

Abstract

Spine surgery planning involves many decisions for which the surgeon has not enough information, and for which biomechanical models might be helpful. Research studies show a high variability in decision making in the planning of scoliosis surgery by experienced groups of surgeons. That variability is problematic because it may cause complications for the patient, such as revision surgery or an increased limitation in spine mobility. Therefore, a biomechanical model for spine surgery planning may be useful in providing the surgeon the information needed to propose the best treatment. In this context, intervertebral efforts represent an essential input in assisting diagnosis and subsequently guiding the surgical planning of scoliosis.

The long-term motivation of this thesis consists in assisting surgeons in obtaining quantitative - kinematic and dynamical - information that will allow them to improve the surgical planning of scoliosis, by specifying the set of vertebrae to be fused, while preserving the spinal mobility. In regards to this final motivation, the thesis seeks to develop a clinical protocol based on experimental data and on a multibody model of the upper body, to quantify the intervertebral efforts for idiopathic scoliotic adolescents in standing up position (statics) and during moderate gait (dynamics). The estimation of intervertebral efforts is based upon four interwoven topics: patient physiology, spine geometry, spine and pelvis kinematics, as well as muscular forces. In line with this, if this work's final objective is to be met, three targeted contributions must be achieved:

- The elaboration of a clinical protocol focusing on assessment of the scoliotic patient's parameters: necessary anthropomorphic data, spine shape and kinematics, and muscle force calibration;
- The development of a physiologically-based multibody model of the upper body, able to predict the spine's kinematics and dynamics during gait;
- In terms of internal efforts, using the multibody model and experimental input conjointly should allow exploration and discussion of plausible solutions, thanks to the high potentials offered by both models and computer simulations.

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Introduction

1

Computer-assisted surgery (CAS) refers to the use of computer technology for surgical planning or assistance during surgical interventions. CAS offers several advantages such as a more accurate preoperative diagnosis and a well-defined and optimized surgical planning. Indeed, those technologies allow surgeons to rely on quantitative data of the patient, in addition to their experience and intuition, reducing the risk of surgical morbidity or a lifelong side effect.

In the particular case of the scoliosis surgery, the planning involves many decisions for which the surgeon does not have enough information, and for which CAS might be helpful. This has been illustrated by a study carried at the École Polytechnique de Montréal [79], which shows a high variability in decision-making in the planning of scoliosis surgery for an experienced group of surgeons. This variability is problematic because it could cause complications for the patient such as a revision surgery or an excessive limitation of spine mobility. Therefore, CAS for spine surgery planning might be useful in giving the surgeon sufficient information to propose the best treatment. This thesis focuses on the intervertebral efforts quantification in scoliotic spines, during walking, in perspective of spine arthrodesis. These efforts represent an essential input to quantify the severity¹ of the deformation, to help in the diagnosis and subsequently to guide surgical planning of scoliosis.

1.1 Spine anatomy

The spine is composed of several bones and muscles. The main bones of the spine consist of the 33 vertebrae: seven cervical, twelve thoracic, five lumbar, five sacral and four coccygeal. The size and the shape of each vertebra vary according to the region

¹In the context of the present study, it is assumed that the severity of the scoliosis is correlated with the magnitude of the intervertebral efforts.

of the upper body [24]. The muscles of the spine could be classified in two groups: extrinsic, for the external muscles involved in the movements of the upper limb and thoracic wall, and intrinsic, for the internal muscles associated to the support and movement of the vertebral column.

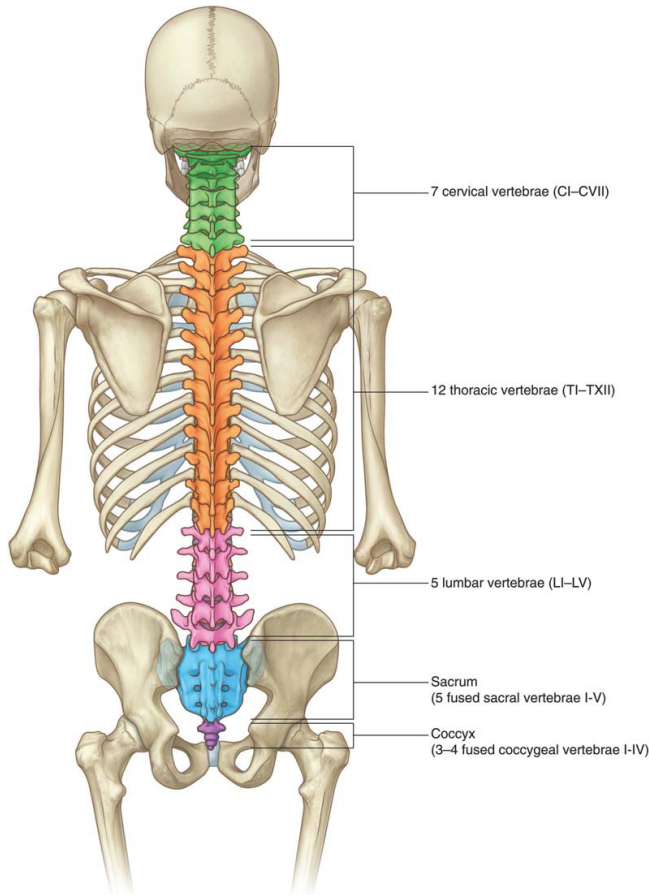


Fig. 1.1: Spine structure and groups of vertebrae [24]

The spine ensures three main functions:

- It supports the upper body, jointly with the muscles of the back;
- It allows mobility of the upper body: forward (flexion) and backward (extension), lateral (bending) and torsion (rotation). This mobility is possible thanks to the relative motions between vertebrae — mostly in the lumbar region — due to the deformability of the intervertebral discs;
- It protects the spinal cord.

1.2 Scoliosis

The term scoliosis was first established by Galen (130-201 AD) and comes from the Greek word *skolios* which means “curved” [48]. Nowadays, scoliosis is defined as “*an appreciable lateral deviation in the normally straight vertical line of the spine*” [94]. According to [94], there are several known causes of scoliosis:

- Alterations of the osseous structure like rickets, tumors or bone fractures;
- Alterations of the ligaments structure such the Marfan’s disease, mucopolysaccharidosis or surgery ;
- Neuromuscular and postural problems such as polio, cerebral palsy or muscular dystrophy;
- Failure on the formation or the segmentation of the vertebrae (congenital scoliosis).

Nevertheless, the origin of most of the cases of scoliosis — about 85 to 90% — is still unknown; it is then referred to *idiopathic* scoliosis [85, 94].

1.2.1 Adolescent idiopathic scoliosis (AIS)

According to the Spine Research Society, idiopathic scoliosis of adolescents (a.k.a. AIS) is defined as a structural lateral curvature of the spine that creates a thoracic and/or a lumbar asymmetry on forward bending, (a.k.a. the Adams forward bend test¹ combined with an angular deformation of at least 10 degrees, as measured by the Cobb² technique on a standing radiography of the spine (Fig. 1.2). It is worth noting that AIS always implies a torsion in the vertical direction, denoted *vertebral rotation* in the medical community. Scoliosis is thus an actual three-dimensional problem. This thesis focuses on the case of AIS.

Possible causes of AIS As its name suggests, the reason for this kind of scoliosis is still unknown. However, as AIS often affects several persons of the same family [91], some authors suggest that there is an important genetic component. For example, [45] shows a notable correspondence of 73% in severity and curve patterns in identical twins.

¹The Adams Bending test is an examination method to measure scoliosis. It consists in asking the patient to bend forward in order to measure his trunk deformation via an scoliometer [69].

²According to [69], the Cobb angle (in degrees) is *the angle between lines drawn along the upper end plate of the most tilted vertebrae above the curve’s apex and the lower end plate of the most tilted vertebrae below the apex*.

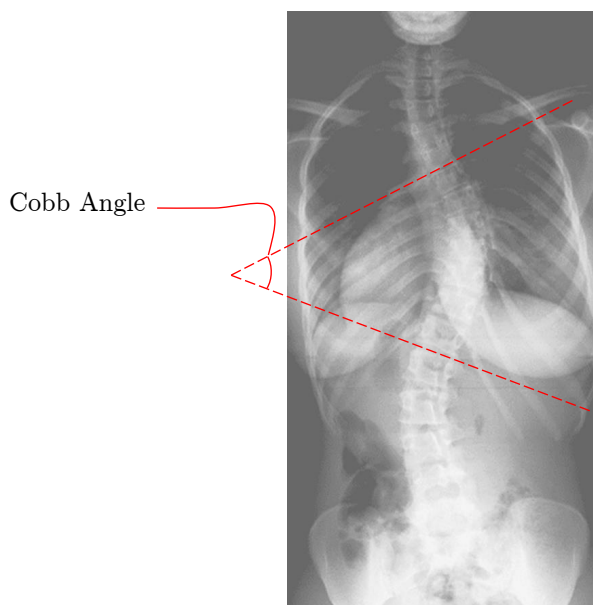


Fig. 1.2: The Cobb angle allows quantifying the severity of a scoliosis (Or.: Cliniques universitaires Saint-Luc, UCL).

Prevalence of AIS According to [48, 96], AIS is a common disease with a prevalence estimated to 0.47-5.2% of the population. AIS affects more women than men: the female to male ratio of incidence of scoliosis ranges from 1.5:1 to 3:1. This ratio increases in patients with higher Cobb angle: it goes from 1.4:1 in curves between 10° and 20° to 7.2:1 in curves between 20° and 40° .

1.2.2 Treatment

The treatment of scoliosis ranges from medical surveillance to surgery. It will mostly depend on the type of scoliosis as well as on the location, severity of the deformation and the growth potential of the patient. As explained at the beginning of this chapter, this treatment varies significantly from one surgeon to another (Sect. 1.2.2). In most of the cases, scoliosis does not require surgery and can be treated with non-surgical treatments such as a medical follow up, a physical therapy or the placing of a brace (a.k.a. bracing) [45, 67].

Medical follow up In the case of mild scoliosis, there is no treatment required [4, 91]. Nevertheless, in most of the cases, the patient will be supervised every six months to analyze the evolution of the spinal deformation (Fig. 1.3 left). This follow-up relies essentially on the Cobb angle measurement and on the Adam's bending test.



Fig. 1.3: Non-surgical treatment of scoliosis: the Adam's bending test (left) and Bracing (right) (or.: Cliniques universitaires Saint-Luc, UCL)

Physical therapy In addition to the medical follow up, some patients will be asked to follow a program (series of exercises) and/or a physical therapy. According to [7] the use of exercises to treat scoliotic patients is controversial.

There are plenty of programs recommended to the patients, namely the Scientific Exercises Approach to Scoliosis (SEAS), the physiotherapeutic scoliosis-specific exercises (PSSE), Schroth, Dobomed, Barcelona Scoliosis Physical Therapy School (BSPTS), Side Shift, Functional Individual Therapy of Scoliosis (FITS) and Lyon. Those programs aim at stabilizing the spine, slowing down the progression of scoliosis and limiting the use of braces or the resort to surgery [7]. The most used programs are the SEAS and PSSE programs. The SEAS one follows an individualized exercise program adapted to each scoliotic patient, and is based upon a self-correction technique without external aid. This method aims at training the neuromotor function to stimulate — by reflex — a self-corrected posture during daily activities [81]. The PSSE program consists in a series of exercises that are individually adapted to a patient's deformation: curve site, magnitude and clinical characteristics [7].

Bracing In the case of a moderate curve — i.e. a Cobb angle between 40° and 50° [90] — some physicians recommend the use of bracing (Fig. 1.3 right), in order to reduce the progression of the scoliosis in the case of a growing child.

Nowadays bracing is widely used for patients with a moderate deformation of the spine; however there is still a disagreement on its effectiveness [91]. On one hand, some studies claim a significant reduction of the progression of AIS using a brace [4,59,92]. On the other hand, others show that bracing has a negative impact on the function and activity level of the patient [11].

Surgical treatment: Spinal fusion In the case of spine curves higher than 45° or if its progression could not be stopped without a non-invasive treatment, surgery becomes necessary [4]. Spinal fusion (a.k.a. arthrodesis) is the standard procedure to treat spinal deformities, although over the past years, various technologies for AIS surgery that are not based on fusion (such as growing rods, the vertebral prosthetic expandable rib, vertebral staples, etc.) have been developed [5,78].

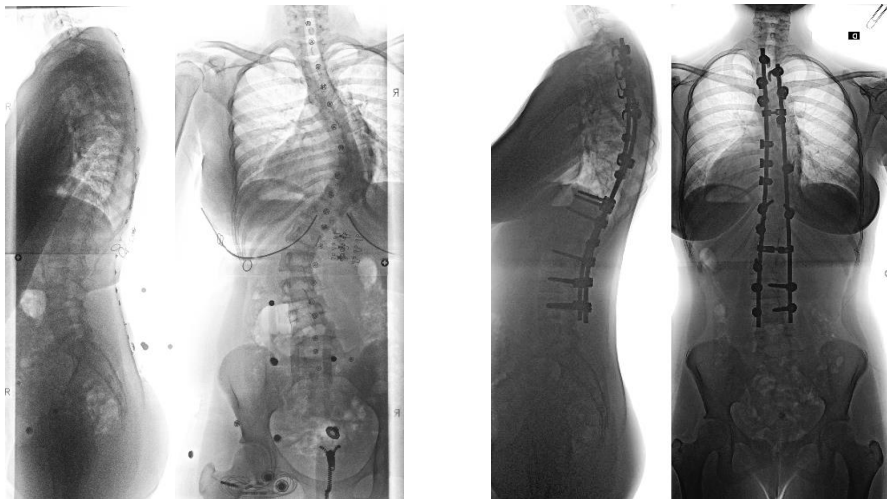


Fig. 1.4: Surgical treatment of the scoliosis; X-rays of a patient before and after the spinal fusion (or.: Cliniques universitaires Saint-Luc, UCL)

Spinal surgery consist in realign the scoliotic spine using metallic rods (Fig. 1.4 right) and then to induce fusion between vertebrae using a bone graft. Metallic rods are connected to the spine using hooks or screws, and will hold the spine until fusion takes place.

The use of spinal instrumentation to correct spinal deformities was first experimented by Hadra in 1891 [36]. It consisted in using wires to correct the spine deformities in the spine of a patient with the Pott's disease. After that, a method to fuse short segments of the vertebral column by using bone grafts from the tibia was independently developed by Fred Albee [1] and Russell Hibbs [39] in the year 1911 [86].

Since its beginnings, several advances in fusion techniques have emerged thanks to the development of various instrumentation devices or the use of new graft materials for vertebral fusion. From all of them, two past developments have largely contributed to the spinal surgery that we know today:

1. the use of rods, developed by the Dr. Harrington [37], to allow segmental correction of the spinal deformity and maintain fixed the spine to facilitate the fusion;
2. the use of screws, proposed by Tourney in 1943 [88] and then King in 1944 [46], to achieve greater stability and better fixation on the vertebrae [86].

Variability and risks of spinal fusion In general, spinal fusion can bring substantial clinical benefits for the quality of life of patients [35]. However, nowadays there are still controversies regarding the needs for such treatment [93].

As pointed out by past studies, it exists an important variability in the treatments for scoliosis proposed by different specialized surgeons. For example in [41], it is already stated that 5 out of 30 surgeons were against surgery to treat a degenerative scoliosis. [79] showed that among 160 planing cases (i.e. 5 different cases proposed to 32 surgeons), 156 different instrumentation strategies were proposed.

The variability in surgical planning can affect the quality of the surgery and cause livelong complications for the patients such as a reduced mobility [58] or a re operation. The need for a re-operation after spinal surgery was studied by [34], whose authors showed that about 37% of the patients who have undergone a lumbar spinal fusion develop some alterations of the adjacent motion segment. Along the same lines, [10] showed that several patients develop adjacent instability such as the breakdown of a neighboring unfixed motion segment.

Surgical planning As explained before (Sect. 1.2.2), there exists an important variability in the treatments of scoliosis proposed by different specialized surgeons. In this context, there are several works suggesting how to solve this problem. In this section, we briefly describe the decision criteria of surgical planning, as well as analyze how our work can contribute to this domain in the future.

According to [94], the King-method was the first to offer well-presented guidelines for determining fusion levels in patients with thoracic scoliosis [46]. This method was based on the study of 405 patients for whom the authors established five types of curves and a recommended treatment for each of them. The King-method however had a poor intra and inter-observer reliability, and for that reason [52] developed a more accurate technique, aiming at classifying scoliosis and standardizing its surgical planning. The natural history of different curve patterns was then analyzed by We-

instein [90]. In his work, he studied the progression of the spinal deformation after skeletal maturity for different kinds and levels of scoliosis.

From a biomechanical point of view, other criteria used during the surgical planning of scoliosis are those presented by Jean Dubousset [25]. In his work, he studied several issues connected to the surgical planning such as: the cone of economy, the intervertebral torsion and the importance of the trunk balance after surgery. However, his work is only based on a static equilibrium criteria. In this context, the main contribution of the present thesis consists in adding to the analysis the impact of the spinal deformation on the intervertebral forces, not just from a static point of view as carried by [25] but also using a dynamical approach where the effect of scoliosis is quantified during walking.

1.3 Biomechanics of the spine

According to [94], clinical biomechanics is defined as *the body of knowledge that employs mechanical facts, concepts, principles, terms, methodologies, and mathematics to interpret and analyze normal and abnormal human anatomy and psychology*. We are convinced that biomechanics can bring significant benefits for the surgical planning of scoliosis. Indeed, in addition to the knowledge of the biomechanical data of patients, kinematic and dynamical inputs (such as the spine deformation during gait or the intervertebral efforts under dynamical loads) should assist surgeons to rely on new quantified results that are complementary — as we believe — to their own experience and intuition.

Several works have already analyzed the intervertebral efforts in the spine. Despite serious ethical issues, some authors measured — on humans — the intradiscal pressure using a pressure sensor located at the L4-L5 disc [95]. In this respect, the present research has a more deontological incentive: the quantification of the intervertebral efforts in a non-invasive way. For that purpose, there exist various approaches based on a multibody model of the human body and a companion software dedicated to biomechanics such as LifeMOD, OpenSim, AnyBody, Apollo [61] and ROBO-TRAN [23] in our case. For instance, [12] develops a musculoskeletal model of the lumbar spine [72] and uses bending to calculate the mechanical properties of scoliotic spines. In [21], they use a biomechanical model to predict — preoperatively — the geometrical correction of an anterior spinal surgery of a scoliotic patient.

Intervertebral efforts strongly depend on four pillars: spine geometry, spine kinematics, patient anthropometry and muscular forces. The next sections, in addition to some technological aspects, present the state of the art developed in those areas.

1.3.1 Introduction to biomechanics

Biomechanics consists in applying the formalisms and methods of mechanics to analyze the structure or the function of a biological system [38]. Biomechanical studies can focus at different levels of biological systems, from the inner work of a cell to the musculoskeletal system [55], and can be applied to a wide range of applications such as surgery, physiotherapy, ergonomics, etc. [49].

Several mechanical and computational approaches can be envisaged in biomechanics according to the applications and to the pursued objectives. For instance, the study of the mechanical properties of living tissues can resort to the Finite Element Method (FEM); predicting the blood flow within the arteries will exploit fluid mechanics theory and numerical algorithms [49]; analyzing the intersegment forces and the body accelerations of a human body in motion is more an issue for the multibody approach (MBS).

1.3.2 Biomechanical analysis using the multibody approach

The characterization of the human motion is a major importance in medicine, sports, physical therapy as well as for engineering applications such as vehicle [2,3]. In this context, there are plenty of works based on multibody dynamics for the development of biomechanical models to study different pathologies. Without seeking to be exhaustive, here are a few representative examples related to well specific domains:

Sport: [15] used a multibody model for the analysis of the muscular and joint forces of a gymnast during ballistic motion. This sophisticated model includes 43 muscle groups for each leg, for which the force are estimated using a Hill type muscle model.

Car Crash: [30] developed a multibody model of the cervical spine to study the whiplash injury during car crashes. This work shows that a multibody model can reproduce the motion characteristics of the head and neck during a rear-end impact.

Shoulder Pathology: [76] developed the multibody dynamics approach to study shoulder pathologies. This model includes the sternoclavicular, acromioclavicular and glenohumeral joints, modeled using spherical joints, and 21 muscles described by 37 individual segments using the obstacle-set method. The activation of the muscles is estimated using a Hill type muscle model.

Gait and ankle orthosis: [31] used a 2D simplified multibody model, consisting on four rigid links for each leg (thigh, shank, foot and toes), a rigid body that represent the upper limb and a foot-floor contact model based on three sphere-plane interac-

tions. This study shows that this model — although simple — is able to reproduce the ankle kinematics measured from the gait lab, with and without orthosis.

1.3.3 Spine geometry

Geometrical identification of the spine The *geometrical* identification consists in obtaining the 3D information of the spine deformities (i.e. scoliosis) using one of the technologies described below. Among the works in this field, the following contributions have drawn our attention:

- [8] achieved the reconstruction of the spine of standing up patients using skin markers and splines interpolants, but without taking the intervertebral torsion along the spine into account, which represents a non negligible limitation in our case;
- [75] computed the intervertebral angles using the SpinalMouse for flexion and extension positions. Those data were mainly used to estimate the range of motion (ROM) of 111 subjects, being thus limited to a geometrical analysis;
- In the present work, the spinal geometry is measured through the identification of anatomical landmarks (disc center and spinous process) from bi-planar X-rays. Stereoradiography, which consists in the use of several X-rays to identify the tridimensional shape of the spine, has been already studied by other authors. For example, the works in [20], [73] and [40] exploit this principle to identify several anatomic landmarks of the spine, then extrapolate those data to reconstruct the complete shape of the spine. In case of [73], they proposed a semi-automated method to identify the shape of the spine, based on a little information from radiographies as well as from a morphometric database. The calibration process during stereography is an important step to obtain a precise 3D reconstruction. This process has been studied by some authors; for example: the work of [42] describes how this calibration process could be used to correlate bi-planar X-rays as well as to estimate the X-ray magnification along the spine. Then, [26] describes several calibration methods used for stereophotogrammetry¹ and develops a calibration method for stereoradiography based on bi-planar X-rays with frontal and lateral incidences. The method described in this work exploits the principle of stereoradiography to identify the position of the intervertebral discs and of the spinous processes, in a similar way as in [44] and [26].

¹Stereophotogrammetry consist in estimating the three-dimensional coordinates of a point using two or more photographic images.

1.3.4 Spine kinematics

Intervertebral range of motion There are several studies dealing with the intervertebral mobility of the spine. Some authors analyzed the intervertebral motion of cadaveric lumbar spines *in vitro* [97]. However, the mobility of the spine has also been studied *in vivo* using different methods such as magnetic resonance [32], X-rays [29] and finally, using three reflective markers positioned on each lumbar vertebrae, via Kirsher wires, to measure the spinal motion during gait [83]. In particular, the results of [83] will be exploited in this research for the computation of the intervertebral deformation during gait.

Data acquisition during gait According to [63], the first quantitative studies about human locomotion begun in the early XIX century. Then, in the mid of the XIX century, Marey [57] and Muybridge [64] were the first in quantifying the patterns of human movement using photography. Nowadays, there exists different techniques for motion capture such as stereoradiography, optoelectronic motion analysis, inertial measurement units (IMU) [80], etc.

Independently from the acquisition method, gait motion data are usually noisy. The treatment of the latter has been studied in [16, 17], whose authors have analyzed different kinds of filtering methods in biomechanical analysis. The one used in the present research consists in a Zero-phase¹ digital filtering as described in Chapter 2 [66].

In this thesis, data acquisition is achieved using optoelectronic motion analysis², which consists of 8 cameras and a set of reflective markers stuck on the skin of the patient.

1.3.5 Mechanical and anthropometric data

Mechanical and anthropometric parameters of the human body are essential inputs for a biomechanical analysis. While geometrical data can be measured or estimated easily (see Sect. 1.3.3), the internal mechanical properties, as well the dynamical data of the patient remain difficult to identify. In the present section, we briefly address two important parameters for our dynamical study: the disc stiffness and the inertia of body segments. Data used in this study are detailed in Appendix B.

Inertia properties of the body segments The inertia properties of human body segments (mass, center of mass, inertia matrix) can be estimated using different tech-

¹This filter process data in both forward and reverse direction in order to avoid phase shift

²BTS GaitLab technology

niques. For instance, some of them directly work with cadavers, as in [9] who measured the mass distribution of the human body using six adult male corpses. Measurement of the inertia properties can also be analyzed *in vivo* using a volumetric estimation as proposed in [70]: using Magnetic Resonance Imaging (MRI) or a gamma-ray scanning technique as in [98], it is possible to estimate the inertia properties of body segments. MRI as well as gamma-ray showed similar results except for the estimation of the mass properties of the patient (the thorax in particular).

In this research for which dynamics is involved (moderate gait), masses have been estimated using the anthropomorphic data from a database referring to the study of Zatsiorsky [19,50], with the exception of the abdominal cavity, in so far as the masses of its constitutive organs are not directly connected to the lumbar spine (Sect. 3.6.1).

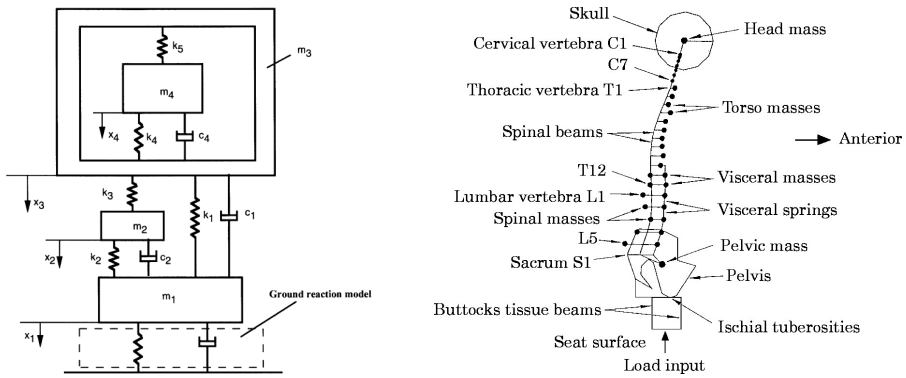


Fig. 1.5: Biomechanical models proposed to simulate the organs inside the abdominal cavity. Left: an equivalent spring-dashpot-mass based model proposed by [53]. Right: a more morphological model proposed by [47]

In the case of the abdomen, there exist various methods to take the mass of its internal organs into account and to model their connections with other elements (diaphragm, vertebrae, etc.). Some approaches assimilate the abdominal cavity to a stack of slices rigidly connected to the lumbar vertebrae [27,28]. Doing so, those formulations neglect the internal motion of the visceral organs. To take these motions into account, some models have been proposed. [53,65] simulated the internal organs motion via a simplified spring-damper model to analyze the repercussion of the impact force during running. [47] developed an equivalent model (i.e. not based on the patient physiology) that reproduces the resonance modes of the visceral organs. Finally, [14,71] used a model based on [47], but their goal is quite different since they specifically analyze vehicle comfort (Fig. 1.5).

A more detailed analysis of the impact of the wobbling masses on the intervertebral

effort is carried out in Chapter 3.

Anthropomorphical data of children Most of the information cited above about the mechanical and anthropometric properties of the human body are mainly based on studies carried out for adults. Two different studies [33, 43] analyzed the difference of the anthropomorphic proportions of the mass, center of mass location and moment of inertia, between those extrapolated from adults and those measured directly on boys aged between 7 and 13. Both of them found that the difference between the measured and the extrapolated data were not significant. This study was corroborated by [6] who repeated a similar analysis on girls aged between 8 and 10.

Disc properties According to [24, 94], the intervertebral discs make up to 33% of the height of the vertebral column. They are formed from three different parts: the nucleus pulposus, the annulus fibrosus and the cartilaginous end-plates, each of them with different mechanical properties: the nucleus pulposus fills the center of the intervertebral disc, is gelatinous and absorbs compression forces between vertebrae, and the annulus fibrosus consists in an outer ring of collagen, and aims at limiting the rotation between vertebrae.

Several studies focus on the estimation of the stiffness and damping properties of the intervertebral discs. For example in [42], they performed static and dynamical compression tests to evaluate the stiffness and damping of eight intervertebral discs. Nevertheless, according to [94], care must be taken for the estimation of the intervertebral disc properties because they can significantly vary depending on which technique is used for the test. Furthermore, disc properties also vary with the age, sex, disc disorders, making virtually impossible to know the disc characteristics with precision for every patient. Fortunately, for the moderate gait envisaged in this work, the impact of this parameter uncertainty is very limited as we will show later on (Chapter 3).

The mechanical properties of the intervertebral disc play an important role in the calculation of the inner stress of the intervertebral disc (Chapter 5). However, the analysis proposed in Chapter 3, based on the intervertebral disc stiffness obtained from [94], shows a rather low sensibility of the intervertebral efforts to the mechanical properties disc for the case of moderate gait.

1.4 Objectives and contribution

As explained at the beginning of this chapter, the planning of the spinal surgery involves many decisions for which the surgeon has not enough information. In this context, the identification of the intervertebral efforts of scoliotic patients should

help him to identify the severity of scoliosis and to refine his diagnosis accordingly.

Facing such a biomechanical issue, engineers, while remaining cautious or even modest in their approach and results, can at least suggest "Cartesian" processes and develop engineering tools whose goals are surely not to tell the truth, but to allow exploring avenues and possible solutions, to submit to (and discuss with) the medical community.

It is with this in mind that the present research work has been thought and achieved.

1.4.1 Research question

The long-term motivation of the present research aims at providing the surgeons with quantitative kinematic and dynamical information that help them to select the best-suited surgery for each patient. In other words, the final goal is:

- to find the surgery that minimizes the intervertebral effort in the adjacent vertebral discs —during daily base activities— while preserving a maximum mobility i.e. by limiting the number of vertebrae to fuse.
- to add more information for surgical planning.
- to identify most adapted surgery for the patient i.e. the anterior or the posterior approach. As mentioned before, until now, most criteria for surgical planning are based on a static analysis of the patient. These criteria are not sufficient since they do not take the spinal motion into account, and can lead to a non-optimal planning of the surgery, considering the daily human body activities.

1.4.2 Final objective

With regards to the long-term motivation of this research i.e. an optimal surgical planning, the final objective of this thesis consists in developing a clinical protocol based on experimental data and on a multibody model of the upper body, to quantify the intervertebral efforts for idiopathic scoliotic adolescents in standing up position (statics) and during moderate gait (dynamics).

1.4.3 Targeted contributions

The main contribution of our work is to add kinematic and dynamical information as input for the surgical planning. To that purpose, a complete study of the spine

is carried out, ranging from X-rays measurement to gait dynamics analysis . The estimation of intervertebral efforts is based upon four interwoven topics: patient physiology, spine geometry, spine and pelvis kinematics and muscular forces. In line with this, to reach the final objective of this work, three targeted contributions have been achieved:

- The elaboration of a clinical protocol focusing on the assessment of the scoliotic patient parameters: necessary anthropomorphic data, spine shape, spine kinematics and muscle force calibration (Chapter 2);
- The development of a physiologically-based multibody model of the upper body, able to predict the kinematics and the dynamics of the spine during gait (Chapter 3);
- The joint exploitation of the multibody model and of the experimental input, to allow exploring and discussing plausible solutions, in terms of internal efforts, thanks to the high potential offered by models and computer simulations (Chapter 4).

This chapter explains the methodology used to reach our final objective, i.e. the computation of the intervertebral efforts in scoliotic patients during gait. It is divided in three sections: The first summarizes the general protocol, the second explains the experimental protocol, and finally the last section focuses on the modeling tools. [13]

2.1 General protocol

The ultimate objective of the research to which this thesis contributes is the planning of the arthrodesis of scoliotic patients. For this purpose, we aim at developing a methodology to quantify the severity of the scoliosis for a particular patient (Fig. 2.1).

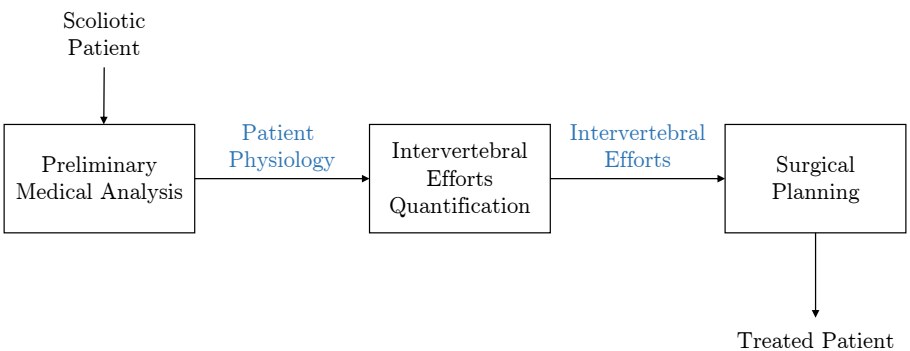


Fig. 2.1: Patient's medical path

As mentioned in Chapter 1, the intervertebral efforts quantification in scoliotic spines during motion, ideally measured before and after spine arthrodesis, seems to be a good indicator of the severity of the deformation and might be subsequently

helpful to improve the surgical planning. An increase of 30% of the energetic cost of adolescent idiopathic scoliosis patients [56], as well as its consequences in ordinary living, suggest that gait is a relevant situation to be considered as a reference motion for the present study.

The accurate computation of the intervertebral efforts strongly depends on four pillars: patient physiology, geometrical identification, spine and pelvis kinematics and muscular forces of the upper body¹. The proposed protocol addresses each of these problems, from both the clinical and the modeling points of view, as depicted in Fig. 2.2, which details the central block of Fig. 2.1.

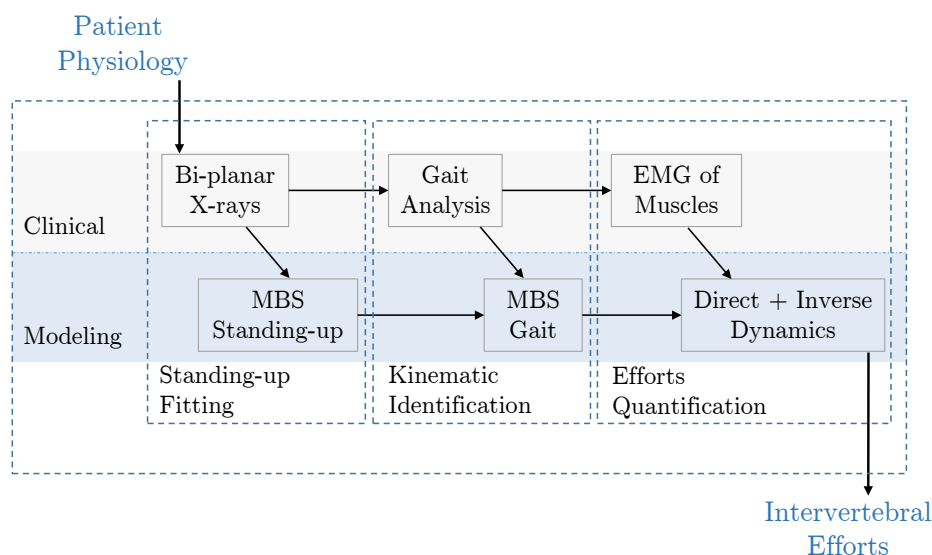


Fig. 2.2: The general protocol is based on both clinical and modeling processes.

2.2 Clinical assessment

As a first step, a complete clinical analysis is achieved to identify the health condition of the patient. This analysis is carried out by an orthopedic surgeon and a physiotherapist and comprises the medical history, the anthropomorphic data collection, a postural analysis, a neuromuscular evaluation and a cardiorespiratory test of the patient.

Some information collected during this clinical assessment is not directly useful for the analysis; however, it allows excluding from the study patients with muscular

¹As explained later on in Sect.3.3, the lower body can be completely ignored in the analysis.

and/or spinal abnormalities — different from scoliosis — such as the presence of a sixth lumbar vertebra for instance.

Finally, the surgeon will manually identify, using metallic markers, the iliac crests and the spinous processes from c7 to sacrum. These metallic markers are visible with the X-rays, and will serve as geometrical references for the next steps of the project (Sect. 2.4).

2.3 Geometrical identification

This section focuses on the geometrical identification of the spine. It is worth noting that the upcoming multibody model will only deal with rigid bodies. First, considering that the intervertebral disc mass can be neglected, a valuable hypothesis consists in lumping the disc elasticity to a single point, located at the disc center and assimilated to the two vertebrae connection (Fig. 2.3).

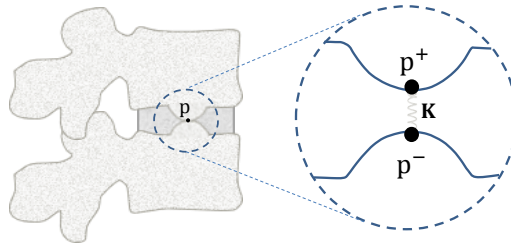


Fig. 2.3: The intervertebral disc lumped in a single massless point p . $\mathbf{K}$ represents the stiffness matrix of the disc

Doing so, each vertebra n is characterized by three non co-linear points (Fig. 2.4): the disc center between vertebrae n and $n - 1$, the disc center between vertebrae $n + 1$ and n , and the spinous process of vertebra n . These points will allow to unambiguously locate the vertebrae in the three-dimensional space.

Given the importance of the geometrical identification for the following steps of the protocol, three issues will be addressed in this section:

- a) A comparative analysis of the image acquisition;
- b) The estimation of the geometrical identification accuracy requirement via a sensitivity analysis;

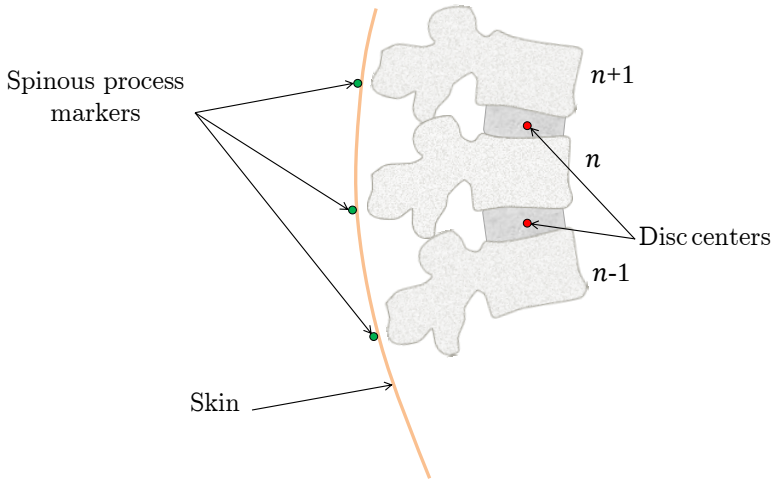


Fig. 2.4: Characteristic points of vertebrae - spinous process and intervertebral disc centers

- c) The development of an experimental protocol and design of a user interface, that allows to identify the spine geometry of any patient.

a) Image acquisition : medical imaging technologies Over the years, several technologies have focused on the creation of a visual representation of the interior of the human body to identify physical abnormalities. This section mainly focuses on two techniques, based on surface analysis or using X-rays.

Surface analysis can give an overview of the internal shape and deformation of the spine. The principle has been exploited by the developers of SpinalMouse™, a measuring device that positions the coordinates of several points in the back and uses them to predict the internal shape of the spine. Using this technique, it is assumed that there is a strong relationship between the surface line on which the recording is done and the mid-line of the spine in terms of alignment [75].

- Advantage: non-ionizing radiation.
- Disadvantages: limited information about intervertebral torsion, which is particular important in the case of AIS.

X-rays are high frequency electromagnetic waves¹ that due to its small wavelength (5 pm to 10 nm) can go through the matter [22]. The amount of matter that X-rays can go through mostly depends on the following three parameters:

- the amount of photons emitted by the X-rays source;
- the density of the matter (i.e. bones are denser than soft tissues);
- the thickness of the matter.

Since X-rays produce ionizing radiation², they can produce alterations into the DNA³ of the patient [22].

Due to their properties, X-rays have been widely used to identify the spinal shape using several technologies such as radiography or CT scan. Following an analysis of those technologies:

Bi-planar X-rays:

- Advantages: low cost (most hospitals have it), standing up position.
- Disadvantages: quality of the images, ionizing radiation.

EOS system:

EOS is an imaging device that is able to take simultaneously antero-posterior and lateral X-rays from the whole body [60].

- Advantages: high quality 3D representation of the spine, standing position, low-ionizing radiation (about 50 to 80% less than conventional X-rays [60]).
- Disadvantages: cost of this technology, currently not accessible for us.

X-ray computed tomography (CT scan)

- Advantage: high quality 3D representation of the Spine.
- Disadvantages: lying position, high-ionizing radiation.

The following table gives the criteria that we took into account for our choice.

¹The electromagnetic spectrum ranges from long radio waves and the visible light to the high energy gamma rays observed in cosmic radiation [54].

²Ionizing radiation refers to random interaction between X-rays photons and the matter such as the photoelectric or the Compton effects.

³DeoxyriboNucleic Acid

Technique	Precision	Image quality	Patient irradiation	Position	Cost
Spinal Mouse	No vertebral shape info	no image 3D-model	None	Standing position	Low
Bi-planar X-rays	Medium: X-ray magnification	Medium	Medium	Standing position	Low
EOS	Good	Good	Low	Standing position	High
Scan: Laid	Good	Good 3D Image	High	Lying position	Medium
Scan: Standing-up	Good	Good 3D Image	High	Standing position	High
IRM	Low for hard tissues	Low for hard tissues	None	Lying position	Medium

Table 2.1: Decision-making criteria for the image acquisition

Our comparative study suggested that the best options for data acquisition are: the EOS system and the bi-planar X-rays. However, due to budget limitations, the EOS system was finally excluded.

b) Sensitivity analysis To identify the accuracy required by the geometrical identification — although it is a bit premature — a sensitivity analysis is already carried out at this stage to assess the effect of these inaccuracies on the expected inter-vertebral efforts.

For this purpose, several simulations were carried on via a vertical 2Hz sinusoidal motion —which is somewhat representative of the gait— imposed to the pelvis (Fig. 2.5, left), in order to estimate the intervertebral efforts via an inverse dynamics model (Fig. 2.6).

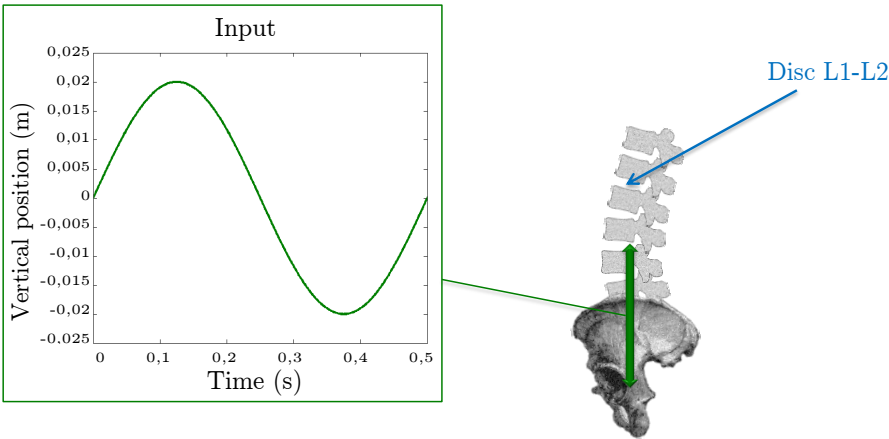


Fig. 2.5: Simplified motion used for sensitivity analysis of geometrical inaccuracies

For each simulation, we introduced a randomly-emulated measurement error in the multibody model for the lateral absolute position of all disc centers. Then, we compute the normal and shear forces acting into the intervertebral disc L1-L2 via a dynamical model (Chap. 3). Each simulation produces a different value for the intervertebral force; however, after several (about 25) simulations, we arrive to identify the range of variation of the force for a given maximum measurement error value of 1 mm, 3 mm and 5 mm (Fig. 2.6).

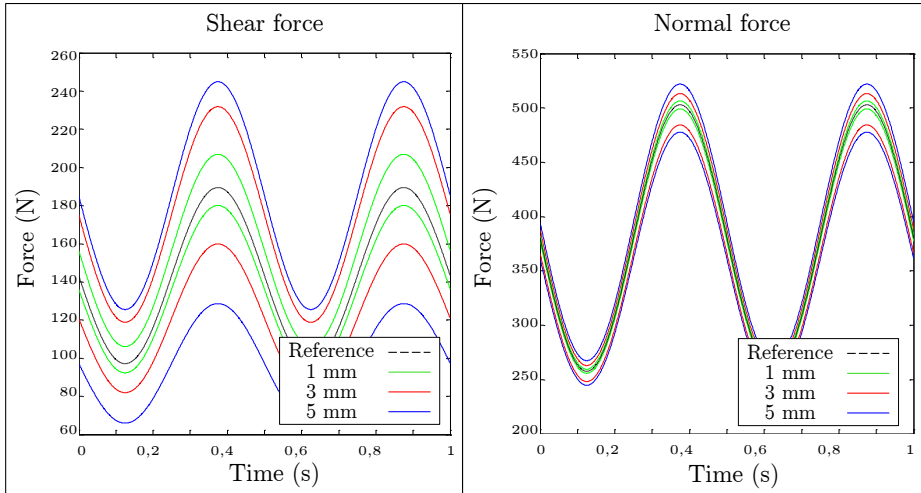


Fig. 2.6: Sensitivity analysis further to a geometrical random error of 1 mm, 3 mm and 5 mm on the geometrical identification of the spine

We can observe in (Fig. 2.6) that those inaccuracies have significantly affected the intervertebral forces in the L1-L2 disc during the simulated gait, compared to the reference spine when no random errors are applied (the central black curve in Fig. 2.6).

- In the case of the shear force, a measurement error of 1 mm, 3 mm and 5 mm causes a variation — with respect to reference curve — of 17N (about 9%), 44N (about 23%) and 55N (about 30%) respectively.
- In the case of the normal force, the same error causes a variation of 3.5N (about 1%), 11N (about 2%) and 19N (about 4%) respectively.

This results could vary from one patient to another. However, the above curves suggest that geometrical identification is a crucial step for the computation of the intervertebral efforts. Since shear forces represent one of the most important expected results, a rigorous protocol must be developed, to reach a maximum measurement error of 1mm.

c) X-ray measurement protocol As highlighted in the previous analysis, a protocol that minimizes errors induced by the bi-planar X-rays should be proposed. The following sources of inaccuracies must be considered:

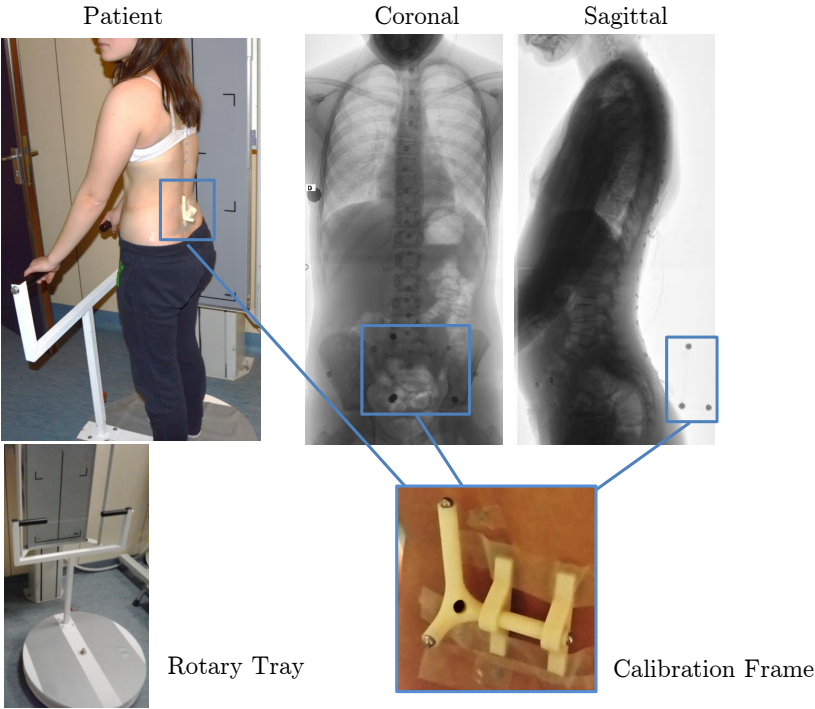


Fig. 2.7: Geometrical identification protocol

1) To ensure a good sagittal and coronal correlation between bi-planar X-rays, we have designed and built a rotary tray, which helps and forces the patient to stand still between the two radiographies (Fig. 2.7 left). The design of this rotary tray (Fig. 2.8) was developed based on the advises of physicians and radiographers, and taking into account the following specifications:

Functions:

- Rotation the patient of 90 degrees for bi-planar X-rays;
- Limitation of the patient's movements between the two radiographies;
- Orientation of the arms at 45° downwards.

Constraints:

- The Pelvis and spine must be totally visible on the X-rays;

- Use of materials that are compatible with the medical environment;
- Transportable.



Fig. 2.8: Rotary tray concept (CAD model)

2) As regards X-ray magnification, the problem has been solved by sticking a calibrated reference frame on the patient's back (Fig. 2.7 middle), visible on the radiography (Fig. 2.7 right). However, since the magnification is different along the body, depending on the distance to the X-rays source (Fig. 2.9), two geometrical phenomena have to be taken in account.

2.1) The magnification factor error induced by a lateral offset of the reference frame with respect to the spine axis (Fig. 2.9), where:

- s is the horizontal distance from the source (X-rays) to the spine;
- Δs denotes the transverse offset between the reference frame and the spine axis;
- p is the height of the frame projection on the detector (film);
- Δp represents the magnification error induced by Δs .

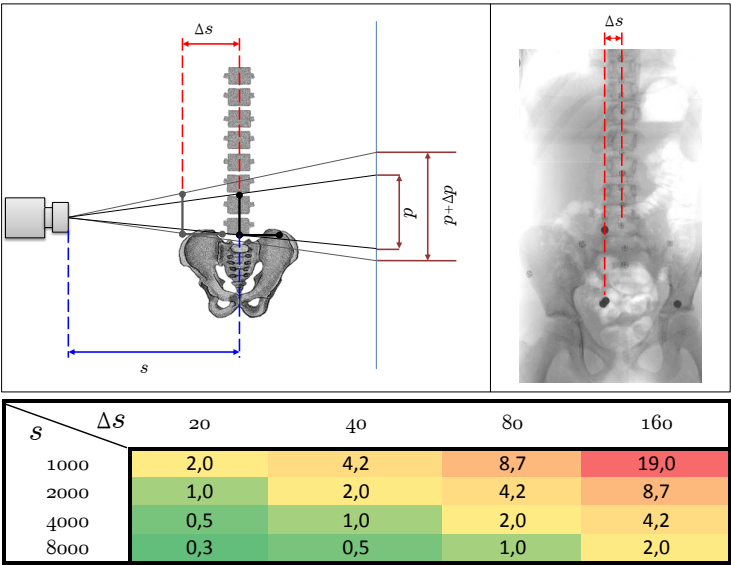


Fig. 2.9: Magnification factor error $\Delta p(\%)$. The values s and Δs are expressed in mm

The table in Fig. 2.9 gives the magnification error (Δp) induced by Δs . It is important to notice that, knowing the exact value of s , we could reduce this error further. However, this process would require to complexify the X-ray protocol and its effect will be minimal for the results. In practice, with enough operation precautions, it is possible to limit Δp to 1%.

Let us note that the magnification does not change along the spine axis whatever the angle of incidence (Fig. 2.10). This is straightforwardly demonstrated via the rule of similar triangles. However, in the case of scoliosis, Δs will vary from one vertebra to another, leading to a slightly different magnification along the spine.

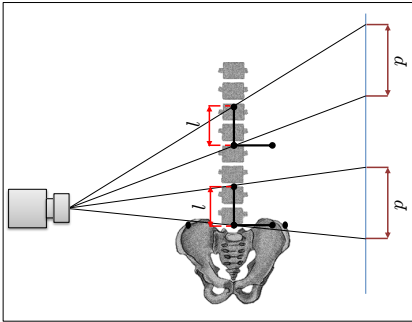


Fig. 2.10: Magnification factor does not change along the spine axis.

2.2) In the case of the landmarks on the pelvis located out of the spinal axis, the left/right magnification discrepancy (Fig. 2.11 left) induces a vertical measurement error Δh (Fig. 2.11 right), where:

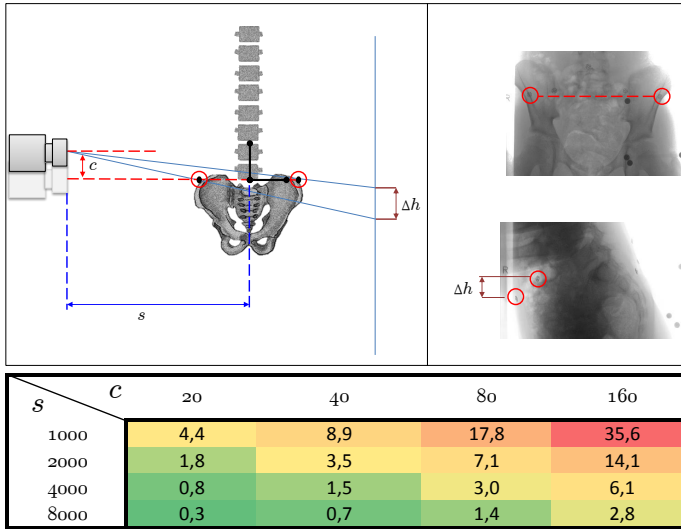


Fig. 2.11: Left/right measurement discrepancy $\Delta h(mm)$. All values are expressed in (mm)

- s denotes the horizontal distance from the source (X-rays) to the spine;
- c is the vertical distance between the source axis and the physiological landmark;
- Δh represent the vertical measurement error induced by c .

The table of Fig. 2.11 shows the vertical measurement error (Δh) induced by c . Once again by experience, with enough operational precautions, when positioning the X-rays source with respect to the iliac crests, it is possible to limit the error to less than $1mm$ (See green cells in the table)

Finally, a user-friendly¹ measurement interface was developed (Fig. 2.12). This interface allows the user, by pointing on the screen, to identify the position of the iliac crests, the intervertebral discs centers and the spinous processes² from the bi-planar X-rays and, at the same time, to verify that the protocol was correctly applied. In

¹No programming knowledge required

²In the case of the iliac crests and the spinous processes, their positions refer to the metallic markers placed on the skin during the clinical assessment.

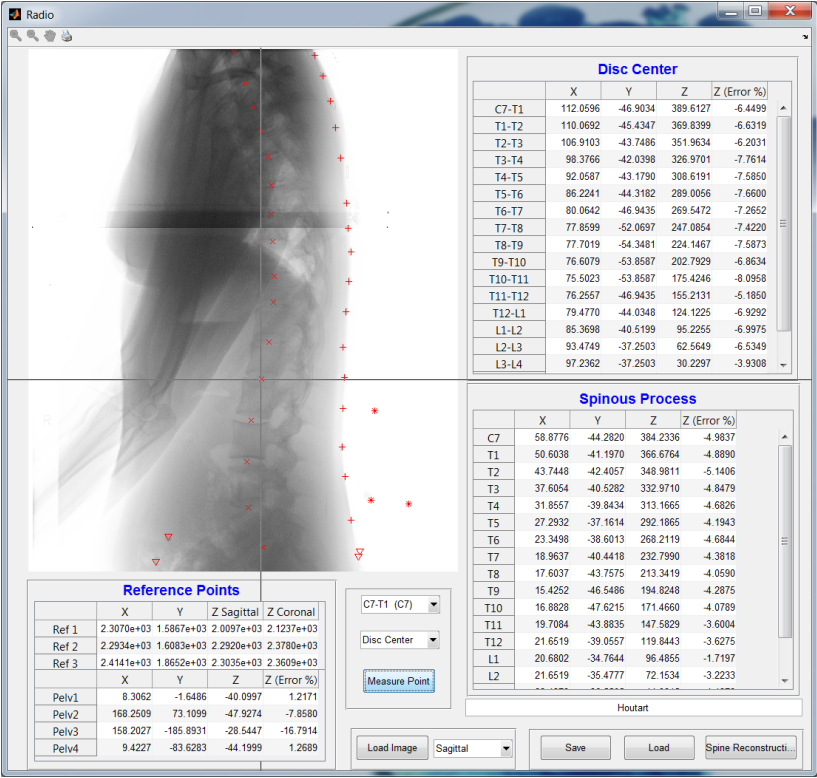


Fig. 2.12: Spine geometry acquisition: user Interface

order to explain how this verification is done, let us describe the measuring process of the calibration frame and of the spine (Fig. 2.13).

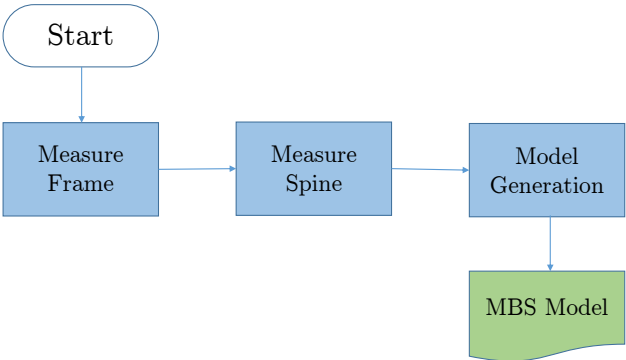
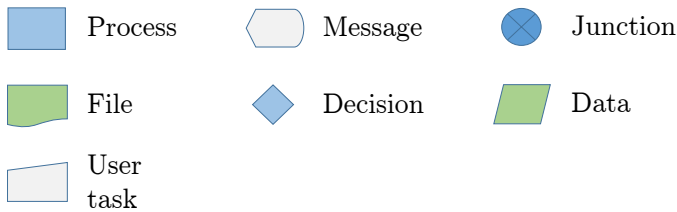


Fig. 2.13: Spine geometry: the measurement process

The notation of Fig.2.13 will be used throughout the text:



where each color represents:



The first step of the geometrical identification consists in measuring the calibration frame to calculate the scaling factor Δp . As explained in Fig. 2.14, the user will load the patient X-rays and point on the screen its 3 reference points. Then for each X-ray, coronal and sagittal, the program computes a scaling factor in the two orthogonal directions. The program will then verify if a difference of less than 1% is reached between the scaling factors in both directions. If so, the final scaling factor for each X-ray will be the average of them. If the scaling factor is different in both direction, an error message, asking the user to repeat the procedure, will be displayed on the screen.

The second step of the geometrical identification consists in pointing the landmarks located in the spine and the pelvis (iliac crests, intervertebral discs and spinous processes). Both sagittal and coronal radiographies are treated separately. Sagittal and coronal radiographies give complementing information. However, both should give the same information about the z coordinate of each reference point of the spine. This redundant information can be advantageously exploded to verify the correlation between both X-rays. This correlation will be verified by the user (Fig. 2.15) and will allow him estimating the precision of his measurement.

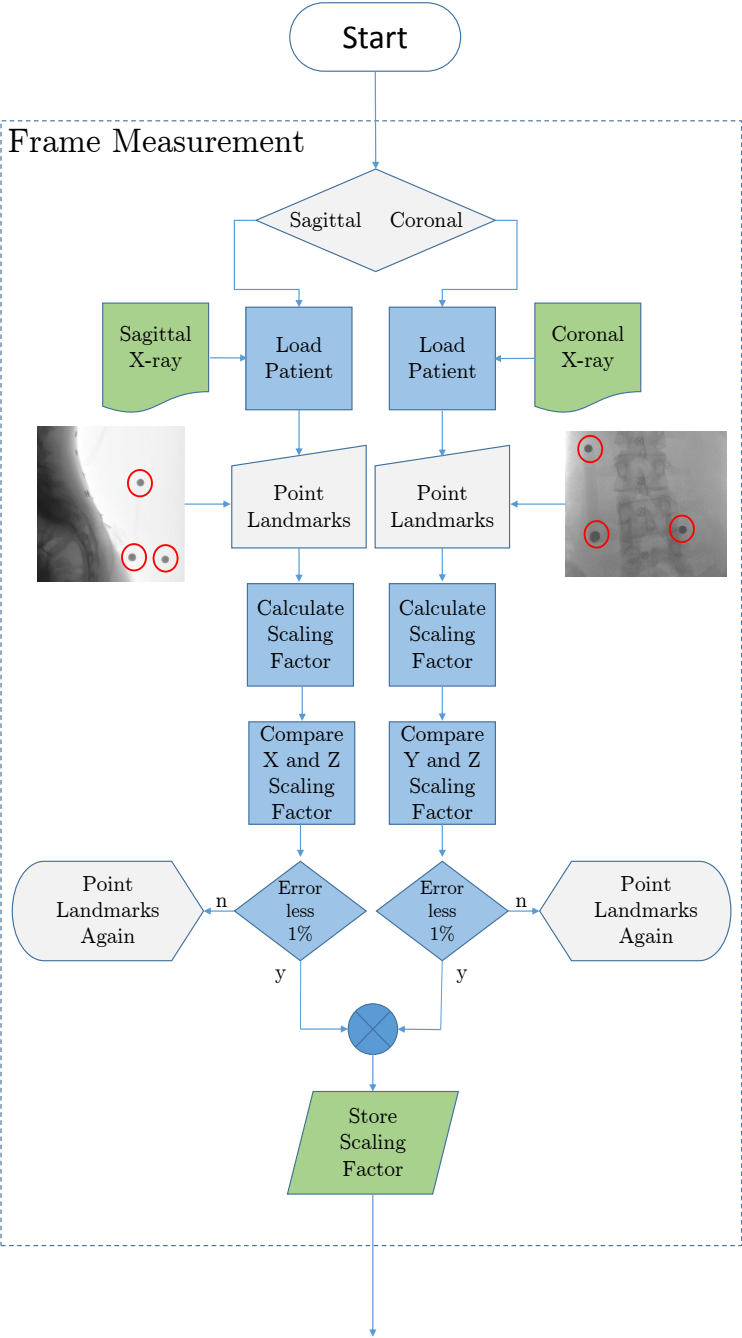


Fig. 2.14: Measurement process of the calibration frame

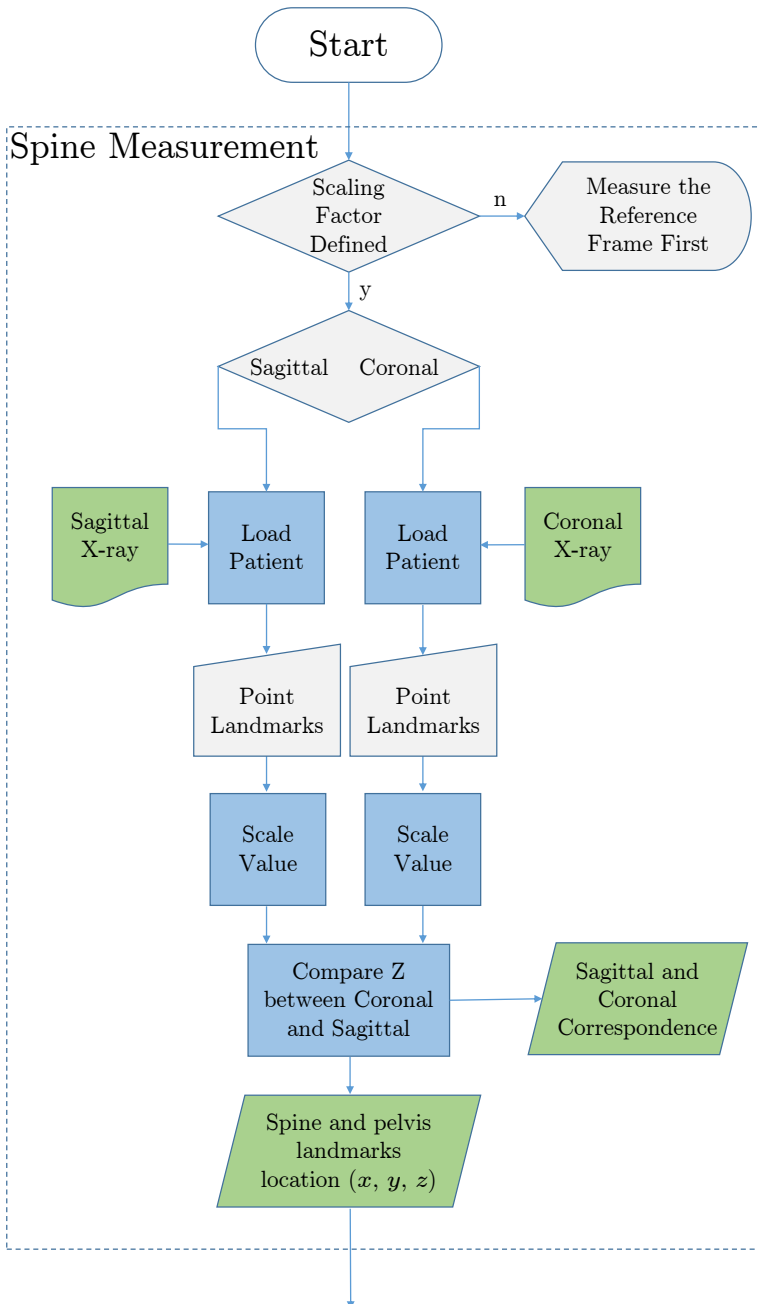


Fig. 2.15: Measurement process of the spine and pelvis landmarks

2.4 Gait analysis

As explained in Sect.2.1, the severity of scoliosis could be also quantified via the knowledge of the intervertebral efforts during gait. In order to have an idea of the importance of the dynamical effects, Fig. 2.6 (Sect.2.3) reveals that the intervertebral efforts are highly influenced by the gait motion: the intervertebral effort between L1-L2 is about $375N$ in a static position, compared to almost $500N$ in a dynamical situation. Although this comparison was based on a simplified motion, in the real life, this effect might be worse due to impacts during gait.

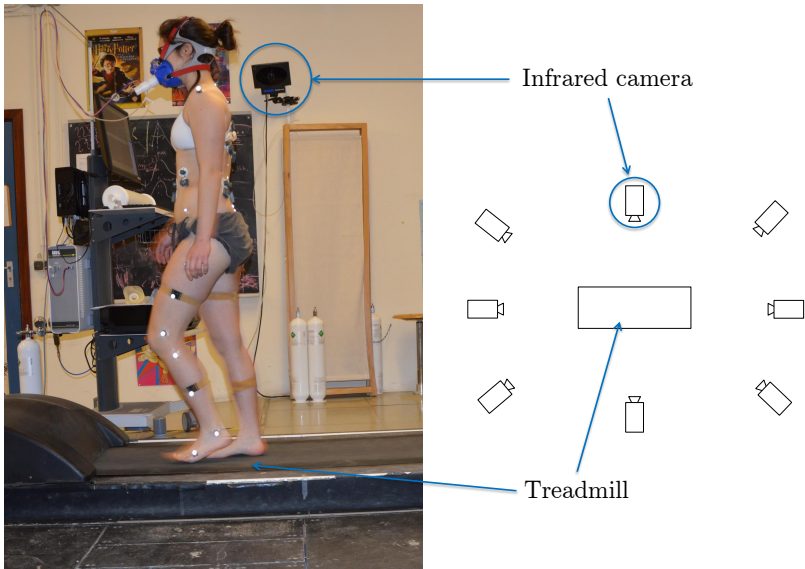


Fig. 2.16: Gait analysis laboratory: Cliniques universitaires Saint-Luc, UCL.

For the present study, the real kinematics of the spine and pelvis is captured during gait via a motion capture system (Fig. 2.16), consisting of 8 cameras, reflective markers and a treadmill. The reflective markers are stuck on the arms, pelvis and back of the patient. In the case of the back and the pelvis, they are stuck exactly at the same place (Fig. 2.17) as the metallic markers used for the geometrical identification of the spine (Sect. 2.2 and 2.3)

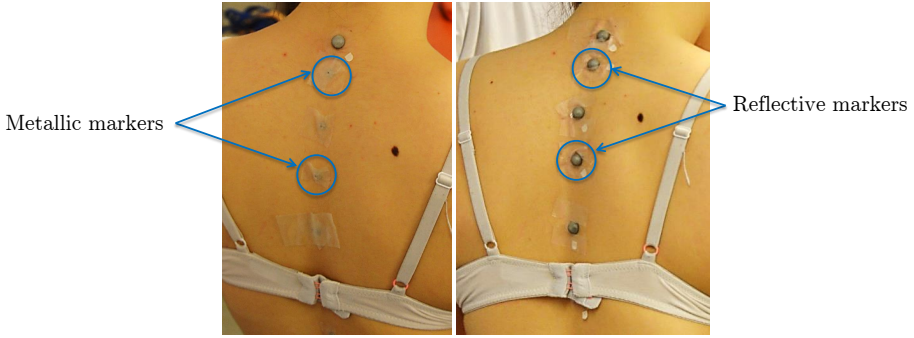


Fig. 2.17: Metallic and reflective markers stuck on the patient

It is important to remind that all the motion information of the lower body are useless as soon as the full kinematics of the pelvis is captured. In other words, the motion of the pelvis being known¹, it does not matter the source of this motion to compute the induced intervertebral efforts.

2.4.1 Data processing

In order to exploit the gait analysis data, they should be treated as depicted in Fig. 2.18.

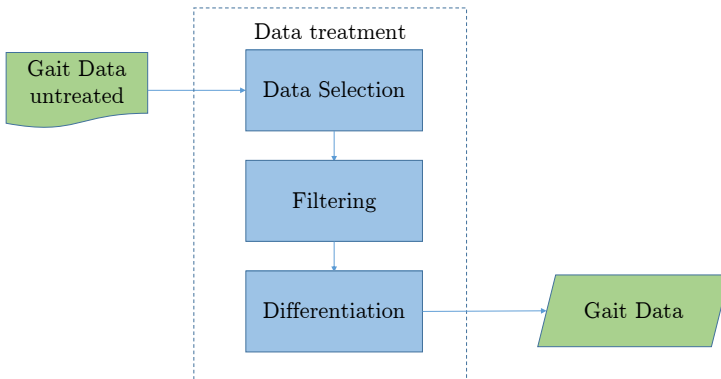


Fig. 2.18: Kinematic data processing path.

Data selection One of the biggest limitations during gait analysis is the amount of data lost in the process. Indeed, sometimes, the reflective marker are switched off by the cameras causing several losses in the data. This problem is worsened when

¹See the inverse dynamics approach in Sect. 3.1.

locating reflective markers close to each other, limiting their number to one per two vertebrae. Nonetheless, there are still data losses that have to be treated. As shown in Fig. 2.19, we developed a computational process to find the best set of data from a given full gait capture (left side of the graph). Then, some minor data losses are linearly interpolated (Red-Lines).

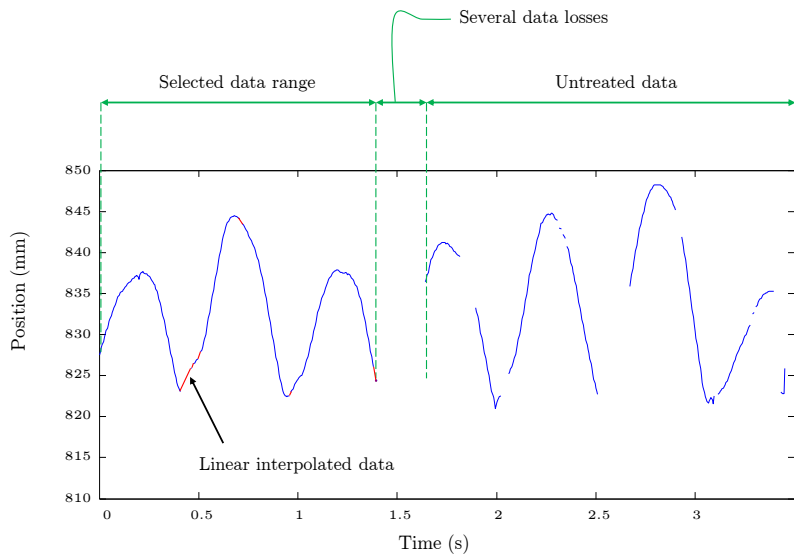


Fig. 2.19: Reflective markers: data losses during gait

In order to select the best data range, we have analyzed two parameters over time which characterize the quality of the data: (1.) the number of data lost at each time frame, and (2.) the time interval during which data are lost. Fig. 2.20 allows to best understand how the selection algorithm works. The abscissa axis represents the time, and the ordinate axis represents the amount of data lost. The four horizontal lines characterize the data intervals selected by our algorithm. The lowest bar represents high quality data, with only two data losses, equivalent to 1/100 second. For most of the tests, we impose a minimum interval of two complete gait cycles, corresponding to the target we have set for gait analysis. If the algorithm cannot manage to find this minimal interval, it will be satisfied with less quality data and will find the associated time interval within the global time window. The maximum amount of data losses allowed by this algorithm is 16, what is equivalent to 8/100 of second (the black horizontal bar in Fig. 2.20).

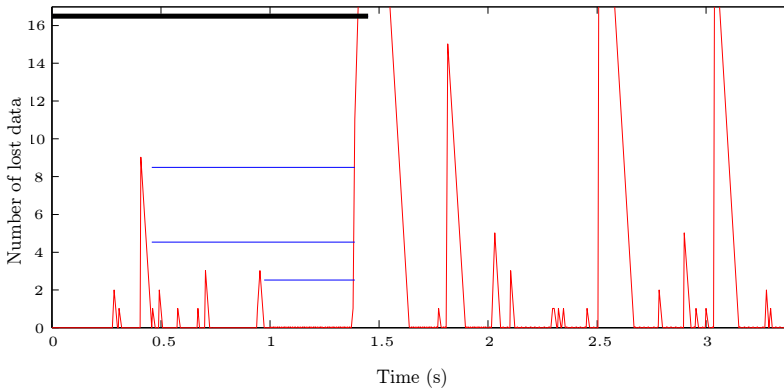


Fig. 2.20: Reflective markers: data selection algorithm

Filtering The filtering of the data was done via a Zero-phase digital filtering. This filter process data of the ordinate in both forward and reverse direction. [66]. The parameters of the filter were the following: Cut-off frequency of $8Hz$ and an order of 6.

Differentiation In order to calculate the velocity and the acceleration of the reflective markers, we first fit their filtered position using cubic splines, which can be analytically differentiated to obtain the velocity and the acceleration. Compared to the finite difference technique, this proposed interpolation method never induces singular points or discontinuities.

2.4.2 Validation

As explained at the beginning of this section, all the kinematic information issuing from the lower body (feet and legs) is useless since only the kinematics of the pelvis is relevant for the spine dynamical analysis. The position of each vertebra can be described as the addition of the vectors $\mathbf{P}$, representing the position of the pelvis, and $\mathbf{p}$, representing the position between the pelvis and the vertebrae. This also applies to velocity, $\dot{\mathbf{P}} + \dot{\mathbf{p}}$ and acceleration, $\ddot{\mathbf{P}} + \ddot{\mathbf{p}}$. From a dynamical point of view, the biggest contribution to the intervertebral efforts will obviously be the — linear and angular — accelerations of the pelvis $\ddot{\mathbf{P}}$.

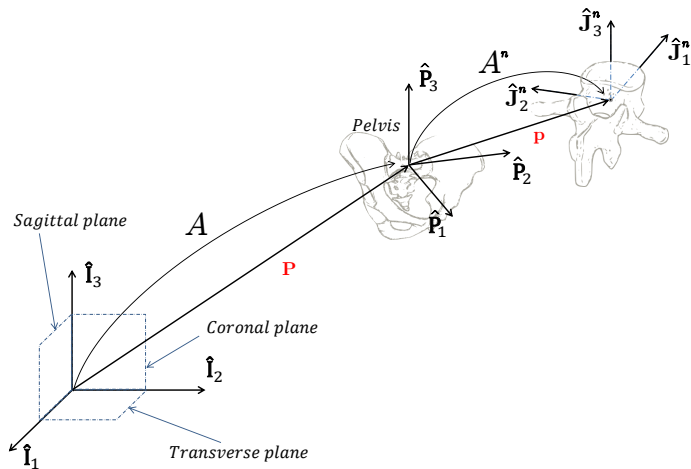


Fig. 2.21: Vertebra n : absolute position vector

In order to verify that this acceleration is well-captured, we measured it using two different approaches: using reflective markers, and comparing them with accelerometers stuck on the pelvis. Fig. 2.22 successfully compares the two approaches which have been used for a moderate but also a jerky gait, the goal being to assess for the robustness of the reflective markers measurement and of the time differentiation of their position.

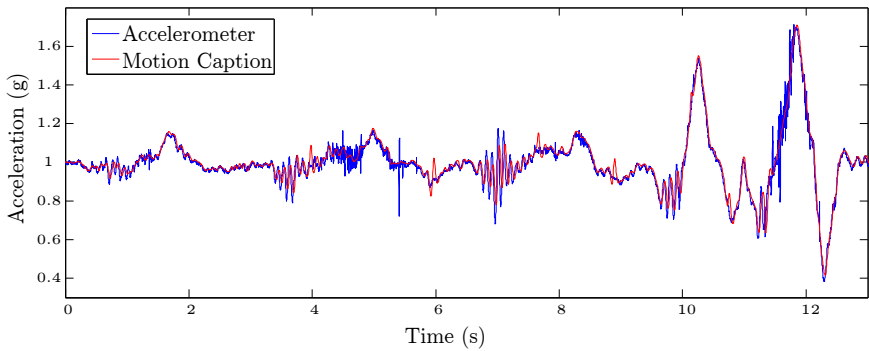


Fig. 2.22: Comparison between the pelvis accelerations measured via an accelerometer and by time differentiation of position vectors

Fig. 2.22 illustrate the pelvis acceleration (magnitude) during an up and down motion. One may hopefully observe that there are no important differences between both measurements.

2.5 Muscular force estimation

Several muscles are in charge of the support and stabilization of the spine. As illustrated in Fig. 2.23, each muscle acts as an elastic cord, provoking additional forces inside the intervertebral discs. Therefore, their knowledge is fundamental to estimate those forces. This relationship between muscles and intervertebral forces will be assessed in the part dedicated to results (Sect. 4.2.2)

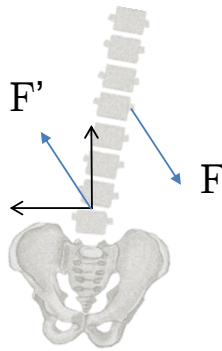


Fig. 2.23: Muscle forces ensure bending rigidity but induce intervertebral forces in the discs (only one is illustrated).

2.5.1 Muscles inventory and calibration

As explained before, there are several groups of muscles connected to the spine and they can be classified as extrinsic — for the superficial muscles — or intrinsic — for the inner muscles — [24]. The function of the muscles varies according to their position: while intrinsic muscles are mostly in charge of the spinal support, extrinsic muscles are in charge of the movement of the upper limbs and the thoracic wall.

Rather than focusing on individual muscles, the present study will consider four muscles *groups* according to their physiological function: flexion, extension, rotation and lateral bending. The estimation of the muscular forces associated to each of those physiological functions is achieved using EMGs, despite the well-known limitation of these measurements¹. EMGs are then calibrated using the David Back test which helps us to correlate the muscular forces with the EMG values for a well-defined static configuration (Sect. 3.7).

¹but which have at least the merit to exist...

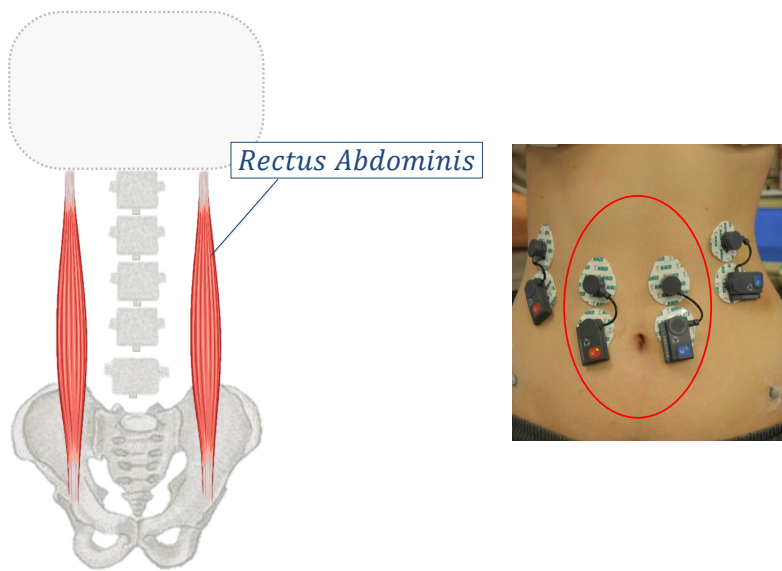


Fig. 2.24: Muscle involved: Rectus abdominis



Fig. 2.25: David Back test: flexion

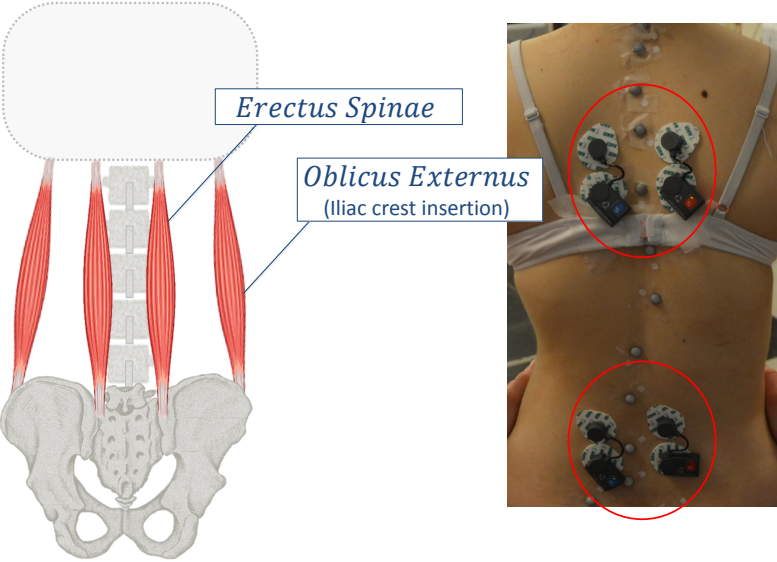


Fig. 2.26: Muscles involved: Erector spinae and Obliquus externus (Iliac crest insertion)

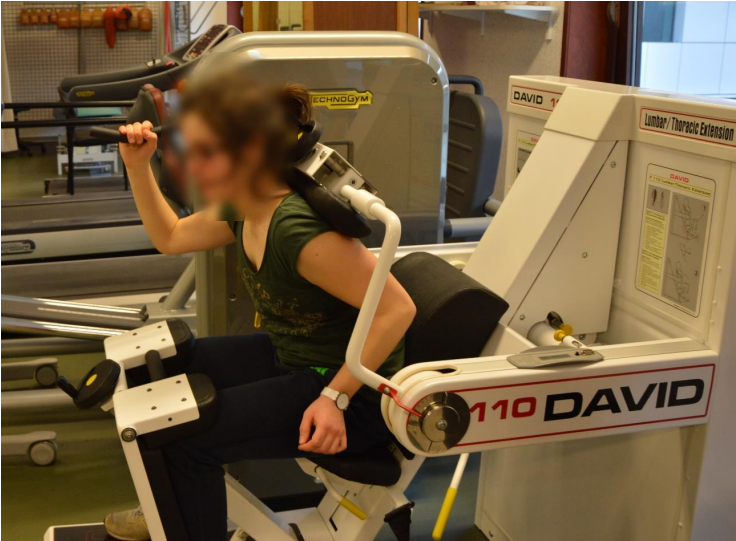


Fig. 2.27: David Back test: extension

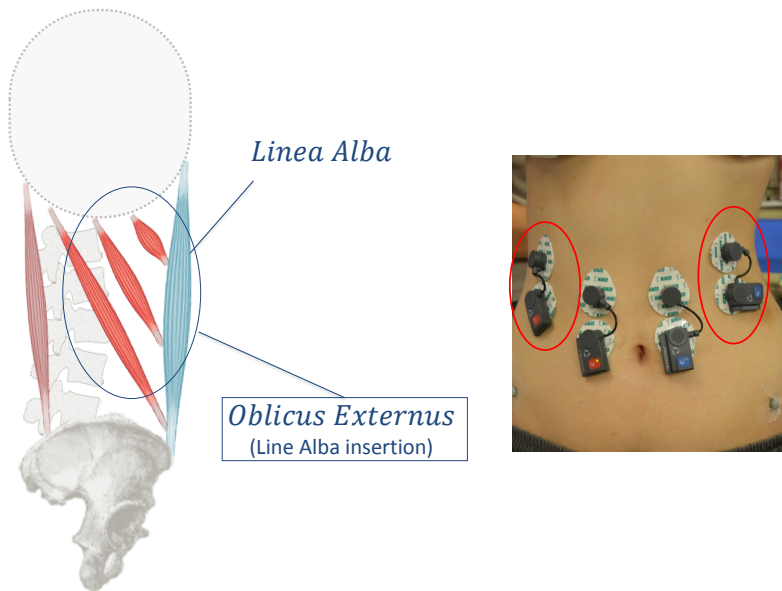


Fig. 2.28: Muscle involved: Obliquus externus (Linea Alba insertion). These muscles are not taken into account for the final multibody model due to their limited influence on the intervertebral forces.



Fig. 2.29: David Back test: torsion

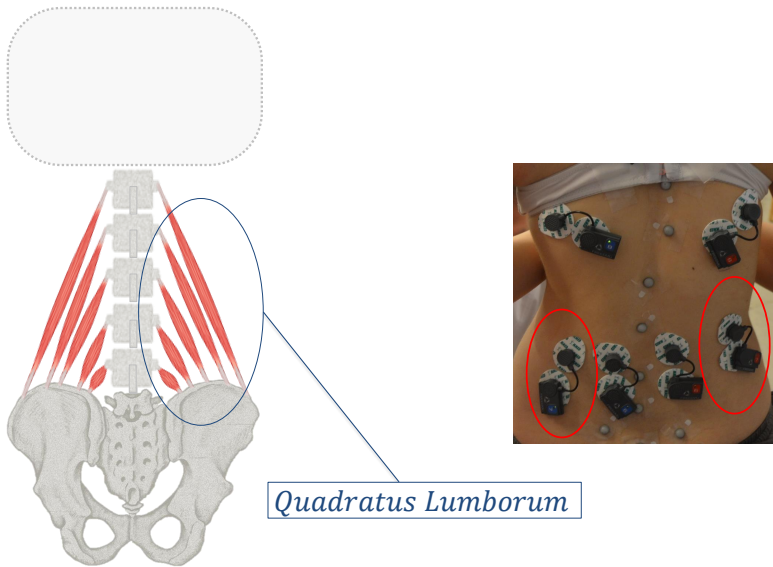


Fig. 2.30: Muscles involved: Quadratus lumborum



Fig. 2.31: David Back test: lateral flexion (i.e. bending)

2.5.2 Data Treatment

The treatment of the EMG data is carried in three steps, following the process proposed in [82], as illustrated in Fig.2.32:

- First, applying a high pass Zero-phase digital filter, with a cut-off frequency of $30Hz$. This will allow reducing the noise provoked by the measurement chain (signal perturbations) and by the movement of the body (i.e. skin effect).
- then, rectifying the signal (getting the envelope of the signal) using the Hilbert transform.
- Finally, smoothing the signal using a moving average filter.

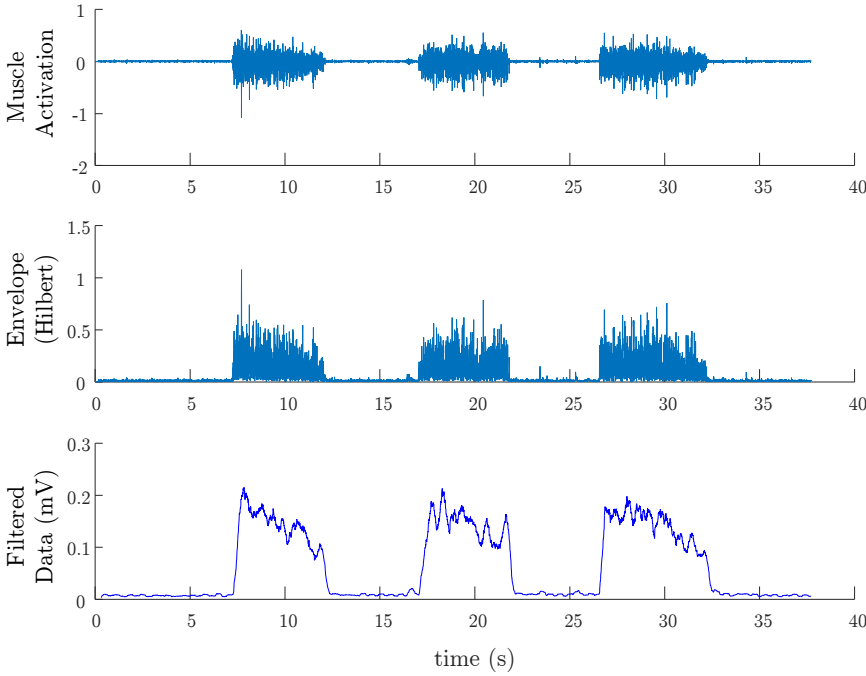


Fig. 2.32: Illustration of the EMG Treatment: the illustrated data come from a David Back test aiming at calibrating the EMGs with the force applied by the patient (see Fig.2.25).

Biomechanical model

3

3.1 Biomechanics and multibody dynamics

In the case of the musculoskeletal system, two different approaches can be considered for modeling systems composed of rigid or flexible bodies: Multibody Dynamics (MBD), Finite Element Method (FEM) or a combination of both. On the one hand, FEM is well-suited for static analyzes or low dynamics of continuous media, for instance to compute the mechanical behavior of bones and soft tissues. Nevertheless, the method relies on complex formulations and requires huge computational time when dealing with motion dynamics. On the other hand, MBD is far more efficient for situations involving large translational and rotational displacements, like gait analysis, limb rehabilitation or sports movements. However, in comparison with FEM, MBD is clearly less adapted to continuum mechanics.

In this research, a rigid MBD approach has been selected for the two following reasons:

- gait, being at the heart of the project, *de facto* involves large motions with moderate — but non negligible — dynamical effects inside the human body,
- the focus is put on the intervertebral efforts resultants rather than in the internal stress and strain within the vertebrae or any other organ.

Modeling difficulties inherent in Biomechanics Compared to multibody systems like vehicles, robots or mechanisms, the human body (as a musculoskeletal system) raises important issues regarding data completeness, definition and accuracy. Without being exhaustive, the following elements are directly concerned:

- the estimation of the mass, center of mass and inertia matrices of segments and organs,
- the kinematics of the joints that most often consist of a combination of translations and rotations subject to possible geometrical constraints,
- the knowledge of muscle forces (amplitude and direction), which probably represents the most foggy data for the biomechanics of the human movement,
- the data acquisition by itself, because the system under consideration is a person who is unique, raises ethic problems and is far from having a perfect "mechanical design" (e.g. bone shape, skin effects, evolutive morphology, etc.)

3.2 Multibody formalism

A multibody system (MBS) is a mechanical system composed of N bodies¹ connected by joints. A tree-like structure (Fig. 3.1 left), in which there are no loop of bodies, is always used to describe the topology of a MBS, i.e. the way the bodies are connected (more details can be found in [84]. Serial robots, cranes, human body are examples of tree-like multibody systems. Each time a loop of bodies is present in the system (Fig. 3.1 middle), a particular procedure must be used to restore a temporary tree-like structure either by cutting the loop (e.g. a body or a joint) or by removing a body whose mass is negligible and is just a force transmitter (e.g.:a muscle) (Fig. 3.1 right). Parallel robots, vehicle suspensions, slider-crank mechanisms are examples of closed-loop MBS.

In this work, the global skeleton model under consideration is a tree-like MBS that starts from the ground (inertial frame), continues to the pelvis (reference floating frame), extends through the spine (trunk) and ends with the upper limbs (branches). When muscles are taken into account (Sect. 3.7), they create loops of bodies leading to a closed-loop MBS. A technique to restore a tree-like topology will be carried out to deal with the associated musculoskeletal dynamical model.

¹considered as rigid, i.e. non deformable in our case

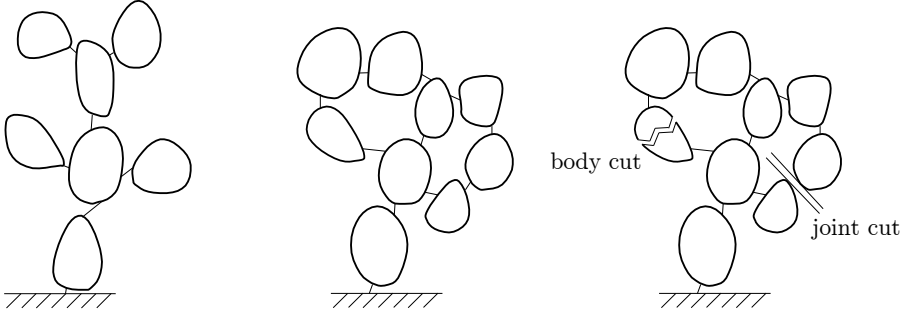


Fig. 3.1: Schematic representation of a multibody system, i.e. a set of bodies (potato shapes) connected by joints (straight lines). Left: Tree-like structure; Middle and Right: closed-loop structure - loop cuts

3.2.1 Frames, vectors and tensors

As shown in Fig. 3.2, an orthonormal body-frame $\{\hat{\mathbf{X}}^i\}$ is rigidly attached to each body i , its origin being the body center of mass CM^i . In a similar way, the inertial frame $\{\hat{\mathbf{I}}\}$ is attached to the fixed body (ground); Its unit vectors are oriented as follows x points forward, z upward and y to the left.

The following notations are used to define vectors and tensors:

A vector $\mathbf{a}$ is described in a given body frame $\{\hat{\mathbf{X}}^i\}$ by its three components a_x, a_y, a_z :

$$\mathbf{a} = a_x \hat{\mathbf{X}}_x^i + a_y \hat{\mathbf{X}}_y^i + a_z \hat{\mathbf{X}}_z^i \quad (3.1)$$

Writing the frame $\{\hat{\mathbf{X}}^i\}$ under the form of a 3×1 column matrix of unit *vectors*:

$$[\hat{\mathbf{X}}^i] \triangleq \begin{pmatrix} \hat{\mathbf{X}}_x^i \\ \hat{\mathbf{X}}_y^i \\ \hat{\mathbf{X}}_z^i \end{pmatrix} \quad (3.2)$$

and applying classical matrix multiplication rules, we can write vector $\mathbf{a}$ in the following form:

$$\mathbf{a} = [\hat{\mathbf{X}}^i]^t \mathbf{a} \text{ where } \mathbf{a} \triangleq (a_x \ a_y \ a_z)^t \quad (3.3)$$

In terms of orientation, each moving frame in a MBS is linked to the inertial frame via a 3×3 orthogonal rotation matrix A . In particular:

$$[\hat{\mathbf{X}}^i] = A(q) [\hat{\mathbf{I}}] , \quad [\hat{\mathbf{I}}] = A(q)^t [\hat{\mathbf{X}}^i] \quad (3.4)$$

where q designates angles or any kind of rotation parameters.

3.2.2 Body characterization

In this work, the parameters of bodies (and later on, of joints) are described using the ROBOTRAN software terminology and conventions [84]. Let us consider a body i (i.e. a vertebra) connected to its parent body h (i.e. the previous vertebra) via a joint characterized by the vector $\overrightarrow{O^i O'^i}$, and to its child body j via the joint vector $\overrightarrow{O^j O'^j}$ (Fig. 3.2).

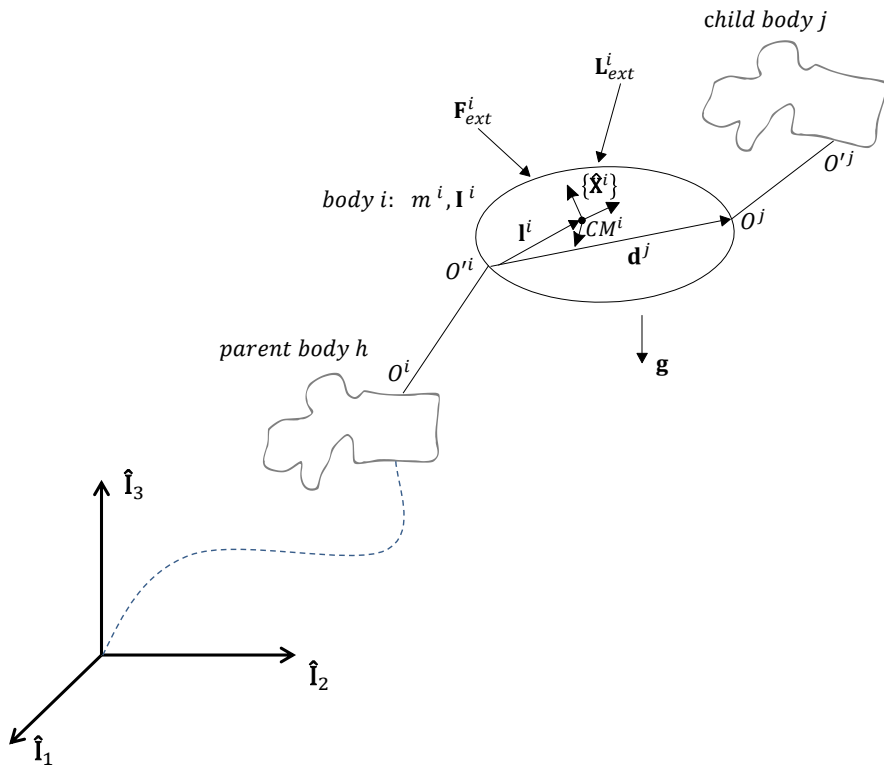


Fig. 3.2: Characterization of a body i of a multibody system

- O^i, O'^i denote the two reference connection points of joint i , on body h (parent of body i) and body i respectively. The $\overrightarrow{O^i O'^i}$ vector represents any kind of physical joint (prismatic, revolute, spherical, etc.) that allows a relative motion of body i with respect to body h (For instance, the relative displacement between two consecutive vertebrae).
- $d^j = [\hat{X}^i]^t d^j$, called the joint vector, locates on body i the connection point O^j of joint j with respect to O^i of joint i . Body i being rigid, the components d^j are constant in the $\{\hat{X}^i\}$ frame.

- $\mathbf{l}^i = [\hat{\mathbf{X}}^i]^t \mathbf{l}^i$ is the vector that locates the center of mass CM^i of body i with respect to the connection point O^i of joint i . The components l^i are constant in the $\{\hat{\mathbf{X}}^i\}$ frame.
- m^i is the mass of body i .
- I^i denotes the symmetric inertia matrix of body i with respect to its center of mass CM^i and to its body-fixed frame $\{\hat{\mathbf{X}}^i\}$.

3.2.3 Joint characterization

In the multibody approach, joints between bodies represent devices such as telescopic arms, hinges, universal joints, ball joints, etc. In biomechanics, shoulder, elbow, knee, ankle, intervertebral discs, etc. represent the joints of human multibody models. Relative motions allowed by these joints are described in ROBOTRAN by *relative* generalized coordinates $q(t)$, also called joint coordinates: they are responsible of the motion *degrees of freedom* (d.o.f.) of the MBS. The absolute configuration¹ of a MBS can be computed in terms of these relative coordinates q and their time derivatives $\dot{q} \triangleq \frac{dq}{dt}$ and $\ddot{q} \triangleq \frac{d^2q}{dt^2}$.

According to the ROBOTRAN conventions, a joint possesses only one degree of freedom, either *prismatic* (denoted T) or *revolute* (denoted R). More elaborate joints (spherical, universal, or even a general 6 d.o.f. joint) can be straightforwardly modeled via a succession of these two elementary joints.

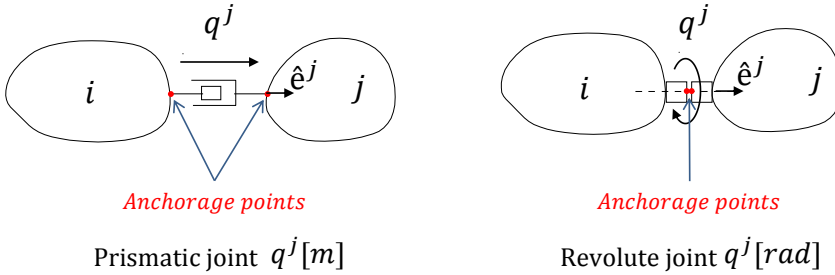


Fig. 3.3: Elementary joint characterization

Each joint i is assumed to be massless and connects body i to its parent h at the anchorage reference points, O^i , O^h respectively (Fig. 3.2). These points, which can

¹The configuration of a multibody system not only refers to its position/orientation but also to its instantaneous linear/angular velocities and accelerations.

be arbitrarily located on the joint axis, are distinct for a prismatic joint and identical for a revolute joint. For practical reasons, the joint unit vector $\hat{\mathbf{e}}^j$ is introduced; it is aligned with the joint axis (Fig. 3.3). In joint j , the generalized coordinate q^j represents:

- a relative displacement [meter] if j is prismatic, such that $\overrightarrow{O^j O'^j} = q^j \hat{\mathbf{e}}^j$ (q^j is thus positive in Fig. 3.3_a). In a human body model, such a joint can be useful to represent the intervertebral shear or compression.
- a relative angle [radian] if j is revolute, characterizing the rotation, along the unit vector $\hat{\mathbf{e}}^j$, of body j with respect to its parent body i . In the human body, most of the articulations can be modeled by a revolute joint or a well-suited combination of these.

3.2.4 Forces and torques

According to the ROBOTRAN conventions, forces and torques acting on bodies, are splitted into four different categories (Fig. 3.2):

1. The *gravitational* force. This force, considered separately in the formalism, is computed from the gravity vector $\mathbf{g}$ expressed in the inertial frame: $\mathbf{g} = [\hat{\mathbf{I}}]^t g$.
2. The *joint* forces/torques. A joint force [resp. torque], $\mathbf{Q}^i = Q^i \hat{\mathbf{e}}^i$, acts from body h on body i through the prismatic [resp. revolute] joint i . In the present study, these forces will be used to simulate the intervertebral efforts acting between vertebrae.
3. The *link* forces. These forces act between two materials points of two distinct bodies i and j of the MBS, independently from the presence and location of the joints. In other words, being not associated with any joint, these link forces are considered as two "action-reaction" internal forces, $\mathbf{F}_{lnk}^{ij}$ and $\mathbf{F}_{lnk}^{ji} = -\mathbf{F}_{lnk}^{ij}$. They are of particular interest since their use allows to artificially restore a tree-like MBS (Fig. 3.1), physically closed by the device that produces them. In so far as the mass — or more exactly the dynamics — of this device can be reasonably neglected, the latter can be simply removed from the MBS and replaced by a link force. In this study, the latter are very suitable to represent muscles, whatever their underlying constitutive law, ranging from a simple spring to a Hill-based model.
4. The *external* forces/torques. Besides joint forces, link forces and gravity, all other forces and torques are considered as external to the MBS and gathered together, for each body i , into a resultant force vector $\mathbf{F}_{ext}^i$ and a resultant

torque vector $\mathbf{L}_{ext}^i$ with respect to body i center of mass. Foot/ground contact forces are typical external forces in biomechanics; they will not be necessary in our model.

With regard to forces and torques, there is an important distinction to highlight in this work, referring to their known or unknown nature.

Known forces/torques refer to those whose value is available from well-established constitutive laws or experimental data. Among these are, for instance, the gravitational force, visco-elastic forces, forces measured from strain gauge sensors or muscle forces estimated from EMG data and/or a specific model. Whatever their measurement accuracy or model complexity, from a multibody point of view, those forces represent model input and do not belong to the set of unknown of the dynamical equations.

Unknown forces/torques are those for which there is neither constitutive model nor experimental input available. If they stem from an imposed — local — motion (i.e. a relative position/orientation and a possible imposed velocity and acceleration), they are called *constraint* forces in a multibody formalism, simply because they refer to a constrained motion. Those forces are completely part of the available results of the multibody dynamical equations.

In the present study for instance, the imposed kinematics of the pelvis is associated with three constraint forces and three constraint pure torques arising from the legs, the latter being intentionally absent from the model (Sect. 3.2.5). Intervertebral shear and normal forces, whose determination is one of the goal of this work, also refer to constraint forces, insofar as we decided to block the associated deformations (shear and normal strains).

3.2.5 Multibody models and analyzes

Multibody dynamics offers different possibilities of analyzes of MBS depending on the application and of the expected results, namely: direct kinematics/dynamics, inverse kinematics/dynamics, dynamical identification, static and quasi-static equilibrium, modal analysis, etc. In this section, we will mainly focus on direct and inverse approaches, both of them being at the root of the expected results. In particular, a method combining the direct and inverse dynamics models will also be presented, since it will allow us to compute the muscular and the intervertebral forces.

Direct dynamics This model aims at computing the generalized accelerations $\ddot{q}$ for a given configuration $(q, \dot{q})$ of the system to which forces and torques are applied. Direct dynamics is quasi exclusively used to predict the motion of a MBS

(i.e.: $q(t), \dot{q}(t)$), starting from an initial configuration $\{q(t = 0), \dot{q}(t = 0)\}$, by time-integrating the accelerations $\ddot{q}(t)$ via a suitable numerical technique (e.g. Runge-Kutta, Euler or Gear methods, ...).

The equations describing the direct dynamics of a MBS with n degrees of freedom (d.o.f.) can be written in the following form:

$$M(q, \delta)\ddot{q} + c(q, \dot{q}, \delta, g, \dots) = Q(q, \dot{q}) \quad (3.5)$$

where

- $q [n * 1]$ denotes the relative generalized coordinates,
- $M [n * n]$ is the symmetric generalized mass matrix of the system,
- $c [n * 1]$ is the non linear dynamical column-vector which contains the Coriolis, gyroscopic and centrifugal terms as well as the contribution of the gravitational force, the link forces $\mathbf{F}_{lnk}$ and the external forces $\mathbf{F}_{ext}$ and torques $\mathbf{L}_{ext}$,
- $\delta [10 n * 1]$ gathers together the dynamical parameters of the bodies: masses m^i , centers of mass vectors $\mathbf{l}^i$ and inertia matrices I^i ,
- $Q [n * 1]$ represents the generalized joint forces (torques).

The accelerations $\ddot{q}$ can be obtained by solving the linear system (3.5). Then, their double time integration allows to predict the motion of the MBS and, incidentally, to visualize it via a virtual animation of the bodies.

Inverse dynamics In this case, the goal is to compute the generalized joint forces or torques Q to be applied in the joints for a given (set of) configuration(s), $q(t), \dot{q}(t), \ddot{q}(t)$, of the MBS to which external and other internal forces and torques are applied. Inverse dynamics reads:

$$Q = \phi(q, \dot{q}, \ddot{q}, \delta, \dots) \quad (3.6)$$

where ϕ is nothing else than the LHS of the direct dynamics equations (3.5) but in which the explicit — and costly — computation of the mass matrix is avoided, being useless here¹. Inverse dynamics (3.6) is classically used to control robot manipulators, by predicting the actuator torques Q required for satisfying a desired trajectory $(q(t), \dot{q}(t), \ddot{q}(t))$. In biomechanics, it allows computing internal efforts, like inter-segment forces or other contact forces resulting from any general dynamical load

¹The inverse dynamics model computed via equation (3.6) is computationally more efficient than (3.5) since it has only an $O(n)$ complexity.

(i.e. ϕ in equation (3.6)).

As far as we are concerned, intervertebral shear and normal forces will be obtained via an inverse dynamics computation performed within a process of time integration of the direct dynamics, thanks to the use of driven joints, as explained in the next paragraph.

Coupled direct and inverse dynamics For specific applications, some of the generalized joint coordinates q^i are no longer considered as variables because their motion is either fixed or forced to obey a prescribed law function of time t . The corresponding joint is then qualified as *driven* and characterized by a driven coordinate denoted q_d in our formalism. One then writes:

$$q_d(t) = f(t) ; \dot{q}_d(t) = \dot{f}(t) ; \ddot{q}_d(t) = \ddot{f}(t) \quad (3.7)$$

where $f(t)$ is the prescribed law¹.

To take driven joints into account, the set of generalized coordinates q in partitioned in two categories: the independent coordinates q_i and the driven coordinates q_d :

$$q = \begin{pmatrix} q_i \\ q_d \end{pmatrix} \quad (3.8)$$

For sake of conciseness, let us define $F \triangleq c - Q$ in the direct dynamical model (3.5) which can be rewritten as follows:

$$M(q)\ddot{q} + F(q, \dot{q}) = 0 \quad (3.9)$$

Then, once partitioned according to (3.8) and using the Lagrange multipliers technique, the previous system of equations becomes:

$$\begin{pmatrix} M_{ii} & M_{id} \\ M_{di} & M_{dd} \end{pmatrix} \begin{pmatrix} \ddot{q}_i \\ \ddot{q}_d \end{pmatrix} + \begin{pmatrix} F_i \\ F_d \end{pmatrix} = \begin{pmatrix} 0 \\ \lambda_d \end{pmatrix} \quad (3.10)$$

in which λ_d denote the Lagrange multipliers associated with the driven coordinates q_d , or more exactly with the constraints equations (3.7) that involve the driven coordinate solely. From a physical point of view, λ_d^i exactly represents the force [N] (resp. the torque [Nm]) that the i^{th} driven joint must produce during the MBS motion (e.g. the intervertebral efforts in a kinematically driven spine during walking.).

Direct kinematics This process, well-known to roboticists, aims at computing the absolute kinematics of any body in terms of the joint relative coordinates.

¹ $f(t) = cst$ in case of a locked/fixed joint.

Practically, by locating a virtual sensor S on a given body of a MBS, it allows obtaining the expression of the forward kinematics of any sub-chain of the MBS (i.e. the position $\mathbf{p}(q)$, the orientation $A(q)$, the Jacobian matrix $J(q)$, the linear/angular velocities $\mathbf{v}(q, \dot{q})$, $\boldsymbol{\omega}(q, \dot{q})$ and the linear/angular accelerations $\mathbf{p}(q, \dot{q}, \ddot{q})$, $\boldsymbol{\omega}(q, \dot{q}, \ddot{q})$) ending with point S and with the body on which it is placed. Direct kinematics will be particularly useful in this work to carry out the geometrical and kinematic identifications of the spine (Sect. 3.4 and 3.5).

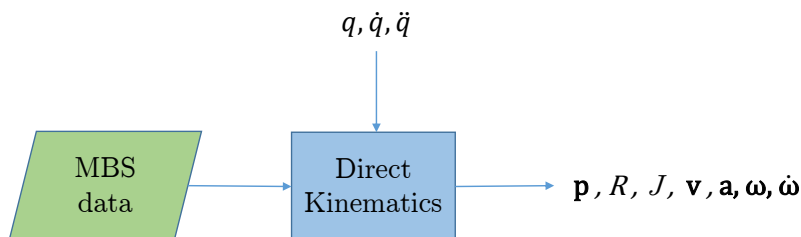


Fig. 3.4: Direct kinematics in ROBOTRAN

3.3 Biomechanical multibody model

In this work, the reference inertial frame is oriented according to the anatomical planes of the human body in the standing up position (Fig. 3.5). The planes formed by $\{\hat{\mathbf{I}}_1, \hat{\mathbf{I}}_2\}$, $\{\hat{\mathbf{I}}_2, \hat{\mathbf{I}}_3\}$ and $\{\hat{\mathbf{I}}_1, \hat{\mathbf{I}}_3\}$ denote the transverse, coronal and sagittal planes respectively. For the scoliosis analysis, the proposed multibody model contains 18 bodies: the 17 vertebrae and the pelvis. The latter plays an important role in our study, being the reference floating body for all kinematic and dynamical analyzes: it possesses a full mobility, i.e. 6 d.o.f. with respect to the inertial frame.

To take large scoliotic deformations into account in a proper way, we suggest to represent the intervertebral kinematics by 9 relative coordinates q . The reason is a practical one: in addition to the 6 d.o.f. representing the relative deformations of each intervertebral disc (in which the viscoelastic strains take place), we have added 3 fixed rotations — via 3 driven coordinates q_d — to capture the initial deformation of the spine in the standing up position (these angles are somewhat similar to the neutral position of torsional springs in mechanical devices). The value of these constant angles will be obtained by an optimization process (Sect.3.4) intended to fit at best the multibody reference configuration to the X-ray measurements described in Sect.2.3.

As explained in Sect. 3.2.1, the body-fixed frames $[\hat{\mathbf{J}}^n]$ and $[\hat{\mathbf{P}}]$ are rigidly attached to

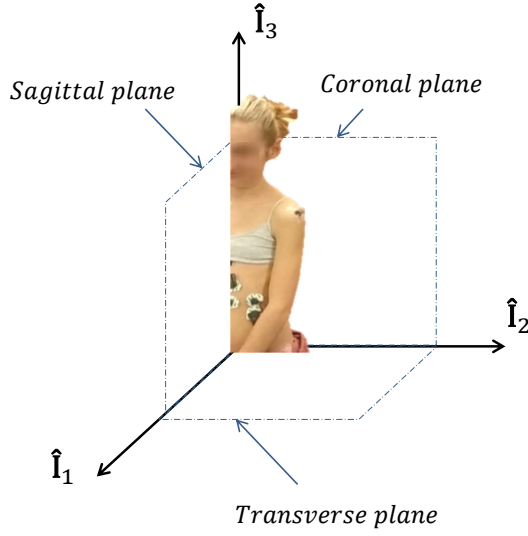


Fig. 3.5: The human body is characterized by three reference planes

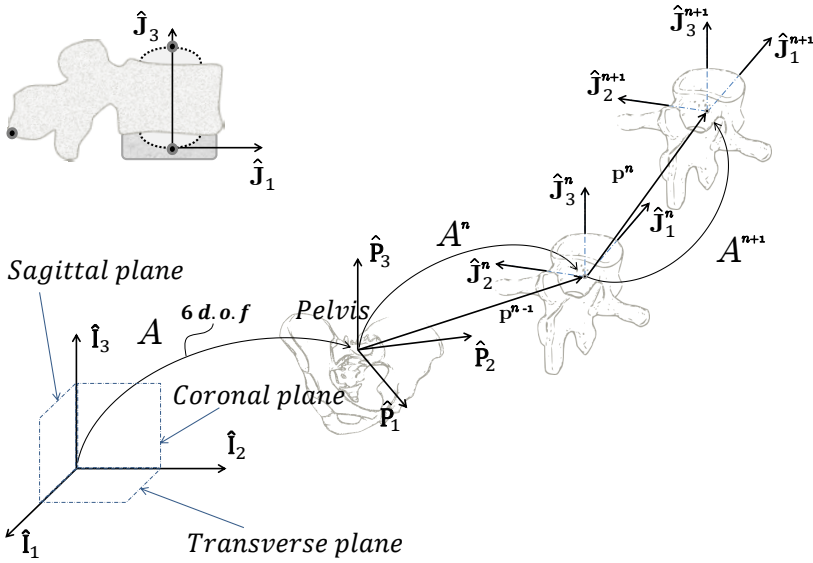


Fig. 3.6: Spine multibody model - main geometrical and kinematic data

each vertebra n and to the floating pelvis (Fig. 3.6). They are located and oriented according to the following rules:

- Their origin is located at the lower intervertebral disc center point¹.
- $\hat{\mathbf{J}}_3^n$ points towards the upper intervertebral disc center point (between vertebrae n and $n + 1$), the one which is virtually fixed to vertebra $n + 1$.
- $\hat{\mathbf{J}}_1^n$ belongs to the plane defined by the 3 characteristic points: the spinous process and the two disc centers.
- $\hat{\mathbf{J}}_2^n = \hat{\mathbf{J}}_3^n \times \hat{\mathbf{J}}_1^n$, according to the requirement of orthonormal frames.

Starting from the inertial frame, the first body of the system is the floating pelvis. As for the vertebrae, an orthonormal frame, denoted $\{\hat{\mathbf{P}}\}$, is attached to it and located, similarly to the vertebrae, between the $L5$ and the $S1$ sacrum vertebrae.

To understand why legs are intentionally not included in the model, it is important to remember the difference between the direct and the inverse dynamics approaches with respect to the pelvis motion during walking (Sect.3.2.5). In one hand, direct dynamics requires the knowledge of the forces arising from the ground and transmitted through the legs, to predict the motion of the pelvis and of the upper body.

On the other hand, the inverse dynamics approach is based upon the *imposed* pelvis kinematics regardless its source motion. In our protocol, the kinematics of the pelvis is experimentally identified during gait (Sect. 2.4) and is imposed to the multibody model via six driven joints (eqs. (3.7)) to which correspond the three constraint forces and the three constraint torques $\lambda_d^{1 \rightarrow 6}$, equivalent to the resultant of those transmitted by the legs to the pelvis (Fig. 3.7). Finally, as regards the motion of the upper body (i.e. spine, thorax, head and arms), it can be classically predicted via the direct dynamics, for the remaining d.o.f. of the MBS.

¹As mentioned earlier, this point, while being physically inside the visco-elastic disc, is assumed to be virtually fixed relative to the upper vertebra and thus, geometrically speaking, belongs to this body.

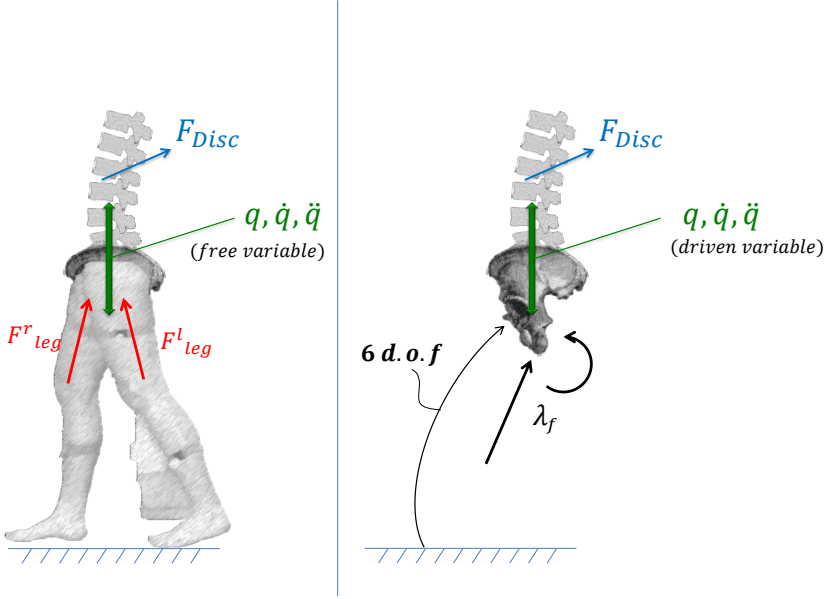


Fig. 3.7: Left: direct dynamics approach. The pelvis motion is predicted by considering the leg forces as model input. Right: inverse dynamics approach. The knowledge of the pelvis kinematics allows considering it as a constrained motion to which Lagrange multipliers (i.e. constrained forces and torques) are associated.

The proposed inverse dynamics approach (coupled with the direct dynamics of the MBS) greatly facilitates the modeling process by alleviating the model (absence of legs) but also, and above all, the experimental protocol: indeed it is not necessary to measure the foot-ground contact forces anymore. However, in turn, the pelvis kinematics (position, linear/angular velocities and accelerations) must be known with sufficient precision: the experimental setup and its validation was the subject of Sect. 2.4.2.

3.4 Geometrical identification

The identification of the 3D shape of the spine for the purpose of building the multi-body model can be carried out via several approaches (Tab. 2.1). This section focuses on the geometrical identification of the spine from bi-planar X-rays, as presented in Chapter 2.

Starting from data measured via the user interface Fig.2.12, the first step consists in calculating the distance d^n between each successive pair of disc centers.

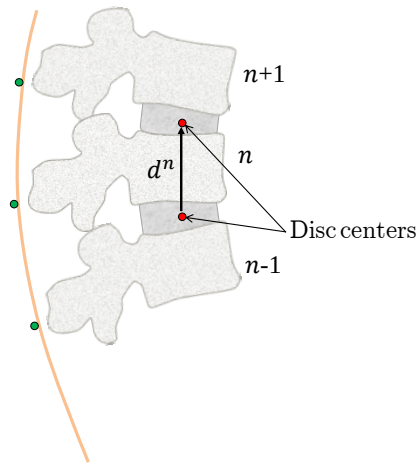


Fig. 3.8: The distance d^n between two successive disc centers can be measured from the X-ray images Sect. 2.3

Then, knowing these intervertebral distances, a numerical algorithm is implemented to minimize two cost functions.

The first one, f_{cost}^b , relates to the spine bending and reads:

$$f_{cost}^b = \sum_{s=1}^{n_{disc}} \|\mathbf{X}_{mod,s}(q) - \mathbf{X}_{exp,s}\|^2 \quad (3.11)$$

where:

- q are the $3 \times n_{disc}$ driven coordinates representing the intervertebral rotations associated with scoliotic bending and torsion; they are the optimization unknowns.
- $\mathbf{X}_{mod,s}(q)$ is the absolute position vector of the intervertebral disc center, calculated by the multibody model via the direct kinematics of sensors Sect.2.4.
- $\mathbf{X}_{exp,s}$ is the absolute position vector of the intervertebral disc measured from radiography (Sect. 2.12).

In the previous cost function, the spinal torsion — which is an important characteristic of scoliosis — is not yet taken into account. The associated intervertebral torsion angles (which are part of the optimization unknowns q) are computed by projecting on the horizontal plane $\{\hat{\mathbf{I}}_1, \hat{\mathbf{I}}_2\}$ the vector $\overrightarrow{VV'}$ joining the center of the upper

disc to the spinous process (Fig. 3.9). The orientation of $\overrightarrow{VV'}$ in this plane is clearly representative of the vertebral rotation θ , indeed:

$$\tan \theta(q) = \frac{\overrightarrow{VV'} \cdot \hat{\mathbf{I}}_2}{\overrightarrow{VV'} \cdot \hat{\mathbf{I}}_1} \quad (3.12)$$

The resulting set of angles θ_s ($s = 1, \dots, n_{disc}$), respectively assessed from the experimental data (θ_{exp}) and computed via the multibody model (θ_{mod}), can be compared and their difference minimized to evaluate the torsion along the spine, according to the following "torsion" cost function f_{cost}^t :

$$f_{cost}^t = \sum_{s=1}^{n_{disc}} (\tan(\theta_{mod,s}(q)) - \tan(\theta_{exp,s}))^2 \quad (3.13)$$

where:

- $\theta_{mod,s}$ is the angle between the unit vector $\hat{\mathbf{I}}_1$ and the projection of the vector $\overrightarrow{VV'}$ onto the $\{\hat{\mathbf{I}}_1, \hat{\mathbf{I}}_2\}$ plane, calculated by the multibody model.
- $\theta_{exp,s}$ is the same angle but measured from radiography

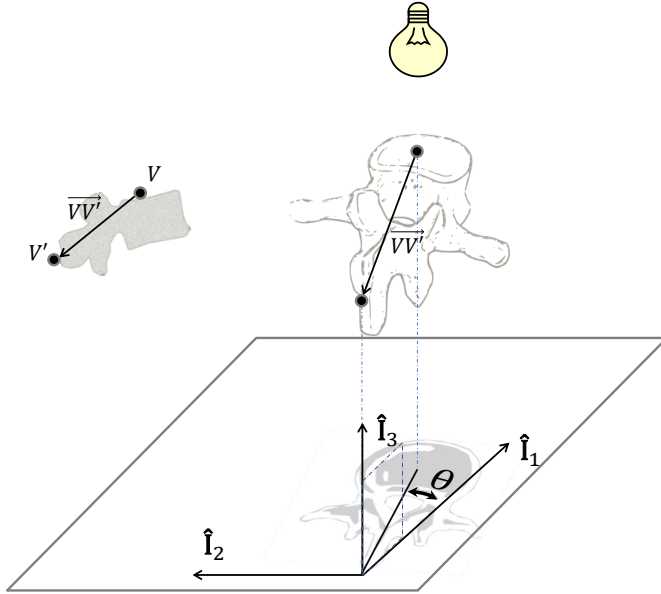


Fig. 3.9: Spinal torsion: projection on the horizontal plane of a characteristic geometrical vector $\overrightarrow{VV'}$

In case of severe scoliotic deformities¹, the inertial frame $\{\hat{\mathbf{I}}\}$ could be inappropriate to perform this projection. According to Fig. 3.10, to compute the torsion angle between vertebrae, we suggest to introduce a temporary floating frame $\{\hat{\mathbf{X}}\}$, and to project the $\overrightarrow{VV'}$ vector associated to vertebra n onto the plane $\{\hat{\mathbf{X}}_1, \hat{\mathbf{X}}_2\}$,

$$\tan \theta' (q) = \frac{\overrightarrow{VV'} \cdot \hat{\mathbf{X}}_2}{\overrightarrow{VV'} \cdot \hat{\mathbf{X}}_1} \quad (3.14)$$

The torsion angle is denoted θ' to indicate that it refers to the floating frame $\{\hat{\mathbf{X}}\}$ instead of $\{\hat{\mathbf{I}}\}$.

This new frame is defined as follows (Fig. 3.10): let us first introduce a vector $\mathbf{u}$ which is representative of the direction of the local spine centroidal axis: $\mathbf{u}$ joins the centers of the upper disc of vertebra n and of vertebra $n - i$ ($i = 2$ is a reasonable choice).

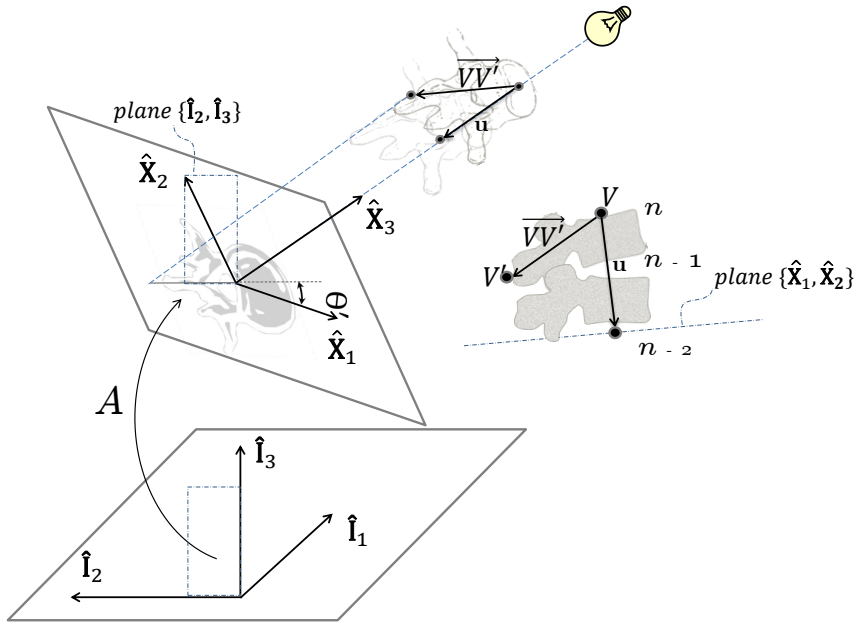


Fig. 3.10: Spinal torsion for severe scoliosis: projection on a floating plane

To position the $\{\hat{\mathbf{X}}_1, \hat{\mathbf{X}}_2\}$ plane in a normal direction with respect to $\mathbf{u}$, $\hat{\mathbf{X}}_3$ reads:

$$\hat{\mathbf{X}}_3 = -\frac{\mathbf{u}}{\|\mathbf{u}\|} \quad (3.15)$$

¹Cobb angle can reach up to 90 degrees.

Then, based on physiological considerations, we suggest orienting the $\{\hat{\mathbf{X}}\}$ frame (around $\hat{\mathbf{X}}_3$) in such a way that $\hat{\mathbf{X}}_2$ belongs to the $\{\hat{\mathbf{I}}_2, \hat{\mathbf{I}}_3\}$ plane (Fig. 3.10):

$$\hat{\mathbf{X}}_2 \cdot \hat{\mathbf{I}}_1 = 0 \quad (3.16)$$

which could lead to two solutions : $\hat{\mathbf{X}}_2$ pointing to the left or to the right. To eliminate the second one, the unilateral constraint $\hat{\mathbf{X}}_2 \cdot \hat{\mathbf{I}}_2 > 0$ must be satisfied for obvious physiological reasons.

For the sake of completeness, let us note that the constraint (3.16) would lead to an infinite number of solutions in case of a spine centroidal axis (i.e. vector $\mathbf{u}$), perfectly aligned with the gait longitudinal direction $\hat{\mathbf{I}}_1$ ($\hat{\mathbf{X}}_3 = \hat{\mathbf{I}}_1$). Fortunately, this configuration cannot be encountered in practice during gait.

From equations (3.15) and (3.16), that provide the components of frame vectors $\hat{\mathbf{X}}_2$ and $\hat{\mathbf{X}}_3$ in the inertial frame, the rotation matrix A between the inertial frame $\{\hat{\mathbf{I}}\}$ and the floating frame $\{\hat{\mathbf{X}}\}$,

$$[\hat{\mathbf{X}}] = A(q)[\hat{\mathbf{I}}],$$

can be computed easily. The $\{\hat{\mathbf{X}}\}$ floating frame is thus unambiguously localized for each vertebra.

The cost function (3.13) can be rewritten in terms of the new projection angles θ' :

$$f_{cost}^{t'} = \sum_{s=1}^{n_{disc}} (\tan(\theta'_{mod,s}(q)) - \tan(\theta'_{exp,s}))^2 \quad (3.17)$$

Bending and torsion being function of the three intervertebral rotation angles (i.e. the three driven generalized coordinates q), the cost functions (3.11) and (3.17) must be gathered and processed together:

$$f_{cost}(q) = f_{cost}^b(q) + f_{cost}^{t'}(q) \quad (3.18)$$

The minimization¹ of (3.18) leads to the static configuration² that fits at best with the radiographic data. For this *reference* configuration, the generalized coordinates will be denoted q_{ref} .

¹This optimization process is based on the Levenberg-Marquardt algorithm [62]

²This optimization process leads to a deterministic solution $f_{cost}(q) = 0$.

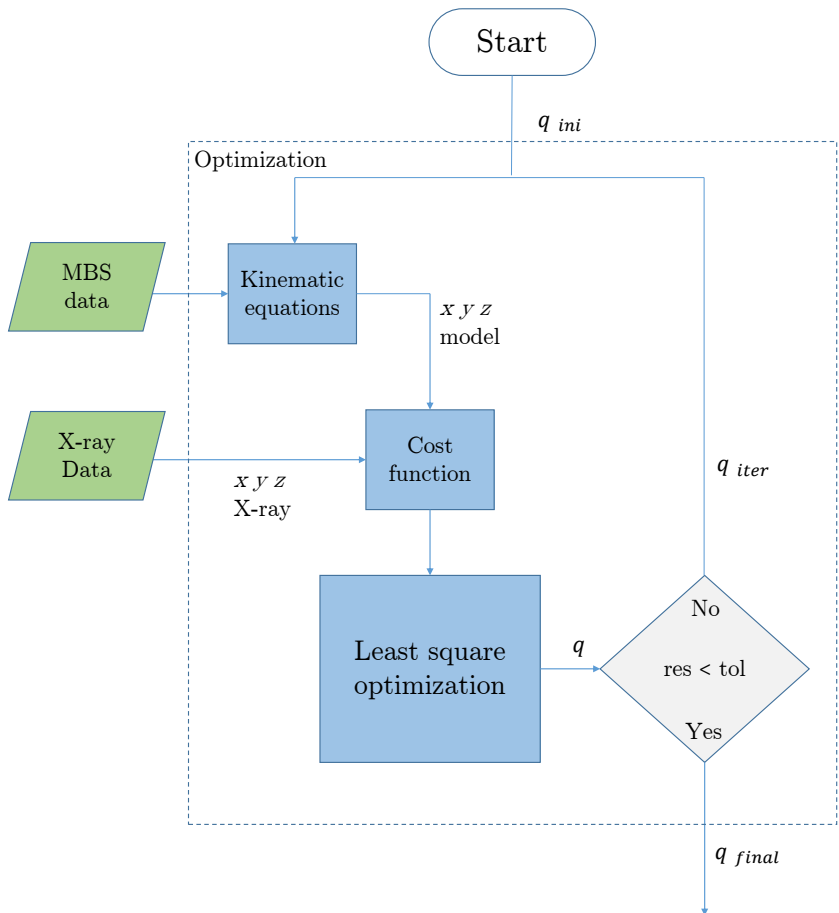


Fig. 3.11: Least square optimization iteratively finds the reference configuration of the scoliotic spine that best fits with the X-ray data.

At this stage of the process it is worth noting that, given the radiographic data, the previous minimization algorithm, once the convergence is acquired (Fig. 3.11), is followed by a process that automatically generates a complete morphological ROBOTRAN multibody model (Fig. 3.12) of the spine reference configuration, ready for any subsequent kinematic and/or dynamical analyzes.

3.5 Kinematic identification

Due to experimental limitations during gait, it will be impossible to obtain the same amount of information as from radiography, leading to an under-determined problem.

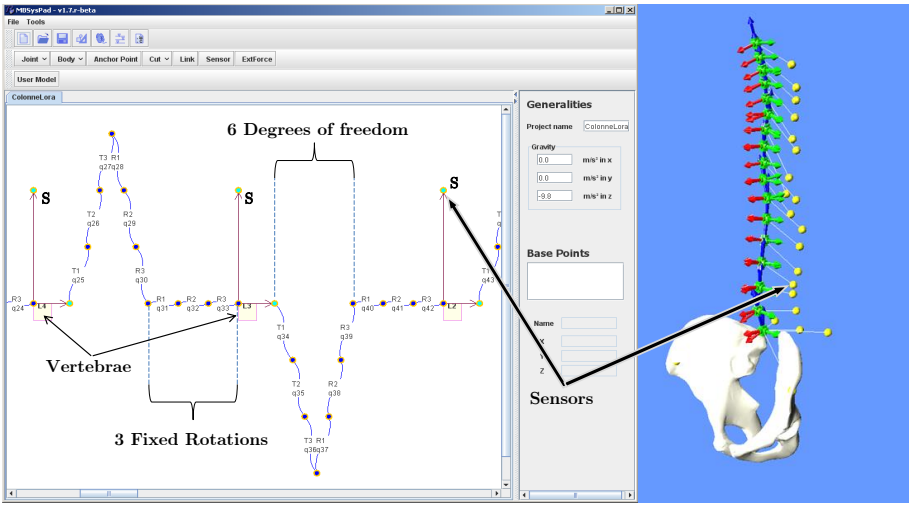


Fig. 3.12: The patient multibody model is automatically generated from X-rays data.

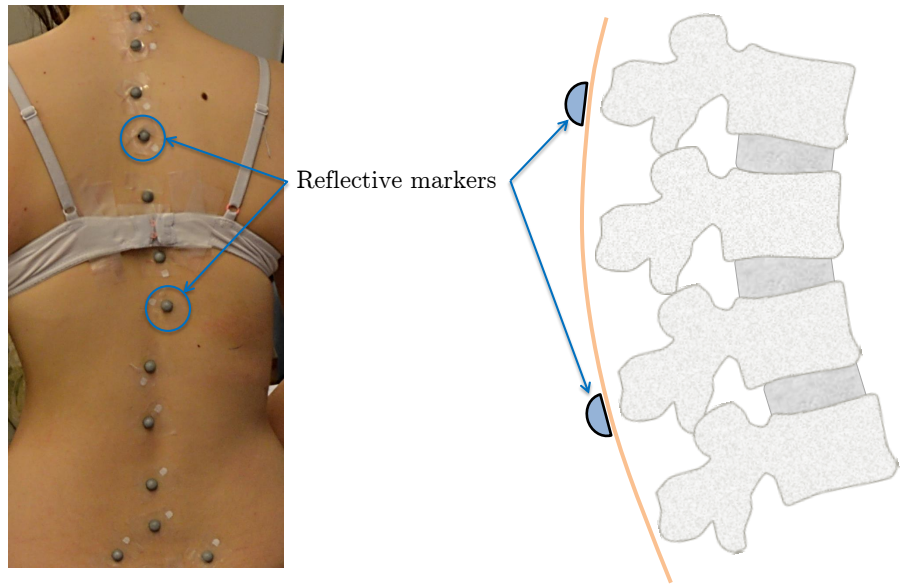


Fig. 3.13: During the experimental phase, the physiotherapist sticks a reflective marker on one spinous process out of two vertebrae.

Indeed in the current protocol, we are able to stick a reflective marker on one spinous process out of two vertebrae (Fig. 3.13). As explained in Sect. 2.4, those markers are

positioned at the same place as the metallic markers used to identify the spinous process from the bi-planar X-rays (Fig. 3.14).

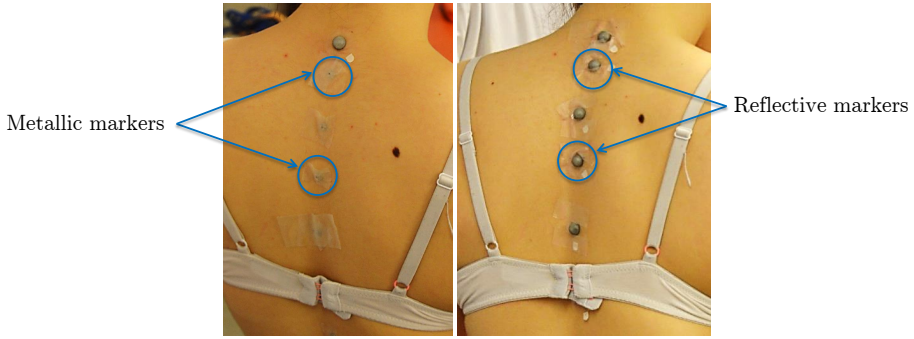


Fig. 3.14: Left: metallic markers; Right: reflective markers, stuck on the patient.

Due to the lack of information, the optimization process based on the previous cost functions (3.11) and (3.17) is thus not deterministic anymore. For this reason, in order to calculate the spinal deformity of the spine during gait, we suggest reconstructing its geometry based on the experimental interpolation results proposed by Rozumalski et al. [83] and our lacking kinematic information (i.e. $n_{markers}$ points) obtained from the reflective markers. For this kinematic reconstruction, we also exploit an above-mentioned hypothesis, that the spine motion relative to the *current* configuration of the pelvis does not deviate too much with respect to the reference configuration defined in Sect. 3.5.

For bending reconstruction, the global cost function of the optimization process is now composed of two weighted terms:

$$f_{cost} = w_1 * f_{cost1} + w_2 * f_{cost2} \quad (3.19)$$

The first cost function computes the distance between the virtual sensors located on the spinous processes in the multibody model $\mathbf{X}_{mod}$, and the real reflective markers used for the gait analysis $\mathbf{X}_{exp}$,

$$f_{cost1} = \sum_{s=1}^{n_{markers}} \|\mathbf{X}_{mod,s}(q_{ref} + \Delta q) - \mathbf{X}_{exp,s}\|^2 \quad (3.20)$$

where q_{ref} refers to the — known — standing up intervertebral bending and torsion angles and Δq (unknowns) stands for their deviation during walking.

The second cost function consists in minimizing the difference between the bending angles in the multibody model, with the angles predicted by [83]. It is important to

notice that, due to the high level of deformation of scoliotic patients, our bending angle variations Δq_{bend} are defined according to the multibody joints orientation and not to the physiological reference planes (Fig. 3.5) used in [83]. For that reason, to compute and compare them coherently, we project the virtual spine frame unit vectors onto those reference planes (Fig. 3.15). The cost function then reads:

$$f_{cost2} = \sum_{i=1}^n \|\Delta\varphi_{mod}^i(\Delta q) - \Delta\varphi_{exp}^i\|^2 \quad (3.21)$$

where:

- $\Delta\varphi$ denotes the (instantaneous) variation of the intervertebral rotations – in the sagittal plane – with respect to the static configuration,
- $\Delta\varphi_{mod}(\Delta q)^1$ refers to these angles computed via the multibody model,
- $\Delta\varphi_{exp}$ refers to the gait angles predicted by [83].

In sum, minimizing f_{cost} in (3.19) allows finding values for Δq that best fit with the spinous processes location measured experimentally (f_{cost1}) and with the angular variations predicted by [83] (f_{cost2}). The weights w_1 and w_2 can be adjusted to enforce one criteria with respect to the other². The global minimization process is summarized in fig. 3.16.

3.5.1 Evaluation: “one ball out” method

Since it is experimentally complicated to place a marker on each vertebra, in order to estimate the impact of the lack of kinematic information, we suggest to work in the opposite direction. In other words, we have fixed one reflective marker each 2 vertebrae from L4 to C7, then 2 different optimizations have been carried out: the first one includes all the reflective markers and the second one uses 1 reflective marker for each 4 vertebrae only i.e. neglecting the information of one every two markers. Finally, both models are compared to analyze the sensitivity of the kinematics to this lack of information. (Fig. 3.17)

¹ Δq_{mod} represents the variables q_{37} , q_{38} , q_{39} in Fig. 3.12-left.

²To privilege our experimental data, w_1 has been chosen 6 times larger than w_2 .

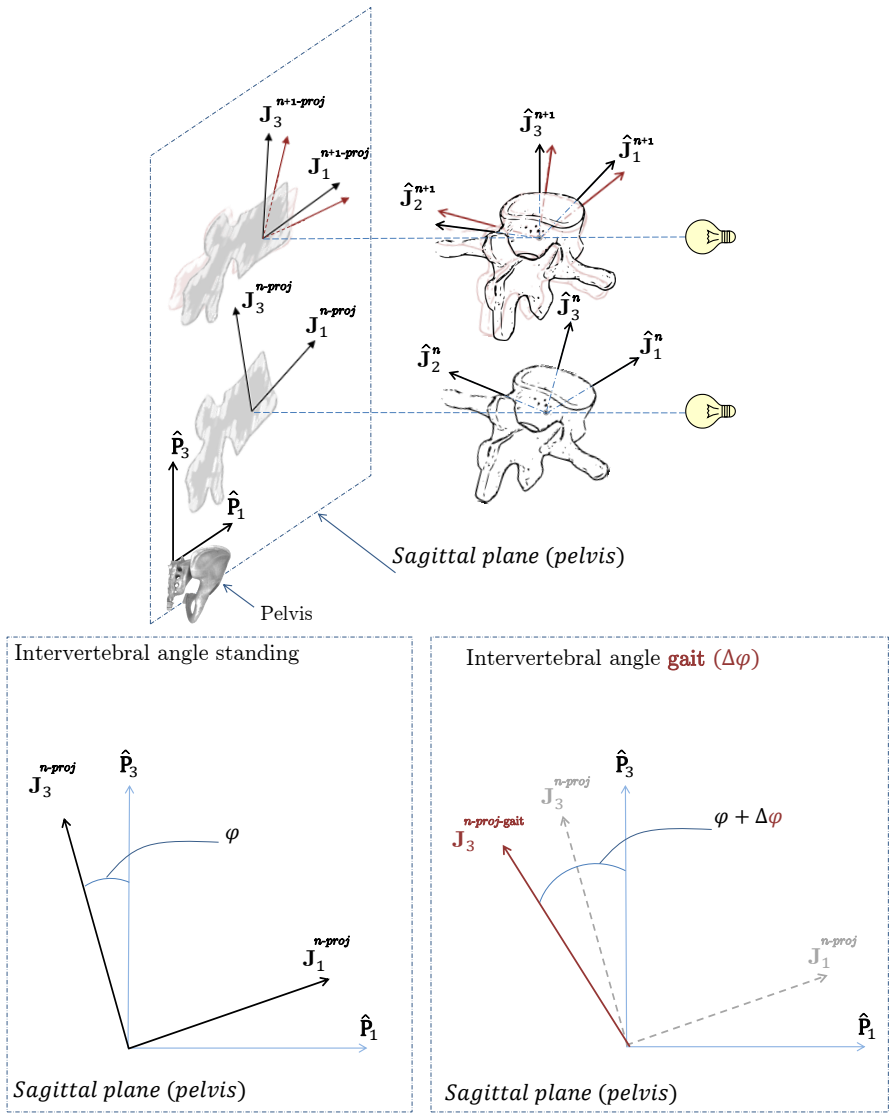


Fig. 3.15: Projection of the vertebral frame $[\hat{\mathbf{J}}^n]$ on the pelvis sagittal planes

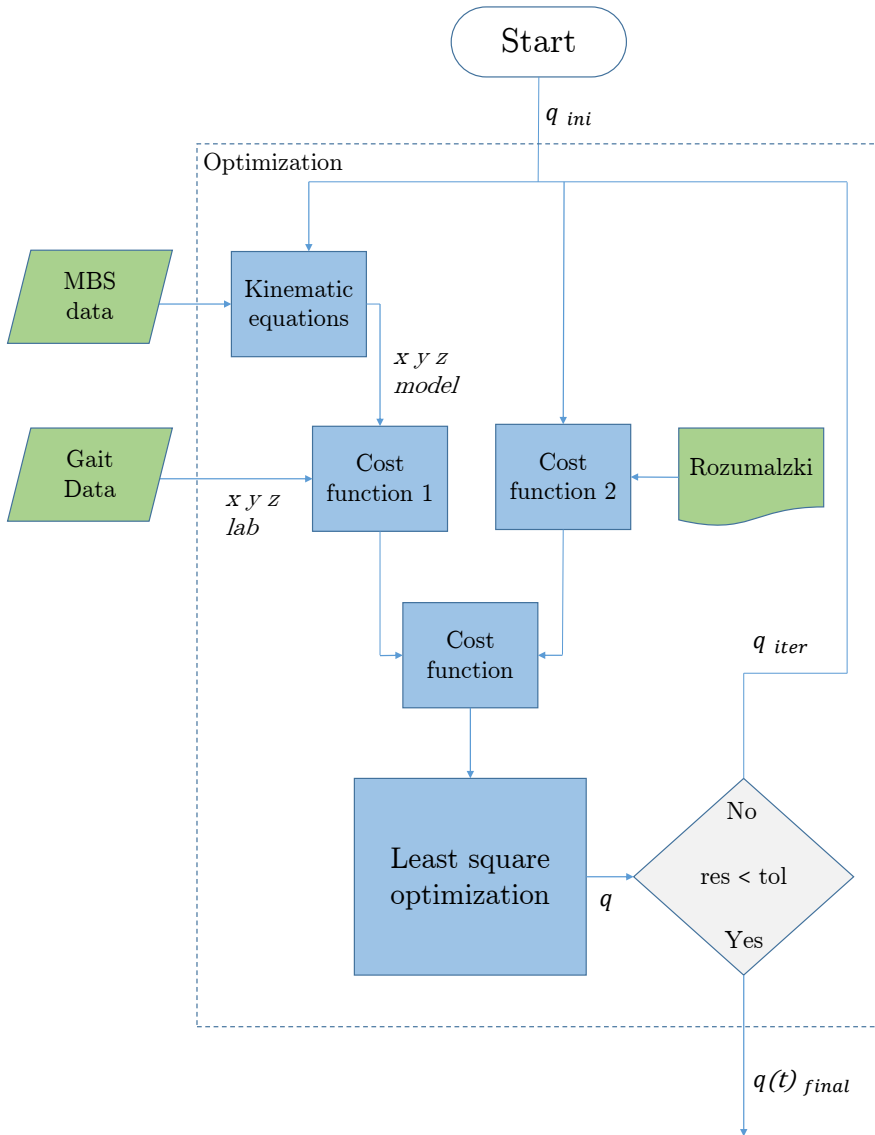


Fig. 3.16: Least square optimization iteratively finds the kinematic configuration of the scoliotic spine that best fits with the gait analysis at each time step t .

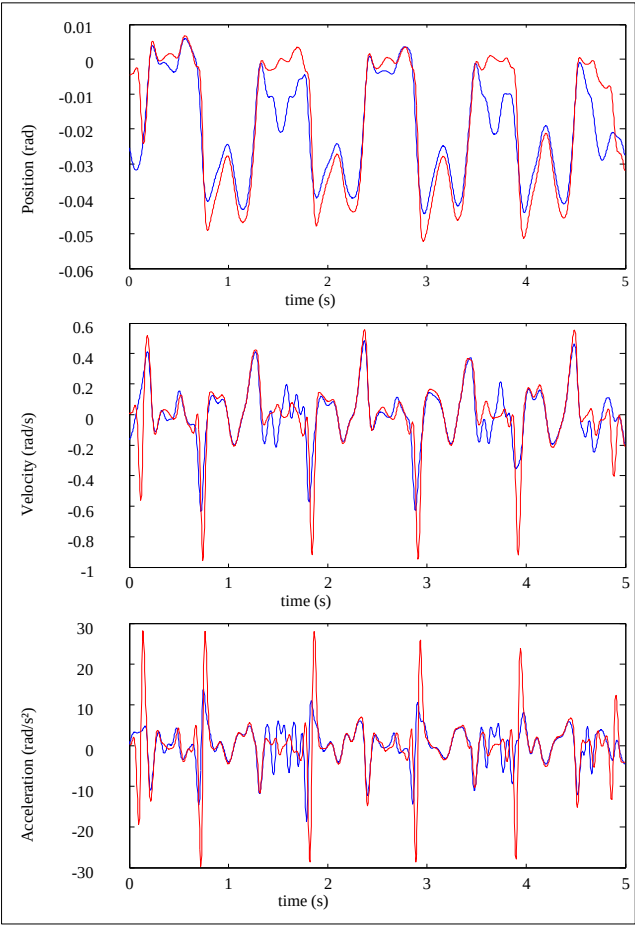


Fig. 3.17: Correspondence between the two multibody models: a model using all markers (blue) and a model using half of the markers (red).

3.6 Patient Physiology

This section focuses on the impact of the patient physiology on the intervertebral efforts computation during gait. Two aspects are discussed: the first relates to the mass distribution in the multibody model, taking special care to the motion of the organs inside the abdominal cavity. The second concerns the stiffness of the intervertebral discs.

3.6.1 Mass distribution

To study the effect of the mass distribution of the human body, via a multibody model [23], we have subdivided it into distinct parts: head, thorax, upper limbs, abdominal cavity, spine, pelvis and lower limbs (deliberately absent from the model Sect. 3.5. During moderate gait, the head, the thorax and the upper limbs could be reasonably considered as a sequence of polyarticulated rigid bodies. However, this is not the case for the abdominal cavity, where the organs are wobbling masses and not directly connected to the spine.

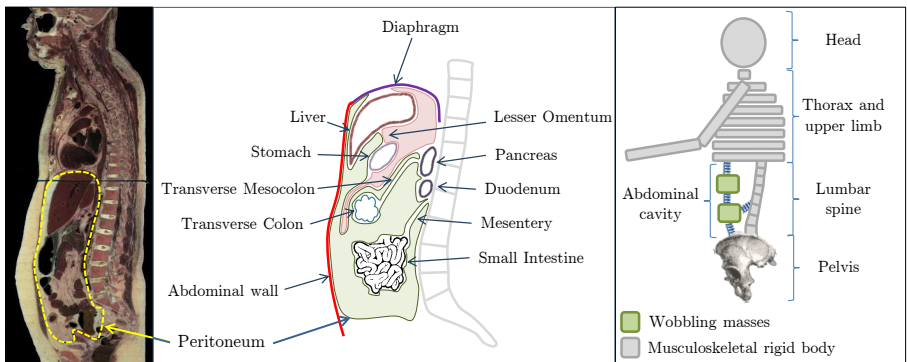


Fig. 3.18: The development of the multibody model of the abdominal cavity (right) is based on the patient physiology (left, middle).

In [47], an equivalent model¹ that reproduces the resonance modes of the visceral organs is proposed and could have been used in the present work. However, this model, originally designated for comfort analysis is not suitable to analyze the impact of the wobbling organs on the intervertebral efforts.

¹From University of Michigan: <http://vhp.med.umich.edu/browsers/female.html>

¹not based on the real mass distribution of the organs

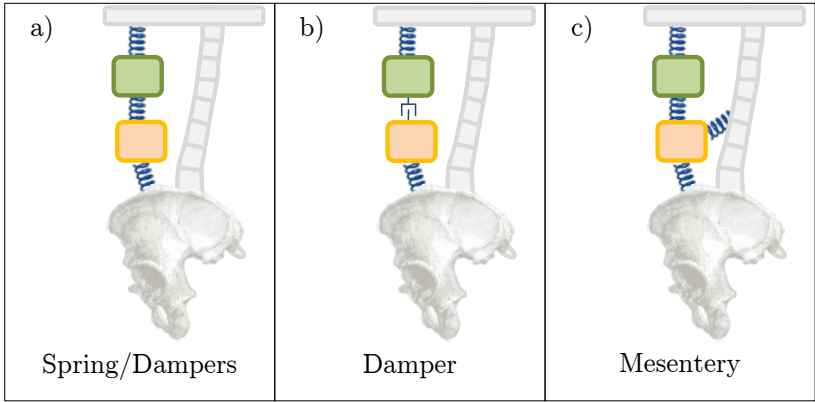


Fig. 3.19: Multibody model variants

Taking into account of visceral organs in our study is *a priori* a delicate task that requires judgment and common sense from both physiotherapist and engineers. As we can observe in Fig. 3.18-middle, it is possible to identify three kinds of attachment in the abdominal cavity:

1. "Strong" anchor to the diaphragm and/or to the spine; this is the case for the liver, the pancreas and the duodenum;
2. "Seak" connection between bodies maintained by the peritoneum; this is the case for lesser omentum, which connects stomach and liver, the transverse mesocolon, which connects the pancreas and the transverse colon, and for the mesentery, which connects the small intestine to the lumbar spine;
3. Finally, the direct surface contact between organs and the abdominal wall (Fig 3.18-left).

Although it is difficult to accurately identify the location and the mechanical properties for each of those attachments, a macro multibody model is proposed (Fig 3.18-right) in order to identify the impact of the abdominal wobbling motion on the spine efforts during gait. The proposed variants of the model are the result of fruitful exchanges with physiotherapists and anatomy professors of UCL, to avoid the development of unduly complicated models (in terms of size and parameters to be identified) regarding our project issue.

Regarding the objectives of the latter, the visceral wobbling masses are subdivided in two: the upper mass represents the liver and the stomach, and the lower mass refers to the intestines. Then, in order to analyze the sensitivity of the attachment types, three variants have been proposed after interdisciplinary exchanges and a few iterations! In the variant a., three vertical parallel springs/dampers are used to connect

two equivalent masses between the thorax and the pelvis (Fig 3.19-a). For the variant b., the same, but without the spring – just a damper – between the two wobbling masses (Fig 3.19-b). In variant c., we have included an additional spring/damper (Fig 3.19-c) to represent the mesentery connection (see Fig 3.18-middle). Finally, a reference model in which the motion of all visceral masses are neglected is also considered for this study.

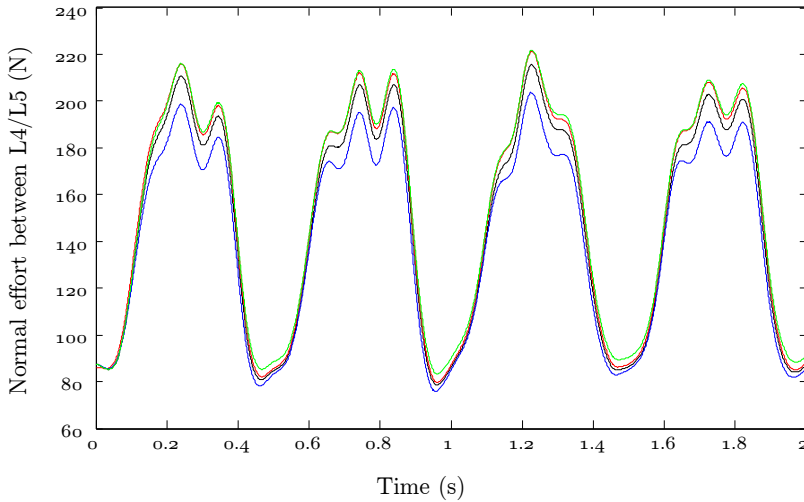


Fig. 3.20: Vertical force between L4/L5 for different model variants: for variant a (black), b (green), c (red) and the reference model (blue)

Model b. appears to be the best choice from a physiological point of view. Moreover, results reveal a rather low sensitivity of the vertical intervertebral efforts with respect to the organs dynamics and their modeling parameters; in Fig. 3.20, a maximum difference of around 10% is observed for the vertical intervertebral force between L4 and L5 between model variant b. and the reference model.

3.6.2 Intervertebral discs stiffness

The analysis of the impact of the intervertebral disc characteristics has been studied via two levels of stiffness conferred to the intervertebral discs in the multibody model: a physiological-based stiffness disc [94] and a rigid disc.

Considering a sinusoidal displacement of the pelvis as a representative input of the

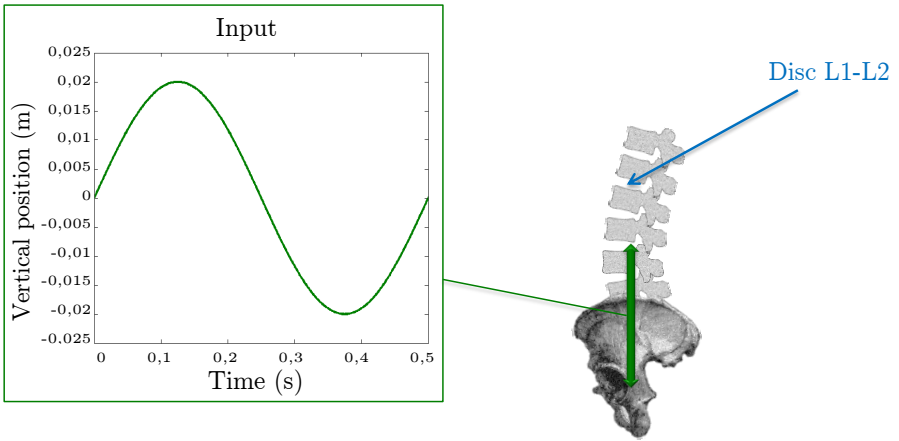


Fig. 3.21: A sinusoidal excitation of the pelvis allows to compare intervertebral forces (i.e. in L1-L2) for two different disc stiffnesses.

gait, the maximum difference observed for the normal force between the L1 and L2 vertebrae was about 20N between the two situations, the first one using an infinite intervertebral stiffness (rigid disc), and the second one a physiological stiffness (i.e. of a real disc) (Fig. 3.22). In the worst case : this difference represents less than 10% of the normal force magnitude.

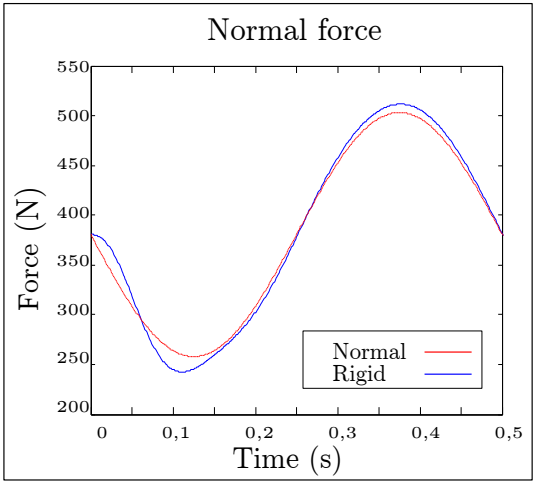


Fig. 3.22: Variation of normal force acting between L1 and L2 vertebrae for two levels of stiffness of the intervertebral L1-L2 disc. The blue curve represents the normal force for a rigid disc. The red curve represents the normal force for a "normal" disc.

These observations suggest that the intervertebral normal forces acting between ver-

tebrae during the simulated motion are slightly influenced by the level of stiffness given to the intervertebral discs in the multibody model of the spine. This suggests that modeling intervertebral discs with rigid elements will be sufficiently reliable to compute the intervertebral efforts acting between vertebrae during normal gait.

3.7 Muscle modeling

One of the most challenging issues of the present work is the assessment of the intervertebral efforts. Above all, this is due to muscle redundancy (i.e. the presence of more muscles than d.o.f.) and to experimental limitations and difficulties to measure the activation of each of them individually.

Referring to the literature which is abundant on the subject, estimating muscular forces — among an infinite number of solutions — could be addressed using several approaches such as: force minimization, energy expenditure, Electromyographic data (EMG), etc. or a combination of these. On one hand, it would be quite simple to compute the muscular force distribution using an energy-based criteria: unfortunately this approach is not always realistic, due to the phenomenon of muscle co-contraction, and is probably not suited in case of a walking scoliotic patient. On the other hand, it is very difficult and often impossible to obtain the information of every muscle using EMG's, although it appears that this lack of information could be partially circumvented via the muscle synergy principle [51, 87].

In this work the computation of the muscular forces will be computed in two steps:

1. First, using a reverse engineering approach to compute the minimal¹ muscular forces of well-targeted muscular groups for a —known— spinal kinematics.
2. Then, adapting the muscular forces to the data measured from EMG² during gait as explained in Sect. 2.5.

Muscle kinematics and forces Since the spine kinematics is already identified (Sect. 3.5), we might have taken the muscular groups into account via a closed loop of bodies composed of the pelvis, the spine and a muscle (or a group of muscle) in order to consider muscular forces as constraint forces (Fig.3.23 top-left) and to compute them via an inverse dynamics model (Sect.3.2.5). However, this approach is unnecessarily complicated since, in case of the human body, muscles only act in tension using the agonistic/antagonistic principle; from the modeling point of view, this

¹More exactly with no co-contraction at all

²EMG data is calibrated using the David back protocol defined in Sect. 2.5

would lead to loops of bodies with unilateral constraints (3.23 bottom-left) which are really cumbersome to treat.

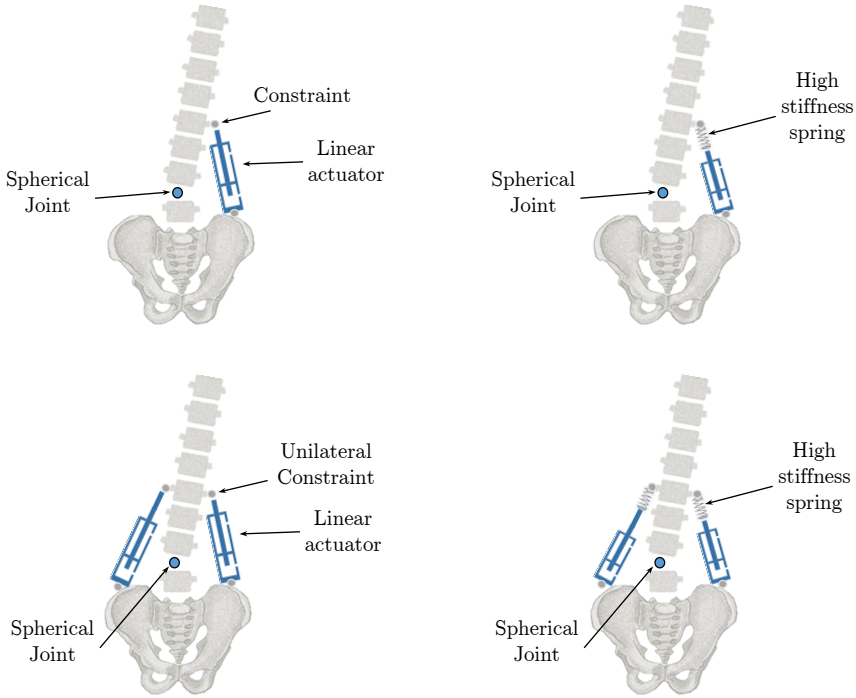


Fig. 3.23: Muscular force model: unilateral link forces.

In the present work, instead of considering muscles as constraint forces, we suggest to place them in series with a *virtual* high stiffness/damping element (Fig.3.23 top-right). Doing so, from a multibody modeling point of view, any loop of bodies is simply closed by a unilateral link force rather than by a kinematic constraint (Fig.3.23 bottom-right). Muscles kinematics is then taken into account in the model by controlling the neutral length of those springs according to the link forces anchor points kinematics that results from the gait optimization (Sect. 3.5).

In this respect, two consecutive simulations will be carried out:

- An inverse kinematics simulation, for which the intervertebral disc motions are imposed according to the gait kinematics: this allows to accurately compute the distance between all muscles connection points during gait.
- A direct dynamics, for which muscle length are imposed according to the previous simulation. Muscles forces are retrieved from the virtual spring/damper series elements. To draw a parallel with engineering, muscles are considered as linear actuators whose length is controlled according to the system motion,

	K = 100K [N/m]	K = 1M [N/m]	K = 10M [N/m]	K = 50M [N/m]	K = 100M [N/m]
Erector spinae Left	14	10	8,6	8,4	8,5
Erector spinae Right	0,38	20	29	30	30
Rectus abdominis Left	20	20	20	20	20
Rectus abdominis Right	20	20	20	20	20
Quadratus lumborum L5 Left	0	0	0	0	0
Quadratus lumborum L5 Right	7,2	12	15	15	15
Quadratus lumborum L4 Left	0	0	0	0	0
Quadratus lumborum L4 Right	49	93	105	106	106
Quadratus lumborum L3 Left	0	0	0	0	0
Quadratus lumborum L3 Right	63	151	186	190	191
Quadratus lumborum L2 Left	0	0	0	0	0
Quadratus lumborum L2 Right	7,0	3,0	5,7	6,1	6,2
Quadratus lumborum L1 Left	8,9	88	124	128	129
Quadratus lumborum L1 Right	0	0	0	0	0
Quadratus lumborum Thorax Left	0	0	0	0	0
Quadratus lumborum Thorax Right	26	19	15	15	15

Table 3.1: Intervertebral forces $[N]$: Impact of the muscular virtual stiffness K to the muscular forces estimation

and on top of which a force sensor (a virtual spring damper in the present context) has been placed in series: those actuators (the muscles) are in charge of the dynamical motion of the mechanical system (the spine and the upper body in our case).

For the direct dynamics simulation, the choice of the value for the virtual stiffnesses K is critical. Indeed, on the one hand, a very high stiffness is recommended to not move too far from the gait kinematics, which is the reference input to be respected at best. On the other hand, it is a well known fact that very high the stiffness can drastically slow down the time integrators. For that purpose, a preliminary sensitivity analysis has been carried out to identify a value (or more exactly a range of value) for the stiffness K for which:

- the gait kinematics is well respected;
- the muscle and intervertebral forces reasonably stabilize and converge for increasing values of K ;
- the direct dynamics CPU time remains decent and acceptable.

Referring to Tables 3.1 and 3.2, as the virtual stiffness K of the muscles increases, both the estimated muscular forces, and the intervertebral efforts converge towards a stabilized value. This preliminary sensitivity analysis shows that a stiffness of

	K = 100K [N/m]	K = 1M [N/m]	K = 10M [N/m]	K = 50M [N/m]	K= 100M [N/m]
L5-S1 Shear	126	227	268	272	273
L5-S1 Normal	375	552	625	634	635
L4-L5 Shear	90	102	108	109	109
L4-L5 Normal	376	576	657	667	668
L3-L4 Shear	101	143	163	166	166
L3-L4 Normal	329	482	550	558	559
L2-L3 Shear	132	198	226	229	230
L2-L3 Normal	261	329	363	367	368
L1-L2 Shear	40	50	58	59	59,44
L1-L2 Normal	280	375	416	421	422
T12-L1 Shear	144	147	149	149	149
T12-L1 Normal	253	262	264	265	265

Table 3.2: Intervertebral forces [N]: Impact of the muscular virtual stiffness to the intervertebral normal and shear forces for an scoliotic patient

$50 e^6 \text{ N/m}$ is sufficient for our study. This proposition is reinforced by the comparison between the gait kinematics (input) and that found by the time simulation (output). In Fig. 3.24 for instance, those two values are compared for the bending angle in the L5 – S1 disc, with a maximum error of $2 e^{-3}$ rad.

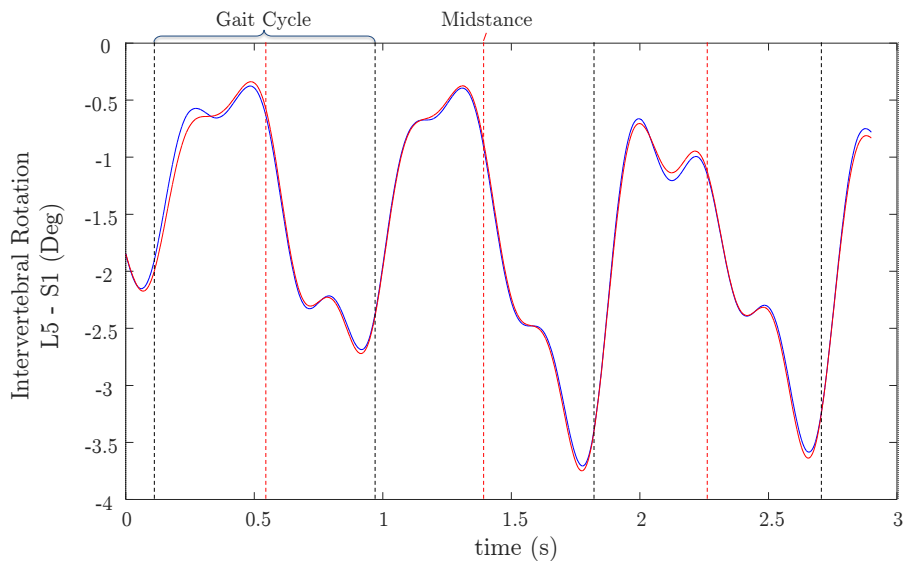


Fig. 3.24: Intervertebral rotation L5-S1: the bending measured from the patient gait kinematics (blue) is dully complied with by the bending resulting from the muscle forces via a direct dynamics computation (red)

Residual muscle tension To correlate muscular forces with the muscular activation measured by the EMG signals (Sect.2.5), we have deliberately decided to achieve it in a rather simplistic way, that is, without resorting to any dynamical constitutive law¹ (ex. Hill or Hill-based muscle models). However, at least to estimate the force of well-targeted muscular groups, the use of a static calibration with the David Back machine (Sect.2.5) will allow us to roughly calibrate some muscle groups, such as the Rectus abdominis (Fig.3.25).

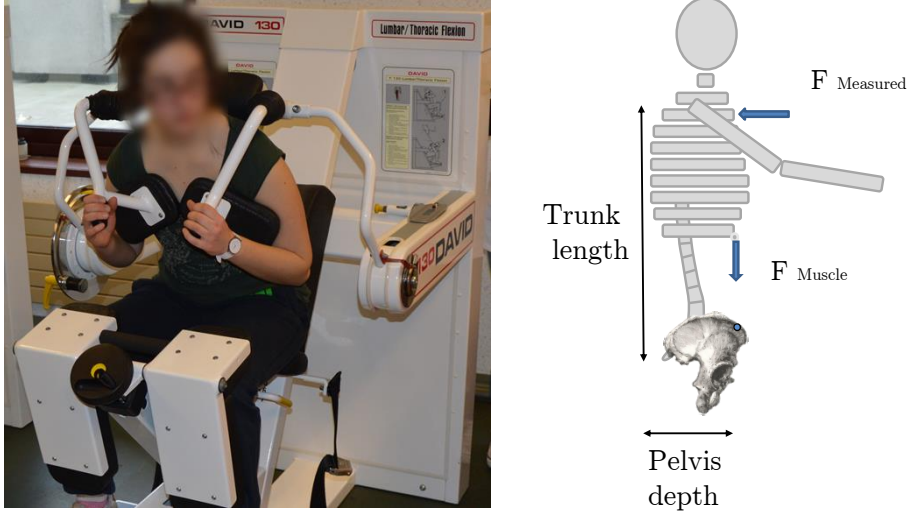


Fig. 3.25: Estimation of muscular activation during flexion: David back protocol.

Knowing (via measurements and via physiological data) the static configuration of the patient being performing a David Back test (Fig.3.25) allows estimating an algebraic ratio Ψ between the filtered and rectified EMG (Fig.3.26) and the Abdominal Muscular forces.

$$\Psi = \frac{F_{Muscle}}{EMG_{Muscle}} \quad (3.22)$$

with

$$F_{Muscle} = F_{Measured} \frac{L_{trunk}}{D_{pelvis}} \quad (3.23)$$

where:

¹We are clearly aware of the underlying simplifications: this point will be discussed in the last section dealing with the perspectives of this work.

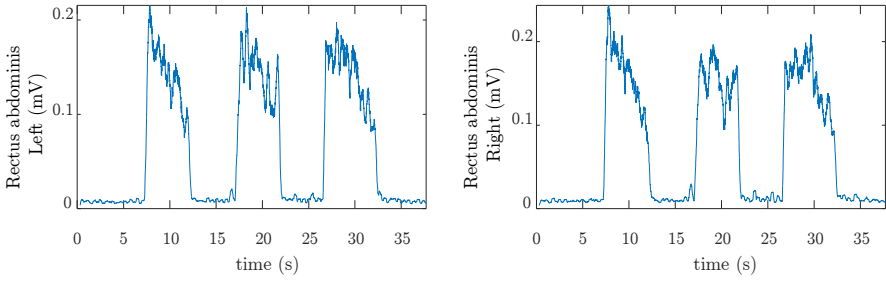


Fig. 3.26: Abdominal muscles activation during flexion.

- L_{trunk} denotes the trunk length,
- D_{pelvis} represents the pelvis depth.

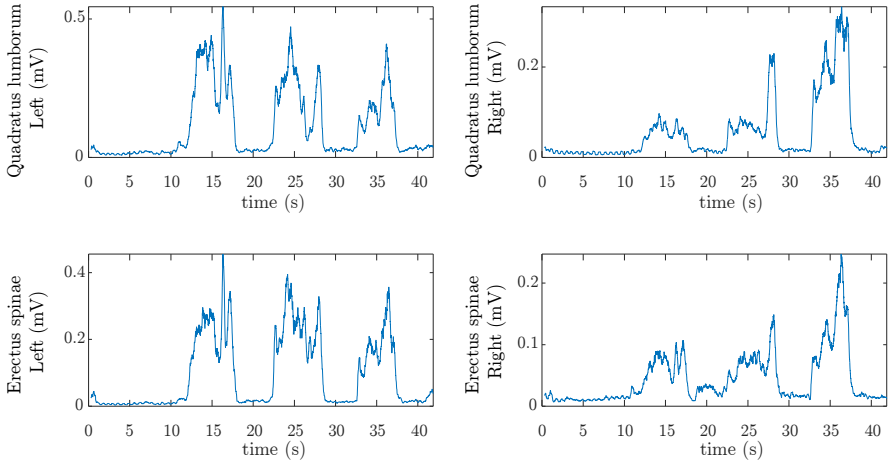


Fig. 3.27: Back muscles activation during extension.

Unfortunately, back muscles could not be analyzed in the same way. Due to muscle redundancy, it is not possible to measure the force of an isolated muscle using the David Back protocol. Furthermore, due to the amount of muscles overlapped on the back, the muscle activation measured via the EMGs contains mixed information arising from several muscles.

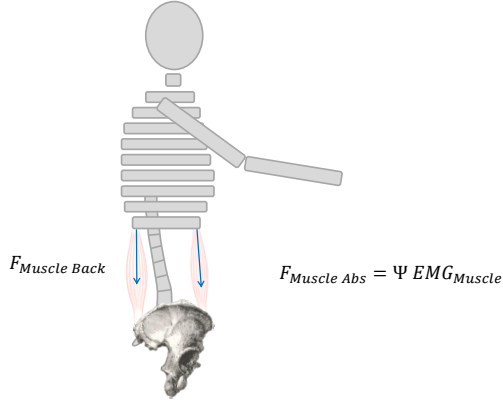


Fig. 3.28: Estimation of the Back Muscular Forces via a static equilibrium.

To overcome this problem, once the force of both Rectus abdominis are estimated via eq. (3.23), a static reasoning is carried out to infer the minimal residual tension (a.k.a. Muscle tone) of the back muscles (Fig.3.27 and 3.28) for given Rectus abdominis forces.

Results and discussion

4

As explained in Sect.1.4, the final objective of the thesis is the computation of the intervertebral efforts in scoliotic adolescents. The musculoskeletal multibody model as well the experimental protocol used in this study were developed in close collaboration with physiotherapists, surgeons and engineers; it was always intended to inspire the modeling approach and to feed the resulting models from the human body knowledge and the medical community expertise, and not the contrary¹. However, as mentioned in Sect. 3.1, it is actually impossible to identify all the components that will affect the intervertebral forces, and hence the present work has *de facto* several limitations that must be preliminary established. Despite some repetition, let us list them here below.

- Geometrical inaccuracies due to X-rays quality and accuracy,
- Lacking kinematic information during gait analysis,
- Rough identification of inertial parameters of the human body,
- Modeling assumptions such as the use of rigid bodies and ideal articulations,
- Limited number of muscles in the multibody model,
- Experimental difficulties to measure muscles activation,
- Other anatomical unknowns such as the interaction between internal organs and the spine.

It is obvious that all those limitations significantly reduce the precision in the calculation of the intervertebral efforts. Nevertheless, the biomechanical models presented

¹Indeed, as engineers, it is often tempting to transpose an expertise from one domain to another ... but a human is far more being a robot.

in this work can be used to better predict the effect of scoliosis on the spine as a mechanical support of the upper body undergoing static or dynamical loads. In other words, although there are clearly not enough data to accurately compute the value of the intervertebral forces, the multibody model, jointly with the experimental input, can help to assess the change in force induced by scoliosis.

This chapter aims at analyzing the results of our study and to indicate how the latter should broaden the current knowledge of surgeons and physiotherapists and, subsequently via further developments, feed their decision-making for a surgery planning.

4.1 Patient Data

The analysis of the intervertebral forces developed in this section relies on the study of one scoliotic patient, for whom the protocol described in Chapter 2 has been followed. It is important to notice that although the current analysis is based on a single patient, the complete development and adjustment of the protocol have been carried out via the study of several patients. However, since the present protocol has evolved all along the development of the project, most of those experiences have been performed to improve and validate subsets of the final protocol and thus, they are not suitable to be discussed in this chapter.

4.1.1 Diagnosis

The medical report¹ of the patient reveals the following:

- In standing position, the patient complained of low back pain, mainly in the left lumbar paravertebral area,
- The patient did not practice any sportive activity on a regular basis,
- The anterior bending test showed a lumbar gibbosity of about 15 degrees,
- The spine is flexible for the lateral bending test,
- Normal gait,
- Abdominal reflexes were asymmetrical,
- The right thoracic curve measured 69 degrees between T5 and T11,
- The left lumbar curve measured 55 degrees between T11 and L4,
- There was a high intervertebral torsion.

¹Complete medical report can be found in Appendix. C

4.1.2 Medical data

Date of the study	12/04/2016
Age of the subject	13 years old

Patient Data		
Weight	54.2kg	
Height	159cm	
Distance pelvis to floor	95cm	
Trunk length	53cm	
Pelvis depth	Right	Left
Total leg length	10cm	10,5cm
Modeling data (Appendix B)	88cm	88,5cm

Upper-body mass	21,7 kg
Upper-body mass inertia	
$I_{trunk}^G(x-axis)$	0,44 kg.m ²
$I_{trunk}^G(y-axis)$	0,33 kg.m ²

4.1.3 X-rays of the patient

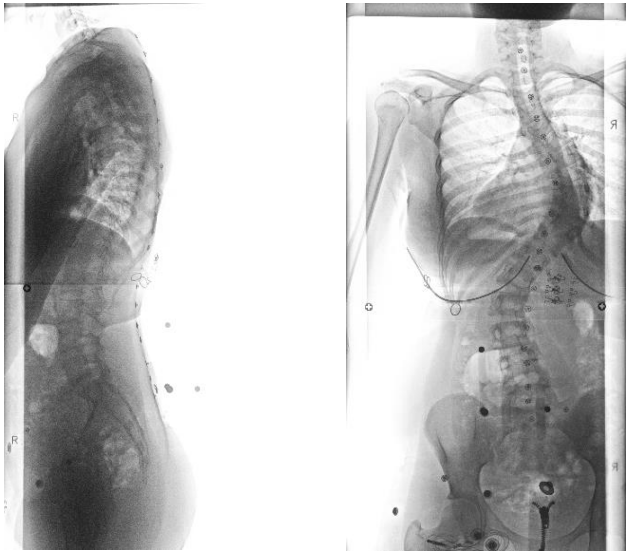


Fig. 4.1: Patient X-rays: one clearly observes a thoracic scoliosis for this adolescent

4.2 Intervertebral forces computation

This section focuses on the computation of the intervertebral forces for an scoliotic patient. They are analyzed for two different situations: standing-up, and during moderate gait. The intervertebral forces will be divided into 2 components: the normal force, Q_{normal} , perpendicular to the disc plane $\{\hat{\mathbf{J}}_1^n, \hat{\mathbf{J}}_2^n\}$ and the shear force, Q_{shear} , acting in the same plane (Fig. 4.2).

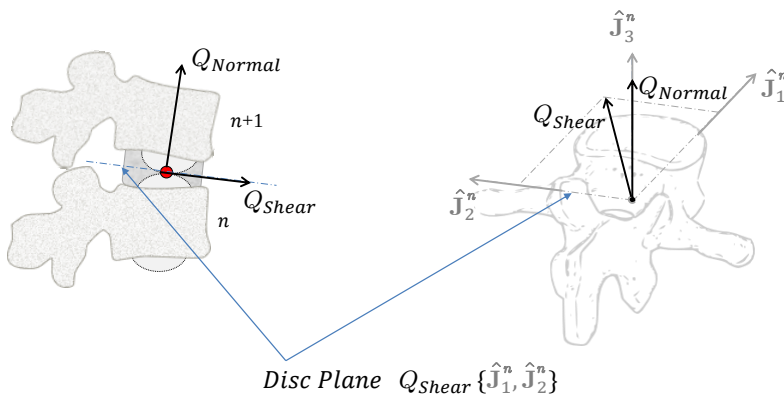


Fig. 4.2: Intervertebral forces components: normal force, perpendicular to the disc plane, and shear force, acting in the disc plane $\{\hat{\mathbf{J}}_1^n, \hat{\mathbf{J}}_2^n\}$.

4.2.1 Bi-planar X-rays and mutibody model of the patient

The first step of the analysis consists in identifying the geometrical shape of the spine using bi-planar X-rays, following the protocol described in Sect. 2.3.

Then, from the sagittal and coronal X-rays jointly exploited, and thanks to the computer acquisition of the markers' location (Sect. 3.5), a multibody model is automatically generated (Fig. 4.4) according to the minimization process described in Sect. 3.4.

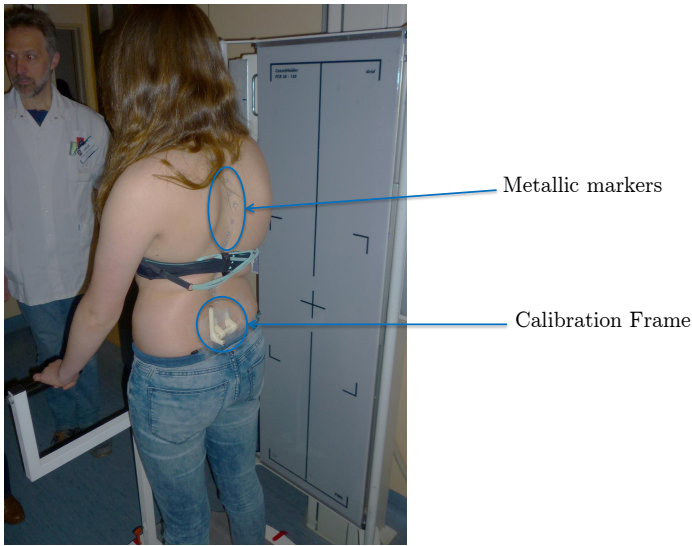


Fig. 4.3: The physiotherapist prepares the patient for the bi-planar X-rays (Cliniques universitaires Saint-Luc, UCL)

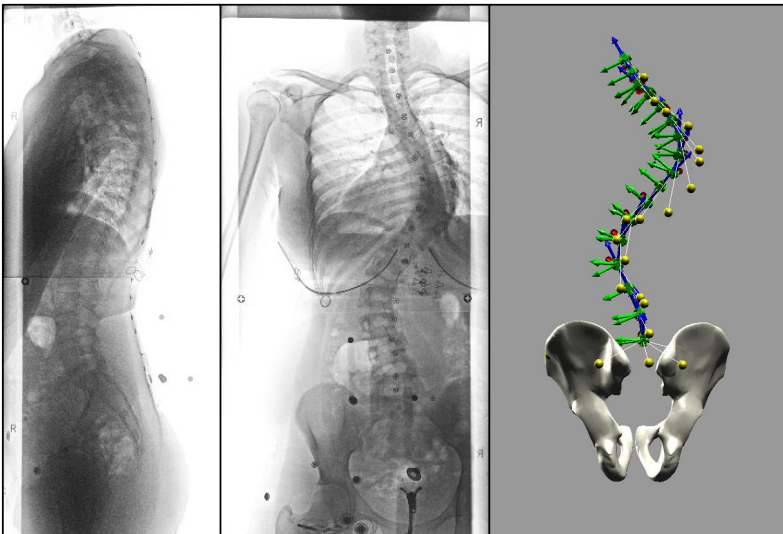


Fig. 4.4: Multibody model of the spine obtained from X-rays(right), resulting from a geometrical minimization process(left)

4.2.2 Static analysis

Once the multibody model is generated, a static analysis can be carried out. As said above, the biomechanical analysis of the spine contains several unknowns. A static analysis will allow us to better understand the impact of those *physiological* unknowns on the computation of the intervertebral forces. The main goals can be assimilated to the following tasks:

- study the impact of some muscle groups,
- analyze the effect of the residual muscle tension of some muscular groups,
- analyze the sensitivity of the intervertebral stiffness,
- estimate the effects of scoliosis,

on the intervertebral efforts, the normal and shear forces in our case, and also on the [other] muscles.

Influence of muscles A study on the impact of scoliosis on the intervertebral efforts was already outlined in [77]. That preliminary study used a simplified multibody model in which the spinal motion was neglected and the muscles were absent.

This section aims at highlighting the importance to take muscles into account when studying the spine from a mechanical (static and dynamical) points of view. To this end, two different models issuing from our general multibody representation (Sect. 3.3) are compared:

- a *simplified* biomechanical model of the spine, in which all muscles are deliberately neglected. In this non physiological model, all the intervertebral rotations are locked using the driven joints technique(Sect.3.2.5);
- a multibody model in which the intervertebral rotations are made free, the spine being now physiologically maintained by the muscles as described in Sect. 3.7.

To this end, the static analysis is carried out using the muscular groups shown in Fig. 4.5, carefully chosen thanks to the advise of surgeons and physiotherapists.

As expected, this comparative study clearly shows that muscles have a significant effect on the intervertebral forces. Indeed, as illustrated in Sect. 2.5, each muscular force, like a "bungee cord", maintains the spine in flexion, but unavoidably induces additional forces in the set of intervertebral discs.

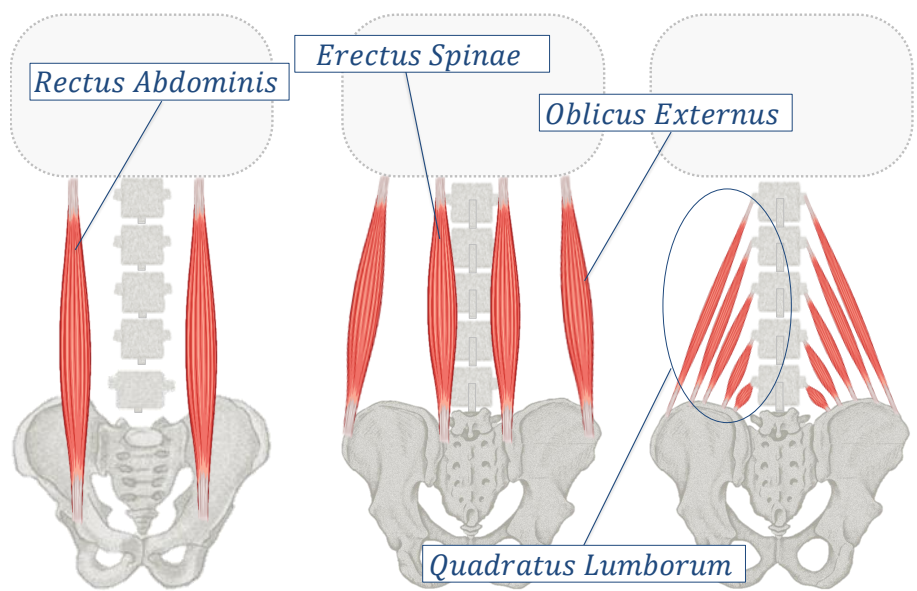


Fig. 4.5: Muscles groups included in the multibody model

	Self-maintained Spine	Muscle-maintained Spine
T12-L1 Shear	101	104
T12-L1 Normal	186	189
L1-L2 Shear	29	38
L1-L2 Normal	212	307
L2-L3 Shear	94	166
L2-L3 Normal	196	278
L3-L4 Shear	85	119
L3-L4 Normal	204	419
L4-L5 Shear	85	86
L4-L5 Normal	207	500
L5-S1 Shear	24	196
L5-S1 Normal	225	472

Table 4.1: Intervertebral forces [N] defined in Fig.4.2: Influence of the muscles to the intervertebral shear forces

It is important to notice that, in addition to the increased intervertebral forces, there is a significant change in the way forces are distributed along the spine. This means that a simplified model does not only underestimate those forces, but also shows a

skewed behavior compared to a physiological model. As explained at the beginning of the chapter, it is not easy to estimate the precision of this result, since it is hard to identify all the components affecting the muscular forces. However, this result at least reveals the importance to take the muscular forces into account in the study.

The difference of the intervertebral forces between the two models could be understood thanks to a simple equilibrium reasoning: for each muscular force F acting on the spine, there is an equal and opposite force increase F' induced in the intervertebral disc (Fig. 4.6).

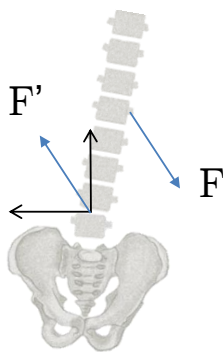


Fig. 4.6: Muscle forces ensure bending rigidity but induce intervertebral forces in the discs (one illustrated).



Fig. 4.7: Instrumentation of the patient for gait analysis: Cliniques universitaires Saint-Luc, UCL.

Residual muscle tension As explained in Sect.3.7, the human body is overactuated and hence, mechanically speaking, there is an infinity of combinations of muscular forces that can satisfy the same motion. This section follows the method described in Sect. 3.7 to estimate the muscular forces of the patient from the David Back test data, based on the EMG measurement (Fig. 4.7).

$$F_{measured} = \frac{Q_{David}}{L_{trunk}} \quad (4.1)$$

where:

- Q_{David} represents the maximum torque (72 Nm) measured during the David Back test;
- L_{trunk} is the trunk length (0.53 m).

Considering a static equilibrium (Fig. 3.28), one can write:

$$F_{Muscle} = F_{Measured} \frac{L_{trunk}}{D_{pelvis}} \quad (4.2)$$

where $D_{pelvis} = 0.1m$ represents the pelvis depth.

Thus:

$$F_{Muscle} = \frac{Q_{David}}{D_{pelvis}} = 720N \quad (4.3)$$

Taking the maximum muscular EMG activation (Fig. 4.8) into consideration results in:

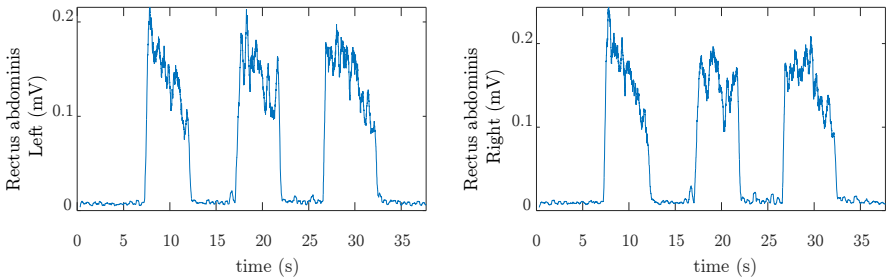


Fig. 4.8: EMG of the left and right Rectus abdominis during flexion (Sect. 2.5). A muscular activation: $EMG_{Muscle} \approx 0.2mV$ is observed, for a muscle force $F_{Muscle} = 720N$

$$\Psi = \frac{F_{Muscle}}{EMG_{Muscle}} \approx 4kN/mV \quad (4.4)$$

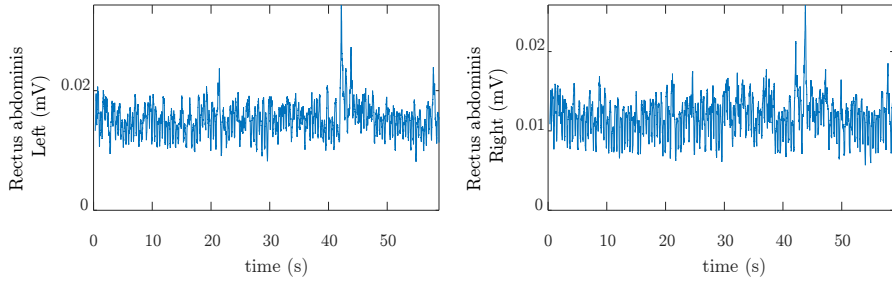


Fig. 4.9: EMG of the left and right Rectus abdominis during gait.

Finally, taking into account the minimal muscle activation (EMG) of about $0.01mV$ measured during gait¹ (Fig.4.9), the average residual muscle tension during gait is:

$$F_{Muscle} = \Psi EMG_{Muscle} \approx 40N \quad (4.5)$$

Influence of residual muscular tension : muscular co-contraction Once the residual muscle tension of the Rectus abdominis is estimated, we suggest to perform the following sensitivity analysis — by taking different values of this residual muscle tension, namely: 2, 40, 80 and 160 N — to observe their impact on the other muscle groups (Tab. 4.2) and on the intervertebral forces (Tab.4.3).

¹the left/right difference of the EMG (Fig. 4.9) is not taken into account in the model.

	2 N	40N	80N	160N
Erector spinae Left	0	8,5	13	33
Erector spinae Right	2,8	30	60	119
Rectus abdominis Left	2	20	40	80
Rectus abdominis Right	2	20	40	80
Quadratus lumborum L1 Left	99	128	161	227
Quadratus lumborum L1 Right	0	0	0	0
Quadratus lumborum L2 Left	0	0	0,37	17
Quadratus lumborum L2 Right	13	6,1	0	0
Quadratus lumborum L3 Left	0	0	0	0
Quadratus lumborum L3 Right	143	190	242	352
Quadratus lumborum L4 Left	0	0	0	0
Quadratus lumborum L4 Right	81	106	133	191
Quadratus lumborum L5 Left	0	0	0	0
Quadratus lumborum L5 Right	1,6	15	31	60
Obliquus externus Left	0	0	0	0
Obliquus externus Right	2,4	15	31	57

Table 4.2: Muscular forces [N] in muscles described in (Fig.4.5) for different residual muscle tension

	2 N	40N	80N	160N
T12-L1 Shear	106	149	195	290
T12-L1 Normal	198	265	343	501
L1-L2 Shear	39	59	82	128
L1-L2 Normal	313	421	541	782
L2-L3 Shear	170	229	295	436
L2-L3 Normal	280	367	465	685
L3-L4 Shear	122	166	215	322
L3-L4 Normal	424	558	708	1038
L4-L5 Shear	87	109	133	183
L4-L5 Normal	506	667	845	1239
L5-S1 Shear	198	272	356	540
L5-S1 Normal	478	634	808	1186

Table 4.3: Intervertebral forces [N] defined in Fig.4.2: sensitivity of the muscular forces to different residual muscle tension

As expected, the muscular forces (Tab.4.2) and intervertebral forces amplitudes (Tab.4.3) grow as the Rectus abdominis muscular residual muscle tension increases. However, force distribution (for both disc and muscular forces) does not seem to be affected, showing a rather low impact of the residual muscle tension to the overall

loading distribution of the spine.

Spinal stiffness influence This study consists in analyzing the effect of the intervertebral bending stiffness on the value of the muscular and intervertebral forces. As stated in [94], there is a significant variation in the data obtained from different sources. This variation, still according to [94], could be explained by the complexity of the mechanical properties of the intervertebral discs (nonlinear, biphasic and viscoelastic).

	Stiffness from literature (K)	Reduced Stiffness (K/5)	Increased Stiffness (K x 5)
Erector spinae Left	8,4	8,4	8,6
Erector spinae Right	30	30	29
Rectus abdominis Left	20	20	20
Rectus abdominis Right	20	20	20
Quadratus lumborum L1 Left	128	129	124
Quadratus lumborum L1 Right	0	0	0
Quadratus lumborum L2 Left	0	0	0
Quadratus lumborum L2 Right	6,1	6,2	5,8
Quadratus lumborum L3 Left	0	0	0
Quadratus lumborum L3 Right	190	191	186
Quadratus lumborum L4 Left	0	0	0
Quadratus lumborum L4 Right	106	106	105
Quadratus lumborum L5 Left	0	0	0
Quadratus lumborum L5 Right	15	15	14,5
Obliquus externus Left	0	0	0
Obliquus externus Right	15	14	15

Table 4.4: Muscular forces $[N]$ in muscles described in (Fig.4.5) for different intervertebral bending stiffness $[Nm/rad]$ obtained from [94]

As shown in Tab. 4.4, the effect of the intervertebral stiffness has a very limited impact on the computation of the muscular forces.

	Stiffness from literature (K)	Reduced Stiffness (K/5)	Increased Stiffness (K x 5)
T12-L1 Shear	149	149	149
T12-L1 Normal	265	265	264
L1-L2 Shear	59	60	58
L1-L2 Normal	421	422	417
L2-L3 Shear	229	230	226
L2-L3 Normal	367	368	364
L3-L4 Shear	166	166	163
L3-L4 Normal	558	560	551
L4-L5 Shear	109	109	108
L4-L5 Normal	667	668	658
L5-S1 Shear	272	273	268
L5-S1 Normal	634	636	626

Table 4.5: Intervertebral forces $[N]$ defined in Fig.4.2 for different intervertebral stiffness $[Nm/rad]$

Tab.4.5 reveals that the effect of the intervertebral stiffness has also a very limited impact on the intervertebral discs efforts. It is important to notice that this limited effect (of the intervertebral stiffness) could be explained by the muscular model. However, this result was corroborated in a previous study — in which muscular forces were neglected — that showed that intervertebral stiffness is clearly not sufficient to withstand the spine (+ the upper body) by itself.

Influence of scoliosis Following the goal of this research, the last static study consists in simulating the effect of the scoliosis on the intervertebral and muscular forces. To that purpose, an artificial model in which the spinal deformation of the patient is *virtually straightened*¹ is simulated.

¹N.B.: but the pelvis remains skewed.

	Original Scoliotic	Virtually straightened
Erector spinae Left	8,4	13
Erector spinae Right	30	10
Rectus abdominis Left	20	20
Rectus abdominis Right	20	20
Quadratus lumborum L1 Left	128	0
Quadratus lumborum L1 Right	0	9
Quadratus lumborum L2 Left	0	0
Quadratus lumborum L2 Right	6,1	45
Quadratus lumborum L3 Left	0	0
Quadratus lumborum L3 Right	190	18
Quadratus lumborum L4 Left	0	0
Quadratus lumborum L4 Right	106	14
Quadratus lumborum L5 Left	0	14
Quadratus lumborum L5 Right	15	0,03
Obliquus externus Left	0	16
Obliquus externus Right	15	16

Table 4.6: Muscular forces [*N*] in muscles described in (Fig.4.5) for an scoliotic vs an artificially non scoliotic patient

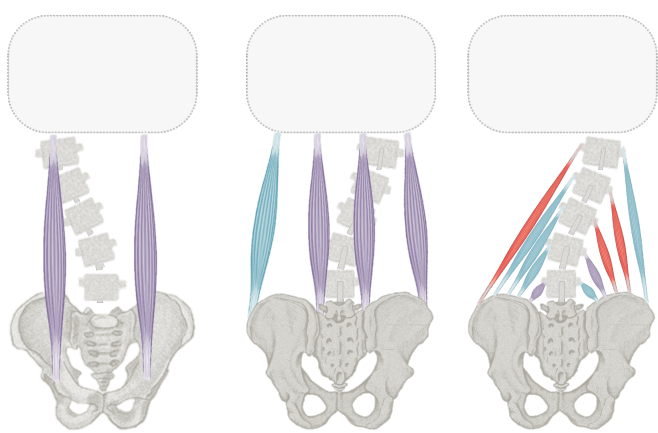


Fig. 4.10: Muscular activation for an scoliotic patient. Muscles with high activation in red (activation superior to 100N), low activation in purple (less than 50 N) and non activated muscles in blue

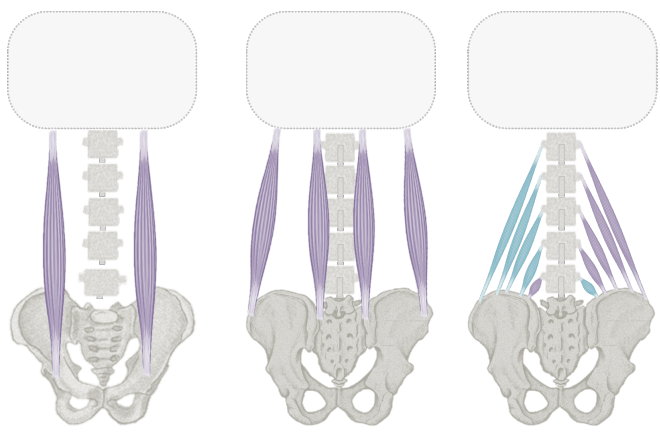


Fig. 4.11: Muscular activation for an virtually straightened patient. Muscles with high activation in red (activation superior to 100N), low activation in purple (less than 50 N) and non activated muscles in blue

As shown in Tab. 4.6 , the muscular force asymmetry is significantly higher (Fig. 4.10) for the scoliotic patient compared to the virtually straightened patient. Similar asymmetries of the muscular activation has already been observed in other studies [89], which analyzed back muscles activation during isokinetic exercises.

	Original Scoliotic	Virtually straightened
T12-L1 Shear	149	23
T12-L1 Normal	265	304
L1-L2 Shear	59,	69
L1-L2 Normal	421	371
L2-L3 Shear	229	120
L2-L3 Normal	367	441
L3-L4 Shear	166	70
L3-L4 Normal	558	469
L4-L5 Shear	109	66
L4-L5 Normal	667	482
L5-S1 Shear	272	17
L5-S1 Normal	634	497

Table 4.7: Intervertebral forces [*N*] defined in Fig.4.2 for an scoliotic vs a virtually straightened patient

As shown in Tab.4.7, the asymmetry in the muscular force distribution increases the shear forces significantly (up to 17 times). While logical, this result might be exaggerated and should be considered cautiously. Nevertheless, it illustrates how spine curvature can induce a significant increase of intervertebral efforts in scoliotic

patients.

4.2.3 Dynamical analysis

The dynamical analysis is carried out in two steps:

- First, the intervertebral and pelvis motions are measured from the walking patient at 6km/h (Fig. 4.12), then computed using the kinematics minimization described in Sect. 3.5. Then, this motion is imposed to the model, as explained in Sect. 3.7 and this kinematic process allows us to accurately compute the distance between all muscles connection points during gait. Let us remind that the muscles considered here are the muscles groups presented in Sect. 2.5.
- A direct dynamics, for which muscular lengths are imposed according to the previous simulation. For this step of the process, the entire motion of the spine is driven by the muscles whose force can be evaluated from the virtual spring/damper serial elements, as explained in Sect. 3.7.

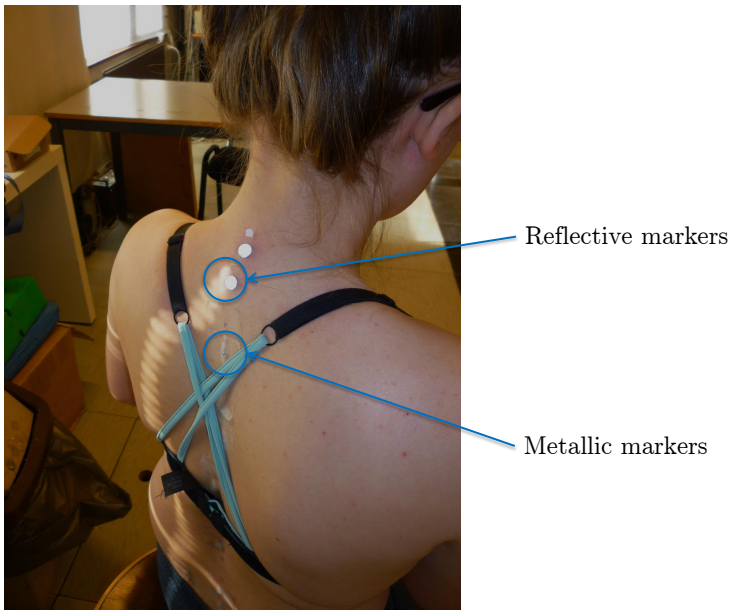


Fig. 4.12: The reflective markers are located on the spinal processes of the patient's spine (Cliniques universitaires Saint-Luc, UCL)

Three different models will be considered in this study:

- A multibody model based on an inverse dynamics approach (Sect. 3.2.5), where intervertebral motions are directly imposed within the discs. In this respect, this "lumped" model is similar to these presented in [77];
- A multibody model for which the spine internal torques are passive, but which is now actuated by all the muscle groups used for the static analysis, i.e. Quadratus lumborum, Rectus abdominis, Erector spinae and Obliques;
- Finally, a simplified model for which the motion between the thorax and the lumbar spine is actuated by a subset of muscles, but where the lumbar intervertebral motion is imposed via driven joints and constraint torques, similarly to the first model.

Inverse dynamics approach: no muscle-model As explained above, in this first model aimed at quantifying the intervertebral normal and shear forces, the intervertebral motion is imposed directly within the discs using driven joints (Sect. 3.3). Doing so, the spine bending withstanding is artificially ensured by constraint pure torques (i.e. Lagrange multipliers) lumped inside the spine, being the equivalent torque resultant of the — deliberately absent — muscles.

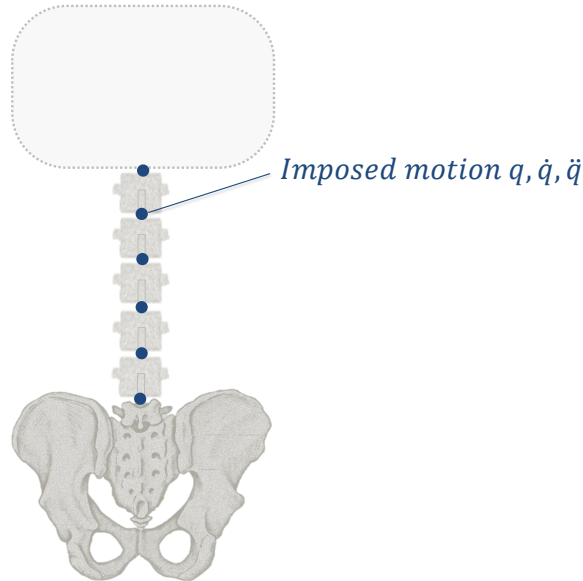


Fig. 4.13: The muscle action is lumped inside the spine via constraint pure torques (small blue discs) that withstand the spine in bending.

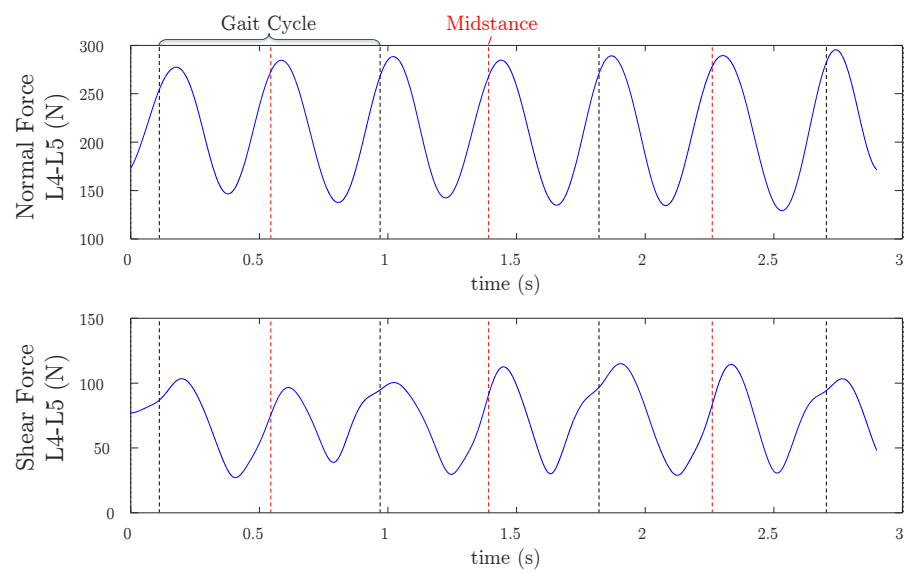


Fig. 4.14: Intervertebral forces into L4-L5 for the no muscle-model: normal force (up) and shear force (down)

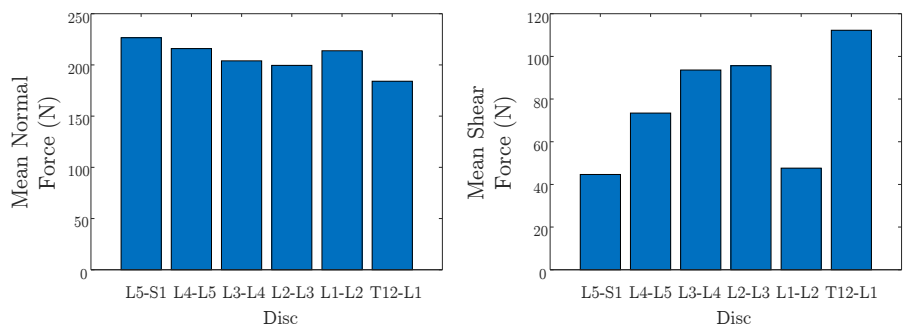


Fig. 4.15: Mean intervertebral lumbar forces for the no muscle-model: normal forces (left) and shear forces (right)

Even if muscles are absent from this first model, one may at least observe that there exists a significant dynamical component inside the intervertebral forces (Fig.4.17). For example, the intervertebral normal force in L4-L5 is increased from about 206 N in the static analysis (Tab. 4.1) to a maximum force of approx 300N.

Direct dynamics approach: full muscle-model For this second approach, the motion of the spine is physiologically actuated by the full set of muscles according to the method described in Sect. 3.7. This model includes the following muscles groups: Erector spinae, Quadratus lumborum, Obliquus and Rectus abdominis (Fig. 4.16).

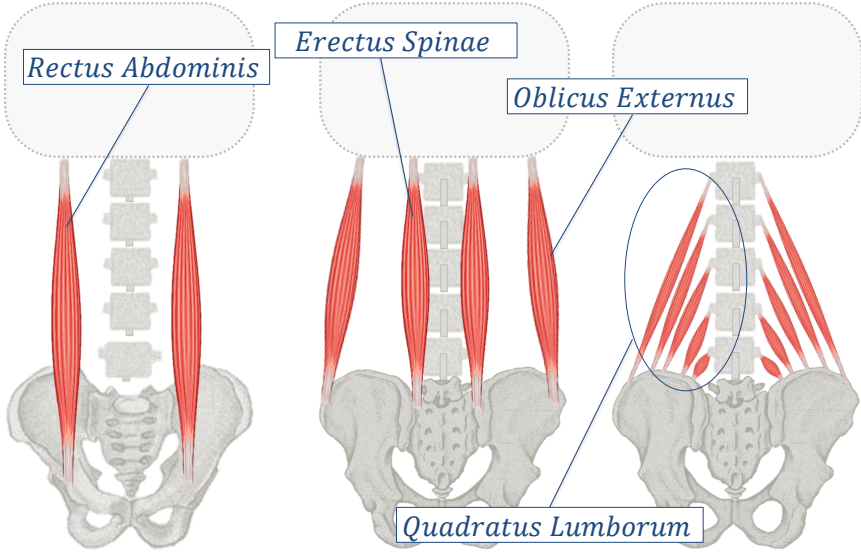


Fig. 4.16: Muscles groups included in the full-muscle multibody model

Unfortunately, it must be admitted that, contrary to the static analysis, forces do not seem to be representative of the reality. As revealed by Figs. 4.17 and 4.18, the intervertebral forces are extremely high.

The origin of those large no physiological force values, may be explained by the very small lever arm of some muscles connected to the lumbar spine (i.e. Quadratus lumborum). Fig 4.20 illustrates this fact with the muscular force measured at the Quadratus lumborum (L5).

It is however important to notice that the model used for this simulation is accurate *from a mechanical point of view*. The inaccuracies come from the physiological data. The results of this section — although being unrealistic — reveal that the muscles of the Quadratus lumborum alone, are not be sufficient to maintain and move the lumbar spine during gait, and that these mechanical functions obviously depend on physiological elements such as other muscles or the pressure inside the abdominal cavity due to the abdominal compression as pointed out in [68]. This results could also be explained by an overestimated inertia of the trunk, that has been modeled as a rigid body.

Another reason for this exaggerated forces could be explained by a simplified muscle and insertion model. While in reality muscles groups are continuous elements connected to spread bones areas, the muscular model used in this study consist in a reduced number of point to point forces. This simplification can lead to inaccurate force moments and subsequently to a non physiological muscular force distribution.

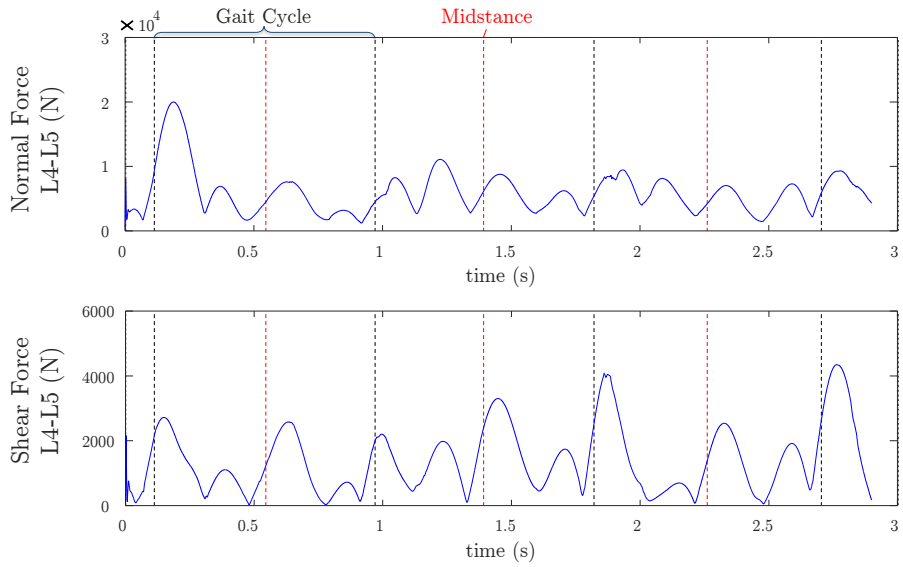


Fig. 4.17: Intervertebral forces into L4-L5 for the full-muscle model: normal force (up) and shear force (down)

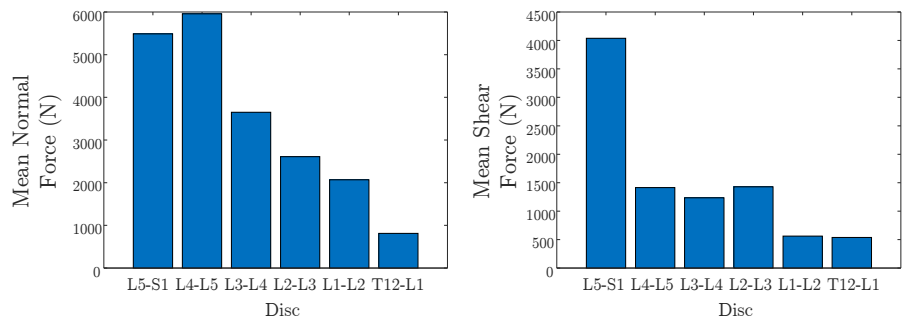


Fig. 4.18: Mean intervertebral lumbar forces for the full muscle-model: normal forces (left) and shear forces (right)

Direct/Inverse dynamics approach: hybrid actuation-model Following the results presented here above, an intermediate simplified model is now suggested, in

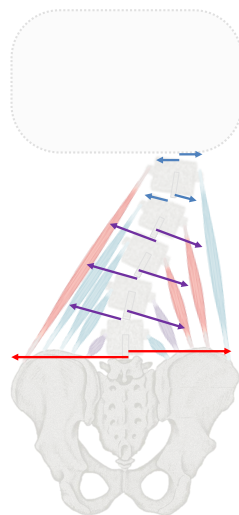


Fig. 4.19: Action/reaction intervertebral shear forces distribution along the spine. red arrows represent forces of about 4000 N, purple arrows represent forces between 1000 and 1500 N and finally, blue arrows represent forces of about 500 N.

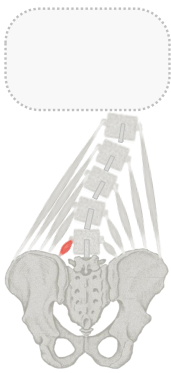
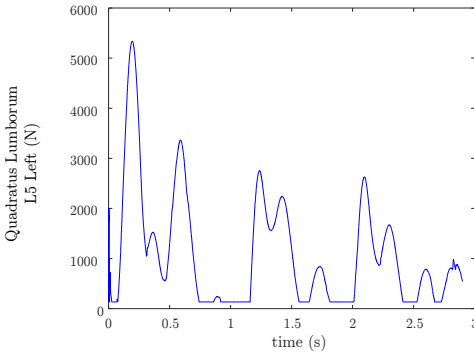


Fig. 4.20: Force in Quadratus lumborum (L5, in red)

which the motion between the thorax and the lumbar spine is ensured by some muscles (*direct* dynamics with known input forces) as in the second model, while lumbar intervertebral bending motions are kinematically imposed (*inverse* dynamics, with unknown constraint torques) as in the first model.

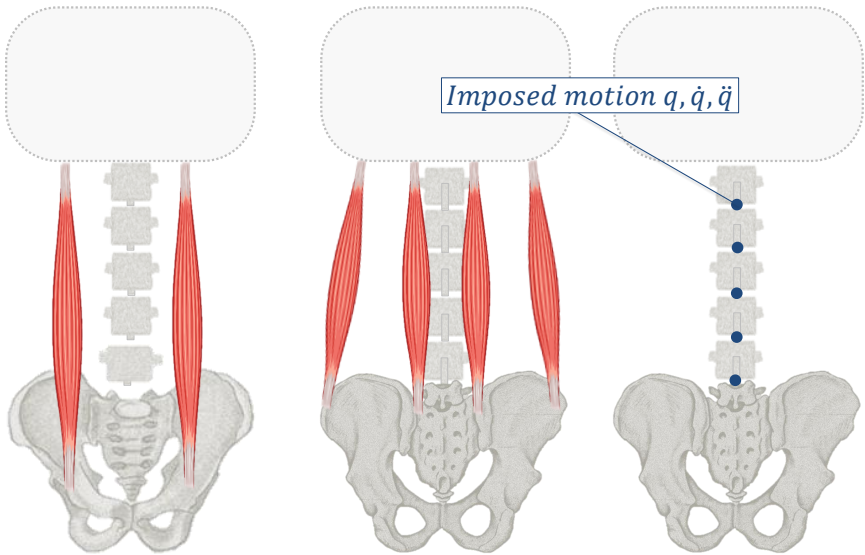


Fig. 4.21: In the hybrid multibody model, a subset of muscles is responsible for the motion of the thorax, the head and the arms), while the lumbar spine flexion is controlled by equivalent constraints torques (blue discs)

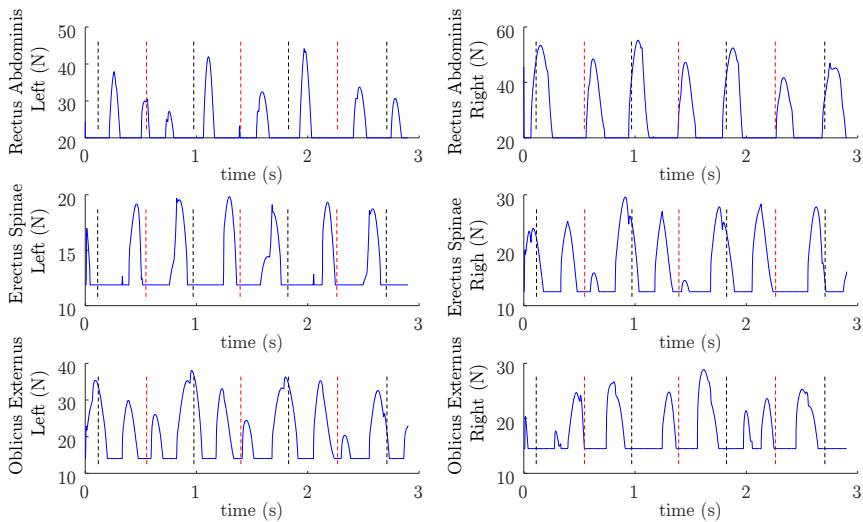


Fig. 4.22: Virtual muscles activation during the simulation.

Fig.4.22 reveals a right/left asymmetry of the muscular forces. The muscular force of the Rectus abdominis and Erector spinae is about 50% more on the right side than on the left side, except for the Obliquus externus.

The magnitude of the intervertebral forces calculated using this intermediate model seems to be more realistic, and in particular, when compared on average, to the values obtained for the static configuration.

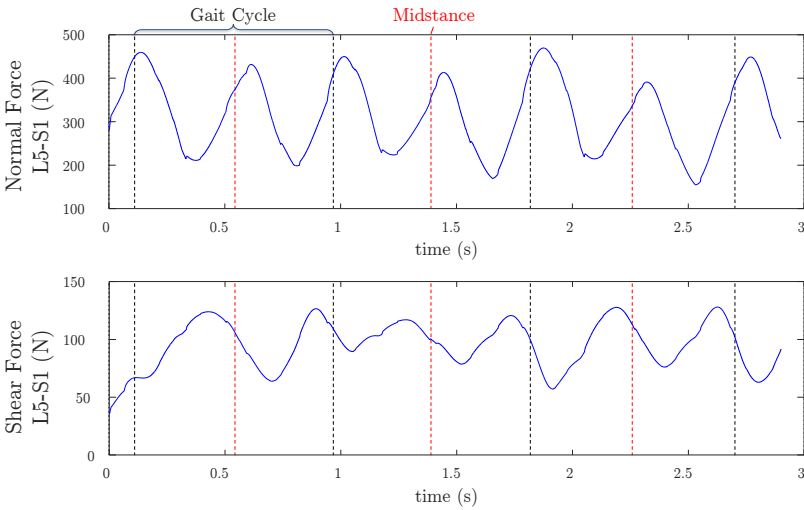


Fig. 4.23: Intervertebral forces into L5-S1 for the no muscle-model: normal force (up) and shear force (down)

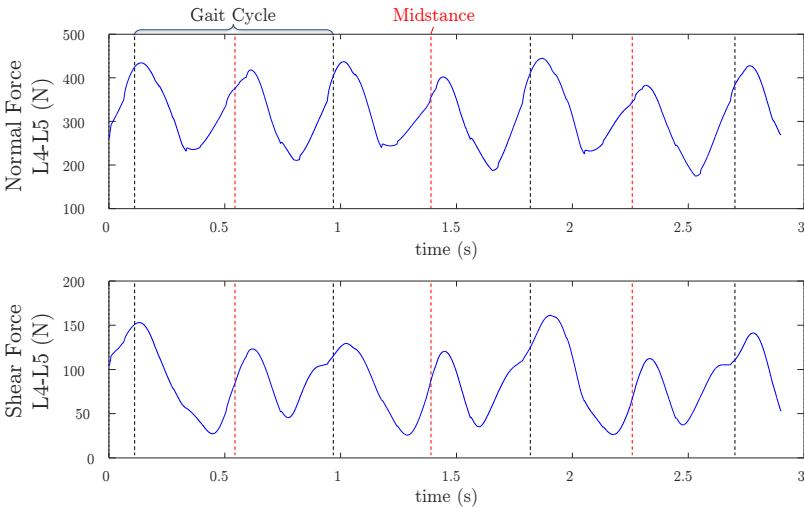


Fig. 4.24: Intervertebral forces into L4-L5 for the no muscle-model: normal force (up) and shear force (down)

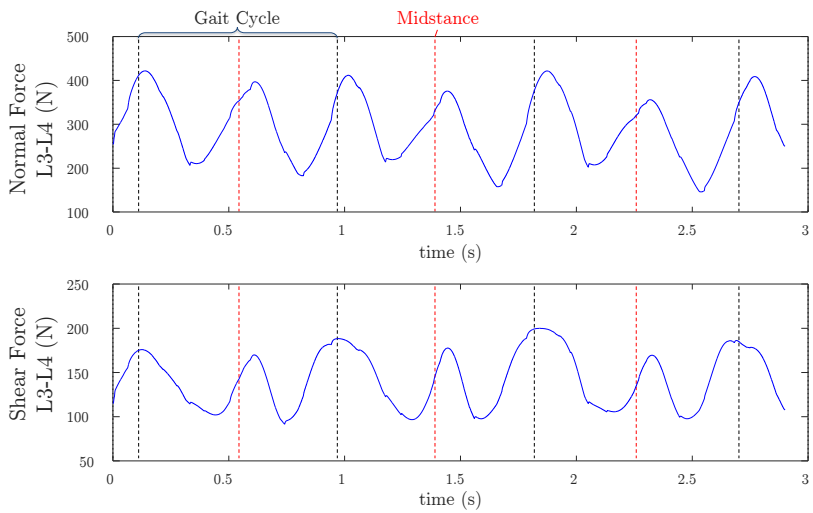


Fig. 4.25: Intervertebral forces into L3-L4 for the no muscle-model: normal force (up) and shear force (down)

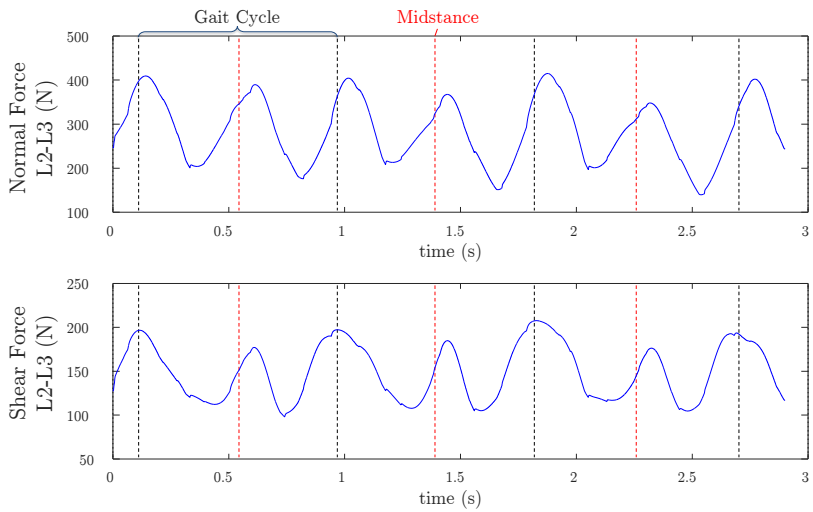


Fig. 4.26: Intervertebral forces into L2-L3 for the no muscle-model: normal force (up) and shear force (down)

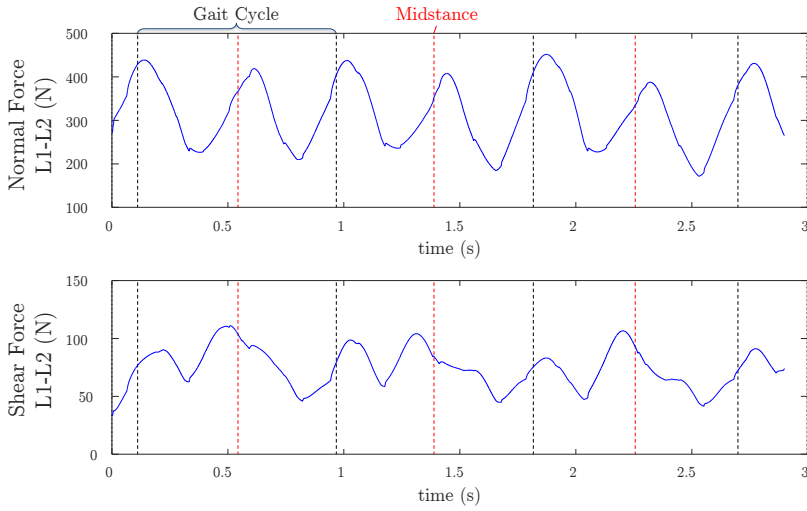


Fig. 4.27: Intervertebral forces into L1-L2 for the no muscle-model: normal force (up) and shear force (down)

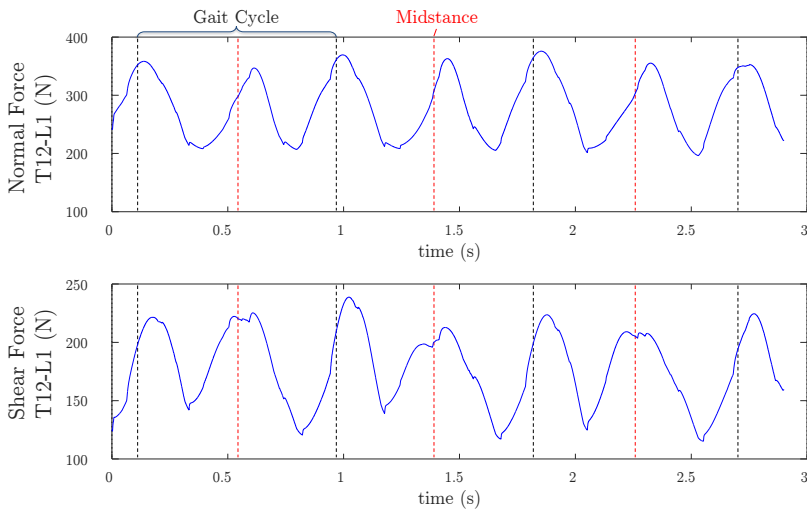


Fig. 4.28: Intervertebral forces into T12-L1 for the no muscle-model: normal force (up) and shear force (down)

As shown in Fig. 4.28, an asymmetry — mostly for the shear force — is observed between the first double limb support (black vertical line) and the second limb double support of the gait cycle (red vertical line).

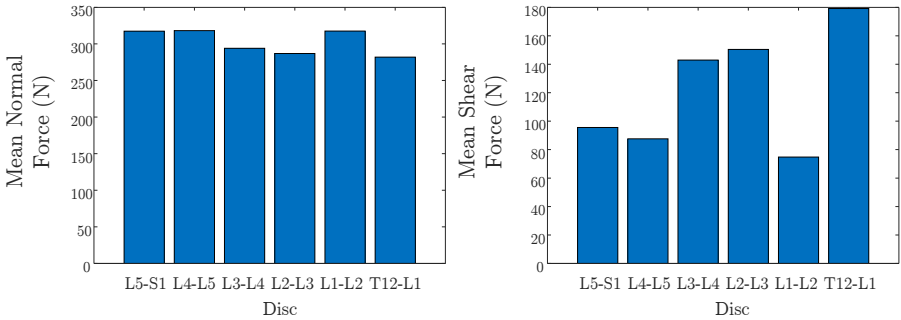


Fig. 4.29: Mean intervertebral lumbar forces for the full muscle-model: normal forces (left) and shear forces (right)

Fig. 4.29 shows the mean intervertebral discs forces. The intervertebral normal forces estimated using the hybrid model are about 40% higher than the no muscle-model (Fig. 4.15). Although, once again, the force distribution along the spine appear to be the same as in the no muscle-model.

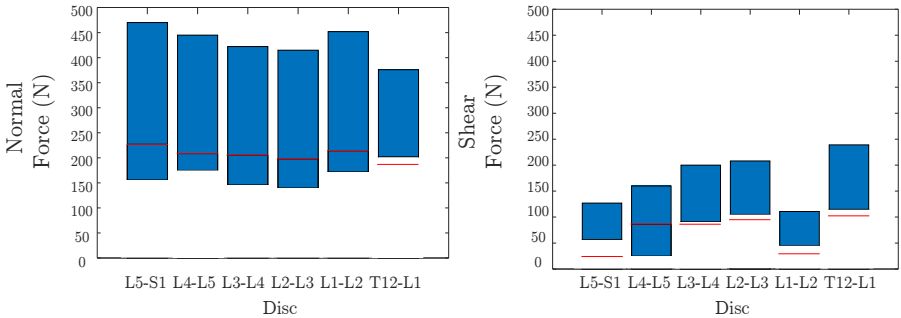


Fig. 4.30: Blue bars represent the Max/Min intervertebral lumbar forces for the full muscle-model: normal forces (left) and shear forces (right). Red lines represent the intervertebral forces obtained from the static simulation.

Fig. 4.30 shows a significantly higher intervertebral force during gait than in a standing up position (up to 125%). This result shows the importance of a dynamical analysis compared to an statical analysis, even for a moderate gait.

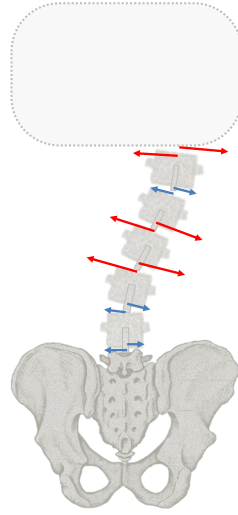


Fig. 4.31: Action/reaction intervertebral shear forces distribution along the spine. Red arrows represent forces of more than 100 N, and blue arrows represent forces of less than 100 N.

4.2.4 Final discussion

General consideration As explained at the beginning of this chapter, it is a rather difficult task to identify the physiological components that have a predominant impact on the intervertebral forces. Those limitations are particularly penalizing for the gait analysis in which dynamics intervenes and highly complicates the modeling approach as well as the measurement process.

The different analyzes developed in the previous sections mainly aimed to characterize the biomechanical models and their parameter sensitivities. However, they also allow identifying some biomechanical behaviors of scoliosis and some quantified information for spinal surgery. Although the initial objective of this work was the computation of the intervertebral efforts during gait, this probably does not represent its main contribution. As explained before, the biomechanical model presented in this study is based on several measurement bias and modeling hypotheses, and for that reason, the dynamical results are probably not yet satisfactory with respect to the reality of the patient. Rather than being the final product of our study, the obtained results and the associated models should be considered, instead, as a first reference dynamical tool to explore this new thematic: the biomechanical analysis of scoliosis using multibody dynamics.

Static analysis The results in Sect.4.2.2 have highlighted the importance to take muscles into account in the study of the intervertebral forces. This results showed that the intervertebral forces computed in the model without muscles (Fig.4.13) were not just underestimated in magnitude compared to the forces calculated with the model that includes all the muscles — a quantitative change — but also the force distribution along — a qualitative change —.

This result is particularly instructive for surgeons, since it suggests that intervertebral forces do depend on other parameters than purely geometrical ones (associated with spinal deformation) like Gibbosity and/or Cobb angle. In addition, they are also influenced by the residual muscular tension, the location of the deformation, the effect of an abnormal muscle insertion, and probably by other physiological parameters (not yet studied here) such as the disc bending stiffness or the interaction of the spine with the organs of the abdominal cavity.

Dynamical analysis The results of the dynamical analysis using the hybrid model confirms the importance of the muscular forces in the biomechanical model of a walking patient. As expected, the analysis points out that there is a significant increase of the (maximum) intervertebral forces due to gait dynamics. This observation seems instructive since nowadays most of the surgeons rely more on purely static considerations.

With respect to the non physiological unrealistic forces of the full-muscle model, we think that they could be explained by an overestimated inertia of the trunk. This phenomenon is not only model issue, but is also relevant from a surgical point of view; indeed, surgery can affect the *inertia* parameters of the upper body; for example, following a spinal fusion that reduces the spine mobility, the corresponding morphological — and thus inertial — modification might require an increase in the muscular and disc forces in the lower levels of the spine to equilibrate bending dynamical motion.

Conclusions and perspectives

5

5.1 Conclusions

This thesis aimed at quantifying the intervertebral efforts for idiopathic scoliotic adolescents in a standing up position and during moderate gait. To this end, the present research has addressed all the aspects needed for this purpose going from the development of an experimental protocol to the modeling of the upper body using the multibody dynamics approach thanks to a multidisciplinary collaboration between surgeons, physiotherapists and engineers. In parallel to the major contribution of this thesis stated above, some minor contributions that might be also useful medical applications, were also developed:

- the development of a user-friendly measurement tool and of a protocol that allow us to identify the 3D spinal shape,
- the implementation of a process able to automatically generate a biomechanical multibody model of the patient,
- the identification of the internal motion of the spine via a limited number of markers placed on the back of the patient,
- the analysis of the impact on the intervertebral efforts of physiological parameters such as: the muscles insertions, the presence of internal organs and the stiffness of the intervertebral discs,
- the computation of the muscular forces, and their impact on the intervertebral efforts.

- the identification of all the ingredients required to achieve an accurate analysis of the intervertebral efforts to suggest further improvements of the protocol and to better characterize the main sources of uncertainty (Tab. 5.1).

Process	Source of uncertainty / Quantification (when available)
X-rays	Errors induced by the X-ray magnification. This error could be reduced to less than 1% by following the protocol proposed in Sec. 2.3
Identification of spinal landmarks	Manual identification of spine and pelvis landmarks. Up to 30% error in the intervertebral force estimation for an error of 5mm.
Placing of reflective markers	Manual location of the reflective marker.
Gait analysis data	Errors connected to the precision during data acquisition: - Data interpolation due to data losses, - Accuracy of the system: marker location/permutation.
Physiological and muscular Data	- Experimental limitations to measure muscles activation, poor signal, lack of access, etc. - Anatomical unknowns such as the interaction between internal organs and the spine.

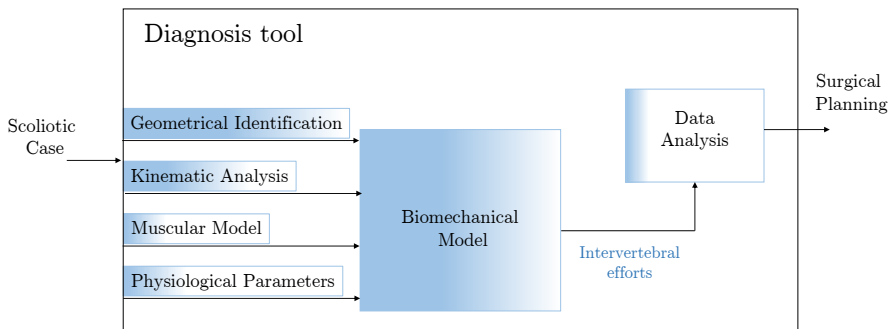
Table 5.1: Experimental sources of uncertainty

As stated before, it is difficult to identify all the physiological components necessary to precisely estimate the intervertebral forces. Furthermore, due to experimental and even ethical issues, it was impossible for us verify whether the intervertebral forces obtained from our biomechanical model are correct. Nevertheless, a preliminary comparison of the results obtained with our approach with the experimental studies carried out by [18] in which the intervertebral forces are measured using an instrumented vertebral body replacement, shows that our model prediction seem consistent with the reality. For example, the results in [18] show peak forces during that range from 21 to 83% patient bodyweight for a of gait at $4km/h$ compared to 50 to 80% of the total bodyweight obtained from our different models for a gait of $6km/h$. We observe a similar order of magnitude, for both the maximum force and the dynamical range. However, this comparison cannot be considered as a validation, since the results in [18] are based on different parameters from those used in our study, namely: non-scoliotic patients, different body mass and a different gait velocity.

Our current biomechanical model obviously needs more maturation. However, the results presented in this work can already give an idea of the main impacts of scoliosis on the intervertebral efforts. In particular, intervertebral forces do not only depend on spinal deformation (Gibbosity and Cobb angle) but also on various other phenomena, specifically linked to the muscle force distribution and to dynamical effects.

5.2 Perspectives

The long-term objective of this research was to contribute to the planning of spinal surgery. Although the present contribution gives guidelines to estimate the intervertebral efforts during gait, the results are not yet exploitable for surgical planning. To reach this ultimate goal, there are still several issues that should be addressed, not only to improve the accuracy estimating the intervertebral efforts but also to make the process more accessible in a clinical environment. In this context, the following scheme gathers the steps that are required to develop a diagnosis tool for surgical planning. The blue background represents the level of maturity, following our study, of each of those processes, suggesting some possible research to be pursued in the next future.



Geometrical identification The geometrical identification of the spine has been studied in this thesis; however some improvements are still necessary. For example, in the present protocol, the position of the X-rays source must be positioned at the level of the pelvis, and for that reason, the legs of the patient are irradiated unnecessarily. To address this problem, further researches should focus on the development of a more flexible procedure, including a new calibration frame, that will allow to precisely identify the position of all the pelvic and vertebral landmarks no matter

where is the position of the X-rays source with respect to the patient (Sect. 2.3). It is worth noting that this issue has been already widely studied by several research groups (Chapter 1), and therefore could be alleviated using existing technologies, such as EOS.

Spine and pelvis kinematics The analysis carried out in Sect. 2.3 points out that the proposed method can reasonably reproduce the internal kinematics of the spine. However, some limitations remain in the protocol, such as the maximum number of reflective markers that could be placed in the back, or worse still, the difficulty in accessing a gait analysis laboratory for a great number of surgeons. Thus, to simplify and accelerate the process, it is envisaged to develop a method to acquire the kinematics of the spine on the basis of inertial measurement units (IMU), that currently prove to be a cheap, accurate and easy to use.

Patient physiology The muscular model proposed in this thesis — although being simple — gives an idea of the importance of the muscles to study a pathology such as the scoliosis and for the calculation of the intervertebral efforts, in particular. Nevertheless, our study also suggests that we sometimes overestimate the importance of some muscles such as the Quadratus Lumborum to maintain of the spine, leading us unrealistic results as shown in the “full muscle-model” presented in Sect. 4.2.3. The complex functioning of muscles should be the subject of further studies such as a deeper investigation of those that are really requisitioned during gait and the refinement of the muscular model: in particular, to refine the force distribution of parallel groups of muscles, as explained in Sect. 2.5.

Further researches should look for other physiological elements that are responsible of the support of the spine. For example, it seems interesting to investigate the possible effect of the pressure inside the abdominal cavity, due to the abdominal compression.

Validation This is obviously the most important task to carry out in the near future. A model remains useless until it has not been confronted with reality.

As mention before, there are several experimental and ethical issues that make difficult to validate our model with patients. However, there exists real research avenues, namely:

- The achievement of a huge process of a sensitivity analyze of the model for well-targeted parameters, associated with the vertebral shape, the position of the muscular insertions or the mechanical properties of the bodies and the intervertebral discs. Nowadays, computer efficiency allows multi-dimensional parameter investigation, for which modern data analysis techniques can bring

high level outputs.

- The adding of extra information to our multibody model model, such as the energy consumption of the patient or more subjectively an image of his perception of pain as in [74] for sport activities,
- The calibration of our model with a series of scoliotic patients with identical and or distinct pathologies, and the validation of our results with patients who posses instrumented implants [18].

Publications

A

DYNAMIC ANALYSIS OF INTERVERTEBRAL EFFORTS IN SCOLIOSIS: A MULTIBODY MODELING APPROACH

Gabriel Abedrabbo¹, Paul Fisette¹, Maryline Mousny², Philippe Mahaudens², Christine Detrembleur², Jean-Claude Samin¹, Carl-Eric Aubin³, Maxime Raison⁴

¹Center for Research in Mechatronics (IMMC/CEREM)
Université catholique de Louvain (UCL)

Place du Levant 2, 1348 Louvain-la-Neuve, Belgium
e-mail: gabriel.abedrabbo@uclouvain.be, jc.samin@uclouvain.be,
paul.fisette@uclouvain.be

²READ Unit at the Faculty of medicine (UCL) and Orthopaedic Research Laboratory, Cliniques universitaires St-Luc Clinics (UCL)

Tour Pasteur, Avenue Mounier 53, 1200 Bruxelles, Belgium
e-mail: maryline.mousny@uclouvain.be, philippe.mahaudens@uclouvain.be,
christine.detrembleur@uclouvain.be

³Department of mechanics, École Polytechnique de Montréal
& Center for Research of Ste-Justine Clinics
2500 Chemin de Polytechnique, H3T 1J4, Montréal QC, Canada
e-mail: carl-eric.aubin@polymtl.ca

⁴École Polytechnique de Montréal & CRME-CHU Sainte-Justine
5200 rue Bélanger Est, H1T 1C9, Montréal QC, Canada
e-mail: maxime.raison@polymtl.ca

Keywords: Scoliosis, Multibody Dynamics, Kinematic Identification, Inverse Dynamics.

Abstract. *A biomechanical model, using the multibody dynamics approach, is developed to calculate the intervertebral efforts in spine. The latter are computed, via inverse dynamics, using kinematic information from gait. Optimization process is required in order to convert that information, expressed in absolute coordinates from the optometric sensors, into relative joint coordinates, in order to calculate joint forces and torques.*

Significant differences were found in the intervertebral efforts between scoliotic and healthy subjects, but also before and after spinal fusion of scoliotic patients. This work shows the changes in spine internal efforts before and after surgery. The perspective of this research is the development of a complementary tool for the orthopedic surgery planning and the evaluation in rehabilitation of scoliotic patients.

1 INTRODUCTION

The evaluation of internal forces and torques in the human body in motion could provide valuable information for the evaluation and follow-up of subjects with musculo-skeletal pathologies, such as scoliosis. But these data are still difficult to measure accurately. In this challenging context, the objective of this study is to check the hypothesis that there exist different dynamical behaviours and effects in the spine between a healthy subject and a scoliotic patient before and after spine fusion surgery. The chosen motion for this study is gait, because of the existing state-of-the-art survey enabling to compare healthy and scoliotic subjects during this motion (e.g. [1]). The approach to compute the joint efforts along the spine is a dynamical multibody modeling of the human body, of which refinement is presently investigated through an ongoing research.

This work will analyze and compare the mechanics of normal and scoliotic spine and aims at developing a method to assess the intervertebral efforts.

2 BACKGROUND

The SRS (Spine Research Society) definition of adolescent idiopathic scoliosis (AIS) is a structural lateral curvature of the spine that creates a thoracic and/or lumbar asymmetry on forward bending (a.k.a. the "Adams forward bend test") combined with a curve of at least 10° as measured by the Cobb technique on a standing radiograph of the spine with associated vertebral rotation along the vertebral direction (torsion). When scoliosis is defined as a Cobb angle of at least 10 degrees, epidemiologic studies estimate that 1% to 3% of children aged 10 to 16 years will have some degree of spinal curvature, with the vast majority of curves requiring no intervention [8, 9, 10]. Some scoliosis will nevertheless be rapidly evolutive and surgical intervention — spine fusion — will be proposed. Nowadays there are still controversies regarding the need for such treatment [11], the ideal surgical instrumentation, and the long-term outcome of untreated and fused scoliotic patients. Figure 1 shows a spine with idiopathic scoliosis before and after surgery

The envisaged model aims at calculating the intervertebral forces and torques acting in the spine. There are 6 degrees of freedom between each pair of vertebrae, but, during gait, the displacement between them will be considered as small around the initial configuration.

3 MATERIALS AND METHODS

The intervertebral forces and torques are calculated according to the global procedure of (Figure 2).

3.1 Subjects analyzed

Gait analysis was obtained from one healthy subject and one scoliotic patient (before and six months after spine surgery) walking on a treadmill at 4 km/h . Three situations were thus analyzed:

1. The healthy subject (69 kg) ;
2. The scoliotic patient (50 kg) before spine surgery, with a Cobb angle of 41° between T10 and L2, and an apical axial rotation of 30° at T12 ;
3. The same scoliotic patient (50 kg) six months after spine fusion surgery over the T11-L2 segment. The Cobb angle was corrected to 10° and the maximum axial rotation to 5° .

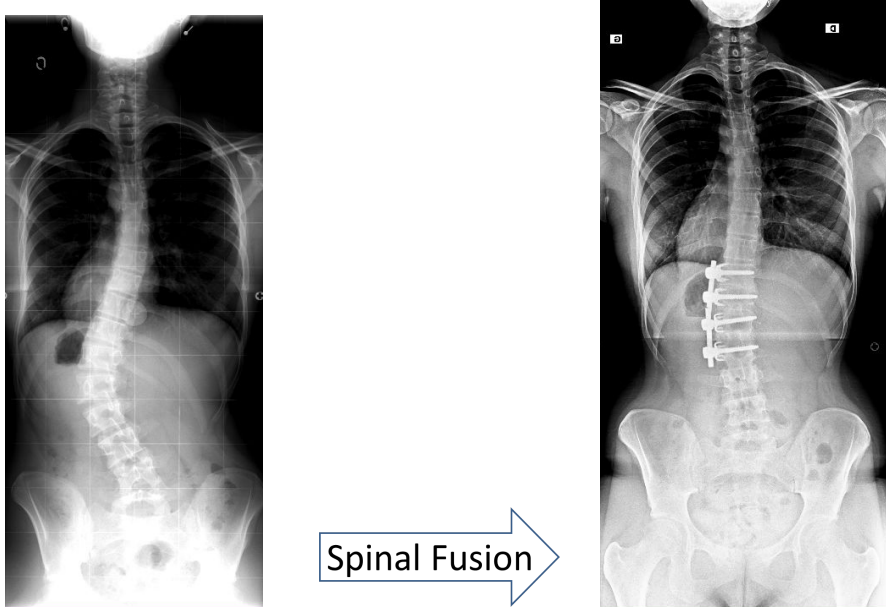


Figure 1: Idiopathic scoliosis before and after spinal fusion.

3.2 Data acquisition

For this work the following acquisition system was used:

1. Optokinetic sensors (*Elite-BTS*), which are reflective markers, whose localization in space is recorded at a frequency of 100 Hz;
2. A treadmill equipped with strain gauges, measuring the 3D external forces applied to each foot ;
3. Postero-anterior and lateral radiographs of the spine, enabling a static 3D reconstruction of the spine geometry [2].

3.3 Joint kinematic identification

The joint coordinates q , velocities $\dot{q}$ and accelerations $\ddot{q}$ are numerically determined via an optimization process that estimates the joint coordinates q of the multibody model that best fit the experimental joint kinematics as in [7]. The process, summarized in (Figure 3), must also include an estimation of the “non-sensored” vertebrae displacement, as well as a filtering and interpolating module to capture velocities $\dot{q}$ and accelerations $\ddot{q}$. This task, which is fundamental for the present research, is the core activity of our ongoing work. The optimization problem is formulated as a nonlinear least-square problem. The cost function for an instant t_k can be written as follows:

$$f_{cost}(t_k) = \sum_{s=1}^{n_{sens}} |X_{mod,s}(q(t_k)) - X_{exp,s}(t_k)|^2 \quad (1)$$

where

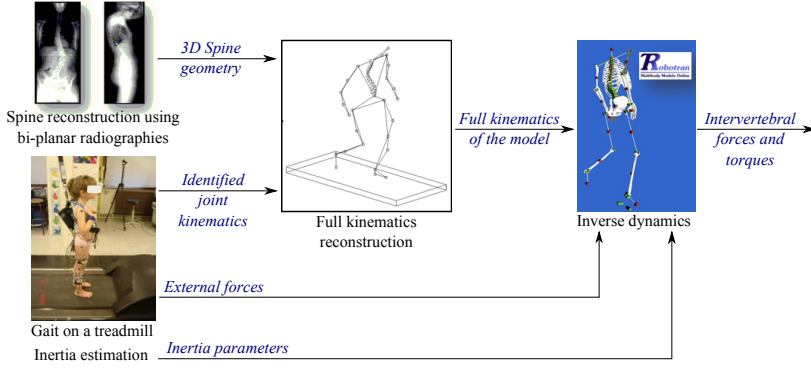


Figure 2: General method to quantify the intervertebral efforts during gait: data are in *italic* type; processes and protocols are in normal type.

- the index $s = 1, \dots, n_{sens}$ indicates the optokinetic sensor;
- $q(t_k)$ is the vector of joint relative coordinates of the MBS at the time instant t_k ;
- $X_{mod,s}(q(t_k))$ is the Cartesian coordinate of the s^{th} optokinetic sensor at time instant t_k , obtained from $q(t_k)$, using the direct kinematic model of the MBS ;
- $X_{exp,s}(t_k)$ is the Cartesian coordinate of the s^{th} optokinetic sensor at time instant t_k , provided by the experiments.

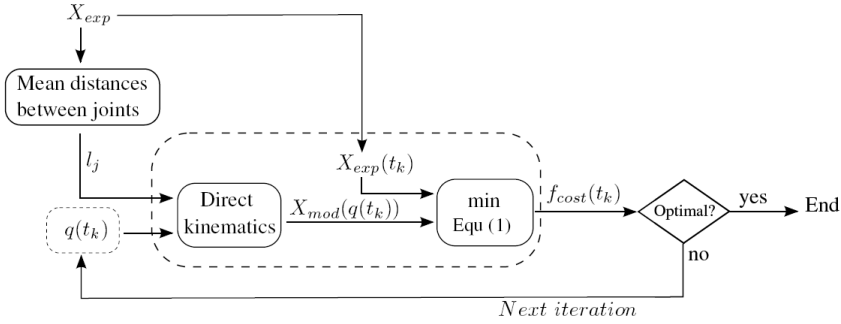


Figure 3: Kinematic optimization process at time instant t_k [7].

3.4 Inverse dynamics

Using a recursive Newton-Euler formalism [4], a 3D multibody model provides the vector Q of the joint forces and torques — in particular those acting between the vertebrae — during gait:

$$Q = f(q, \dot{q}, \ddot{q}, F_{ext}, M_{ext}, g), \quad (2)$$

where f is a function of the kinematics ($q, \dot{q}, \ddot{q}$) and represents the inverse dynamic model of the human body, subject to external forces F_{ext} , torques M_{ext} and gravity. These equations are symbolically generated by the Robotran multibody software [4], developed at the CEREM (UCL), which allows to straightforwardly interface these equations with any numerical process, such as the kinematic identification process presented above and any simulation of the system (direct dynamics, optimization,...).

4 PRELIMINARY RESULTS

Significant differences in kinematics and Ground Reaction Forces (GRF) were observed between the subjects, leading to different internal effort behaviour in magnitude, maxima and minima when normalized to the subject mass. Considering for example the normalized antero-posterior torques (NAPT) between vertebrae, Figure 4 shows significant differences between the three cases : the maximal absolute magnitude of the NAPT was higher for the scoliotic patient before surgery (2.5 Nm/kg) than after surgery (1.3 Nm/kg) and than the healthy subject (1.5 Nm/kg); the NAPT maxima and minima were rather symmetrical for the healthy subject (-1.5 and 1.5 Nm/kg, resp.), while the scoliotic patient before surgery presented a large shift of NAPT maxima and minima (2.5 and 1.9 Nm/kg, resp.), image of the disequilibrium in the left-right balance of the measured GRF; this shift was partially rectified for the scoliotic subject after fusion surgery (-1.3 and 1.1 Nm/kg, resp.); finally, the NAPT general patterns presents clear “crenel-type” left-right swings for the healthy subject, a more perturbed and more “ramp-type” behaviour for the scoliotic patient before surgery, and finally a rather “crenel-type” swing and smoother gait for the scoliotic patient after surgery.

5 DISCUSSION AND CONCLUSION

In this study, different dynamic behaviours — in magnitude, min, max and pattern — were found between a healthy subject and a scoliotic patient before and after spine fusion surgery. Particularly, the magnitudes, maxima and minima of the NAPT values were found higher for the scoliotic patient before instrumentation. These results suggest that intervertebral forces could be modified in scoliotic spine. The importance of this phenomenon on the evolution process of the scoliosis has yet to be investigated. Presently, two investigations are performed in parallel:

- The refinement of the present lower member multibody modeling and processing, which will be replaced by an equivalent model for which the pelvis kinematics will be constrained (at position, velocity and acceleration level) using an accurate kinematic identification, in order to avoid measuring the GRF applied to the feet. The latter will be used for an “a posteriori” model validation (Inverse dynamics consistency).

- The clinical validation — on a larger number of subjects — of the hypothesis that there exist different spinal dynamic behaviours between healthy and scoliotic subjects.

6 FUTURE WORK

Up to now, the movements between vertebrae, considered as negligible, have been constrained. Thus, the internal force/torque become reaction forces which can be straightforwardly calculated by the inverse dynamics. A new model is currently developed in order to consider a more refine intervertebral kinematics of the column. As already mentioned, some difficulties arise since the amounts of degrees of freedom exceed the sensors information. Radiographies in different positions and stress conditions will allow us to overcome this issue.

A validation of the model in scoliotic and non scoliotic subjects will be considered in the future.

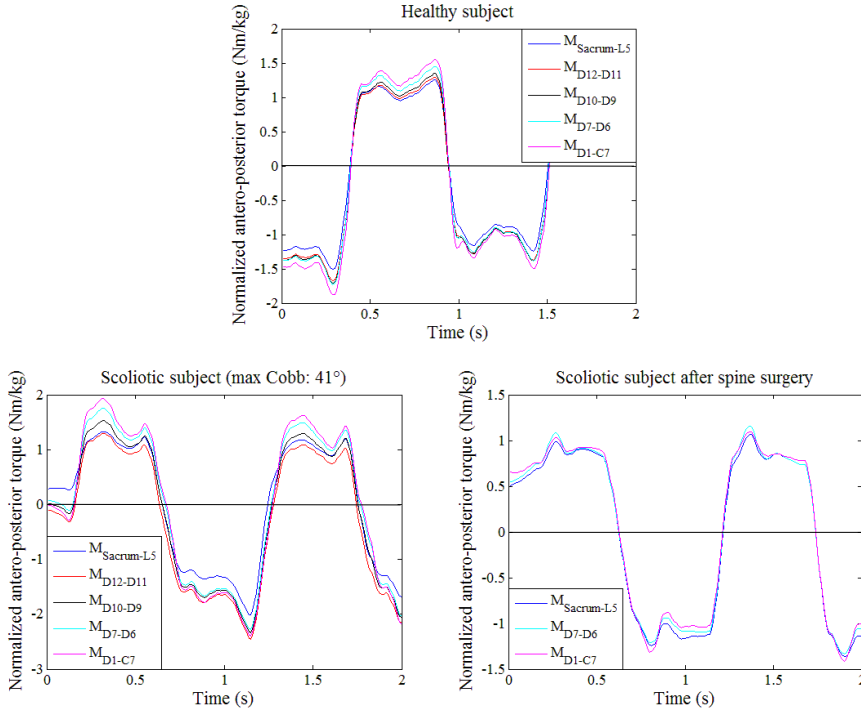


Figure 4: Considering the healthy subject and the scoliotic patient before and after instrumentation surgery, all walking at 4km/h on a treadmill: comparison between antero-posterior intervertebral torques corresponding to the spine extremities ($M_{\text{sacrum-L5}}$ and $M_{\text{T1-C7}}$), the apices ($M_{\text{T7-T6}}$ and $M_{\text{T12-T11}}$) and the maximal lateral inflexion ($M_{\text{T10-T9}}$) of the scoliotic patient.

It will include the walking analysis previously done, but it will also consider static bending and traction configuration to reinforce the model validity.

Finally, as previously done in [7] for the arm, one of the final goals of this research is to include and calculate, via an EMG-based optimization process, muscular forces which intervene in the spine dynamic analysis.

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A multibody-based approach to the computation of spine intervertebral motions in scoliotic patients

Gabriel Abedrabbo^{*}, Pierre-Antoine Absil[#], Philippe Mahaudens[†], Christine Detrembleur[†],
Maxime Raison[&], Maryline Mousny[§], Paul Fiset^{*}

^{*} Centre for Research in Mechatronics
Institute of Mechanics, Materials, and Civil
engineering
Université catholique de Louvain
Place du Levant, 1, B1348, Belgium
[gabriel.abedrabbo](mailto:gabriel.abedrabbo@uclouvain.be), [paul.fisette](mailto:paul.fisette@uclouvain.be)@uclouvain.be

[#] Institute of Information and Communication
Technologies, Electronics and Applied
Mathematics
Université catholique de Louvain
Avenue Georges Lemaître, 4, B1348, Belgium
pa.absil@uclouvain.be

[&] Institute Research Chair in Pediatric Rehabilitation
Engineering
École Polytechnique and CRME – Sainte-Justine
5200, rue Bélanger Est, H1T 1C9, Canada
maxime.raison@polymtl.ca

[§] Service d'Orthopédie et de Traumatologie de
l'appareil locomoteur
Cliniques universitaires Saint-Luc
Avenue Hippocrate 10, B-1200, Belgium
maryline.mousny@uclouvain.be

[†] Faculté des sciences de la motricité
Institute of NeuroSciences
Université catholique de Louvain
Avenue Mounier 53, 1200, Belgium
[philippe.mahaudens](mailto:philippe.mahaudens@uclouvain.be), [christine.detrembleur](mailto:christine.detrembleur@uclouvain.be)@uclouvain.be

ABSTRACT

In idiopathic scoliotic patients, dynamical intervertebral efforts may be correlated with the spinal deformity. The quantification of these efforts may be useful for the orthopedic surgeons to plan their surgical procedure, and is the final objective of the present research. We first focused on the geometrical reconstruction of the spine and especially on the rotation sequences between vertebrae using a multibody approach. The reconstruction process was performed in the standing position and in bending postures, using an optimization process based on geometrical data reconstructed from radiographs. The obtained results will serve as input for the subsequent gait motion analysis, for which the limited set of geometrical information must be compensated. Preliminary results related to intervertebral forces are presented for a simple body motion, comparing inverse and direct dynamical approaches.

1. INTRODUCTION

1.1. Medical context

The SRS (Spine Research Society) definition of adolescent idiopathic scoliosis (AIS) is a structural lateral curvature of the spine that creates a thoracic and/or lumbar asymmetry on forward bending (a.k.a. the "Adams forward bend test") combined with a curve of at least 10 degrees as measured by the Cobb technique on a standing radiograph of the spine with associated vertebral rotation along the vertebral direction (torsion). When scoliosis is defined as a Cobb angle of at least 10 degrees, epidemiologic studies estimate that 1% to 3% of children aged 10 to 16 years will have some degree of spinal curvature, with the vast majority of curves requiring no intervention [4, 6, 11]. Some scoliosis will nevertheless rapidly evolve and surgical intervention—spine fusion—will be proposed. Nowadays there are still controversies regarding the need for such treatment [9], the ideal surgical instrumentation and the long-term outcome of untreated and fused scoliotic patients. Figure 1 shows a patient with adolescent idiopathic scoliosis before and after surgery. The final goal of this research is the quantification of the intervertebral forces and torques acting in the spine. There are 6 degrees of freedom between each pair of

vertebrae; however, during a smooth gait, the displacement between them will be considered as small around their static configuration.

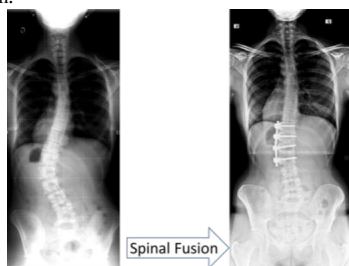


Figure 1. Idiopathic scoliosis before and after spinal fusion [1]

1.2. Project issues and objectives

During gait, a 30% increase of the energetic cost was observed in adolescent idiopathic scoliosis patients (AIS). A possible reason would be their additional efforts to maintain their spine and control the upper body when walking [5]. The quantification of intervertebral effort during gait could help to better understand this excessive cost; pilot studies already suspected a substantial difference in intervertebral efforts between AIS and healthy subjects [7]. These observations were however based on a model with no intervertebral motions and on a limited set of experimental data. Therefore the present research project aims at developing a predictive multibody model and a pragmatic protocol to quantify the intervertebral efforts (forces/torques) in idiopathic scoliotic patients during standing position and smooth gait..

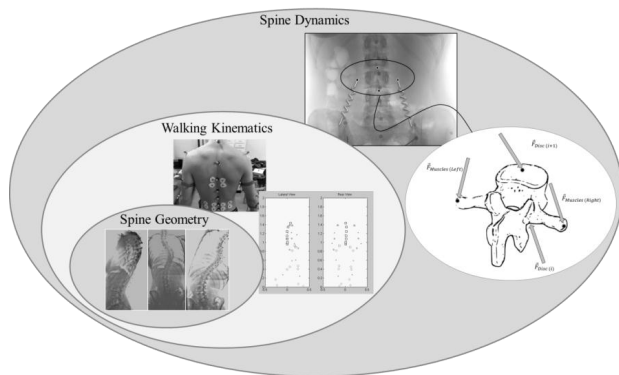


Figure 2. General process — from the acquisition of the spine geometry (subject of the present paper) as input for the next two steps: the kinematics and then the dynamics during gait, to quantify intervertebral forces.

As shown in Figure 2, the computation of the intervertebral and muscular forces will follow three main steps:

1. the reconstruction of the spine geometry using the standing posteroanterior xray and the xray in the bended position;
2. the prediction of the complete kinematics of the spine during gait, via a *kinematic* multibody model;
3. the computation of the intervertebral forces and torques via an inverse *dynamic* model in which the main muscles action during gait will be included.

The present study will focus on the *geometrical* reconstruction of the spine, with a particular attention on the rotation sequence between vertebrae, being a fundamental input for the subsequent kinematic gait analysis (see Figure 2).

This paper is organized as follows.

- Section 2 describes the data acquisition process (2.1), the multibody model (2.2) and finally the spine reconstruction process for standing-up and bending configurations (2.3).
- In Section 3, the spine reconstruction results are analyzed and compared for a fully or partially available set of data (3.1). Then, considering a simulated “up and down” body motion, two dynamic approaches are compared to assert for the advantage of an inverse dynamic model with respect to a direct dynamic one, for the calculation of the intervertebral forces (3.2).

2. METHODOLOGY

2.1. Materials

- Three healthy subjects and one AIS studied. Posteroanterior standing and bending spine xrays were performed in each subject..
- The acquisition system included a bi-planar radiograph of the spine in standing position, from which we extracted the absolute position of the center of each intervertebral disc (see Figure 3).
- The Spine 3D program [3] is used for the measurement of the intervertebral discs position on the bi-planar radiograph.
- Finally, ROBOTRAN [8] was used for the generation of the multibody model in symbolic form

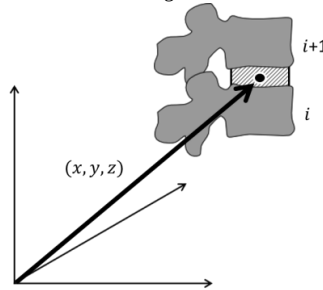


Figure 3. Position of the center of the disc expressed in the inertial frame

2.2. Multibody model

Starting from the static configuration of the scoliotic patient in a standing position, a multibody representation of the spine is automatically generated by the ROBOTRAN program [8]. The model contains 18 bodies: the moving pelvis and the 17 vertebrae. To take large scoliotic deformations into account, we suggest representing the intervertebral kinematics by 9 relative coordinates (visible in Figure 4-left). The reason is that, in addition to the 6 degrees of freedom representing the relative motion of each intervertebral disc via generalized coordinates q , we have added 3 fixed rotations to capture the initial deformation of the spine in the standing position. These rotations can be rather precisely computed from the bi-planar radiography process.

For the gait modeling, the following kinematic hypothesis has been proposed in collaboration with the surgeons: relative to the pelvis, the vertebrae motions remain small with respect to their configuration computed for the standing up posture. This hypothesis (which has an obvious interest for modeling) is pertinent for the surgeons and physiotherapist since they only consider smooth gait on treadmills.

Thus, to fit the multibody model to the gait postures, absolute sensors (denoted “S” in Figure 4) characterizing the reflective markers are introduced into the Robotran model, according to their location also visible on the radiography. They will be the basis of the future kinematic optimization process.

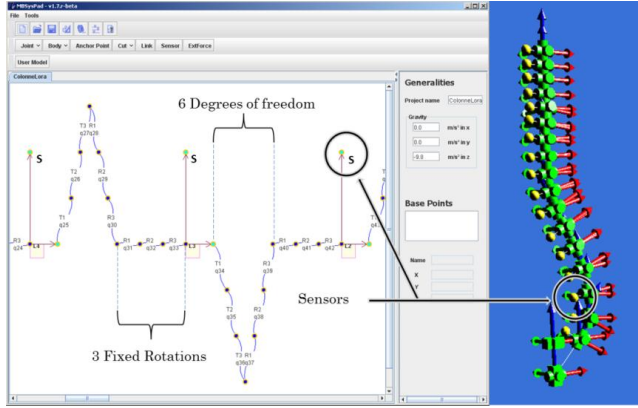


Figure 4. ROBOTRAN interface – Left : 2D topology (3 vertebrae), Right : corresponding 3D model [8]

2.3. Spine reconstruction

2.3.1. Deterministic solution

Let us note that in case of a full data information (i.e. all the intervertebral disc location), the problem is deterministic, i.e. it corresponds to a full set of measurement data. As we shall see later on, this will not be the case anymore during gait for which additional relations or constraints will be added.

First, from the measured data (Figure 5), we can easily calculate the distance between each successive pair of disc centers.

Then, knowing this intervertebral distance, an optimization algorithm allows us to fit the geometrical multibody model to the absolute position of the intervertebral disc centers measured via Spine 3D, as shown in the snapshot of Figure 5 (right), by minimizing the following cost function:

$$f_{cost} = \sum_{s=1}^{n_{vertebrae}} ||\vec{x}_{mod,s}(q) - \vec{x}_{exp,s}||^2 \quad (1)$$

where:

$\vec{x}_{mod,s}$ is the position vector of the intervertebral disc calculated by the multibody model,

$\vec{x}_{exp,s}$ is the position vector of the intervertebral disc measured from radiograph.

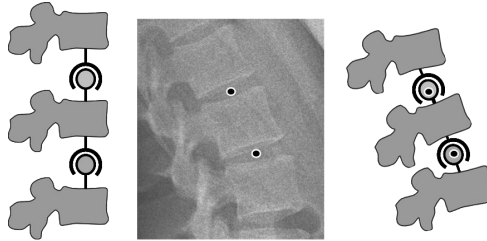


Figure 5. Spine bending reconstruction

In the previous bending calculation, the spine torsion has been deliberately disregarded (think about the centroidal axis of a flexible beam). Intervertebral torsion angle Θ is now computed via a second step by projecting on the inertial frame $\{\hat{I}\}$ the line joining the center of the disc and the spinous process (Figure 6). The resulting angles Θ_{exp} and Θ_{mod} , respectively computed from the experimental data and via the

multibody model, can be compared and their difference minimized via optimization to evaluate the torsion along the spine according to the following cost function:

$$f_{cost} = \sum_{s=1}^{n_{vertebrae}} ||\tan(\Theta_{mod,s}(q)) - \tan(\Theta_{exp,s})||^2 \quad (2)$$

where:

$\Theta_{mod,s}$ is the angle between the projection of the vector $\overrightarrow{VV'}$ onto the plane $\{\hat{l}_1, \hat{l}_2\}$ calculated by the multibody model and the unit vector $\hat{l}_1$.

$\Theta_{exp,s}$ is the angle between the projection of the vector $\overrightarrow{VV'}$ onto the plane $\{\hat{l}_1, \hat{l}_2\}$ measured from radios and $\hat{l}_1$.

$\overrightarrow{VV'}$ is the vector joining the centre of the disc and the spinous process.

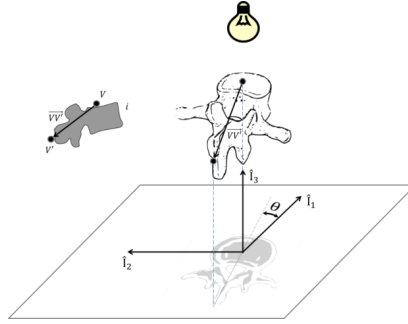


Figure 6. Spine torsion reconstruction: projection on the inertial frame

However, in case of severe scoliotic deformities, the inertial frame is not suitable to perform this projection. Thus, we propose to introduce a new floating frame $\{\hat{X}\}$ according to Figure 7 and to project the $\overrightarrow{VV'}$ vector onto the plane $\{\hat{X}_1, \hat{X}_2\}$ normal to vector $\vec{u}$ which is representative of the direction of the local spine centroidal axis: vector $\vec{u}$ joins the upper center of the disc of vertebra i and vertebra $i-2$ (see (Figure 7)).

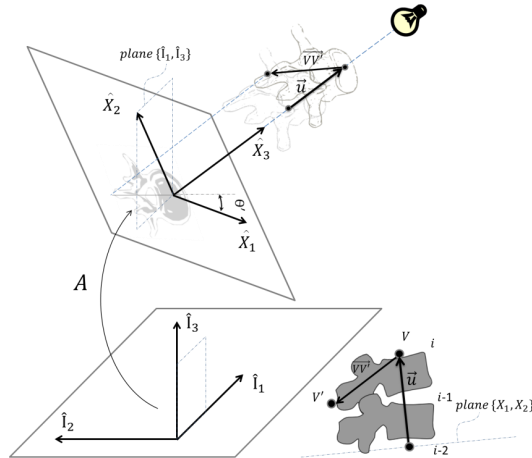


Figure 7. Spine torsion reconstruction: projection on a floating frame

Thus, for each torsion angle calculation, the 3x3 rotation matrix A between the inertial frame $\{\hat{I}\}$ and the floating frame $\{\hat{X}\}$ can be computed

$$[\hat{X}] = A[\hat{I}] \quad (3)$$

by choosing

$$\hat{X}_3 = \frac{\vec{u}}{\|\vec{u}\|} \quad (4)$$

and by orienting the frame (around $\hat{X}_3$) in such a way that $\hat{X}_2$ belongs to the $\{\hat{f}_1, \hat{f}_3\}$ plane, i.e.:

$$\hat{X}_2 \cdot \hat{f}_2 > 0. \quad (5)$$

Finally, the cost function (2) can be rewritten in terms of the angles Θ' , referring now to frame $\{\hat{X}\}$.

$$f_{cost} = \sum_{s=1}^{n_{vertebrae}} \|\tan(\Theta'_{mod,s}(q)) - \tan(\Theta'_{exp,s})\|^2 \quad (6)$$

The static bended configuration being necessary for the subsequent gait kinematic analysis, additional radiographs are performed in lateral bended-position as shown in Figure 8 for a non scoliotic patient (left) and a scoliotic one (right).

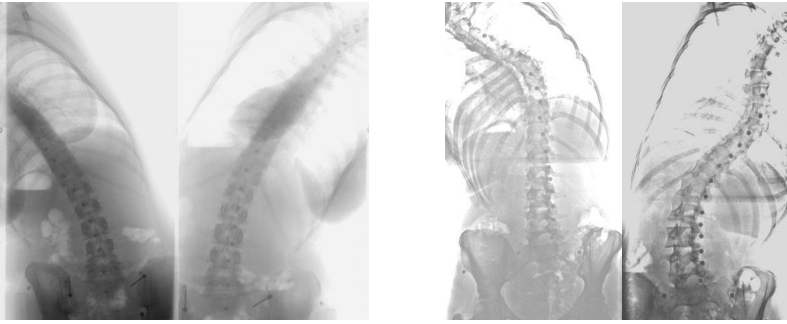


Figure 8. Lateral bending radiographs

The reconstruction of those bended spines (static posture) can also be done in a deterministic way, using the same cost function (1) as previously done for the standing up reconstruction.

2.3.2. Non deterministic solution

During gait, due to experimental limitations, it will be not possible to obtain the same amount of information we obtain from radiograph, leading to an underdetermined problem. In our last protocol we used maximum 1 reflective marker every two vertebrae (see Figure. 9): the optimization process based on previous cost functions is not deterministic anymore.

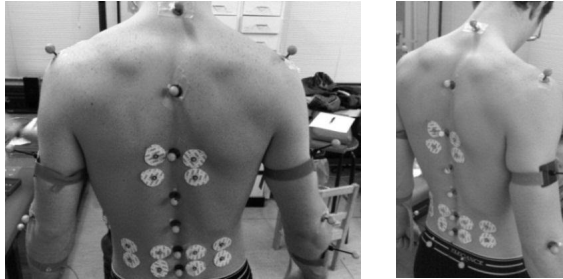


Figure 9. Reflective markers on spinal process

For this reason, in order to establish a predictive model to calculate the kinematics of the spine, we reconstruct its geometry based on experimental data [2] and a limited number of points n_{points} measured from the bending radiographs. The global cost function is now composed of two weighted terms:

$$f_{cost} = w_1 * f_{cost1} + w_2 * f_{cost2} \quad (7)$$

assume that the position measurements are available for vertebrae number $s_1, \dots, s_{n_{points}}$

$$f_{cost1} = \sum_{s=1}^{n_{points}} ||X_{mod,s}(q) - X_{exp,s}||^2 \quad (8)$$

and

$$f_{cost2} = \sum_{i=1}^{n_{dof}} ||\Delta q_{mod}(i) - \Delta q_{exp}(i)||^2 \quad (9)$$

In these expressions,

Δq_{mod} denotes the variation of the generalized coordinate q between the standing up position (q_{stn}) and the bended position (q_{bend}), in the multibody model

$$\Delta q_{mod} = q_{bend} - q_{stn} \quad (10)$$

and Δq_{exp} are evaluated from the work of Christophy et al. [2] which suggests interpolated values of the total bending angle β to estimate the intervertebral angular deviation Δq_{exp} :

$$\Delta q_{exp} = f(\beta) \quad (11)$$

Due to the limited number of measurement data, there exist an infinite number of solutions which minimize cost function f_{cost1} (8) for the generalized coordinates q . A favorable consequence of the addition of f_{cost2} in f_{cost} is to reduce the set of minimizers to a single q , solution of an overdetermined problem based on function (7). Intuitively, minimizing f_{cost} amounts to finding q that is as compatible as possible with the available position measurements and with the angular variations predicted in [2].

3. RESULTS

3.1. Geometrical reconstruction

Using the sequence described in Section 2.3.1 in which flexion (Figure 5) and torsion (Figure 7) are decoupled, the mathematical problem is deterministic: insofar as the numerical convergence is ensured, a perfect numerical fit is obtained between the model and the measurements, as illustrated in Figure 10 during the standing position in healthy (Figure 10A) and scoliotic (Figure 10B) subjects. In this figure, circles denote the experimental values and the continuous line segments refer to the numerical solution.

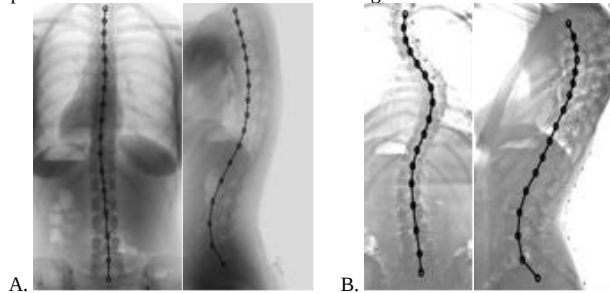


Figure 10. Coronal and sagittal views in standing up position: 3D reconstruction in a healthy subject (A) and scoliotic (B) patient (circles = experimental data, line segments = numerical solution obtained by mathematical algorithm)

Exaggerated bending configurations shown in Figure 11 have been performed for two reasons.

First, large bending has allowed us to validate the above numerical process on extreme configurations, to confirm that the numerical results are reliable: once again, a good fit is observed between circles and continuous line (see Figure 11).

Secondly, as explained in Section 1.2, we need to capture the spine deformation during “smooth gait on treadmill”, for which we assume that additional spine deformations will remain small in bending and torsion. To capture these “dynamical” deformations, reflective markers located on some specific vertebrae along the spinous processes will be used (Figure 9). Since it is rather complicated to place a marker on each vertebra, let us predict the effect of a lack of geometrical information on the spine reconstruction. The static bending in Figure 11 emulates this situation: dark circles denote known points (2 and 4 respectively), whereas empty circles refer to unknown ones. According to Section 2.3.2, the cost function (7) is used to solve this problem.

In Figure 11, we can observe that passing from 2 to 4 known points clearly improves the results (black dots) for both healthy and scoliotic subjects. Although the fit is not as good as the deterministic case (represented by continuous segments), this result is encouraging for two reasons: first, we are able to fix more than 4 markers on the spine process (Figure 9) and secondly, the bending deformation during “smooth gait on treadmill” will be obviously far smaller than the static bending of Figure 11.

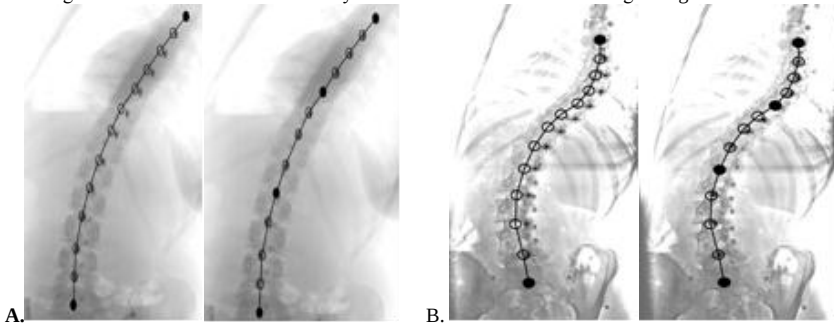


Figure 11 Lateral bending in non-scoliotic (A.) and scoliotic subjects (B.) - circles: measured data, line segments: deterministic solution, black circles: emulated known points, dots: solution based on known data and weighting coefficients from [2].

3.2. Intervertebral Forces

Another issue we would like to clarify relates to the goal of this study (see. Figure. 2, spine dynamics) dealing with the quantification of the intervertebral forces. More precisely, the following questions often arises: what value to choose for the intervertebral stiffness to get a reliable dynamic model of the spine ?

What is the sensitivity of the system dynamics with respect to those parameters?

In this respect, let us present here a preliminary result related to intervertebral normal and shear forces during a simple but realistic motion : a vertical “up and down” movement of the body introduced in the multibody model via an harmonic displacement imposed to the pelvis (amplitude : 0.02 m; frequency : 0.5 Hz and 2 Hz respectively):

Two dynamic approaches are compared to assert for the advantage of an inverse dynamic model with respect to a direct dynamic one, for the calculation of the intervertebral forces.

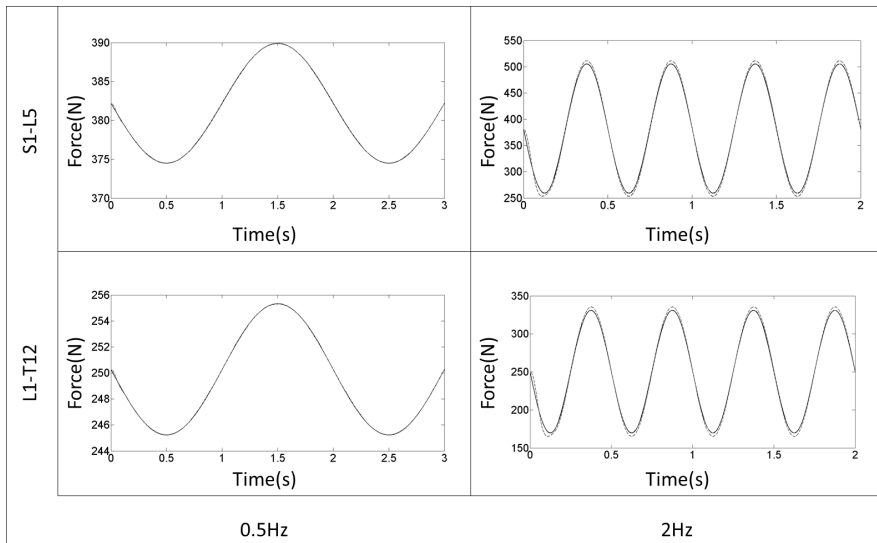


Figure 12. Intervertebral normal forces - continuous line: inverse dynamic model, dotted line: direct dynamic model

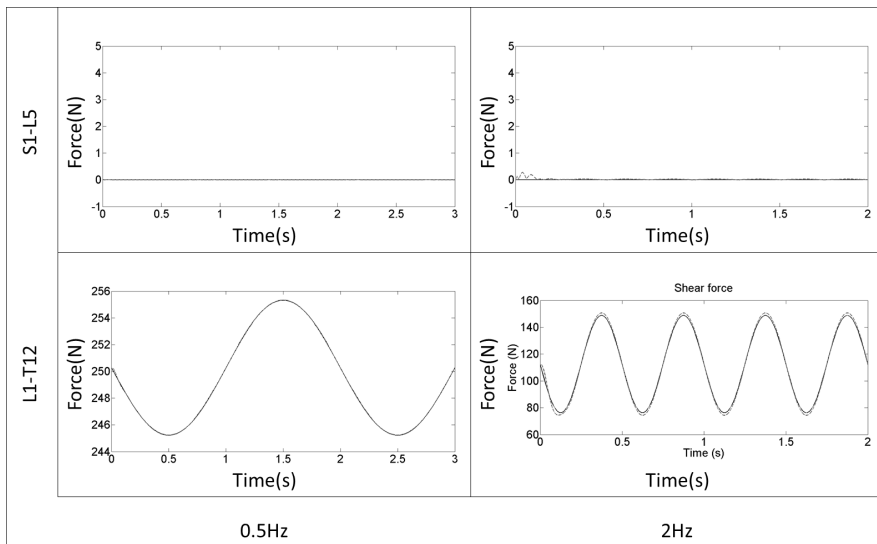


Figure 13. Intervertebral shear forces - continuous line: inverse dynamic model, dotted line: direct dynamic model

Figure 12 and 13 show a comparison between the inverse dynamic results for which rigid intervertebral discs have been considered (infinite stiffness) and those obtained via a direct dynamic model in which the intervertebral disc stiffness has been introduced in the lumbar spine according to the values proposed by White et al [10], as well as a small damping force. We can observe a good match between the two

models, which suggests for future investigations, that inverse dynamics is largely sufficient to quantify those forces for smooth gait walking. This would bring two significant advantages. First, it allows us to do without intervertebral stiffness, whose value is difficult to estimate. Secondly, inverse dynamics is far more efficient than direct dynamics in terms of computational time, which is not negligible in the frame of the development of a future diagnostic tool.

4. CONCLUSIONS AND PERSPECTIVES

In this paper, the geometrical reconstruction of the spine has been addressed. To make the optimization process converge towards a reliable solution, we suggest decoupling the bending and torsion deformations. The method has been applied in healthy and AIS subjects during standing position with possible bending; a very good fit was obtained between the experiment and the multibody model, with a full set as well as with a partial set of experimental measurement data. Some preliminary results have also been shown to assert for the advantage of an inverse dynamic model for the calculation of the intervertebral forces.

Large bending being rather well captured with lacking information, they will serve as interpolated input for the spine reconstruction during gait, the future step of this research, via a standard data acquisition protocol based on infra-red cameras and reflective markers located along the spinous processes. Finally, muscular forces will be included to the model, to quantify the intervertebral forces.

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A multibody-based approach to the computation of spine intervertebral motions in scoliotic patients

Gabriel Abedrabbo ^{a,1}, Paul Fisette ^a, Pierre-Antoine Absil ^b, Philippe Mahaudens ^c,
Christine Detrembleur ^c, Maxime Raison ^d, Xavier Banse ^e, Carl-Eric Aubin ^f,
Maryline Mousny ^e

^a*Center for Research in Mechatronics (CEREM), Institute of Mechanics, Materials, and Civil Engineering, Université catholique de Louvain, Louvain-la-Neuve, Belgium*

^b*Institute of Information and Communication Technologies, Electronics and Applied Mathematics (ICTEAM), Université catholique de Louvain, Louvain-la-Neuve, Belgium*

^c*Institute of NeuroScience (IoNS), Université catholique de Louvain, Brussels, Belgium*

^d*Research Chair in Pediatric Rehabilitation Engineering, École Polytechnique and CRME – Sainte-Justine UHC, Montreal QC, Canada*

^e*Service d'Orthopédie et de Traumatologie de l'appareil locomoteur (ORTO), Cliniques universitaires Saint-Luc, Brussels, Belgium*

^f*Department of mechanics, École Polytechnique de Montréal & Center for Research of Ste-Justine UHC, Montréal QC, Canada*

Abstract. In idiopathic scoliotic patients, dynamical intervertebral efforts acting between vertebrae seem to be correlated with the spinal deformity. The quantification of these efforts, which is useful for the orthopedic surgeons to set up their surgical planning is the final objective of the present research. As a first step, we focus in this contribution on the geometrical reconstruction of the spine and especially on the rotation sequences between vertebrae in a multibody approach. The reconstruction process is performed in the standing position with possible bending, using an optimization process based on geometrical data reconstructed from radiographs. The obtained results will serve as input for the subsequent gait motion for which the limited set of geometrical information must be compensated.

Keywords. Multibody modeling, Idiopathic scoliosis, Spine geometry, Gait

1. Context and Objectives

During gait, an increase of 30% of the energetic cost was observed in adolescent idiopathic scoliotic patients (AIS), a possible reason being their additional efforts to maintain their spine and control the upper body when walking [1]. The quantification of intervertebral effort during gait can help us to better understand this excessive cost; pilot studies already suspected a substantial difference in intervertebral efforts between AIS and healthy subjects [2]. These observations being based on a model with no intervertebral motions and on a limited set of experimental data, the present research work aims at developing a predictive multibody model and a pragmatic protocol to quantify the intervertebral efforts (forces/torques) in idiopathic scoliotic patients during

standing position and gait. As first stage of this project, the objective of the present contribution is to obtain a *geometrical* reconstruction of the spine, with a particular emphasis on the rotation sequences between vertebrae, using a multibody approach [3]. Two static configurations are considered: standing up vertically and bended. Once the spine geometry is modeled, the two following steps can take place: first, the spine reconstruction during gait via a *kinematic* multibody model, and finally the computation of the intervertebral forces and torques via an inverse *dynamical* model in which the main muscles action during gait will be included.

2. Material and Method

- Three healthy subjects and one AIS were standing up vertically and bended. The acquisition system included bi-planar radiographs of the spine in standing position, from which we extracted the x-y-z position of the center of each intervertebral disc (see Fig. 1), via the *Spine 3D* program [4].
- The ROBOTRAN program [3] is necessary for this first step to build the geometrical multibody model of the human body (upper part) on the basis of the position of the intervertebral discs and of the spinous processes previously measured via *Spine 3D*.

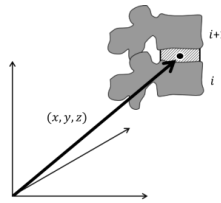


Fig.1. Intervertebral Cartesian position

The 3D reconstruction of the rotation sequences between vertebrae are performed according to the three following steps, using ROBOTRAN [3]:

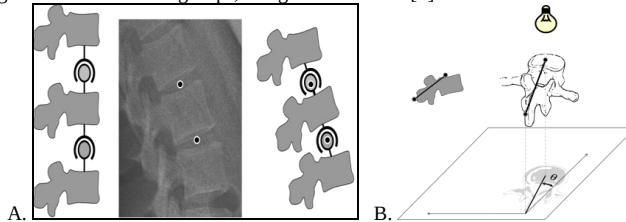


Fig. 2. Reconstruction process – Spine flexion (A.); Spine torsion (B.).

- First, from the measured data (see Fig. 1), we calculate the distance between each successive pair of disc centers.
- Secondly, knowing the intervertebral distance, an optimization algorithm allows us to fit the x-y-z position of the intervertebral disc centers, measured via *Spine 3D*, to the multibody model as shown in the snapshot of Fig. 2A.
- Finally, each intervertebral torsion angle θ is obtained by projecting on a suitable plane (the horizontal one for sake of clarity in Fig. 2B) the line joining the center of the disc and the spinous process (see Fig. 2B); the resulting angles θ_{exp} and θ_{mod} , respectively computed from the experimental data and via the multibody model, can be compared and their difference minimized via optimization to evaluate the torsion along the spine.

3. Results and Discussion

Using the previous sequence in which flexion (Fig. 2A) and torsion (Fig. 2B) are decoupled, the mathematical system is deterministic, namely the number of equations and unknown are equal: thus, insofar as the numerical convergence is ensured, a perfect numerical fit is obtained between the model and the measurements, as illustrated in Fig. 3 during the standing position in healthy (Fig. 3A) and scoliotic (Fig. 3B) subjects. In this figure, circles denote the experimental values and the continuous line segments refer to the numerical solution.

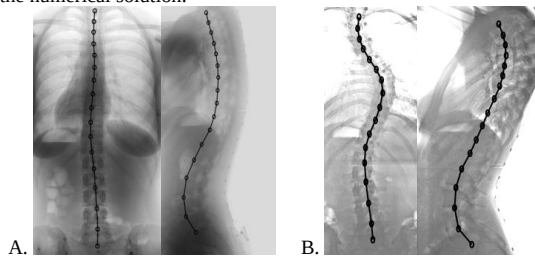


Fig. 3. Lateral and frontal views in standing up position: 3D reconstruction in a healthy subject (A) and scoliotic (B) patient (circles = experimental data, line segments = numerical solution obtained by mathematical algorithm)

The bending situation shown in Fig. 4 has been performed for two reasons. First, large bending has allowed us to validate the above optimization process on extreme configurations, to confirm that the numerical results are reliable: a good fit is observed between circles and continuous line segments.

Secondly, we plan to capture the spine deformation during “smooth gait on treadmill”, for which we make the hypothesis that the additional spine deformation will remain small for bending and torsion. To capture these “dynamical” deformations, reflective markers located on some specific vertebrae along the spinous processes will be used (see Fig. 5). Since it is rather complicated to place a marker on each vertebra, we want to predict the effect of a lack of geometrical information. The static bending in Fig. 4 emulate this situation: dark circles denote known points (2 and 4 respectively), whereas empty circles refer to unknown ones. To solve this undetermined problem, the bending coefficients proposed by Christophy et al. [5] for the lumbar deformation are used in the optimization process.

In Fig. 4, we can observe that passing from 2 to 4 known points clearly improves the results (small stars) for both healthy and scoliotic subjects. Although the fit is not as good as the deterministic case (represented by continuous segments), this result is encouraging for two reasons: first, we are able to fix more than 4 markers on the spine process (see Fig. 5) and secondly, the bending deformation during “smooth gait on treadmill” will be obviously far smaller than the static bending of Fig. 4.

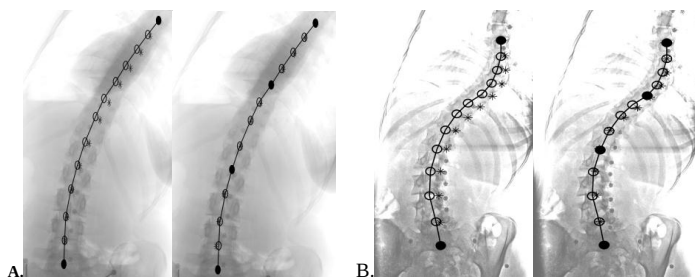


Fig. 4. Lateral bending in non-scoliotic (A.) and scoliotic subjects (B.) - circles: measured data, line segments: deterministic solution, black circles : emulated known points, stars : solution based on the known data and weighting coefficients from [5].

4. Conclusion and Perspectives

In this paper, the geometrical reconstruction of the spine has been addressed. To make the optimization process converge towards a reliable deterministic solution, we suggest decoupling the bending and torsion deformations. The method has been applied in healthy and AIS subjects during standing position with possible bending: a very good fit was obtained between the experiment and the multibody model.

Large bending being rather well captured with lacking information, they will serve as interpolated input for the spine reconstruction during gait (the future step of this research), via a standard data acquisition protocol based on infra-red cameras and reflective markers located along the spinous processes (see Fig. 5).

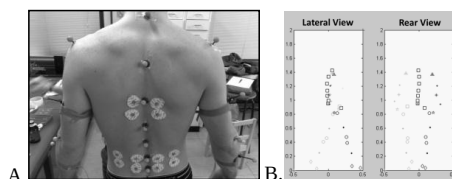


Fig. 5. Data acquisition during gait at UCL – Experimental set up (A.) - Marker motion (B.)

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Kinematic Identification of the Spine: a Multibody Approach

Gabriel Abedrabbo^{*}, Maxime Raison[#], Philippe Mahaudens[†], Christine Detrembleur[†], Maryline Mousny[§], Olivier Cartiaux[§], Paul Fisette^{*}

^{*} Center for Research in Mechatronics (CEREM),
 Université catholique de Louvain
 Place du Levant 2, 1348 Louvain-la-Neuve, Belgium
 gabriel.abedrabbo@uclouvain.be
 paul.fisette@uclouvain.be

[†] Institute of NeuroScience (IoNS)
 Université catholique de Louvain
 Brussels, Belgium
 philippe.mahaudens@uclouvain.be
 christine.detrembleur@uclouvain.be

[§] Computer Assisted and Robotic Surgery (CARS)
 Université catholique de Louvain
 Brussels, Belgium
 maryline.mousny@uclouvain.be
 olivier.cartiaux@uclouvain.be

[#] Research Chair in Pediatric Rehabilitation
 Engineering, École Polytechnique
 and CRME - Sainte-Justine UCH
 Montreal QC, Canada
 maxime.raison@polymtl.ca

ABSTRACT

For idiopathic scoliotic patients, dynamic intervertebral efforts are correlated with the spinal deformity. The quantification of these efforts, which are useful for the orthopedic surgeons to plan their surgical procedure, is the final objective of the present research. The accurate computation of intervertebral efforts strongly depends on: geometrical identification, spine and pelvis kinematics, patient physiology and muscular forces. The present work focuses on the quantification and improvement of the accuracy of the data acquisition for the geometrical identification and for the kinematic analysis of the spine. Finally, A preliminary sensitivity analysis related to intervertebral forces is presented for a simplified *vertical* human body motion, comparing inverse and direct dynamic approaches.

1 INTRODUCTION

1.1 Project Context

The intervertebral efforts quantification in scoliotic spines, before and after spine arthrodesis, appears to be useful for surgical planning. An increase of 30% of the energetic cost for adolescent idiopathic scoliosis patients (AIS) [1], as well as its consequences in ordinary living, suggest that gait is a relevant motion to be considered in this study.

The accurate computation of intervertebral efforts strongly depends on 4 pillars: geometrical identification, spine and pelvis kinematics, patient physiology and muscular forces.

1.2 Previous Works

The *geometrical identification* has been carried out by different authors to obtain 3D information of spine deformities (i.e. scoliosis). For example, [2, 3, 4] reconstruct the complete spine geometry via anatomical landmarks on each vertebra measured from biplanar X-rays, in the same way as in the present work. The correct correlation between biplanar X-rays as well as the X-ray magnification problem has been studied by [5]. Meanwhile, [6] shows that X-ray magnification varies along the body according to the distance to the X-ray source.

The *spine kinematics*, and more particularly the data acquisition, has been widely analysed by many authors. For example, [7, 8] analyse different kinds of filtering methods in biomechanical analysis.

The *spine dynamics* allows to quantify the intervertebral efforts or stresses into the spine. Despite serious ethical issues, intradiscal pressure has already been experimentally measured, using a pressure sensor located at the L4-L5 disc [9]. However, model-based studies using the a multibody formalism [10, 11, 12, 13], suggest that the intervertebral efforts could be quantified in a non-invasive way.

1.3 Objective

The present work focuses on the quantification and improvement of the accuracy of the data acquisition for the geometrical identification and for the kinematic analysis of the spine.

2 Methodology

2.1 Geometrical Identification

The static configuration, in a standing position, is measured from bi-planar X-rays. During this process, three aspects have been addressed:

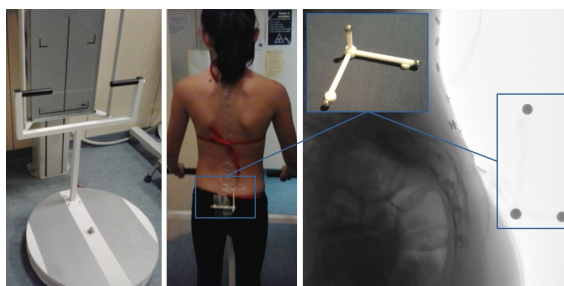


Figure 1. Geometrical identification protocol

To ensure a good Sagittal and Coronal correlation between biplanar X-rays, we have developed a rotary

tray, which helps the patient to stand still to reduce his mobility between the two radiographies (Fig.1 left).

As regards X-ray magnification, as a first step the problem has been solved by sticking a calibrated reference frame onto the patient's back (Fig.1 middle), visible on the radiography, (Fig.1 right). However, since the magnification is different along the body, according to the distance to the X-rays source (Fig.2), two aspects have to be taken in account.

1) The magnification factor error induced by the offset of the reference frame with respect to the spine axis (Fig.2), where:

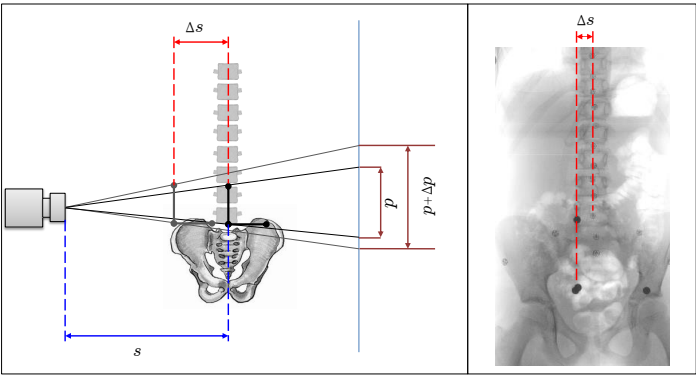


Figure 2. Magnification factor error

- s is the distance from the source (X-rays) to the spine,
- Δs denotes the distance between the spine and the reference frame,
- p is the length of the frame projection on the detector (film),
- Δp represents the magnification error induced by Δs .

Table 1 shows the magnification error (Δp) induced by $\Delta s(\%)$. With enough operation precaution, it is possible to limit the error to 1% (See green cells in Table 1)

$s \backslash \Delta s$	20	40	80	160
1000	2,0	4,2	8,7	19,0
2000	1,0	2,0	4,2	8,7
4000	0,5	1,0	2,0	4,2
8000	0,3	0,5	1,0	2,0

Table 1 . Magnification factor error $\Delta p(\%)$. The values s and Δs are expressed in mm

However, as shown in (Fig.4), the scaling factor does not change along the spine axis, whatever the angle of incidence.

2) The left/right magnification discrepancy (Fig.5 left) induces a vertical measurement error on the physiological landmarks located out of the spinal axis (Fig.5 right), where:

- s denotes the distance from the source (X-rays) to the spine,

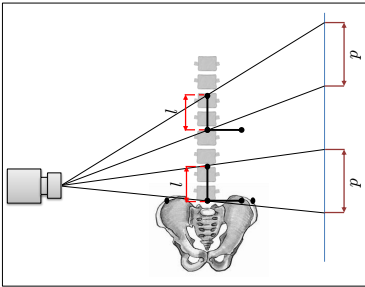


Figure 4. Magnification factor along the spine axis

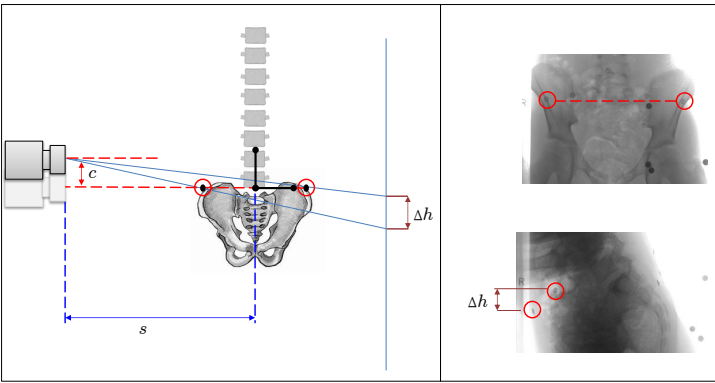


Figure 5. Left/right measurement discrepancy

- c is the vertical distance between the source axis and the physiological landmark,
- Δh represent vertical measurement error induced by c .

Table 2 shows the vertical measurement error (Δh) induced by $c(mm)$. With enough operation precaution, it is possible to limit the error to less than $2mm$ (See green cells in Table 2) Finally, in term of measurement

$s \backslash c$	c				
		20	40	80	160
1000		4,4	8,9	17,8	35,6
2000		1,8	3,5	7,1	14,1
4000		0,8	1,5	3,0	6,1
8000		0,3	0,7	1,4	2,8

Table 2 . Left/right measurement discrepancy $\Delta h(mm)$. All the values are expressed in (mm)

reproducibility, from X-ray, each vertebra is thus characterized by three non co-linear points (Fig.7): the disc center between vertebrae n and $n - 1$, the disc center between $n + 1$ and n and the spinous process of vertebra n , which allows to unambiguously locate the vertebrae in the three-dimensional space [4].

However, since vertebrae landmarks are freehandly identified by an operator, the reproducibility of this measurement has been seriously questioned and is currently analysed by an interclass correlation coefficient

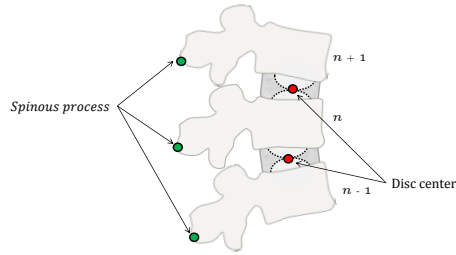


Figure 7. Characteristic points of vertebrae - spinous process and intervertebral disc centers

and a sensitivity analysis. Results and conclusions will be presented during de oral talk.

2.2 Multibody model

Using the methodology described in [4], a multibody model is generated on the basis of the above static configuration. The topology of the spine is built according to the ROBOTRAN data convention [14]. The model contains 18 bodies: the 17 vertebrae and the pelvis which plays an important role in our study, being the reference floating body for all future analyses.

To take large scoliotic deformations into account in a proper way, we suggest to represent the intervertebral kinematics by 9 relative coordinates q . The reason is the following. In addition to the 6 degrees of freedom representing the relative motion of each intervertebral disc (induced by viscoelastic efforts), we have added 3 fixed rotations to capture the initial deformation of the spine in the standing up position (those angles are somewhat similar to the neutral position of torsional springs in mechanical devices). The computation of these constant angles will be achieved by an optimization process [4] in order to fit the multibody reference configuration with the X-ray configuration.

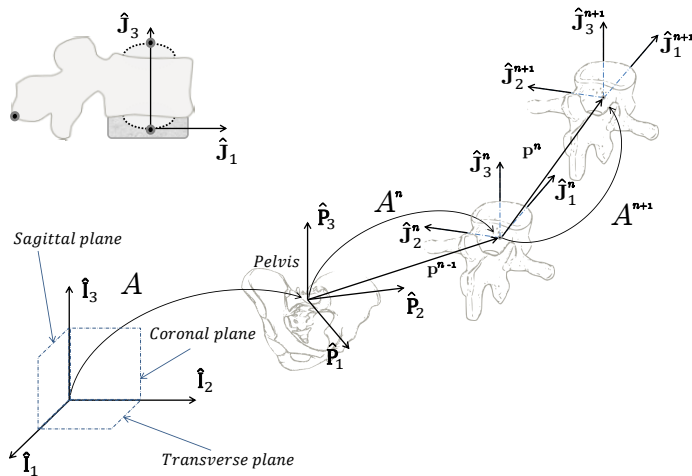


Figure 8. Spine multibody model - main geometrical and kinematic elements

At this level, as we know the location of the vertebrae and of the pelvis via the Cartesian position of the vertebra characteristic points, the necessary kinematic ingredients can be defined according to the ROBOTRAN

requirements. Without entering into details, Fig. 8 summarizes the main elements. In particular, an orthonormal frame $\{\hat{\mathbf{J}}^n\}$ is rigidly attached to each vertebra and is such that: its origin O^n is located at the lower intervertebral disc center point, $\hat{\mathbf{J}}_3^n$ points towards the upper intervertebral disc center point (between vertebrae n and $n+1$), the one which is virtually fixed to vertebra n , $\hat{\mathbf{J}}_1^n$ belongs to the plane defined by the 3 characteristic points: the spinous process and the two disc centers.

Concerning the floating pelvis, the same strategy holds: an orthonormal frame denoted $\{\hat{\mathbf{P}}\}$ is attached to it and located, in a similar way, between the $L5$ and the $S1$ sacrum vertebrae.

2.3 Kinematic Data

Spine and pelvis kinematics during gait on a treadmill is measured by a motion capture system which consists of 8 cameras and a set of reflective markers stuck on the skin. The latter are positioned on the pelvis and along the spine; they are also visible on the X-rays.

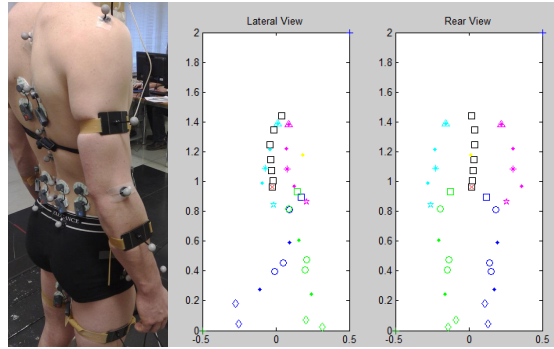


Figure 9. Data acquisition of kinematic data

The data obtained from reflective markers are successively filtered and interpolated using splines, and then differentiated to calculate the joint velocities and accelerations. This method has been validated on the pelvis motion by comparing the absolute acceleration computed from 4 reflective markers with an accelerometer stuck on the patient (See Section 3).

2.4 Sensitivity analysis

While being an anticipated analysis, in order to identify the accuracy required during data acquisition, a sensitivity analysis is carried to see the effect of these inaccuracies on the expected intervertebral efforts. For this purpose, a random – but realistic – error has been introduced for all the disc center location with a maximum value of 1 mm, 3 mm and 5 mm respectively. Then, a 2Hz sinusoidal motion is imposed to the pelvis in order to calculate the intervertebral efforts via inverse dynamics.

3 Results

3.1 Geometrical identification

An analysis to verify the reproducibility of the measurements, as well as the correlation between sagittal and coronal radiographies is now carried out. For illustration purposes, (Fig.11) superimposes the geometrical reconstruction on the X-rays. Preliminary conclusion confirms that the rotary tray significantly improves the correlation between the coronal and the sagittal X-rays.

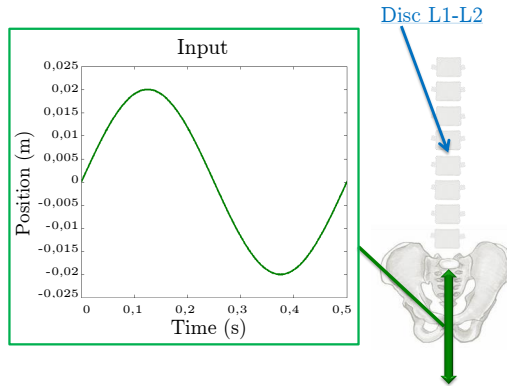


Figure 10. Simplified motion used for sensitivity analysis

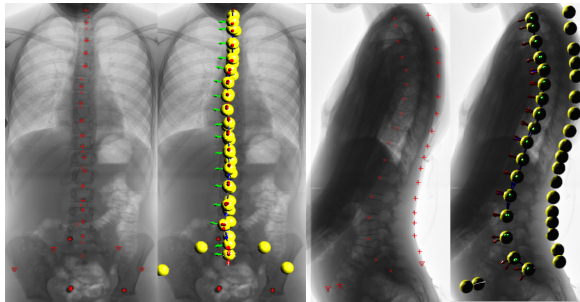


Figure 11. Characteristic points of vertebrae - spinous process and intervertebral disc centers.

3.2 Sensitivity analysis

Simulated inaccuracies, explained in Section 2.4, in the manual identification of anatomical landmarks in the biplanar radiographs induced maximum variations of about 120N for the shear force and 50N for the normal force acting at the L1-L2 intervertebral disc (Fig.12). These variations represent about 66% (120/180) and 10% (50/500), respectively, of the maximum shear and normal forces computed at the L1-L2 disc during the simulated gait when no random errors are applied (the black curve in Fig.12).

3.3 Pelvis kinematics

A very good correspondence between acceleration measured with accelerometers and a motion capture system was found for the pelvis (Fig.13). This is an important result since the pelvis kinematics is the model input, its role being to impose the motion to the spine, when walking. The analysis also revealed that a cut-off frequency of 10Hz for data filtering was suitable for an accurate spine reconstruction.

4 Conclusions

The sensitivity analysis demonstrated that the intervertebral efforts acting between vertebrae during simulated motion are significantly influenced by the inaccuracies in the manual identification of anatomical

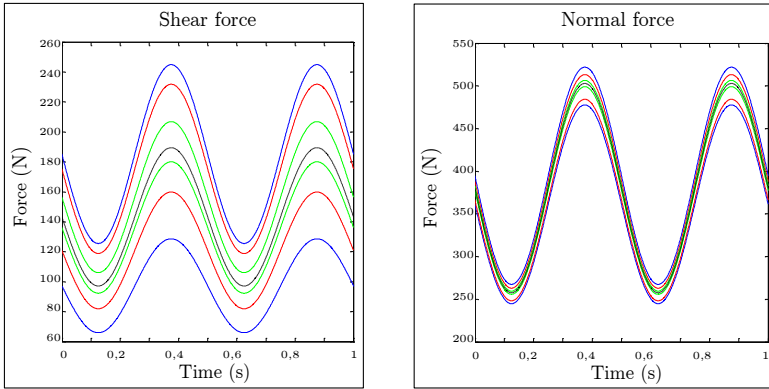


Figure 12. Sensitivity analysis for a random error of 1 mm, 3 mm and 5 mm on the geometrical identification of the spine

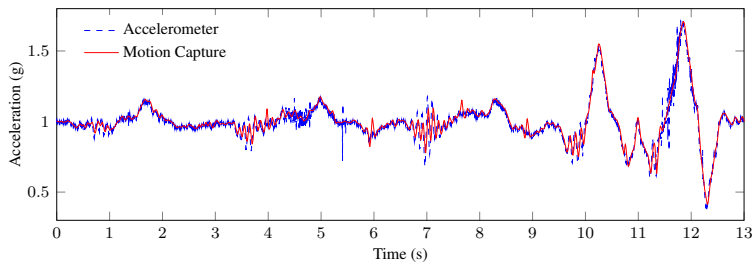


Figure 13. Absolute accelerations of the pelvis measured with accelerometers and a motion capture system match very well.

vertebral landmarks in biplanar radiographs. This suggests that the repeatability of the 3D geometrical multibody reconstruction of the spine has to be thoroughly optimized. It also appears that normal forces are less sensitive than the shear forces, even for a gait simulated with a vertical sinusoidal motion. This study also shows that taking special attention to the protocol during the geometrical identification (particularly during X-rays) reduces significantly (See Table 1 and 2) the error during geometrical identification. Finally, the good correspondence between acceleration measured with accelerometers and a motion capture, confirms a good accuracy of pelvis kinematics measurement. More details about the model, the protocol and the results, will be given during the oral talk.

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Sensitivity analysis of geometric and dynamic variables of the scoliotic spine on the computation of intervertebral efforts

Gabriel Abedrabbo¹, Olivier Cartiaux², Paul Fiset¹, Maxime Raison³,
Philippe Mahaudens⁴, Christine Detrembleur⁴, Maryline Mousny^{2,5}

1 : Center for Research in Energy and Mechatronics, Institute of Mechanics, Materials, and Civil Engineering, Université catholique de Louvain, Louvain-la-Neuve, Belgium

2 : Computer Assisted and Robotic Surgery, Institut de recherche expérimentale et clinique, Université catholique de Louvain, Brussels, Belgium

3 : Research Chair in Pediatric Rehabilitation Engineering, Ecole Polytechnique and CRME-Sainte-Justine UHC, Montreal QC, Canada

4 : Institute of Neuroscience, Université catholique de Louvain, Brussels, Belgium

5 : Cliniques universitaires Saint-Luc, Service de Chirurgie orthopédique et Traumatologique, Brussels, Belgium

gabriel.abedrabbo@uclouvain.be

Introduction

In scoliotic spine surgery, the quantification of intervertebral efforts acting between vertebrae (i.e. on the intervertebral discs) can be useful to improve the preoperative surgical planning and define the instrumentation and correction maneuvers to be performed per-operatively. In idiopathic scoliotic patients (AIS), the intervertebral efforts seem to be correlated with the severity of spinal deformities [1]. Further, a 30% increase in energy consumption during gait suggests that the intervertebral efforts have a high dynamic component [2], and therefore should be dynamically computed during gait.

Considering spine as an articulated multibody mechanical system, the computation of the intervertebral efforts can be implemented in a 3-step process: (1) construction of a 3D multibody model of the spine using standing biplanar radiographs, (2) analysis and prediction of the spine kinematics (i.e. motion) during gait, and (3) computation of dynamic intervertebral efforts in terms of forces (N) and torques (Nm) acting between vertebrae. Step #1 has already been implemented for scoliotic patients using the standing position with possible bending and an optimization process based on geometrical data reconstructed from preoperative biplanar radiographs [3].

The present study aims to investigate the sensitivity of the proposed 3-step process for the computation of intervertebral efforts within the spine. Practically, the work investigated the effect on the intervertebral efforts, of two main factors: (1) the inaccuracies in the identification of the geometrical data (i.e. disc centers and spinous processes of vertebrae) in the preoperative biplanar radiographs, and (2) the level of stiffness given to the intervertebral discs in the multibody model of the spine.

Materials and methods

For this sensitivity study, the 3-step process for the computation of intervertebral efforts within the spine has been implemented using biplanar radiographs of one healthy subject in standing position. Biplanar radiographs were calibrated using a phantom of known geometry previously fixed onto the skin of the subject near the pelvis. The phantom has been manually digitized by an operator on each radiograph, enabling to construct a unique coordinate system. On each radiograph, anatomical vertebral landmarks were also freehandly identified by the same operator, including the centers of the intervertebral discs and the spinous processes.

Given the anatomical landmarks, the geometrical 3D multibody model of the spine was constructed using a symbolic-generator modeling software (Robotran software, Louvain-la-Neuve, Belgium) and mechanical properties (stiffness) of the intervertebral discs were defined.

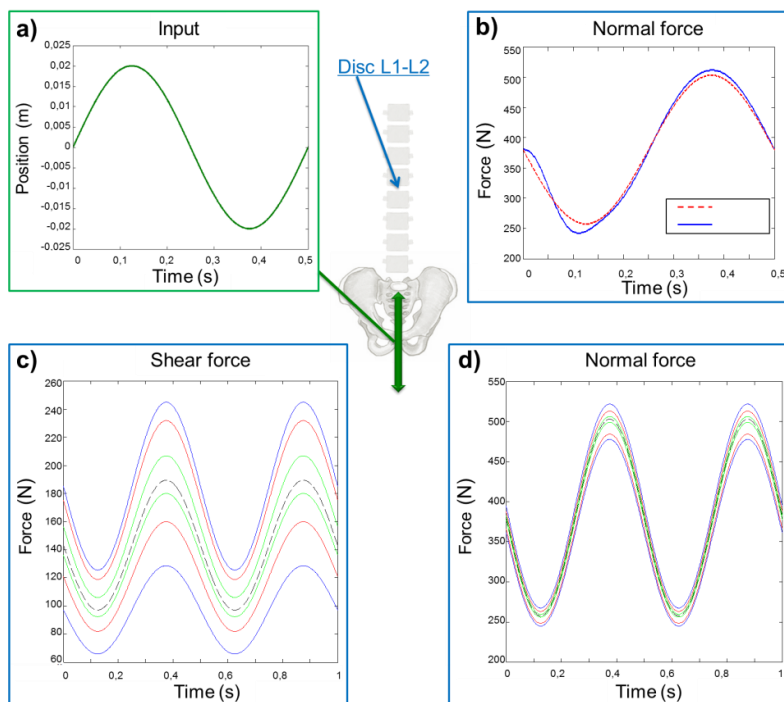


Figure 1 – (a) Sinusoidal vertical motion applied to the pelvis to simulate gait. (b) Variation of normal force acting between L1 and L2 vertebrae for two levels of stiffness of the intervertebral L1-L2 disc. The solid blue curve represents the normal force for a high stiffness level. The dotted red curve represents the normal force for a low stiffness level. (c) Variation of the shear force acting between L1 and L2 vertebrae when random errors are applied on the centers of the intervertebral discs. The dotted curve represents the shear force when no random errors are applied. The other solid curves represent the shear force when 1-mm (green), 3-mm (red) and 5-mm (blue) random errors are applied. (d) Similar to (c) but for the normal force.

Kinematics (position, velocity and acceleration) of the spine has been defined using a vertical 2-Hz sinusoidal motion applied to the pelvis as the boundary input of the multibody model of the spine (Fig. 1a).

Then, given the position, velocity and acceleration of the spine during the simulated gait, intervertebral efforts acting between vertebrae, including the normal (vertical) force (N) and the shear (transverse) force (N), have been computed by applying inverse-dynamics algorithms to the multibody model.

First, sensitivity analyses investigated inaccuracies in the manual identification of the anatomical vertebral landmarks in the biplanar radiographs, using randomly applied errors of 1 mm, 3 mm and 5 mm to the points defining the centers of the intervertebral discs.

Finally, this study investigated two levels of stiffness given to the intervertebral discs in the multibody model: a physiological-based stiffness level and an excessive rigid stiffness level (simulated by a very high numerical value in the multibody model).

Results

Simulated inaccuracies in the manual identification of anatomical landmarks in the biplanar radiographs induced maximum variations of about 120N for the shear force and 50N for the normal force acting at the L1-L2 intervertebral disc (Fig. 1c-d). These variations represent about 66% (120/180) and 10% (50/500), respectively, of the maximum shear and normal forces computed at the L1-L2 disc during the simulated gait when no random errors are applied (the dotted curve in Fig. 1c-d).

The maximum difference in terms of the normal force acting between L1 and L2 vertebrae was about 20N between the two simulated motions performed, first with a high level, then with a low level of stiffness for the intervertebral discs (Fig. 1b). This difference represents less than 10% of the computed magnitudes of the normal force.

Discussion

The sensitivity analysis demonstrated that the intervertebral efforts acting between vertebrae during simulated motion are significantly influenced by the inaccuracies in the manual identification of anatomical vertebral landmarks in biplanar radiographs. This suggests that the repeatability of the 3D geometrical multibody reconstruction of the spine has to be thoroughly optimized. It also appears that normal forces are less sensitive than the shear forces, even for a gait simulated with a vertical sinusoidal motion.

This study also showed that the intervertebral longitudinal forces acting between vertebrae during the simulated motion are slightly influenced by the level of longitudinal stiffness given to the intervertebral discs in the multibody model of the spine. This suggests for further investigations that modeling intervertebral discs with rigid elements could be sufficiently reliable to compute the intervertebral efforts acting between vertebrae during normal gait.

Although all these results clearly have to be validated in a larger number of scoliotic patients during gait, they may be straightaway useful to assess the level of accuracy that the multibody modeling of the spine needs to be provided with in terms of geometric and dynamic properties.

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Dynamic Analysis of the Spine: Impact of the Internal Organs Wobbling Motion During Gait

Gabriel Abedrabbo^{*}, Olivier Cartiaux[§], Philippe Mahaudens[†], Christine Detrembleur[†], Maryline Mousny[§], Paul Fiset^{*}

^{*} Center for Research in Energy and Mechatronics (CEREM)
Université catholique de Louvain
Place du Levant 2, 1348 Louvain-la-Neuve, Belgium
gabriel.abedrabbo@uclouvain.be
paul.fisette@uclouvain.be

[§] Computer Assisted and Robotic Surgery (CARS)
Université catholique de Louvain
Brussels, Belgium
maryline.mousny@uclouvain.be
olivier.cartiaux@uclouvain.be

[†] Institute of NeuroScience (IoNS)
Université catholique de Louvain
Brussels, Belgium
philippe.mahaudens@uclouvain.be
christine.detrembleur@uclouvain.be

Context and Objective

The intervertebral efforts quantification in scoliotic spines, before and after spine arthrodesis, appears to be useful for surgical planning. An increase of 30% of the energetic cost for adolescent idiopathic scoliosis patients [1], as well as its consequences in ordinary living, suggest that gait is a relevant motion to be considered in our study.

The accurate computation of those efforts strongly depends on 4 pillars: geometrical identification, spine and pelvis kinematics, patient physiology and muscular forces. The geometrical identification of the spine, using bi-planar X-rays (Fig 1-left)), as well as the computation of its kinematics from a limited amount of data (Fig 1-right)), has been addressed in previous studies [2], [3]. The present work focuses on the patient physiology, and more particularly on the motion of the organs inside the abdominal cavity and its impact on the intervertebral efforts during gait.

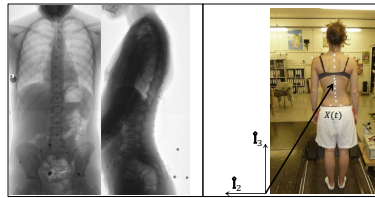


Figure 1: Spine Geometrical and Kinematic identification [2], [3].

Methodology

To study the physiology of the human body, via a multi-body model [4], we have subdivided it into different parts: head, thorax, upper limbs, abdominal cavity, spine, pelvis and lower limbs (deliberately absent from the model [3]). During smooth gait, the head, the thorax and the upper limbs could be considered as a sequence of polyarticulated musculoskeletal rigid bodies. However, this is not the case for the abdominal cavity, where the organs are wobbling masses and not directly connected to the spine.

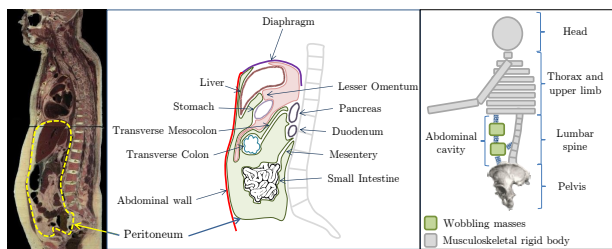


Figure 2: Development of the multi-body model (right) of the abdominal cavity, based on patient physiology¹ (left, middle)

In [5], an equivalent model (i.e. not based on patient physiology) that reproduces the resonance modes of the visceral organs was developed. However, this model was not originally designed to analyse the impact of these wobbling organs on the intervertebral efforts.

¹From University of Michigan: <http://vhp.med.umich.edu/browsers/female.html>

Taking into account of visceral organs for our study is a delicate task that has required judgement from both physiotherapist and engineers. As we can observe in Figure 2-middle, it is possible to identify three kinds of attachment in the abdominal cavity: 1) "strong" anchor to the diaphragm and/or the spine; this is the case for the liver, the pancreas and the duodenum; 2) "weak" connection between bodies maintained by the peritoneum; this is the case for lesser omentum, which connects

stomach and liver, the transverse mesocolon, which connects the pancreas and the transverse colon, and for the mesentery, which connects the small intestine to the lumbar spine. 3) Finally, the direct surface contact between organs and the abdominal wall (Fig 2-left). Although it is difficult to accurately identify the location and the mechanical properties for each of those attachments, a multibody model is proposed (Fig 2-right). In order to identify the impact of the abdominal cavity on the spine efforts during gait. The proposed variants of the model are the result of intense exchanges with physiotherapists and anatomy professors of UCL, to avoid the development of an unduly complicated model (in terms of size and parameters to be identified) regarding our project issue. In this model, the visceral wobbling masses are subdivided in two: the upper mass represents the liver and the stomach, and the lower mass refers to the intestines. Then, in order to analyse the sensitivity of the attachment characteristics, three variants are proposed. In (Fig 3-a), three vertical springs/dampers are used to connect the two masses between the thorax and the pelvis; in (Fig 3-b), the same, but without spring – just a damper – between the two wobbling masses; and in (Fig 3-c), we included a spring/damper connection that represents the mesentery (Fig 2-middle). Finally, a reference model, in which all visceral masses are neglected is also considered for this study.

Preliminary Results and Conclusion

A preliminary analysis reveals a rather low sensitivity of the vertical intervertebral efforts with respect to the organs dynamics and its modelling parameters. In Fig 4, a difference of around 10% is observed for the vertical intervertebral force between L4 and L5 between model b. (Fig 3-b) and the reference model. Although model b. (Fig 3-b) appears to be the most relevant from a physiological aspects, the definitive choice of this variant requires further investigation and identification. Those are currently in progress for a non-scoliotic and for scoliotic patient, walking at different speeds.

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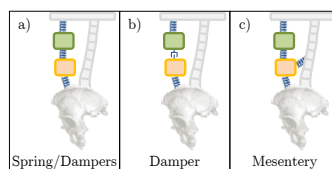


Figure 3: Variants of the multibody model

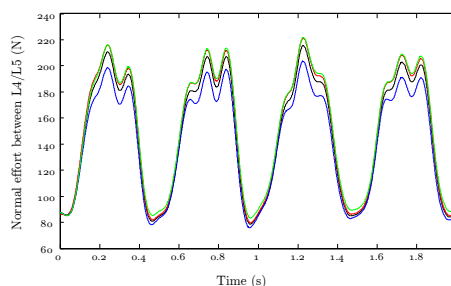


Figure 4: Normal effort between L4/L5 for different variants: for variant a (black), b (green), c (red) and the reference model (blue)

Computation of spine intervertebral motions in scoliotic patients: a multibody approach

G.Abedrabbo^a, O. Cartiaux^b, P. Mahaudens^b, C. Detrembleur^b, M. Mousny^b, P. Fisette^a

^a Center for Research in Energy and Mechatronics, Institute of Mechanics, Materials, and Civil Engineering, Université catholique de Louvain, Louvain-la-Neuve, Belgium.

^b Computer Assisted and Robotic Surgery, Institut de recherche expérimentale et clinique, Université catholique de Louvain, Brussels, Belgium.

Keywords: Multibody Modeling, Spine Kinematics, Scoliosis

1. Introduction

The intervertebral efforts quantification in scoliotic spines, before and after spine arthrodesis, appears to be useful for surgical planning. An increase of 30% of the energetic cost for adolescent idiopathic scoliosis patients (Mahaudens and Mousny 2010), as well as its consequences in ordinary living, suggest that gait is a relevant motion to be considered in our study. The accurate computation of those efforts strongly depends on 4 pillars: geometrical identification, spine and pelvis kinematics, patient physiology and muscular forces.

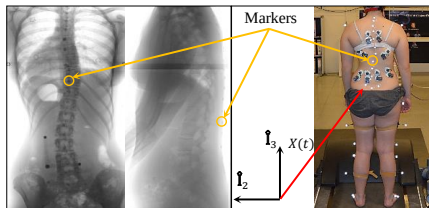


Figure 1: Geometrical and kinematic identification

The geometrical identification of the spine, using bi-planar X-rays (Fig 1-left)), as well as the computation of its kinematics from a limited amount of data (Fig 1-right)), has been addressed in previous studies (Abedrabbo et al. 2012, 2014). The present work focuses on the evaluation of the spine kinematics identification during gait, for a scoliotic patient.

2. Methods

2.1 Geometrical identification

The Geometrical identification (Abedrabbo et al. 2012, 2014) of the spine is based on bi-planar X-rays, whereby the center of the intervertebral discs and the position of metallic markers stuck on the skin at the level of the 18 (from L5 to C7) spinous processes are identified (Fig 1-left). Reflective markers are stuck at the same place (for only 8 spinous processes) for the gait analysis (Fig 1-right, Fig 2-right).

The followings steps are carried out:

1. A multibody model (MBS) (Docquier et al. 2013) is developed based on the bi-planar X-rays.
2. An optimization algorithm is then used to fit the center of the discs in the model, to those measured on the standing-up patient.
3. The position of the spinous processes is then used to compute the axial rotation along the spine. This rotation is particularly important in scoliotic patients.

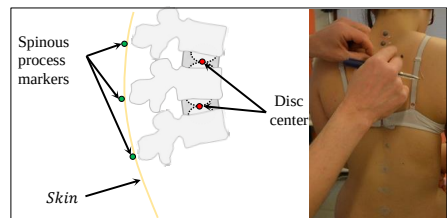


Figure 2 left: Reference points for geometrical identification (red and green) and for the gait kinematic reconstruction (green); right: Positioning of reflective markers.

2.2 Kinematic reconstruction

Due to experimental limitations during gait, it is impossible to obtain the same amount of information as from radiograph, leading to an underdetermined problem. Indeed in the current protocol, we are only able to stick up 1 reflective marker on the spinous process per two vertebrae. For this reason some additional information is required to predict the lacking kinematic data of the spine during gait. To address this problem, the total bending angle of the spine, β , is distributed among the lumbar vertebrae, using a multi-objective optimization process, based on the available kinematic data (8 reflective markers) and previous experimental results (Rozumalski et al. 2008), that provides a good indication of the intervertebral angle distribution $f(\beta)$ (Fig 3).

*Corresponding author. Email: gabriel.abedrabbo@uclouvain.be

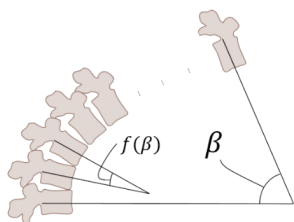


Figure 3 Motion distribution along the spine

2.3 Evaluation: “one ball out” method

Since it is experimentally complicated to place a marker on each vertebra, in order to estimate the impact of the lack of kinematic information, we suggest to work in the opposite direction. In other words, we have fixed one reflective marker each 2 vertebrae from L4 to C7, then 2 different optimizations have been carried out: the first one includes all the reflective markers and the second one uses 1 reflective marker for each 4 vertebrae only i.e. neglecting the information of one every two markers.

Finally, both models are compared to analyze the sensitivity of the kinematics to this lack of information.

3. Results

In this study, two different kinematical reconstructions have been achieved. One using the information from markers stuck on: the pelvis (4 markers), L4, L2, T12, T10, T6, T4 and T2 and another one using the information from markers stuck on: the pelvis, L4, T12, T8, T6 and T2. To analyze the difference between both reconstructions, the relative rotation between vertebrae has been compared. By way of illustration, Figure 3 shows a comparison in the relative rotation (lateral bending) between L2 and L3 during gait. The vertebra rotation results from the above optimization process (Fig 3-upper) that fits the multibody models with the filtered (bandwidth 8Hz) experimental data. Angular velocity and accelerations, also required for the future spine analysis, are obtained via curve fitting (cubic splines) and differentiation.

4. Conclusions

Although some discrepancies are observed for velocity and acceleration, it can be said that the spine kinematics is rather well-captured to predict the overall motion of the spine during gait. The future work will focus on the spine dynamics, for which the present kinematic computation represents a fundamental input. It will be important to keep in mind the sensitivity of the kinematic accuracy, tackled in this study, on the final results, i.e. the intervertebral efforts in the dynamic context of gait.

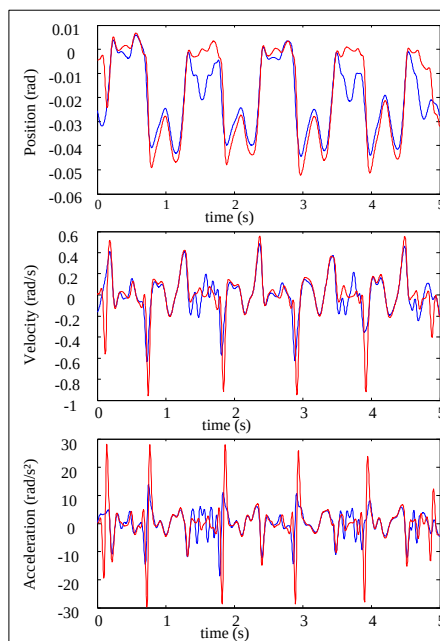


Figure 4 Correspondence between the two multibody models: a model using all markers (blue) and a model using half of the markers (red).

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Mechanical and anthropometric data

B

Date of the study	12/04/2016
Age of the subject	13 years old

Patient Data

Weight	54.2Kg
Height	159cm
Asis breadth	24cm
Distance pelvis to flor	95cm
Trunk length	53cm

	Right	Left
Pelvis depth	10cm	10,5cm
Total leg length	88cm	88,5cm
Knee diameter	10,5cm	10,5cm
Ankle diameter	7,5cm	7,5cm
Thigh length	33,5cm	35,5cm
Calf length	33,5cm	33cm
Foot length	9,5cm	10,5cm

	Mass	CoM	RG(I CoM R1)	RG(I CoM R2)	RG(I Prox R1)	RG(I Prox R2)
Head (VERX-MIDG)	0,0668	0,589	0,33	0,359	0,527	0,545
Upper Trunk (ACR-XYPH)	0,155	0,22	0,734	0,494	0,767	0,541
Lower Lrunk	0,147	0,318	0,305	0,249	0,441	0,404
Arm (ACR-STYL)						
(Upper arm + Forearm)	0,0449	0,56	0,304	0,301	0,637	0,636

Table B.1: General relative values [50]

	RG(I CoM R1)	RG(I CoM R2)	RG(I Prox R1)	RG(I Prox R2)
Head (VERX-MIDG)	0,0594	0,0646	0,0949	0,0981
Upper Trunk (ACR-XYPH)	0,154	0,104	0,161	0,114
Lower Trunk	0,0641	0,0523	0,0926	0,0848
Arm				
(ACR-STYL) (Upper arm + Forearm)	0,122	0,120	0,255	0,254

Table B.2: Relative values for the patient [50]

	Mass	I CoM R1	I CoM R2	I Prox R1	I Prox R2	I CoM R1	I CoM R2
Head	3,62	0,0128	0,0151	0,0326	0,0348	0,241	0,243
Upper Trunk	8,37	0,2	0,0901	0,217	0,108	0,2	0,09
Lower Lrunk	7,94	0,0326	0,0217	0,0681	0,0572		
Total arm	4,87	0,072	0,0706	0,316	0,315	0,416	0,415

Table B.3: Inertia values of the patient with respect to L5/T12

Inertia Upper body		
	I total (R1)	I total (R2)
I Total	1,3	1,08
I (No arms)	0,44	0,33
M total	21,7	

Table B.4: Mass and Inertia Upper body

Medical diagnosis



Diagnosis

The medical report of the patient reveals the following.

- In standing position, the patient complained of low back pain, mainly in the left lumbar paravertebral area,
- The patient did not practice any sportive activity on a regular basis,
- In standing position, the pelvis was level ,
- The anterior bending test showed a lumbar gibbosity of about 15 degrees,
- The spine is flexible for the lateral bending test,
- There was no pain nor muscle contracture at paravertebral muscles palpation,
- Normal gait,
- Neither ischiotibialis muscles nor muscles rectus femoris muscles were retracted,
- A Neurological exam showed symmetrical reflexes at both lower legs,
- Abdominal reflexes were asymmetrical,
- The right thoracic curve measured 69 degrees between T5 and T11 ,
- The left lumbar curve measured 55 degrees between T11 and L4 ,
- There was a high intervertebral torsion,
- The patient was at Risser 0 ,
- Triradiate cartilages were closed .

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