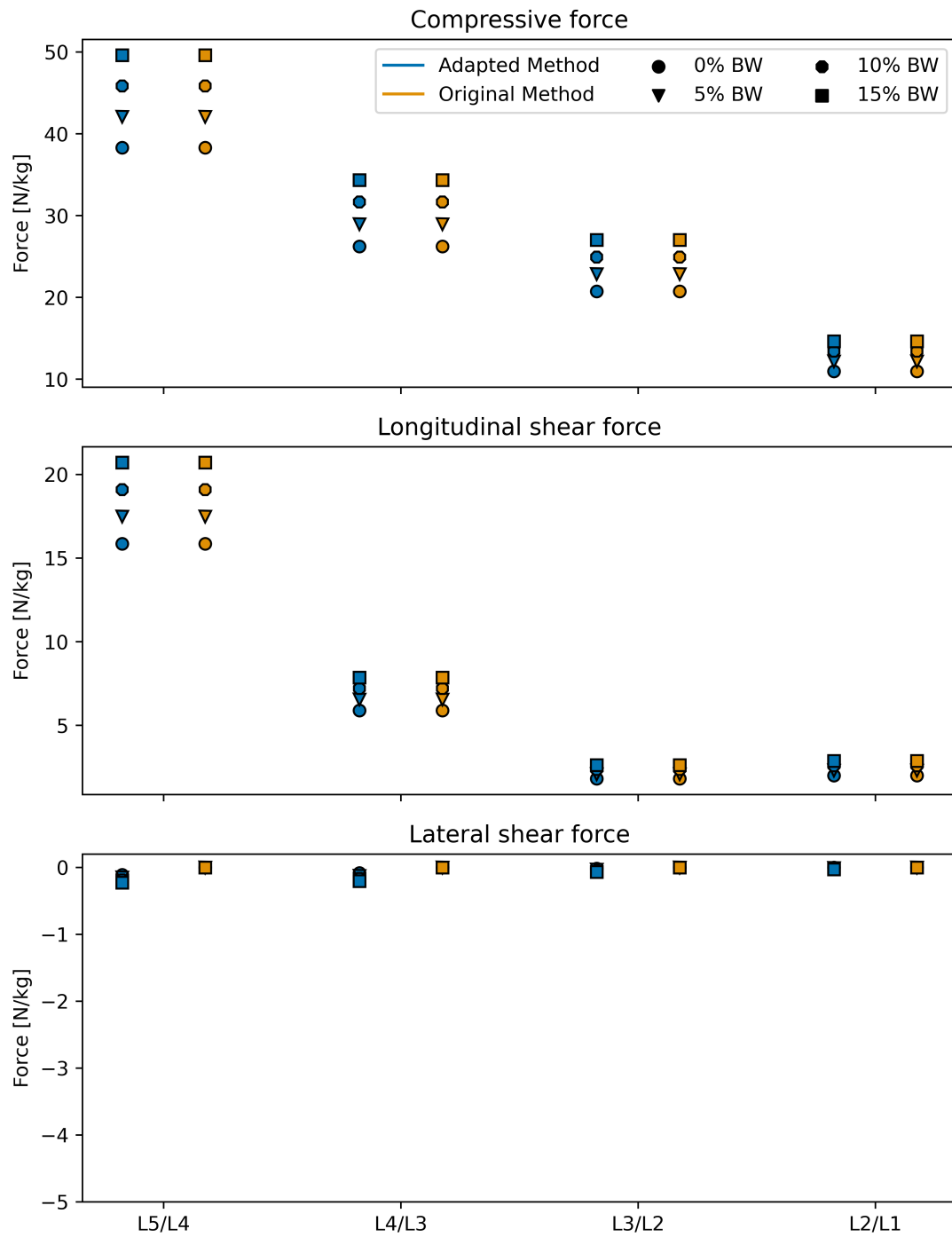


Figure 5.2: Normalized compressive, longitudinal shear and lateral shear forces in [N/kg] in the lumbar IVDs for the healthy subject for the original and the adapted method.

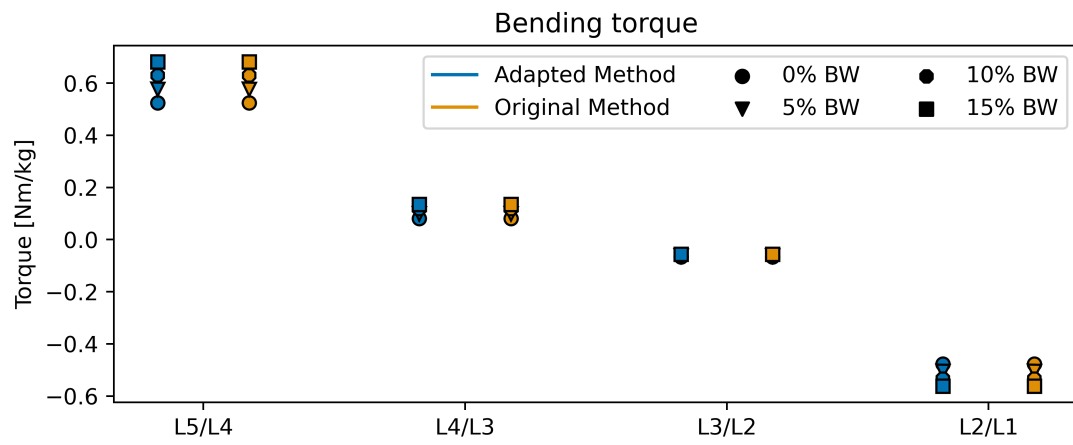


Figure 5.3: Normalized bending torques in [Nm/kg] in the lumbar IVDs for the healthy subject for the original and the adapted method.

Chapter 6

Discussions

6.1 Results

6.1.1 Experimental electromyographic measurements

Maximum voluntary contraction tests

It was somewhat unexpected that the AIS subject was able to generate greater force during the MVC test. This suggested that, despite a geometry altering muscle compositions, the subject with scoliosis was able to produce more force at maximum capacity than the healthy one. However, the difference in force corresponded to approximately 3 kg, which was not a particularly high value. Nevertheless, this paragraph will attempt to understand this result. As the two female adolescents perform the same physical activity on a daily basis, track and field, this should not affect the comparison. Furthermore, both subjects followed the exact same protocol, hence a difference in motivation seemed not realistic. Finally, the scoliotic patient had started physiotherapy only a few weeks before the experiment. Therefore, the effect of physiotherapy should not already impact the results. Consequently, two reasons might explain this result. Firstly, the scoliotic participant was taller. Therefore, her lever arm was longer and this might explain why she was able to produce more force. Secondly, the scoliotic participant, due to altered back muscles, might be forced to use a different strategy, which could explain a difference in the force produced. The hypothesis that seemed to be the most realistic was the second one, meaning that the scoliosis subject used a different muscular strategy. Nevertheless, it is important to note that the literature mainly compared fatigue of subjects, meaning the amount of time a subject could hold a certain force, which has not been studied here. However, literature did not compare the value of the force maintained on a short period of time.

Loaded isometric contraction tests

Firstly, Fig. 3.1a illustrated a lower contraction relative to the maximum capacity in the case of scoliosis. Once more, two hypothesis could explain this. First, the scoliotic subject weighted less than the healthy one. Therefore, at body weight, the force required to hold the torso for the scoliotic participant could be smaller. In addition, it was possible once more that the scoliotic participant used a different muscle strategy, therefore contracting muscles that were not measured during the experiment, such as gluteal muscles.

However, Fig. 3.1b illustrated that for a light increase in the constraint of the exercise, the increase in effort that had to be developed was much higher for the scoliosis subject. This suggested that the difficulty perceived of producing a required effort increased faster when scoliosis was present. This is a typical weakness associated with scoliosis. In addition, this was in line with the answers of the difficulty felt by the subjects on a scale from 1 to 10 (Table 2.1): the scoliosis subject expressed more difficulty to perform the tests.

Secondly, the experiment showed a difference in strategy between the two participants. On the one hand, the healthy subject using less her LD relative to her other muscles seemed logical. Indeed, the LD is a superficial muscle that is more recruited in dynamic movements and not so much in posture maintain. On the other hand, the LD in the case of scoliosis presented a normal level of activation. As this muscle is superficial, it seemed normal that it is not impacted as much as other muscles by the scoliosis. In addition, the curve that worries the most the doctors of the patient is the lumbar curve. Therefore, the fact that the lumbar muscles displayed a lower activity corroborated the concerns of the doctors, as the lumbar muscles seemed to be more impacted by the scoliosis than the thoracic ones and were expected to be weaker. However, it is important to note that the MVC tests in the Sorensen posture were well designed to collect data for lumbar muscles, but less adapted to thoracic muscles. Therefore, this interpretation has to be taken carefully.

Thirdly, the activity of each muscle (Fig. 3.1) highlighted the predicted asymmetry between right and left muscles for the scoliosis case in the lumbar region. As a reminder, the curves of the scoliosis induced a convexity on the left side in the lumbar region and on the right side in the thoracic region. In the literature, with regard to the magnitude of the electrical activity of muscles in AIS, multiple authors, including Fan et al. [23] and Schmid et al. [36], have observed that EMG activity was higher for the muscles located on the convex side of the curve. Therefore, in the lumbar region the left side was supposed to be stronger, and in the thoracic region the right side was supposed to be stronger. Indeed, the muscles

acting in the lumbar region (LP and QL muscles) were significantly more recruited on the left side than on the right side. It also showed that the asymmetry was more important when scoliosis was present.

However, Alves et al. [37] have concluded, although tending towards the same opinion as Fan et al. [23] and Schmid et al. [36] regarding asymmetrical activation, that there was still no real consensus on the matter following a literature review. Therefore, it was difficult to draw a totally valid conclusion. Chwala et al. [38] even stated that the EMG activity could be greater on the concave side in the case of a double curve. In addition, Yeung et al. [16] postulated that the reason of a reduced muscle activity on the concave side was due to the prolonged compression imposed by the concavity, which altered the muscle composition. Furthermore, Fan et al. have also correlated this discrepancy with curve progression. Therefore, the results pointing towards a more active convex side could indicate a progression of the curve, which was confirmed by the physiotherapist of the scoliosis subject.

With regard to the thoracic region, no conclusion could be drawn as the measurements were mainly taken in the lumbar region. However, it is interesting to note that Gaudreault et al. [15] observed that the side difference for AIS was more pronounced at lower levels of the spine regarding EMG parameters comparison.

Regarding the healthy subject, she displayed a gibbosity on the right side in the lumbar region. In this case, the gibbosity probably showed a more developed musculature for the right lumbar side, and therefore a stronger region. Therefore, it seemed accurate that the muscles of this stronger side did not need to be contracted as much compared to the maximum to produce the required force.

When comparing the linear regressions of the two participants, Fig. 3.2 showed that the slopes were always greater in the scoliosis case. This suggested that the scoliosis subject still showed a lower resistance to the effort as soon as the difficulty of the exercise increased.

Finally, the RMS-EMG values at the apex of the major curve was inline with the ones of Fan et al. in the case of curve progression. Indeed, Fan et al. [23] showed a moderate correlation between the paravertebral RMS-EMG ratio and curve progression. They computed this ratio at three levels: the apical vertebra, the upper end vertebra and the lower end vertebra of the major curve. Although the study analysed a position different than the Sorensen test, it still consisted of a static back extension. Their findings showed that curve progression was consistently associated with a high RMS-EMG ratio at the apical level (1.7 ± 0.3), whereas

a non-progression of the curve displayed lower ratios at that level (1.1 ± 0.2). However, it is important to remember that the exercise performed in the study and the one of this study were not completely identical which might lead to some different values. Nevertheless, this closeness in values allowed to validate, on some levels, the measurements.

6.1.2 Quantification of intervertebral efforts and muscle forces

The existing literature on intervertebral efforts, and especially muscle forces, for subjects with scoliosis is considerably less abundant than studies on [EMG](#) results. However, in the following, some comparative analysis could still be made.

Intervertebral efforts

First of all, the results were compared with the ones of Hinnekens. Hinnekens et al. [1] applied their method on healthy subjects and validated their results with the literature. Therefore, a comparison of the healthy subject's result with the ones of Hinnekens allowed to validate the results of this study, especially since the protocol of this study was exactly the same. However, there were some differences between the studies to keep in mind. Firstly, the lateral shear force was not present in the results of Hinnekens as the hypothesis of a symmetry between right and left muscles induced null lateral shear. Secondly, the [LD](#) muscle was not consistently modeled in Hinnekens work. In addition, the nineteen subjects of Hinnekens were all male aged 22.7 ± 2.1 years, weighting 73.9 ± 8.7 kg, and with a height of 1.81 ± 0.05 m. The results of Hinnekens for an 8kg load ($\pm 10\%$ of [BW](#)) at the L5/L4 level were 1.87 MPa for the compressive force, 0.75 MPa for the longitudinal shear force and 2.30 MPa for the bending moment. The values of the efforts of the healthy subject at the L5/L4 segment in Table 3.4 were therefore completely in accordance with the results of Hinnekens [1], validating the values.

On the other hand, the increase in compressive force at the apex were significantly lower in the present case than in the study of Schmid et al. [36]. Schmid et al. [36] provided an analysis of compressive forces in the case of [AIS](#), based on a model of [AIS](#) with and without carrying loads in an upright position. The study found that scoliosis increased compressive forces by 10% at the apex. Furthermore, the study showed that for 10% and 15% [BW](#) loads, the apical compressive forces were further increased by 50-62% and 77-94%, respectively. However, Schmidt et al. provided results for an upright position, where the gravity added to the compressive forces. Therefore, lower compressive forces in the Sorensen posture

were to be expected.

Thirdly, the lateral shear force being the most impacted effort by the scoliosis seemed rational. Indeed, the main characteristic of a scoliosis is the curved profile in the frontal plane, leading to asymmetries between the right and left body parts. As the lateral shear force is the only effort affected by the presence of an asymmetry, it was the most impacted effort. This was also highlighted by the sensitivity analysis. The remainder of the efforts seemed to be impacted in a relatively similar way.

The lateral shear being maximal around the apex appeared as a new piece of information. However, as the analysis were carried on one single scoliosis subject, this trend could not be generalized and no conclusion could be drawn.

Additionally, the profiles of the intervertebral efforts showed that the compressive and longitudinal shear force increased towards the pelvis, and the bending torque was maximum at the most extreme lumbar levels, which was what Hinnekens findings showed as well [1]. Two reasons came to mind to explain this behaviour. Firstly, the mass of the vertebrae increased towards the pelvis as well, inducing more efforts to support them. Secondly, in the Sorensen posture the back had to remain supported by the muscles, with only the ones with fascicles originating from S1/L5 being modeled. Therefore, the closer to this joint, the smaller the lever arm is, and the higher the force of the muscles needed to be to maintain the vertebrae in the desired position. Moreover, the component of forces in play to support the trunk in the plane of lateral shear were not important, which could explain why the lateral shear force did not follow the same trend. In addition, these most extreme lumbar levels (L5/L4 and S1/L5) are problematic discs for surgeons as they often present degeneration like disc slipping [39], but offers the most flexibility to the spine and are therefore desirable to be left in free motion, i.e. not included in the spinal fusion. Therefore, it is reassuring to note that the present results aligned with the statistical observations.

Furthermore, the direction of the efforts was interesting to analyse. It is important to remember that these were the efforts from the lower vertebra onto the upper one. Firstly, the compressive forces tended to push the vertebrae towards the head. However, only compression in the lumbar region was shown. It is likely that the compression forces in the upper thoracic region were in the opposite direction, putting the spine under compression. This could be a mechanism used to increase the capacity of the spine to maintain loads, as it is meant to resist in compression. Secondly, the longitudinal shear force simply showed that every vertebra carried the next against gravity. Thirdly, the lateral shear forces were pointing towards

the concavity. Finally, it seemed like the bending moments simply followed the natural sacral curve and lumbar lordosis.

Finally, an increase in load had more influence on the intervertebral efforts for the scoliotic subject than for the healthy one. This aligned with the observations of the LIC results and again with the answers of the difficulty felt by the subjects on a scale from 1 to 10 (Table 2.1): the scoliosis subject expressed more difficulty to perform the tests.

Muscle forces

Firstly, the fact that the profiles of the fascicle forces were all following the same shapes for the two subjects just illustrated that the same weights, $w_{fa,i}$, for the distribution of the forces at the fascicle level was used for both subjects. Indeed, these weights came from the OpenSim model [26] used by Hinnekens to create the ROBOTRAN [30] model. These weights were based on the maximum isometric force associated with each fascicles on the healthy OpenSim male model, there were a physiological parameters. Therefore, they seemed not adapted to female adolescents, and even less to an AIS subject.

Secondly, the effect of the different muscle groups led to new observations. On the one hand, the non-negligible values of the force emitted by the LD muscle in Table 3.5 probably resulted from fascicles originating outside of the S1/L5 joint that were not modeled in the MBS, but were measured during the experiment. These fascicles seemed to exert a significant influence on maintaining the thoracic part of the spine, as illustrated by the light green fascicle in Fig. 2.9. Therefore, the problematic geometry of the scoliosis may depend more on the impact of the LD.

On the other hand, every muscle force and therefore every fascicle force was higher for the scoliosis subject due to the fact that the global equivalent force F_{equiv} was also higher for the scoliotic subject. This also inevitably impacted the efforts by increasing their values. The more the muscles pulled on the vertebrae, the higher the efforts. This was not surprising. Indeed, the pathology most probably induced a less optimal geometry to support, where the body would have to compensate through new mechanisms such as increased muscle forces and therefore increased intervertebral efforts. In addition, the same results could also be noticed, to a lower intensity, in the healthy subject. Indeed, increasing the load the subject had to carry during the experiment induced more bending and longitudinal shear in the spine, which was consistently accompanied by an increase in the compressive force.

Thirdly, the fascicles producing the most force were the ones at the level of the L4 vertebra, simply because the maximum isometric force associated with these fascicles was the highest one, and therefore the weights attributed to these fascicles were also the highest ones. The same reason applied for the fascicles of the TP muscle inserting at the tenth, eleventh and twelfth ribs. This was therefore just a reflection of the physiological parameter of the fascicles.

Finally, the asymmetry observed in muscle forces and fascicle forces reflected the same pattern as the intervertebral efforts and the muscle tones, but in more detail. Indeed, the strategy of the healthy subject underwent a change between 0 and 15% of BW load. Despite this asymmetry remaining relatively minor, especially when compared to that of the scoliosis subject, it was in accordance with the findings of Chow et al. [40]. The study analysed the gait of AIS and control subjects and identified a difference in strategy at a critical load of 7.5% and 10% of BW, respectively. It is important to note that the study by Chow et al. was based on a different exercise and analysed different parameters. Consequently, a comparison with the present study was a priori unjustified. Nevertheless, it was possible that the change of strategy beyond a critical load remained true for other exercises.

Indeed, the latest statement seemed to be corroborated by the new linear regression of Fig. 3.8. Evidently, a set of tests with more intermediate loads would allow for greater precision in this analysis. Nevertheless, the results of this study indicated that a critical load may exist for the Sorensen posture, which significantly impacted the strategy around 10% of BW for the healthy subject and earlier on for the subject with scoliosis. As Mahaudens et al. [41] observed, this is particularly relevant in the case of a scoliosis student who is required to carry a backpack to school on a daily basis, but this could also be a more general indicator of the recommended maximum load a scoliosis subject could carry. However, this hypothesis needs more in-depth investigation to be confirmed.

New OpenSim model

In general, as the muscle tones were not affected by the new model, and as the same fascicles weights were used for both subjects, the same observations for a comparison between the two subjects still held. The asymmetry remained more marked in the pathological case, and the amplitudes of the forces were also higher for that case. Therefore, no new observations could be made.

6.1.3 Conclusion

Upon observation of the results, many tendencies were highlighted. Firstly, the experiment showed that the scoliotic subject seemed to rely on a different strategy of muscle contraction, and developed more force at maximum capacity. Despite this suggestion, experimental results also demonstrated that the tolerance to the difficulty of the exercise of the scoliotic subject was lower than the healthy one. In addition, **EMG** measurements showed a higher activation of the thoracic paravertebral muscle in the case of the scoliosis, likely due to the altered geometry. Finally, the asymmetrical behaviour in the lumbar region showed more activation on the convex side in the case of scoliosis, and depicted more asymmetry as well when scoliosis is present. Finally, the **RMS-EMG** ratios proposed by Fan et al. indicated, in the present case, a progression of the curve.

Secondly, the intervertebral efforts suggested new pieces of information. Indeed, it was demonstrated that the efforts were overall higher for the **AIS** subject. While the compressive, longitudinal shear and bending stress followed the same profiles for the two subjects, the most evident impact of the scoliotic geometry occurred on the lateral shear by yielding a different profile for the two subjects, with a maximum at the apical vertebra. However, the other efforts showed more impact on the L5/L4 segment, which is a problematic disc as it is associated with pathological disorders to a high frequency in patients. Furthermore, an increase in load seemed to impact every segment of the lumbar spine to the same extent, increased the degree of asymmetry, and had a higher impact on the scoliotic spine.

Finally, the observation of the muscle and fascicle forces showed once more that the **TP** muscle was the greatest contributor for the two subjects. The same profiles of forces over the fascicles were observed for the two subjects, but the magnitudes differed, with the greatest force coming from the fascicles of the **LP** originating from the L4 vertebra. However, the analysis also allowed to suppose the presence of a critical load, beyond which the strategy for both subjects changed significantly. However, this hypothesis needs further analyses to ensure its validity.

6.2 Limitations of the method

The greatest limitation of this study is its subject specificity. Indeed, as the method was applied to only one scoliosis subject and one healthy subject, no general conclusion regarding the impact of the scoliosis can be drawn, but rather subject-specific conclusions. However, this study allowed to adapt and confirm the use of the

method proposed by Hinnekens [1] on scoliosis subjects and remains therefore a relevant study.

6.2.1 Experimental electromyographic measurements

As in every experiment, the quality of the measurements are affected by numerous factors. In this case, the thickness of the subcutaneous tissues could hinder the quality of the measurement by increasing skin impedance, especially since the skinfold thickness is always greater, for scoliotic subjects, on the concave side at the apex [14]. However, thanks to normalization by MVC, the alteration of the measurements due to varying skinfold thickness is minimized.

However, normalization of the signals did not prevent against muscle movement relative to the electrodes during the exercises. Fortunately, in this study, the Sorensen test was a static exercise which reduced the movement of muscles.

Nevertheless, it is important to remember that one of the greatest weakness of surface electrodes is signal interference from surrounding muscles. The electrodes were placed by a physiotherapist expert in scoliosis to minimize this weakness, but it could not be eliminated.

6.2.2 Quantification of intervertebral efforts and muscle forces

Regarding the modeling part, two important limitations are to be noted. Firstly, the model of Hinnekens that was also used in the present study, even though adapted to the maximum possible, was based on the model of a 25 year old male of 175 cm and 78 kg. Therefore, the differences between this model and the reality of female adolescents of 13-14 years old and around 50 kg and 160 cm were numerous and non-negligible.

Secondly, as Chapter 4 demonstrated, the right geometry was a highly important factor in the determination of the intervertebral efforts, and consequently the muscle forces. However, the determination of the geometry was a weakness of this study, and is a path of improvement. Moreover, the Stokes algorithm used to compute the axial vertebral rotation only allowed AVR to be computed for vertebrae L4 through T3, assuming zero AVR for vertebrae L1 through L3, which was not a realistic assumption. Other methods of AVR computation exist, such as

the Perdriolle method which uses a specific measurement tool [42]. This method could provide the AVR of all vertebrae, but the tool was not available during this study, and this measurement technique has a limited precision.

In addition, the weights used to distribute the force among the fascicles were based on a healthy subject and did not take into account physiological alterations. Indeed, scoliosis has proven to affect the composition of muscle fibers [15] and this should be taken into account in those weights, depicting asymmetry between right and left fascicles. However, accounting for this alteration seemed hardly feasible.

Nevertheless, this study still allowed to have some great insights into real phenomenon occurring for both subjects.

Chapter 7

Conclusion

This study successfully quantified, through the hybrid method of Hinneken et al. [1], the intervertebral efforts and the muscle forces of two participants, one of whom was affected by scoliosis. The participants had similar anthropometric data for comparison. The quantities were obtained for a static configuration, the Sorensen posture, which was performed at body weight and with 5, 10, and 15% of BW added load.

Although limitations of the study existed, such as its specific scope, the results obtained enabled the identification of discrepancies between the two participants in terms of perceived difficulty, muscle strategies, asymmetry between right and left muscles, magnitudes of the efforts and of the muscle forces, as well as the impact of the load. It was observed that the scoliosis had the most pronounced effect on the lateral shear. The results indicated that the maximum lateral shear in the scoliosis case appeared in the apical region. Moreover, the results appeared to indicate a progression of the lumbar curve. Finally, the results of this study raised the question of whether there was a critical load for the Sorensen test, after which the muscle strategy of the subject changed. Nevertheless, further in-depth analysis is necessary to confirm this hypothesis.

In conclusion, this study presented a case study of a single scoliosis subject, which allowed for the quantification of the segments of the spine that experienced the greatest mechanical stress. This work also demonstrated that the use of this method required a model specific to each scoliosis subject, a process that may be time-consuming but remains an essential step in developing a quantitative tool.

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