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Quantification of trunk muscle forces through multibody models and non-invasive experimental measurements

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Ph.D. thesis

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Abstract

In biomechanics, spine modelling allows the calculation of intervertebral efforts. The latter reflect the combined effect of spine geometry, spine kinematics and muscle forces. Although the geometry and kinematics of the spine are well known, the estimation of muscle forces is still one of the major challenges in spine modelling.

Muscle forces are usually quantified using either experimental or mathematical approaches. However, non-invasive experimental approaches only provide indirect measurements of the force. Mathematical approaches rely on a model of the spine that includes the muscles and on an optimisation process to compute their forces. However, they face the muscle redundancy problem as the human body is overactuated. Furthermore, the muscle forces obtained with this approach only reflect only a global muscle strategy and not the diversity observed in a population sample.

For the quantification of muscle forces, hybrid approaches may be a good opportunity to take into account the individual muscle strategy, thanks to electromyography (EMG) signals from an upfront experiment as input to the model.

In this work, a novel hybrid approach is proposed. It differs from existing hybrid approaches by focusing on static configurations and consequently neglecting the muscle dynamics. The hybrid approach relies on a well thought experiment and protocol to ensure that the EMG signals are sufficiently reliable to be used. The Sorensen test posture was chosen as a suitable static configuration to quantify back muscle forces. A multibody model associated with the experiment was developed to compute muscle forces and intervertebral efforts. Two main experiments in the Sorensen test posture were considered in this work: a sample of 19 subjects to illustrate the diversity of muscle strategies and a sample of one subject who produced

different muscle strategies.

The hybrid approach was developed in two steps. First, it was proposed to use a relatively simple EMG-based distribution of a global equivalent force, distributed over all the muscle fascicles included in the model. Secondly, it was proposed to embed EMG signals into the optimisation process via constraint functions or by modifying the cost functions. Overall, the hybrid approach showed promising results with intervertebral efforts close to those obtained with purely optimisation-based solutions, while computing muscles forces more representative of the recorded muscle strategies.

To increase the interest in the hybrid approach, additional studies were analysed: application to a slow dynamic configuration, sensitivity analysis of a deep back muscle and validation of the use surface EMG on this back muscle.

This thesis highlights the use of hybrid approaches to quantify trunk muscle forces for static and quasi-static configurations. Hybrid approaches provide a good alternative to mathematical approaches to improve muscle force estimation and account for the diversity of muscle strategies.

Résumé

En biomécanique, la modélisation du rachis permet de calculer les efforts intervertébraux. Ces derniers reflètent l'effet combiné de la géométrie de la colonne vertébrale, de sa cinématique et des forces musculaires. Bien que la géométrie et la cinématique de la colonne vertébrale soient bien connues, l'estimation des forces musculaires reste l'un des principaux défis de la modélisation de la colonne vertébrale.

Les forces musculaires sont généralement quantifiées à l'aide d'approches expérimentales ou mathématiques. Cependant, les approches expérimentales non invasives ne fournissent que des mesures indirectes de la force. Les approches mathématiques reposent sur un modèle de la colonne vertébrale qui inclut les muscles et sur un processus d'optimisation pour calculer leurs forces. Cependant, elles sont confrontées au problème de redondance musculaire, car le corps humain est sur-actionné. En outre, les forces musculaires obtenues avec cette approche ne reflètent qu'une stratégie musculaire globale et non la diversité observée dans un échantillon de population.

Pour la quantification des forces musculaires, les approches hybrides peuvent être une bonne opportunité de prendre en compte la stratégie musculaire individuelle, grâce aux signaux électromyographiques (EMG) d'une expérience réalisée en amont comme entrée du modèle.

Dans ce travail, une nouvelle approche hybride est proposée. Elle diffère des approches hybrides existantes en se concentrant sur les configurations statiques et en négligeant par conséquent la dynamique du muscle. L'approche hybride repose sur une expérience et un protocole bien conçus afin de garantir que les signaux EMG sont suffisamment fiables pour être utilisés. La posture du test de Sorensen a été choisie comme une configuration statique appropriée pour quantifier les forces des muscles du dos. Un modèle multicorps associé à l'expérience a été développé pour calculer les

forces musculaires et les efforts intervertébraux. Deux expériences principales dans la posture du test de Sorensen ont été prises en compte dans ce travail : un échantillon de 19 sujets pour illustrer la diversité des stratégies musculaires et un échantillon d'un sujet qui a produit différentes stratégies musculaires.

L'approche hybride a été développée en deux étapes. Premièrement, il a été proposé d'utiliser une distribution relativement simple, basée sur l'EMG, d'une force équivalente globale, répartie sur tous les faisceaux musculaires inclus dans le modèle. Deuxièmement, il a été proposé d'intégrer les signaux EMG dans le processus d'optimisation par le biais de fonctions de contrainte ou en modifiant les fonctions de coût. Dans l'ensemble, l'approche hybride a donné des résultats prometteurs avec des efforts intervertébraux proches de ceux obtenus avec des solutions purement basées sur l'optimisation, tout en calculant des forces musculaires plus représentatives des stratégies musculaires enregistrées.

Afin d'accroître l'intérêt de l'approche hybride, des études complémentaires ont été analysées : application à une configuration dynamique lente, analyse de sensibilité d'un muscle profond du dos et validation de l'utilisation de l'EMG de surface sur ce muscle profond.

Cette thèse met en évidence l'utilisation d'approches hybrides pour quantifier les forces musculaires du tronc pour des configurations statiques et quasi-statiques. Les approches hybrides offrent une bonne alternative aux approches mathématiques pour améliorer l'estimation de la force musculaire et prendre en compte la diversité des stratégies musculaires.

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Glossary

Abbreviations

3D	Three-Dimensional
AIS	Adolescent Idiopathic Scoliosis
ANOVA	Analysis of Variance
BMI	Body Mass Index
COM	Center of Mass
DoF	Degree of Freedom
EMG	Electromyography
ES	Erector Spinae
F ²	Minimise the sum of Squared muscle Forces
FEM	Finite-Element Model
FT	Functional Test
Hybrid-1	First version of the Hybrid approach
Hybrid-2	Second version of the Hybrid approach
ILpL	Iliocostalis Lumborum pars Lumborum
ILpT	Iliocostalis Lumborum pars Thoracis
IVD	Intervertebral Disc
LBP	Low Back Pain
LD	Latissimus Dorsi
LIC	Loaded Isometric Contraction
LP	Lumbar Paravertebral
LS	Lateral Sorensen
LTpL	Longissimus Thoracic pars Lumborum
LTpT	Longissimus Thoracic pars Thoracis
MBS	Multibody System
mM	Minimise the Largest relative muscle force
MMS	Maximum Muscle Stress

MVC	Maximal Voluntary Contraction
OE	Obliquus Externus
PCSA	Physiological Cross-Sectional Area
QL	Quadratus Lumborum
QL-a	Quadratus Lumborum - anterior fibres
QL-p	Quadratus Lumborum - posterior fibres
RA	Rectus Abdominis
RMS	Root Mean Square
RMSD	Root-Mean-Square Deviation
s^2	Minimise the sum of Squared muscle Stresses
s^3	Minimise the sum of Cubed muscle Stresses
TLL ratio	Normalised ratio between the Trunk height and the Lower Limb length
TP	Thoracic Paravertebral

PART I

Introduction

1

Context

When a Ph.D. thesis is about muscles and human body, it is not uncommon to believe that it is the research work of a physiotherapist or a doctor researcher. So, it is surprising to see the term 'engineer' and even 'mechanical engineer' under the title of the present manuscript. What experience could an engineer bring to the medical field?

Physiotherapists and doctors know muscles, or rather their name, their composition, their localisation and their(s) action(s). However, they have no tool to quantify the action and therefore cannot evaluate their impact on bones, joints or cartilaginous structures to which they are attached. Quantifying the impacts on these structures is not futile. Indeed, quantifying means putting numbers, i.e. objective information, which can be used by physiotherapists to improve their rehabilitation or doctors to improve medical treatments and surgeries and overall to better understand the degradation of these structures. The engineer has then for purpose to bring additional quantified information to the medical staff to improve health service and consequently the quality of patient's life.

From a biomechanical point of view, the human body is usually split and studied according to three main parts: the lower limb (from the hip joint to the foot), the upper limb (from the shoulder girdle to the hand) and the trunk (from the pelvis to the neck). In each of them, there are bones actuated by muscles through tendons. Among other structures, there are ligaments connecting bones between them, fascias wrapping muscles

and organs and nerves, bringing nerve impulse between the brain and the rest of the body. Physically, taking into account bones, muscles and tendons is sufficient to quantify muscles' actions and their impact. Muscle forces and their resulting joint efforts, i.e. forces and torques generated between segments, for both lower and upper limbs have been extensively quantified both in terms of quantity and diversity of tasks. For the upper limb, one can mention elbow flexion-extension [1, 2] and more complex motions (pronation-supination [3] and abduction [4]). For the lower limb, joint efforts have been quantified from simple knee flexion-extension to jump, walking or even running. Muscle forces are often considered as intermediate results intended to evaluate the associated joint efforts. The latter can be afterwards used as indicators to assess the joint stability, e.g. the shoulder for the upper limb and the hip for the lower limb, but also as objective biological markers for return to sport in rehabilitation, e.g. via the evaluation of the peak knee net torque in the case of the anterior cruciate ligament injury [5, 6]. One can consequently admit that quantifying muscle forces and resulting joint efforts is useful for rehabilitation, surgical planning or prosthesis design purposes.

If joint efforts in both lower and upper limbs have been extensively quantified, one cannot say the same for the trunk – and more generally for the spine –, as highlighted by two recent literature reviews on the modelling of the healthy spine [7] and of the scoliotic spine [8]. It is surprising that although the spine is the central component of the human body, its modelling is only in its infancy. Spine integrity is crucial to the proper functioning of the human body. To preserve its integrity, objective biological markers could be the intervertebral efforts, i.e. forces and torques undergone by each intervertebral disc (IVD) [9]. However, these efforts are unknown to the surgeons prior to a spine surgery. In particular, in the case of scoliosis (spine deformity defined later in the manuscript), surgeons often base their practice on their own experience what may result in large numbers of possible surgical strategies for the same patient [10, 11]. Improving knowledge of the spine and of the distribution of intervertebral efforts could help to develop an objective tool for surgical planning [12, 13]. The latter is intended to inform the surgeon on post-operative intervertebral efforts to prevent future spine degeneration and subsequent reoperations [14, 15].

Intervertebral efforts are the reflect of the combined action of spine geometry, spine kinematics and muscle forces. Spine geometry and kinematics are relatively well known but not muscle forces [7, 8, 16]. Indeed, there is still a lack of mastery for muscle force estimation, good enough to represent

the diversity of muscle strategies and to solve the muscle redundancy problem. To capture the diversity of muscle strategies encountered in subjects, one can resort to experimental measurements such as electromyography (EMG) [17], i.e. an experimental approach. However, they are indirect measurements of the muscle' action. To quantify actual muscle forces in N, this approach is relatively limited.

Alternatively, numerical models can quantify muscle forces, i.e. a mathematical approach. But in addition to all inaccuracies related to geometry, kinematic and physiological parameters, they faced the complex muscle redundancy problem. To solve the latter, optimisation process with appropriate cost functions mimicking a chosen muscle strategy is often used [18]. Experimental EMG signals are often required afterwards to validate computed muscle forces on the basis of the muscle activity amplitude and timing in the case of dynamic tasks. Mathematical approaches can hardly represent the diversity of muscle strategies between subjects but also the different muscle strategies produced by one and only one participant.

In this research, it is proposed to combine advantages of experimental and mathematical approaches, i.e. a hybrid approach. For the experimental process, it was decided to take only non-invasive measurements, i.e. surface EMG. The idea was to make the most of EMG signals from a carefully designed upfront experiment as input of the model, rather than a posterior experiment intended to assess the relevancy of the results. Firstly, EMG signals can help to select the only relevant muscle fascicles in the model. Secondly, they can be used in the computation of muscle forces, according to a specific actuation strategy.

1.1 The trunk

1.1.1 Components of the trunk

The trunk segment is defined from the pelvis to the neck, usually excluding both upper limbs. From a posterior view, the spine is covered by back muscles. From an anterior view, the lower trunk, i.e. the abdomen, contains all internal organs surrounded by abdominal muscles. On top of the abdomen, there is the thorax, composed of the rib cage, the two lungs, the heart and muscles responsible of their good functioning as well of the upper limbs' motion. The neck is composed of the cervical spine and surrounding muscles. When modelling the trunk, the head and both upper limbs are usually added to the model but considered as non-articulated segments.

They only add masses and inertias to the model. This thesis focusing on the spine and its surrounding structures, the same hypothesis was used for the head and both upper limbs.

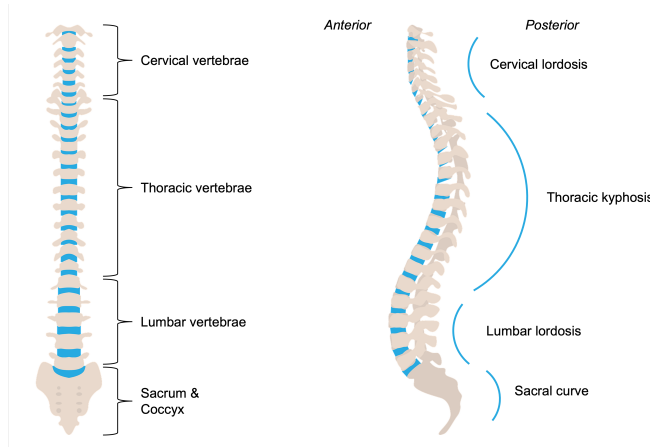


Fig. 1.1 (Left) Anterior view of the spine and vertebrae. (Right) Sagittal left view of the spine and natural spine curvatures. Adapted from [19].

From a mechanical point of view, in addition to the wobbling masses, the trunk is composed of the spine and its surrounding back and abdominal muscles. The spine is made up of 32 to 34 vertebrae depending on the individuals (Fig. 1.1 Left). Starting from the bottom, there are 3 to 5 coccygeal vertebrae (coccyx bone) and 5 sacral vertebrae (sacrum bone). Those coccygeal and sacral vertebrae are fused and cannot move relatively to each other. Coming next are the 5 lumbar vertebrae, the 12 thoracic vertebrae and finally the 7 cervical vertebrae. All these vertebrae can move relatively to each other, but they do not all have the same range of motion. A relative motion is made possible by the *IVD* (see infra for more details) between each pair of vertebrae. From the anterior and posterior views, the spine is relatively straight for an healthy individual standing upright. However, from a sagittal view, the spine is characterised by four natural curvatures, two pointing forwards (anterior concavity) – the sacral curve and the thoracic kyphosis – and two pointing backwards (posterior concavity) – the lumbar lordosis and the cervical lordosis (Fig. 1.1 Right). These four spinal curvatures contribute to the stability of the spine and a good distribution of spinal loads. The spine can move in the three planes: sagittal, frontal

(coronal) and transverse (Fig. 1.2).

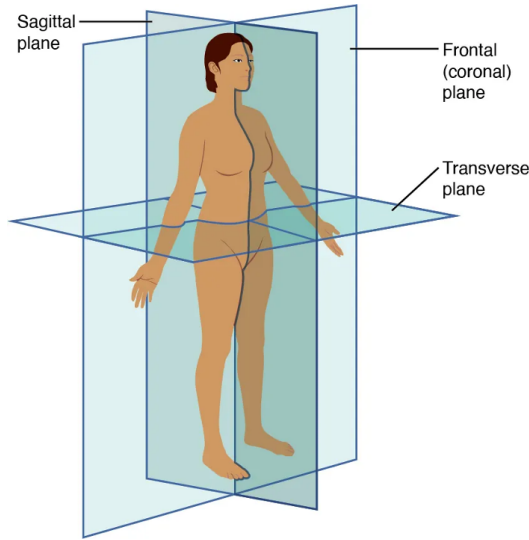


Fig. 1.2 Anatomical planes of the human body. Adapted from [20].

Flexion/extension movements of the spine take place in the sagittal plane. The angle covered by the uppermost lumbar IVD located between L1 and T12 vertebrae goes from 60 degrees for a spine in flexion to 35 degrees for a spine in extension with respect to its neutral position for an upright and standing configuration. The thoraco-lumbar spine can flex towards 105 degrees and extend towards 60 degrees while the cervical spine can flex towards 40 degrees and extend towards 75 degrees. In the frontal plane, the spine can bend laterally with a distribution of 20, 20 and 35 degrees for the lumbar, thoracic and cervical segments respectively. Finally, in the transverse plane, the lumbar spine can barely rotate through an angle of 5 degrees. The thoracic spine can rotate toward 35 degrees and the cervical spine towards 45-50 degrees. In the present thesis, the lumbar spine will be considered as the spine hinge in which all the motion – if there is – is located. The cervical spine is fused with the head body while the thoracic spine is not allowed to move and is modelled as a rigid body.

As mentioned previously, the IVD is a key component of the spine, located between each pair of mobile vertebrae (23 in total). IVD is fibrocartilage and is divided into two parts: the internal nucleus pulposus (central

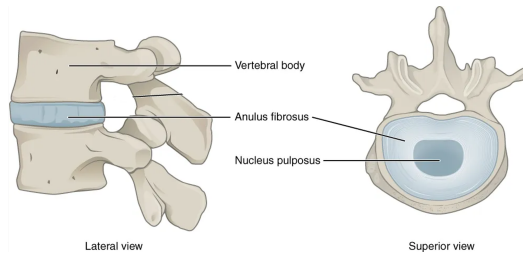


Fig. 1.3 Anatomy of an IVD. Adapted from [21].

part) and the external annulus fibrosus (outer ring) (Fig. 1.3). This cartilage is unique in the human body in terms of its composition. It is avascular with a limited innervation and a low cells density compared to hyaline cartilage. The nucleus pulposus is a highly hydrated gelatinous tissue while the annulus fibrosus is a dense and organised collagen structure. The annulus fibrosus is not homogeneous. Its outer layer is denser and more organised with fibroblast-type cells while the inner layer is less dense and less organised with chondrocyte-type cells. IVDs have several functions. They are shock absorbing components, mainly in weights-bearing situations. They allow the flexibility and the mobility of the spine in flexion/extension and lateral bending. Finally, they count for almost 30% of the spine length. IVD degrade naturally over time due to structural changes related to the age. They can also be affected by genetic, environmental and mechanical factors which may also result in disc degeneration. From a kinematic point of view, in general, IVD have 6 degrees-of-freedom (DoF). Indeed, in addition to the three rotations which reflect the global rotations of the spine, IVD can compress vertically and shear transversally. Consequently, it was assumed that the spine could be assimilated to a beam with 6 DoF according to the Timoshenko-Ehrenfest beam theory. The computation of the associated intervertebral efforts within these IVD is possible thanks to the multibody approach and will be explained in Section 4.2.

Vertebrae and IVD are the mechanical components of the spine as well as muscle, which are attached to the bones (vertebrae and ribs). Trunk muscles are described in a dedicated section of this chapter (see Section 1.1.4). Among other mechanical components of the spine, tendons, ligaments and fascias are not considered – or not directly modelled – in this thesis. Internal organs, lungs and heart are considered as passive masses in to the model.

1.1.2 Spine diseases

The spine is subject to several disorders, which can be local or global. One of the most impressive global disorder of the growing spine is scoliosis. Scoliosis is a spine deformation which happens in the three planes, but mainly a lateral deviation (frontal plane) of the spine along with an axial rotation (transverse plane). Most (up to 80%) of scoliosis are idiopathic, i.e. without identified causes [22, 23]. Adolescent¹ idiopathic scoliosis (AIS) has a higher prevalence in girls than boys. Most of AIS will not deteriorate over time thanks to physiotherapy or bracing. However, 10% will deteriorate until the surgery cannot be avoided [22]. Spine surgery for AIS happens at the end of the puberty (end of the adolescent growth rate). Reasons for surgery are usually preventing and/or stopping curve progression, aesthetic reasons which affects patients psychologically (importance of self-image at puberty) or altered pulmonary malfunctioning due to the thoracic deformation [23, 24]. In female patients, a scoliotic spine might also prevent for epidural when giving birth. This aspect can sometimes justify surgery. The latter is called spinal fusion as the scoliotic spine is straightened up thanks to pedicle screws and rods. The distorted and flexible spine portion becomes a straighten but rigid segment. Spinal fusion consequently reduces spinal mobility. Choosing intervertebral levels to fuse requires a good balance between fusing enough to limit the spine to distort again and not fusing too much to keep enough mobility. It has been showed in the literature that there exist a high variability for the choice of fusion level [11] and a high variability of spine surgeries [25]. Besides, although the surgical treatment of AIS patients is relatively reliable, postsurgical complications exist and might require reoperation [26, 27, 28]. Furthermore, there is a risk after spinal fusion to have adjacent segment degeneration, i.e. presence of arthritis above or below the proximal and distal instrumented vertebrae [14, 15]. There is consequently a need for an objective surgical planning tool to help surgeons choose the most suitable surgery. The development of such a tool requires the mastery of spine geometry and kinematics and estimation of trunk muscle forces, in order to quantify intervertebral efforts. Estimating muscle forces must be improved [7], even for simple static configurations or for healthy subjects. Doing so may help estimate muscle forces for dynamic daily tasks such as walking and for scoliotic patients and improve finally surgical planning of scoliosis surgery.

Among other spine disorders, let us mention low back pain (LBP),

¹diagnosed between after 10 years old

spondylolisthesis and herniated disc. **LBP** mainly affects working-age people in Northern Europe and Northern America, with a prevalence from 49% to 70% [29]. 90% of reported **LBP** are non-specific, i.e. without identified cause. Spondylolisthesis occurs when a vertebra slips forward onto its below vertebra [30]. Its causes might be due to a fracture of the vertebral apex or a severe disc degeneration. Its prevalence is estimated at 50% but is asymptomatic most of the time. Finally, a herniated disc is when the nucleus pulposus goes out the intervertebral disc (rupture of the annulus fibrosus). The nucleus pulposus is likely to press on the nerve, which causes pain and neurological signs such as sensory loss or muscle weakness. People have from 30% to 40% of chance to get a herniated disc but it is asymptomatic in 90% of the cases.

1.1.3 Spine geometry and kinematics

Spine geometry refers to the description of its morphology and of its state at a given time. It begun with cadaveric studies to describe spine components and mainly to quantify parameters of muscles (localisation, insertion points, fibres direction, magnitude) [31, 32, 33, 34], tendons and ligaments and intrinsic parameters of **IVD** [35, 36]. The spine has been studied as a whole but also through its different components: functional spinal units (a vertebra and its two adjacent **IVD**), vertebra alone, **IVD** alone or spinal ligament [37]. Natural curvatures of the spine — lumbar lordosis, thoracic kyphosis and cervical lordosis — could also be measured [38]. With the emergence of medical imaging, the same parameters and geometries could be measured in-vivo [39] or with non-invasive surface instruments [40, 41]. Imaging techniques improved with time, leading to current three-dimensional (**3D**) representations of the spine on the basis of two images, e.g. EOS system [42]. Finally, cadaveric studies and in-vivo measurements led to the development of musculoskeletal models of the spine [7, 8]. The latter — multibody **MBS** or finite-element (**FEM**) models depending on the purpose of the study — are often generic describing the geometry of the spine that can be scaled to different populations or varied spine configurations. Advanced model reconstructions were also developed, namely the **3D** representation of the spine based on bilateral X-rays images of scoliotic patients [43, 44]. One of the main current challenges of the **MBS** approach in biomechanics is to develop subject-specific models and so to use input data even more [7]. To this end, most recent developments investigated to incorporate computed tomography scans [45, 46], EOS images [47] or radiographs ([43, 44] into

generic models.

Kinematics of the spine refers to its motion disregarding the forces that produce it, i.e. sagittal flexion/extension, lateral bending, stretching and axial rotation. This motion results from all the intervertebral deformations occurring at each IVD level. Ranges of intervertebral motion have been studied for all IVD levels in-vivo in a non-invasive way thanks to medical imaging [48, 49, 50, 51] and in an invasive way with wires implanted into the spinous processes of the lumbar vertebrae [52]. Historically, ranges of intervertebral motion were studied in-vitro with experiments on fresh cadaveric lumbar spine specimens [53, 54]. Simple and more complex spine motions can be also be studied based on experiments like walking or basic tasks [55, 56, 57, 58] or quantified with musculoskeletal models mentioned previously, often based on the kinematics obtained from an upfront experiment.

1.1.4 Trunk muscles

Functionally trunk muscles can be categorised as flexors, extensors, lateral benders and rotators according to the motion they impose to the trunk. Muscles often share several functions, namely for axial rotations. Anatomically, they can be studied according to their plane – anterior for abdominal muscles and posterior for back muscle –, their localisation in this plane and their depth starting from the skin.

From a posterior view, on both sides of the spine, there are erector spinae (ES) or paraspinal muscles or paravertebral muscles as they are called in this thesis (Fig. 1.4). These muscles can be further split into lumbar paravertebral (LP) or thoracic paravertebral (TP) muscles according to their localisation in the back, lumbar or thoracic areas. They are responsible of trunk extension and are a collection of different muscles, e.g. iliocostalis (pars) lumborum (ILpL), iliocostalis (pars) thoracis (ILpT), longissimus thoracis (pars) lumborum (LTpL), longissimus thoracis (pars) thoracis (LTpT) among others. This more precised classification result from differences in muscle origins and insertions or a more refined function [60]. These muscles are on the most superficial layer of the back. In deeper layers, there are several muscles which are very localised, often connecting two or more vertebrae. These muscles take part locally in trunk extension. One can cite intertransversarii laterales and mediales, interspinales and multifidus (Fig. 1.4) for the only lumbar spine [61]. In the low back, there is also the quadratus lumborum QL which is a deeper muscle connecting the pelvis and the different lumbar

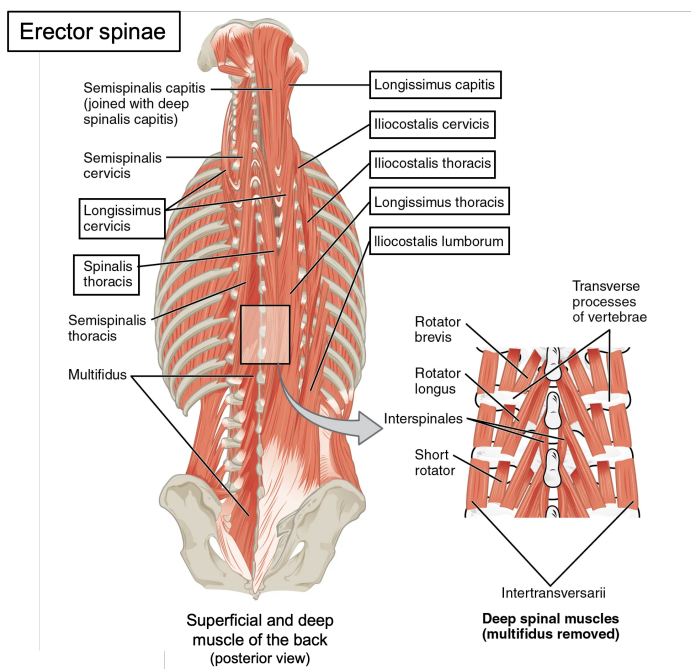


Fig. 1.4 Erector spinae (boxed text) muscles and deep back (non-boxed text) muscles. Adapted from [59].

vertebrae as well as the twelfth ribs (Fig. 1.5). It assists the trunk extension but acts mainly for the lateral bending [60]. The QL muscle will be further detailed in Chapter 10. In the upper part of the back, most of the muscles have actions of upper limbs and shoulders' motion. However, the latissimus dorsi LD (Fig. 1.5) may help the trunk's motion with hyperextension of the spine and anterior tilt of the pelvis as well as spine flexion [60].

From an anterior view, superficial abdominal muscles are more commonly known, namely the rectus abdominis (RA), i.e. the central muscle of the belly and obliquus externus (OE), i.e. the oblique muscle (Fig. 1.6). They both act bilaterally to flex the trunk forward. OE acts unilaterally to rotate the trunk or bend it laterally. A trunk rotation towards the right requires the action of the left OE, i.e. via a contralateral activity of the muscle. On the contrary, the same muscle (left OE) will bend the trunk laterally towards the left, i.e. ipsilateral activity. Deeper there is obliquus internus or internal oblique muscle. Its main functions are maintaining abdominal viscera,

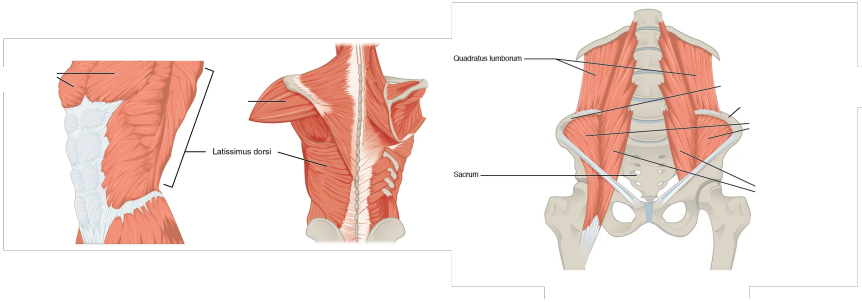


Fig. 1.5 (Left) Latissimus dorsi (Right) Quadratus lumborum. Adapted from [62, 63].

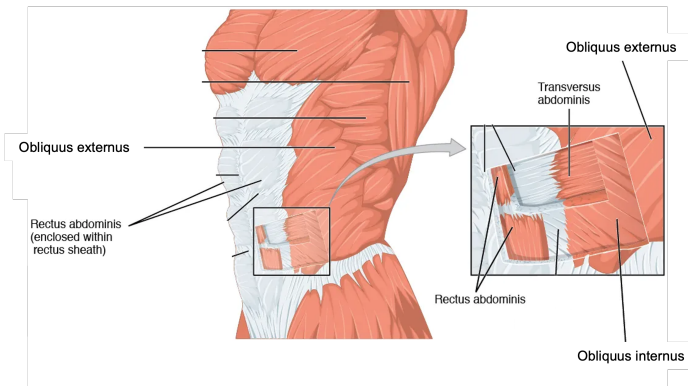


Fig. 1.6 Abdominal muscles. Adapted from [63].

trunk flexion and lateral flexion. Finally, there is the transversus abdominis which is also a deep abdominal muscle and helps to flatten the abdominal wall and to compress the abdomen.

Among all these trunk muscles, as the objective of this thesis was to use surface **EMG**, only superficial muscles were studied. For back muscles, it meant that **LP**, **TP** and **LD** were considered, as well as **QL**, despite its deeper localisation. Indeed, it is believed that some of its fibres are more superficial in a specific area of the back (for more details, see Section 3.2). The only superficial abdominal muscles were **RA** and **OE**. It gave a total of six muscles for which **EMG** signals could be recorded during the experiment.

1.2 Thesis objectives

The main objective of this thesis is to quantify trunk muscle forces:

- In the short term, to improve the quantification of intervertebral efforts;
- In the medium term, to provide the physiotherapist with a better understanding of the actions of the trunk muscles and new information for rehabilitation purposes;
- In the long term, to provide the surgeons with a surgical planning tool.

As already said, three fields must be successively investigated to develop a surgical planning tool: spine geometry, spine kinematics and muscle forces. Nowadays, research developments must be achieved to improve muscle force estimation [7, 16] whose assessment is an essential prerequisite to quantify the intervertebral efforts.

As far as we are concerned, quantifying muscle forces requires the use of mathematical approaches and musculoskeletal models. However, to solve the muscle redundancy problem and to better represent the variety of encountered muscle strategies, we propose to use a hybrid approach mixing experimental measurements and mathematical analyses. This approach is characterised by making the most of [EMG](#) signals obtained and processed a priori from an upfront experiment as input into the [MBS](#) model.

The objectives of this thesis are of different natures, associated with three distinct developments:

- Experimental: well thought out design of an experiment and protocol to extract [EMG](#) signals sufficiently reliable to be used;
- Modelling: development of a [MBS](#) musculoskeletal model of the trunk to compute intervertebral efforts, given muscle strategies;
- Methodology: development of the hybrid approach for static configurations to involve the [EMG](#) signals within the model. This part distinguishes two complementary approaches:
 - The first is based on an EMG-based deterministic distribution of a global equivalent force into muscle fascicles of the model (this step does not require the use of an optimisation process);

- The second is an EMG-assisted optimisation in which EMG signals are added to the optimisation process.

1.3 Thesis structure

The rest of the thesis is divided into two main parts: Part II explains and details the materials and methods used in the frame of this thesis and Part III presents how the hybrid approach was designed, fine-tuned and tested from the origin of the project until its end.

Following this introduction, a thorough state-of-the-art about the quantification of muscle forces and intervertebral efforts is presented in Chapter 2.

The materials and methods in Part II are subdivided into three chapters. In the first one, Chapter 3, the EMG equipment and its calibration are first explained. Then the main experiment, the loaded isometric contractions (LIC) in the Sorensen test posture, carried out during this thesis is presented. All other peripheral experiments (variants of the main experiment or new protocols) will be described into dedicated sections in Part III.

Once the experimental work has been presented, one will move in Chapter 4 to the MBS model of the Sorensen test posture, including the multibody concepts and formalism used in the frame of this research.

Finally, a sensitivity analysis of the model is proposed in Chapter 5 with respect to the different optimisation implementations and parameters.

In Part III, the hybrid approach will be incrementally built through different studies. First, a quick overview of the pilot study about predefined static postures in weights devices will be presented in Chapter 6.

Then, the first version of the hybrid approach, i.e. a purely deterministic distribution of a global equivalent force, will be compared to a classic mathematical optimisation in Chapter 7. This method will be applied first to a sample of 19 subjects to illustrate the diversity of muscle strategies and how they can be discriminated thanks to the incorporation of the EMG in the input of the MBS model. Then, the same method will be applied on one additional subject producing three different muscle strategies, revealing how EMG can help to solve the muscle redundancy problem.

Chapter 8 will present the second version of the hybrid approach in which EMG is considered as input in the optimisation process. Again, this will be applied on the 19-subject sample and on the additional subject capable of producing three different muscle strategies.

Chapter 10, last chapter of Part III, will focus on a deep back muscle already mentioned, the QL muscle (Fig. 1.5). Indeed, it appeared thanks to a parametric study of the Sorensen test model that this muscle had a significant impact on intervertebral efforts. That was the reason why it was considered in the hybrid approach and why its activity was measured with surface EMG although this practice is commonly rejected in the literature as it is a deep muscle. However, in this chapter, we will show how this muscle is important given the fact that it can impact intervertebral efforts, and how its action can still be measured with surface EMG. Finally, we will show that EMG amplitudes of the QL are significantly different between males and females under specific loading conditions.

To end this manuscript, Chapter 11 of Part IV will bring the main conclusions of the thesis with a summary of principal contributions and future perspectives in the field.

2

State-of-the-art

As previously said, whereas both spine geometry and kinematics have been largely studied in the literature, muscle forces are more difficult to quantify. Similarly, there is no much data on intervertebral efforts. There are however more data available for intervertebral efforts than for muscle forces. In this chapter, we will go through the state-of-the-art firstly of the quantification of muscles forces, and secondly of the quantification of intervertebral efforts or loads into the spine.

2.1 Quantifying muscle forces

Muscle forces can be assessed either via experimental, mathematical approaches or the combination of both, i.e. hybrid approaches.

2.1.1 Experimental approaches

Experimental approaches specifically rely on experimental measurements. One of the most used is **EMG**¹. An **EMG** signal is a an electric signal, usually fluctuating between $-5 \mu\text{V}$ and $+5 \mu\text{V}$ before **EMG** processing (-5 mV and $+5 \text{ mV}$ with gain), reflecting the electrical field created by the ionic motion of sodium and potassium ions going inwards and outwards muscle cells.

¹In this thesis, **EMG** refers to passive **EMG**, i.e. measuring the electric signal produced by the muscle. Active **EMG**, or neuro-muscular electrical simulation, is not considered.

EMG is this an image of the muscle activity, i.e. its ability to generate force. The more intense the **EMG** signal is, the more the muscle units are recruited to generate and increase the force. **EMG** has been extensively studied since the second half of the XXth century [17] and is still a field of active research, attesting by three recent publications on this topic about electrode selection [64], amplitude normalisation [65] and terminology [66].

EMG gathers several techniques of recording muscle activity. One can first distinguish surface **EMG**, which is a non-invasive technique, from intramuscular **EMG**, which is invasive. Surface **EMG** relies on electrodes placed on the skin. It is therefore limited to superficial muscles, i.e. the closest to the surface. In this category of electrodes, there are conventional surface electrodes and matrix surface electrodes [64]. Conventional surface electrodes consists in measuring the potential difference between a pair of two electrodes while matrix surface electrodes are grids of highly-condensed electrodes. This second type of surface **EMG** allows a better spatial resolution, thank to principal components analysis, and is less sensitive to the alignment of the pair of electrodes with the fibres' direction [67, 68]. For deeper muscles, intramuscular **EMG** is required. There are two types of electrodes in this category as well: fine-wire electrodes for which a fine diameter insulated wire is placed into the muscle, and needle electrodes for which a needle is placed into the muscle. Intramuscular **EMG** requires a highly-trained experience in surgery and in anatomical structures, a sterilisation procedure and is risky as it can imply trauma, pain or infection. On the other hand, surface electrodes are easy-to-use as they only require a good understanding and knowledge of surface anatomy. However, they require skin preparation to reduce skin impedance and improve the signal quality. Furthermore, they are limited to the study of superficial muscles.

EMG signals being noisy by definition unavoidably require signal post-processing. Depending on the equipment, the studied muscle(s) and the type of activity, raw **EMG** signals are filtered using low-pass, high-pass or bandwidth filters. After filtering, one can compute the **EMG** amplitude as the root-mean-square (**RMS**), the average rectified value or the mean absolute value over the actuation time [66]. The **EMG** amplitude is often normalised by the **EMG** amplitude recorded during a maximal voluntary contraction (**MVC**) test [65, 69]. This normalisation allows subjects to be compared more objectively with each other or even to compare a same subject in experiments carried out on different days. **EMG** amplitudes are often considered to be linearly correlated to the muscle force in isometric situations [68, 70, 71]. Estimating muscle force using **EMG** is computed in

the time domain. On the basis of raw EMG signals, muscle fatigue can also be determined but in the frequency domain; this aspect is not covered in the frame of this thesis.

Regarding EMG, and in particular surface EMG, crosstalk cannot be avoided. Crosstalk refers to the EMG signal produced by a muscle other than the muscle of interest [66]. The EMG signal is contaminated by another muscle's electrical activity. This phenomenon is current when recording EMG signals of thin muscles or of muscles localised deeper from the skin surface. Crosstalk is therefore limited when using intramuscular EMG whereas surface EMG is more prone to it [64]. Crosstalk is considered as one of the most important source of misinterpreting surface EMG because it is difficult to quantify by non-invasive methods and because it can be reduced but not totally removed [72, 73]. To reduce crosstalk, it is advised to use a small electrode size, reduce the inter-electrode distance as possible, i.e. at least smaller than 20mm, and ensure a low skin impedance and a proper electrode localisation, i.e. on the belly of the muscle [72, 73, 74, 75]. It was also hypothesised that crosstalk could be reduced when EMG signals are normalised by isometric MVC [72].

About trunk muscles, EMG has been rarely used to directly estimate muscle force. For this segment of the human body, EMG amplitudes are often used to compare groups of population, e.g. male vs females subjects to identify sex-related differences [76, 77], healthy subjects vs LBP patients to test rehabilitation and therapeutic exercises [78, 79, 80, 81, 82, 83]. Besides, for LBP detection, spectral parameters of EMG signals, similar to those used for fatigue assessment, have been extensively studied either to detect LBP or to improve the knowledge of muscle characteristics of LBP patients [84, 85, 86, 87].

Among the limitations of EMG, one can point out the limited frequency spectrum usable due to the noise, specially in the lower frequencies range of EMG signals. To solve this problem, more recent studies have proposed to use mechanomyography for muscle force estimation. Mechanomyography consists in measuring low-frequency lateral oscillations induced by the contraction of muscle fibres [88]. These oscillations can be measured on the skin surface in a non-invasive way thanks to a transducer, e.g. an accelerometer, a microphone, a piezoelectric contact sensor, a goniometer or a laser distance sensor [89]. Unlike EMG, mechanomyography is not subject to skin impedance and does not require an accurate localisation of the sensor. From a clinical point of view, this technology is therefore preferred compared to EMG. For muscle force estimation, as EMG, there

is no direct measurement of the force. It has been proved that there exists a linear correlation between mechanomyography and force [88, 90]. Similarly to [EMG](#), both temporal and spectral parameters can be extracted from mechanomyography signals. Thereby, mechanomyography is often associated to [EMG](#) to provide more temporal information about force fluctuations [91, 92]. Furthermore, muscle fatigue can be assessed on the basis of spectral parameters of mechanomyography [89]. For trunk muscles particularly, this technology has been validated to assess the fatigue of lower back muscles [89] and to estimate muscle force of respiratory muscles [88], which are difficult to access in a non-invasive way.

A last category of non-invasive techniques to estimate muscle force is medical imaging, and more specifically ultrasonography. It consists in measuring shape, size or structure of a muscle with an external probe placed on the skin surface. While its primary use was dedicated to detect and identify pathology [93], its application is now larger with ability to characterise the muscle in rehabilitation with real-time measurements, even for trunk muscles [94]. In the frame of muscle force estimation, the physiological cross-sectional area ([PCSA](#)) is a variable quantifiable via the measurement on ultrasonography images. Indeed, [PCSA](#) can be related to the maximum isometric force via the maximum muscle stress ([MMS](#)). However, it cannot be used for an instantaneous estimation of the muscle force as ultrasonography is not reliable enough to estimate changes in muscle activity [95, 96]. Its application is therefore limited to isometric contractions in the frame of muscle force estimation. Furthermore, the use of [PCSA](#) to estimate the maximum isometric force via [MMS](#) is open to discussion. Indeed, the value attributed to [MMS](#) in the case of trunk muscles is not constant through studies [97]. Even more, it is believed that the [MMS](#) value could not be a constant at all, as highlighted for the elbow joint by Buchanan [98]. A recent study nevertheless showed that ultrasonography could be associated with bioelectrical impedance analysis to estimate trunk muscle strength in both flexion and extension. The estimated strength is not the muscle force itself but the resultant force of all trunk flexors and extensors [99].

Related to the precedent point, let us mention that it is possible to estimate the trunk strength thanks to (isokinetic) dynamometers. It is not an estimation of the muscle force but it can indicate the strength of all agonist muscles for a given motion (flexion, extension, lateral bending or axial rotation). Often associated with another experimental measurements, e.g. [EMG](#), they can provide additional information for rehabilitation or trunk assessment [100].

Up to now, only in-vivo experimental measurements have been considered. There exist obviously in-vitro measurements for muscle force estimation. After dissection of a cadaver specimen, the targeted muscle can be isolated for characterisation. This method does not allow in-situ characterisations, i.e. it is not realistic with respect to the muscle localisation in the trunk, its orientation, etc. Alternatively, Wilke et al. estimated muscle forces of RA and ES by isolating cadaveric lumbar spines [101]. Muscles were simulated by cables to more realistically represent the lumbar spine and its surrounding structures. In-vitro measurements were approximately close to in-vivo measurements [102] according to the authors.

To conclude about experimental approaches, there are indirect measurements of the muscle force, except in-vitro measurements but which are less realistic. Until recently, there were often invasive, except surface EMG. Recent developments [89, 90, 94] in sensors allowed a wider use of mechanomyography and ultrasonography which are also non-invasive. Whereas they could offer an alternative to EMG, they are often combined with it [91, 92]. All these technologies require experimental measurements and volunteers and are consequently time-consuming and often limited to a small sample of subjects.

2.1.2 Mathematical approaches

Mathematical approaches used to estimate muscle forces often rely on a generic musculoskeletal model. Unlike experimental approaches, they do not require neither experiment nor subjects. They can therefore be applied to a larger population. For that purpose, a generic model is classically scaled to the participant's morphology and net torques/muscle forces are estimated via inverse dynamics, i.e. the computation of internal efforts for a given motion of the human body. In a recent review, Lerchl et al. identified 13 original MBS models over 81 articles found in the literature [7]. These 13 original MBS have been developed on 8 different multibody software but with a greater prevalence of the open-source software OpenSim [103] and of the commercial software AnyBody (AnyBody Technology A/S, Aalborg, Denmark). Besides, three of the original models have been more widely used or extended, two on OpenSim [104, 105] and one on AnyBody [106]. The choice of the model to use depends on how detailed the model needs to be and, therefore, depends on biomechanical functionalities of the software and available numerical methods for analysis. Some models include a fully articulated thoracic spine [105] while others will consider the thorax as a

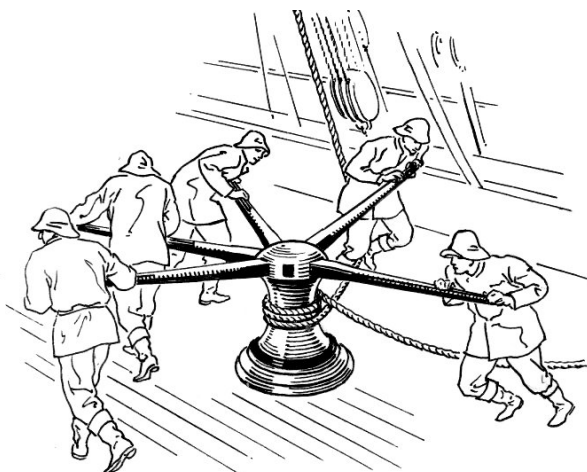


Fig. 2.1 Example of an overactuated system: a capstan actuated by several sailors. Adapted from [110].

rigid segment [104]. Eight of the 13 original models are considered as generic models according to Lerchl et al [7]. It means that their geometry is based on measurements available on databases which derive themselves on imaging data. The other 5 models are considered semi or fully individualised models, deriving from cadaveric measurements [107] or which require computer-tomography scans as input of the model [46, 47].

In addition to the **MBS** models, **FEM** models are also used to study the spine, and more specifically to distinguish in a more detailed way intervertebral efforts at each **IVD**. Unlike **MBS** models, **FEM** models usually do not include muscles [108]. Their actions are rather taken into account by modifying the stiffness of the spine structures. One can nevertheless mention one study in which the resultant actions of bilateral **ES** and **RA** have been modelled into a **FEM** model [109]. In this study, the authors modelled the in-vitro characterisation of a cadaveric lumbar spine [101] mentioned in Section 2.1.1 to estimate muscle forces. Despite a good agreement with in-vitro and in-vivo measurements, authors recognised that they computed only one global trunk muscle on each anterior and posterior sides, which represented the combined action of all the non-modelled trunk muscles.

When modelling the human body, we are automatically confronted to

the problem of muscular overactuation: there are more muscles than motion DoF. More generally, an overactuated system describes any system for which there are more actuators than DoF. For example, a capstan (Fig. 2.1), used in ships to raise the anchor, is a 1-DoF rotating machine which requires the action of several sailors (actuators in this case). Mathematically speaking, it refers to underdetermined systems, i.e. more unknowns than motion (equilibrium) equations, which admits an infinity of solutions. In the human body, this refers to the muscle redundancy problem. There is an infinity of muscle strategies, and so an infinite number of combinations of muscle forces, which produces the same motion or satisfies the same equilibrium configuration. Besides, it is important to understand that muscle forces are rarely the final result in themselves, but rather intermediate results required to assess internal efforts, such as the intervertebral ones in the case of the trunk.

To solve the muscle redundancy problem, Dreischarf et al. described in their review three classes of models [16]. The first class gathers single-equivalent muscle models. For that particular class, all the muscles acting for the motion, i.e. agonist muscles, are grouped into one unique muscle. The mathematical problem then becomes deterministic and easy to solve. However, this class of models is not physiological at all and gives only a global view of trunk force.

The second class are all based on optimisation-based models. In the latter, all muscle fascicles are modelled and the problem of muscle redundancy is solved thanks to an optimisation process. The latter aims at minimising a cost function, which is assumed to represent at best the muscle strategy favoured by the central nervous system. At the level of the muscles, most of the optimisation formulations have the generic form Equation 2.1, whose objective is to

$$\text{minimise } \sum_i \left(\frac{F_{\text{muscle},i}}{\kappa_i} \right)^p \quad \text{for } i = 1, \dots, m \quad (2.1)$$

or the same cost function but including the p-root of the above expression to avoid numerical overflow. In Equation 2.1, $F_{\text{muscle},i}$ is the force (N) of the i^{th} muscle fascicle, κ_i is a characteristic parameter of this muscle fascicle, e.g. its maximum isometric force or its PCSA and p is an exponent whose value, usually greater than 2, is assumed to mimic the physiological strategy of the central nervous system. Obviously, this equation has for common-sense objective to promote the lowest values of muscle forces, which goes hand

in hand with a reduction of energetic cost, and therefore to maximise the muscle endurance [111]. For the trunk muscles, cost functions classically used and based on Equation 2.1 are summarised in Table 2.1. As it was just said, they all tended to minimise forces. It must be noticed that if the value of the p parameter is high, i.e. $p \rightarrow \infty$, it had been showed [112] that Equation 2.1 was equivalent to

$$\text{minimise} \left(\max \left(\frac{F_{\text{musc},i}}{\kappa_i} \right) \right) \quad \text{for } i = 1, \dots, m \quad (2.2)$$

Increasing the value of the p exponent could be seen as a smoothing process on muscle forces in order to spread out the action into all the muscle fascicles and not only the largest. Equation 2.2 has not been tested yet on the trunk according to the authors' knowledge. It had been recently tested in modelling of gait of healthy and spinal-cord-injured subjects [113].

In addition to these cost functions (Table 2.1 and Equation 2.2), it has been also proposed to use the muscle fatigue criterion [114], which consists in maximising the endurance time – non-linear function of muscle forces maximum muscles forces and percentage of slow twitch fibres –, or the double linear criterion [115]. The latter aims at solving two linear programs sequentially; the first one to minimise the resultant maximum muscle intensity (force of a muscle divided by its PCSA) and the second to minimise the sum of muscle forces while using the resultant maximum muscle intensity obtained as a limit of the second problem. The minimisation parameter can also be the muscle activation instead of the muscle force [116]. The muscle force is then considered as a constraint function whose expression can be various. There also exist cost functions non-associated to the muscle forces but to other trunk parameters such as minimising intervertebral forces and/or torques [18, 117, 118, 119], trunk displacements and intervertebral displacements [118]. Stokes and Gardner-Morse also proposed a multi-criterion optimisation by combining cost functions previously cited. They showed that there was a good agreement between experimental data and a combination of minimising the sum of cubed muscles stresses and minimising the intervertebral displacements [118]. Finally, there also exist criteria based on spine stability for which the spine stiffness is usually computed [118, 120]. At that point, one could already ask oneself which of these cost functions is the most physiological, i.e. is the most representative of the human body, e.g. mimic the effect of co-contraction or muscular compensation strategies. In addition to the objective function(s), the optimisation

Objective	Equation	References
Minimise the sum of muscle forces	$\min \sum F_{musc}$	[18, 119]
Minimise the sum of muscle stresses	$\min \sum \left(\frac{F_{musc}}{PCSA} \right)$	[18]
Minimise the sum of squared muscle forces	$\min \sum F_{musc}^2$	[18]
Minimise the sum of squared muscle stresses	$\min \sum \left(\frac{F_{musc}}{PCSA} \right)^2$	[18, 118]
Minimise the sum of cubed muscle stresses	$\min \sum \left(\frac{F_{musc}}{PCSA} \right)^3$	[18, 119, 120]

Table 2.1 Cost functions associated to muscle forces classically used in the literature for the trunk.

process can also include constraint functions. Classical constraint functions are bounds on muscle forces, e.g. positive values, but not exceeding maximum isometric forces.

The last class described by Dreischarf et al. contains models which are driven by experimental input and in particular by **EMG**, i.e. EMG-driven models. The idea behind this category of models is to use parameters of **EMG** signals, such as **EMG** amplitude or muscle activation to estimate the muscle force and help to solve the muscle redundancy problem. This category is strongly based on the existence of an **EMG** amplitude-force relationship, which is often considered as linearly correlated [68, 70, 71]. As example, Granata and Marras proposed in 1995 an EMG-assisted model to compute muscle forces and resulting intervertebral efforts for lifting tasks [121]. They defined forces in each muscle as function of the normalised **EMG** signal of the corresponding muscle multiplied by a gain, the **PCSA** of the muscle and two modulation factors, one depending of the muscle length and the other of the muscle velocity. This class of model is the closest to the physiological strategies recorded during experiment but they may lack to satisfy equilibrium conditions in terms of spine stability. Furthermore, they share some of the limitations of **EMG** technology such as the limited use to superficial muscles and the cross-talk phenomenon. Moreover, the **EMG** amplitude-force relationship might be difficult to evaluate. Therefore, as proposed in this work, there is a tendency to combine **EMG** and

optimisation-based computations, i.e. hybrid approaches or EMG-assisted optimisation approaches, which will be covered in Section 2.1.3.

Until now in this section, the muscle fascicle could be considered as a pure force transmitter. However, its representation can be more complex. Indeed, Hill showed in 1938 that a muscle, or more particularly the muscle-tendon unit, could be modelled as a spring element placed in parallel with a damping element and a tension source, all in series with a second spring element [122]. This model illustrates the length-dependence and the velocity-dependence of the muscle-tendon unit. The Hill model is still valid but has been made more complex over time [123, 124, 125]. This higher complexity, recently reviewed [126], results in many more parameters to estimate, task which is difficult to do in-vivo.

2.1.3 Hybrid approaches

In this work, by hybrid approaches, we mean all these which use input from an upfront experiment to help for the muscle force estimation. It means that not only geometric and kinematic data but also EMG signals are considered as valuable information to develop the model and to compute muscle forces.

As already said in the previous section, hybrid approaches are not new. Granata and Marras proposed what they called EMG-assisted modelling technique to make the most of EMG signals [121]. Their equations were used afterwards and compared with an EMG-assisted optimisation approach at the S1/L5 level of the lumbar spine [127]. Their strategy for the EMG-assisted optimisation approach was to perform the least possible adjustment with respect to the initial estimate of muscle forces given by the equations of Granata and Marras. This approach has been compared a few years later to a FEM model with an optimisation-based approach for muscle forces to compute intervertebral efforts at the different levels of the lumbar spine for isometric load lifting tasks [128]. Both methods produced relatively different muscle strategies. In particular, due to the asymmetry between right and left EMG signals of a same muscle, the EMG-assisted optimisation approach led to non-symmetric muscle forces in contrast to the optimisation-based computation. Both models also produced significant differences in force magnitudes. In particular, the EMG-assisted optimisation approach activated all the muscles, including abdominal muscles (co-contraction phenomenon [129, 130]), whereas the optimisation-based method favoured the activation of the large global trunk muscles.

A more recent study also compared three different EMG-assisted optimi-

sation approaches with a purely optimisation-based solution [120]. The three approaches were relatively close to each other: the EMG-force relationship was a linear relationship, i.e. $F_{muscle,i} = nEMG_i * PCSA_i * MMS$, in which $nEMG_i$ is the normalised EMG amplitude of the i th muscle, $PCSA_i$ its PCSA and MMS the value of the MMS, equal for all the muscles. The approaches differed in their way of considering the passive muscle force, representing the force-length relationship, or how to include the muscle activation estimator as a bound or an inequality constraint in the optimisation process. Each approach also included a gain factor which was specific to each muscle. According to the authors, the large range of gain values and the different manners to include them into the optimisation process led to significantly different muscle forces. Also, the spine stability was not always guaranteed with the hybrid approaches while the optimisation-based solution provided a stable spine for every simulated case.

Hwang et al. also described an EMG-force relationship similar to the one described previously: the muscle force was subdivided into active and passive muscle forces. The active part was very similar to the EMG-assisted equations proposed by Granata and Marras [121] and depended directly on the EMG muscle activity, a gain factor, the PCSA and two dependencies to length and velocity. The passive part was function of the gain factor, the PCSA and had also a muscle-length relationship [131]. The authors have defined this EMG-force relationship in the frame of a larger project aiming at developing a curved muscle model instead of a straight line muscle model [131, 132, 133]. The geometry of the curved muscle model had important impact for complex load lifting tasks with different starting and reaching positions. Their global approach behind the development of the curved muscle model was to make the model more subject-specific in terms of muscle force estimation.

All these studies including hybrid approaches focused on load-lifting tasks when standing or in flexed postures. Besides, these tasks were mainly dynamic, hence dependencies to muscle length and velocity in the force expression. The latter increase the complexity of computations because muscle length and velocity are difficult to estimate and are often subject-dependent. Furthermore, EMG input was globally included in a similar way described by Granata and Marras [121]. The muscle force estimation also depended on the muscle gain which could lead to very different muscle forces as highlighted by Mohammadi et al. [120].

This relatively complex relationship between EMG and muscle force has not helped to develop the use of hybrid approaches. One way to apply them

more appropriately could be to integrate **EMG** into the force expression with an even more simplified manner and to limit the study on well-defined static configurations to eliminate the dependencies to muscle length and velocity.

2.2 Intervertebral efforts

As already stated before, in this thesis, muscle forces are intermediate results. Combined spine geometry, spine kinematics and muscle forces are needed to compute intervertebral efforts acting at each **IVD** level. They will be indicators of excessive exposure or even rupture of the spine [9]. Intervertebral efforts, image of the spinal loads, have been studied in the literature via three approaches: in-vitro characterisation of cadaveric specimens, in-vivo studies or **MBS** and **FEM** models. In most of studies, intervertebral efforts are translated into equivalent stresses given the disc area. Sometimes, authors measured and reported the intradiscal pressure values inside the **IVD** because of the use of a pressure sensor [102, 134]. Although intradiscal pressures and equivalent stresses, in particular the normal stress due to compressive forces, are obtained quite differently, it has been showed that they could reflect each other in a relatively reliable way [135].

2.2.1 In-vitro studies

In-vitro studies consist in isolating a cadaveric specimen of a spine segment, often the lumbar spine, mounted into a spine tester. The lumbar spine is afterwards submitted to external forces, torques and/or displacements while the variables of interest are measured with appropriate sensors. In its simplest form, Adams et al. tested lumbar joints in flexion [136] and in combined torsion and compression [137]. Knowing the applied force, they measured the displacement of the joint and drew typical flexion-displacement and torsion-displacement curves. Besides, they quantified which structures of the joint (facets, ligaments, **IVD**) were responsible for the resistance to the applied forces. In their book, White and Panjani also showed mechanical tests on spine segments [138]. In addition to compression and torsion tests, they reported typical values of **IVD** in traction, bending or shear while damaging or not the disc. The creep phenomenon of the **IVD**, i.e. the time-dependent part of the deformation associated with the application of a load to the **IVD**, has also been studied in order to improve its modelling [36]. Characterising **IVD** could also be situation-dependent such as vibrations

provoked on the disc in the frequency range of a car seat [35].

All these studies focused either on one thoracic or on one lumbar joint, i.e. one IVD between two vertebrae. A more recent in-vitro study reported intradiscal pressures of the whole lumbar spine for various loads [101]. Finally, in order to validate models or report input to develop model, Stokes and Gardner-Morse developed a database reporting the behavior of the whole lumbar spine for different loading cases [139]. Unlike the studies previously cited at the beginning of this section, for this database, cadaveric specimens were equipped with a six-degree-of-freedom load cell to record forces and torques in the three directions. Specimens were submitted to displacements thanks to linear actuators whose motion values were recorded with six (three translations and three rotations) displacement transducers. The database was developed specially for FEM models but could open the way for other studies.

2.2.2 In-vivo studies

In the case of in-vivo studies, intervertebral efforts or intradiscal pressures are directly measured in subjects performing tasks. One of the most known in-vivo study for the lumbar spine was performed in 1999 by Wilke et al. [102]. They implemented an in-vivo pressure transducer in one healthy male subject and recorded intradiscal pressure during different daily tasks, e.g. lying, standing, sitting, walking or lifting loads. This study is often cited as it serves as validation for the development of spine models, e.g. [46, 105, 106, 107] among others. Wilke et al. were not the first to implant a pressure transducer for in-vivo measurements: Nachemson had already done it in the 1960s, also for different daily tasks [140] and Schultz et al. in four subjects with a comparison with estimated compression forces [117].

More recently, a telemetrised vertebral body implant has been developed and implanted in patients which had a specific compression fracture of the vertebral body [141]. Once implanted, patients were asked to come to the laboratory to perform an impressive number of daily tasks during which intervertebral efforts in the six directions were recorded thanks to the implant [142]. All the tasks and associated data have been compiled into an open-access database called Orthoload². Besides, this open-access database also gathers equivalent data for other joints such as the hip, the knee or the shoulder [143]. Among all the daily tasks recorded over time and reported in the literature, the ones producing the highest intervertebral efforts were

²<https://orthoload.com/database/> (consulted on September 21st, 2023)

load lifting tasks combined with a trunk flexion. However, there were large intra- and inter-individual variations and a large dependence on how the tasks were performed [16]. To conclude about in-vivo measurements, they are necessary to validate models of the spine but, being invasive by nature, they can affect the patient health and are often reduced to few subjects, i.e. from one to twenty [16], what does not represent statistically the reality.

2.2.3 Musculoskeletal models

As for the quantification of muscle forces, musculoskeletal models can be an appropriate alternative to estimate intervertebral efforts in order to represent a larger portion of the population. As mentioned previously, a lot of models of the spine have been developed with more or less details, the goal being to compute values of intervertebral efforts [7]. Similarly to in-vivo measurements, greater compressive and shear forces were obtained for load lifting tasks combined with trunk flexion, dynamic exercises or sudden loading conditions, compared to standing or lying conditions [16]. Among models, one recent co-simulation approach, developed to improve the estimation of intervertebral efforts, can be highlighted [144, 145]. This co-simulation approach consists of one **MBS** model combined with a **FEM** one. As muscles cannot be modelled realistically in a **FEM** approach, they are computed in a **MBS** models in order to be afterwards injected into the **FEM** model. Thanks to the latter, intervertebral efforts can finally be computed given the mechanical characteristics of the spine components (**IVD**, ligaments, facets) while taken into account a more realistic action of trunk muscles.

To conclude about intervertebral efforts, one recent study proposed subject-specific regression equations to estimate compressive and shear forces in standing and flexed postures for the two lowermost lumbar disc levels, i.e. S1/L5 and L5/L4 [146]. For that purpose, the authors simulated over 3840 cases using a combined **MBS** and **FEM** model in which they vary different parameters, e.g. body height and weight, distance to the external mass, value of the external mass and flexion angles of the trunk in both sagittal and frontal planes. They concluded that trunk (sagittal and lateral) flexion angles, hand-load mass and moment arm and body weight influenced the most spinal loads unlike the sex or the body height. Their estimated spinal loads were in strong agreement with in-vivo measurements whereas their predicted muscle activities were in moderate to weak agreement with muscle activities recorded in healthy subjects. As far as

we are concerned, this finding might be hazardous as intervertebral efforts depend on the external load and muscles' action mostly. However, the authors recognised that their work could not reflect inter- and intra-subject variability in terms of muscle strategies [146].

2.3 Thesis contributions

In line with the state-of-the-art given about the quantification of muscle forces for the trunk and the need of a better muscle force estimation [7, 16] and given the limitations of both experimental and mathematical approaches, this thesis contributes to the improvement of trunk muscle forces estimation via the use of a hybrid approach, the subsequent goal being to assess intervertebral efforts, in the case of a static configuration.

For that purpose, we have designed a new hybrid approach which relies on an upfront experiment where EMG signals are recorded. These signals are taken into account at two steps of the model. First, they are indicators of which muscles have to be included in the model. Low EMG amplitudes meaning a weak-participation of associated muscles, their modelling is therefore not necessary as it will unnecessarily complicate the optimisation process and slow down the computation time. Secondly, for the muscles present in the model, their force is estimated at best as a function of EMG signals, according to two different approaches and with several precautions.

Two parts were considered for the hybrid approach developed in this thesis. The first one relies on a deterministic distribution of a global equivalent force among muscle fascicles to satisfy a static configuration. This distribution is based on weighted factors, themselves based on EMG amplitudes of the associated muscles. The second part could be named as an EMG-assisted optimisation method as EMG is directly embedded into the optimisation process. For this second part, two methods to embed EMG into the optimisation process are explored.

To eliminate dependencies to muscle length and velocity, a static configuration is favoured in this thesis. Furthermore, to limit the use of weights devices and their modelling, an isometric exercise involving only the gravity force is chosen: we decided to focus on the quantification of muscle forces for the Sorensen test posture.

It is relatively complicated to compare muscle forces directly with the literature. Indeed, there exist no comparable indicators as muscle forces depend on muscle strategies and how their respective contributions are computed. Therefore, intervertebral efforts are favoured to compare our results

with these from the literature. The three intervertebral efforts computed are compressive forces in the spine axis, shear forces in the anterior-posterior direction and bending torques in the sagittal plane. As only healthy subjects are considered in this work, their muscles are assumed to work conjointly and to produce left/right symmetrical forces and torques. In other words, there is no imbalance in the frontal plane and so no shear force in the medio-lateral direction. Similarly, there are no torque in both frontal and transverse planes.

To study the limitations of our hybrid approach, we also study its application for a slow-dynamic exercise: a slow flexion/extension of the trunk in a standing position. This is somehow a validation of the model for the flexion/extension trunk motion. For that study, intervertebral efforts computed with our method are compared to intervertebral efforts obtained in-vivo [102, 142].

Finally, we bring a contribution to the quantification of the QL muscle. We show that its action contributes differently to the quantification of intervertebral efforts. In line with this result and with our conviction of including EMG in the model, we study if the use of surface EMG is however possible for this deep back muscle at a specific location and recognising a certain margin of measurement error. If so, it will improve the knowledge of QL, the estimation of its force and a better quantification of intervertebral efforts.

PART II

Materials and Methods

3

Experiment

In this chapter, we explain how the **EMG** signals were filtered as well as the experiment of the Sorensen test which is the reference experiment developed and analysed in the thesis. Every result related to this experiment will be referenced to this section. Lower-ranking experiments will be described in dedicated chapters of Part **III**.

3.1 Electromyography calibration

3.1.1 Electromyographic equipment

By nature, **EMG** is noisy and its signals must be filtered. The first step of the thesis was therefore to carefully select the filter appropriate to the **EMG** equipment at our disposal.

For recording EMG signals, the equipment was the wireless surface EMG system *BTS FREEMG 1000* developed by BTS Bioengineering (Garbagnate Milanese, Italy). Performance characteristics of the system were:

- Sample rate of 1000Hz;
- Resolution of 16bit;
- Sensitivity of 1 μ V;
- Accuracy of $\pm 2\%$.

The signals were transmitted with a frequency of 2.4GHz Industrial, Scientific and Medical radio band. The system had an input impedance of $100\text{M}\Omega$ and a common-mode rejection ratio greater than 100dB at 50-60Hz.

To limit its impedance, the skin must be cleaned with alcohol and degreased with ether [74]. Oval (35x41mm) surface electrodes (Ag/AgCl, Skintact FS-TC1/10) are stuck to the skin to record the EMG signal from the muscle. One pair of surface electrodes is placed by muscle to measure the potential difference. To maximise the magnitude of this potential difference, the interelectrode distance must be as small as possible. The smallest distance that could be reached with the available surface electrodes was 20mm which is the distance recommended in the literature [74]. The pair of surface electrodes must be ideally aligned with the muscle fibres.

After recording, raw EMG signals must rectified, filtered and smoothed for a better visualisation. The rectified signal is defined as the absolute value of the difference between the raw signal and its mean. The smoothing for the visualisation was performed with a moving mean over 150ms.

The filter must be a low-pass filter with a zero-lag phase, i.e. applied twice (back-forward procedure). Three types of filters were considered: Butterworth 4th- and 5th-order and Chebyshev 8th-order. Butterworth and Chebyshev filters are the most used current filters in the literature, either for simple use or as reference methods for new development in EMG processing [147, 148, 149].

The filter requires to define a cut-off frequency. Values reported in the literature were from 3 to 40 Hz in the case of dynamic applications [72, 73]. Merletti et al. reported the same value of low cut-off frequencies, i.e. 5-10 Hz, below which the surface EMG signals had negligible contributions [149]. Based on these reported values, a good compromise was to use a cut-off frequency of 10 Hz, which was a common value for gait applications [72]. At the time of the EMG calibration, gait analysis was still the objective of the thesis, hence the choice of 10 Hz for the cut-off frequency. It was kept identical for static configurations.

3.1.2 Filter selection: experiment and analysis

For the filter selection, one healthy 19-year old participant (175 cm, 63 kg) took part voluntarily to the experiment. The subject was with good self-reported general health. Two pairs of electrodes (Fig. 3.1) were placed bilaterally on RA, 4cm above the umbilicus and 4cm lateral to the median line [150, 151]. The reference point was at mid-distance from the electrode



Fig. 3.1 Surface EMG electrodes placed bilaterally on RA.

centers. Movement studied was based on the maximal voluntary isometric contraction described by Dankaerts et al. [152]. The subject, in supine position with his legs straight and maintained by a first examiner, was asked to curl-up while the second examiner applied a symmetrical manual resistance with his hands on subject's shoulders. After explaining the test to the subject, he was asked to perform the movement at least once before the actual EMG recording. Then, the subject was asked to perform the movement four times, two lying on the ground and two lying on a table. The subject performed four trials in total, each lasting 5 sec with 1-min rest in between.

The three filters were compared based on the RMS value over the contraction time. The RMS values were computed on the filtered (but non smoothed) EMG signals.

3.1.3 Results and conclusion

As a preliminary remark, it must be noticed that the two trials carried out on the table produced more intense EMG signals. It is believed that the second examiner could produce more resistance on the subject's shoulders when standing up compared to kneeling with the back bent for the two trials on the ground.

The filtered EMG signals for the three different low-pass filters when using the 10-Hz cut-off frequency are presented in Fig. 3.2. Visually, there was clearly no difference. However, when zooming on the curves (see

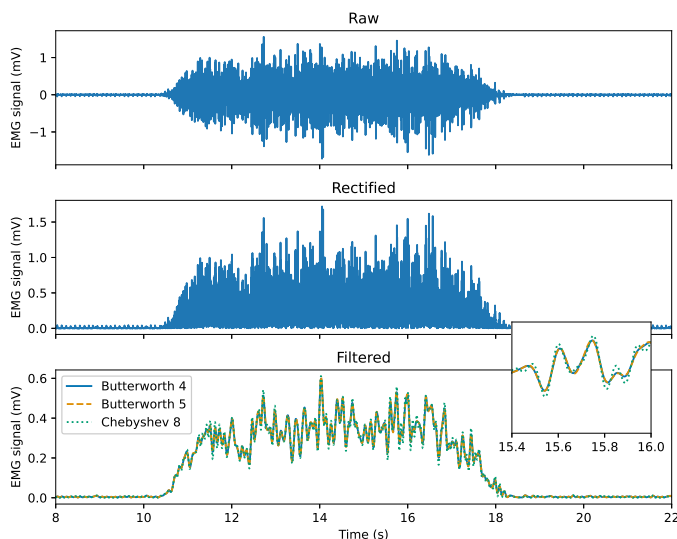


Fig. 3.2 EMG processing from raw (Top), rectified (Middle) to filtered (Bottom) signals with a cut-off frequency of 10 Hz for abdominal contraction. The boxed graph is a zoom on the filtered EMG signals between 15.4 and 16 s.

boxed graph in Fig. 3.2), the 8-th order Chebyshev filters presented a more oscillatory behavior than the two Butterworth filters. According to the curves, the RMS values for the three different filters were very close (less than 1% difference for all the trials). Given the absence of clear difference between the three filters, it was decided to work with the filter already used in the walking laboratory, i.e. the 5-th order Butterworth. The Chebyshev filter was atypical for this application and more often used as a high-pass filter [147].

In sum, for the whole work carried out in this thesis, when using the wireless surface EMG system *BTS FREEMG 1000*, it was decided to filter the raw EMG signals with a 5-th order Butterworth low-pass filter implementing a zero-lag phase and a cut-off frequency of 10 Hz.

3.2 Sorensen test

3.2.1 Description

The main experiment of this thesis was carried out in the Sorensen test posture¹ as illustrated in Fig. 3.3. The Sorensen test is a prone back extension exercise, commonly used to assess isometric endurance of the extensor muscles of the trunk. This test is considered as a valid reference tool for the evaluation of the trunk muscle fatigue in a patient with LBP [153]. The Sorensen test was created by the Danish researcher Fin Biering-Sørensen, based on a one-year research observing the inhabitants of a suburb of Copenhagen. Sørensen realised that twelve months after the test, there was a correlation between the muscle endurance time and the occurrence of LBP [154]. This test was selected for this thesis for two main reasons. First, it was necessary to select an isometric exercise which could produce relatively intense muscle contractions. Callaghan et al. showed obtained intense EMG signals for a position similar to the Sorensen test posture [155]. Secondly, this exercise had to be performed without the use of weights machines, i.e. only the gravity should act as that force is precise and very simple to model. On the contrary, weights devices would have made the model vainly complex and mostly inaccurate due to friction.

3.2.2 Subjects and experimental procedure

Nineteen healthy male participants (age: 22.7 ± 2.1 years; mass: 73.9 ± 8.7 kg; height: 1.81 ± 0.05 m) voluntarily participated in this experiment after signing an informed consent. The experiment took place in a single session, including explanations, warm-up, muscle palpation, placement of electrodes and various tests: three MVC tests followed by five LIC tasks.

At the beginning of the session, the examiner ensured that the subject met the eligibility criteria. For that purpose, biometric measurements like symmetries for equal lengths of lower limbs, balanced pelvis and spine alignment in the frontal plane and lack of concave/convex scoliosis curves were checked. These biometric measurements were selected as they were commonly carried out by physiotherapists at the 'Médecine Physique' department of the 'Cliniques Universitaires Saint-Luc' (Brussels, Belgium). Subjects with major body asymmetries were excluded. Lengths of lower

¹Whereas the original Sorensen test is assumed to hold the position until exhaustion, this thesis focuses on a modified Sorensen test. However, for the sake of readability, the term 'Sorensen test posture' will be used throughout this manuscript.

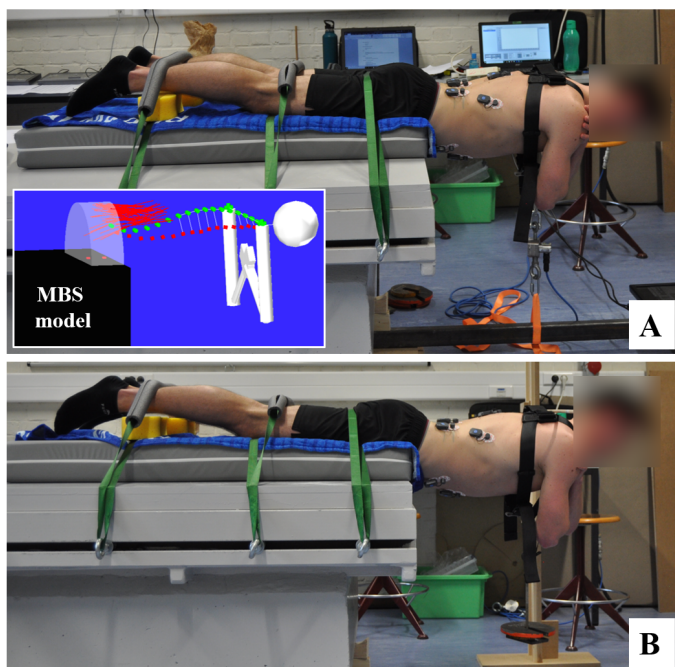


Fig. 3.3 The Sorensen test posture. (A) For maximal voluntary contractions (MVC) tests, the subject was attached to the ground and had to produce as force as he could. (B) For LIC tests, the subject was loaded with external loading. The equivalent MBS model is also showed for illustration.

limbs (distance between the greater trochanter and the lateral malleolus) were considered identical if there was less than 1 cm of difference between both legs. Pelvis and spine were considered aligned in the frontal plane if, when the subject was standing in an upright posture and the rope of a plumb line was located at the seventh cervical vertebra, the plumb line was less than 2 cm away from the sacrum. The anterior flexion test – running a gibbosity tool (Fig. 3.4) along the spine while the subject was bent forward with hand palms held together – ensured the lack of concave/convex scoliosis curves if the angle measured was smaller than 5 degrees. None of the 19 subjects were excluded in this study on the basis of eligibility criteria.

The subject then pursued with a warm-up including a 5-min global warm-up (i.e. cardio-vascular and respiratory) on a stationary bicycle [156]

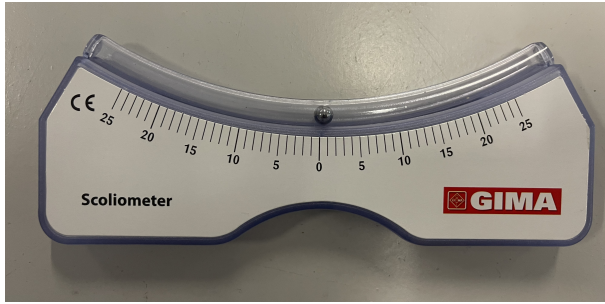


Fig. 3.4 Gibbosity tool or scoliometer (Gima, Gessate, Italy).

followed by two more specific exercises for the back (i.e. trunk muscles activation) (Fig. 3.5): the "Bridge" and the so-called "Short Lever Arm Sorensen Posture on Swiss Ball" exercises. Each of them consisted in holding the posture during 30 sec, repeated three times with a 30-sec rest in between.

MVC tests, used for **EMG** normalisation [65, 69], were performed with the subject positioned, according to the Sorensen test, in prone posture on a table with the antero-superior iliac spines of the pelvis at the edge of the table [153]. The subject's lower limbs were secured with straps at the glutes, on the distal part at the back of the thigh and behind the ankles (Fig. 3.3). The subject was instructed to contract his abdominal muscles by pushing the navel up. To ensure the isometry of the test, the subject was attached to a metal bar on the ground with a vertical slack. Each **MVC** test, during which the subject was instructed to lift the trunk by developing as much strength as possible, lasted six seconds with a 30-sec rest between each test for a total of three repetitions. Both bio- and auditory feedback were given in real time to the subject to ensure the standardisation of the test. Bio-feedback consisted in showing the time history of the force developed by the subject in real time on a screen placed on the ground. Auditory feedback was given by counting the contraction time from zero to six seconds with a neutral but loud voice.

LIC tests were performed in the same posture as for **MVC** tests but without linking the subject to the metal bar anymore. Instead, he was loaded with five different loads (from 0 to 8 kg with a 2-kg step), applied in a random order, for a total of five trials. The posture was held during 15 seconds for each trial. As for **MVC** tests, feedback was implemented. In addition to auditory - time counting - feedback, sensory feedback (rod

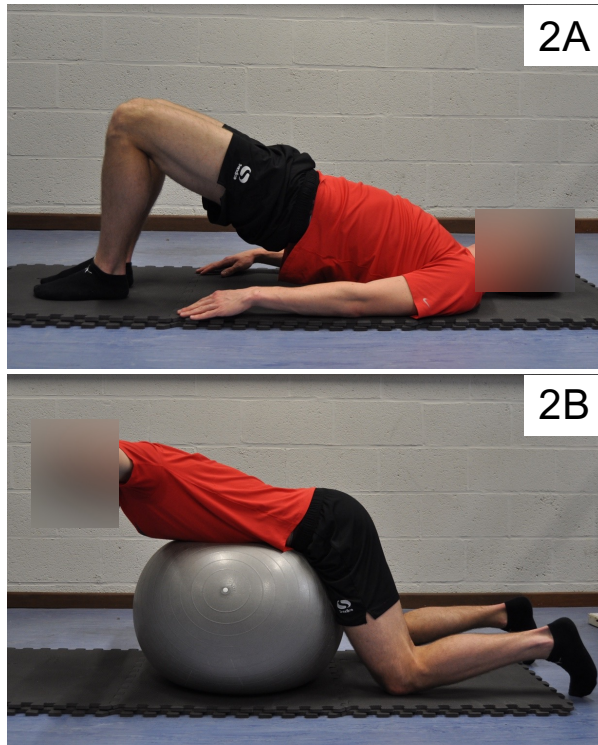


Fig. 3.5 Specific warm-up. (2A) The "Bridge" exercise; (2B) The "Short Lever Arm Sorensen Test Posture on Swiss Ball" exercise.

placed at the level of the subject's shoulder blades) was set up to help the subject maintain his horizontality. The subject was instructed to keep contact with the rod throughout the posture maintenance.

The **EMG** activity of six muscles was recorded bilaterally during the **MVC** and **LIC** tests with the 12-channel wireless surface **EMG** system described in Section 3.1.1. Surface electrodes were placed on the six muscles as follows (Fig. 3.6):

- Lumbar paravertebral muscles: at L3 level, 2 cm away from the spinous process [84, 150];
- Thoracic paravertebral muscles: at T9 level, 2 cm away from the spinous process [84, 150];

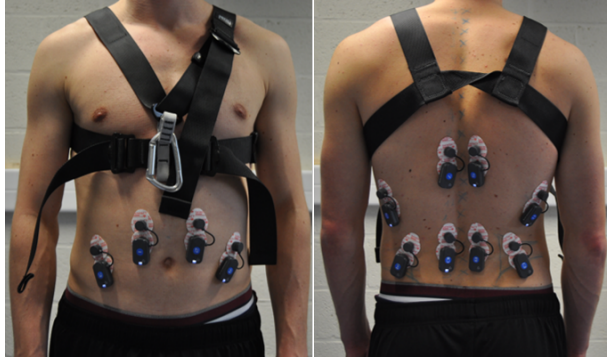


Fig. 3.6 Surface **EMG** electrodes location on (Left) abdominal muscles and (Right) back muscles.

- Quadratus lumborum: medially to the superior lumbar triangle (or Grynfeltt-Lesshaft triangle) [150]. The superior lumbar triangle was localised after identification of **LP**, the iliac crest and the lower line of rib cage which form the three sides of the triangle;
- Latissimus dorsi: at T12 level, along the line connecting the most posterior point of the posterior axillary fold and the spinous process S2 [150];
- Rectus abdominis: 4cm above the umbilicus and 4cm lateral to the median line, on the belly of the muscle [150, 151] ;
- Obliquus externus: at the junction between the straight line passing through the lowest point of the rib cage and the opposite pubic tubercle and the horizontal line located one centimetre above the umbilicus [157].

Raw **EMG** signals were rectified and filtered with a 5th-order Butterworth low-pass filter with a cut-off frequency of 10 Hz. The **EMG** amplitude of a muscle for a given **LIC** test was calculated as the **RMS** value of the filtered signal over the contraction time (15 sec) normalised by the **RMS** value over 6 sec of the **MVC** test of the corresponding muscle.

During the **MVC** tests, the force developed by the subject was measured via a force sensor mounted in series with the slack linking the subject to the ground. A harness (Chest'Air, Petzl, Crolles, France) was used to attach the subject to the metal bar during MVC tests and to load him during **LIC** tests

(load placed in a bag safely attached to the harness). During MVC tests, the force developed by the subject was measured via a force sensor (Sensy 5000 CHBN, capacity 3025 N, sensitivity 0.702 mV/V, Jumet, Belgium) mounted in series with the slack. The force was shown by bio-feedback in real time to the subject (Student Strain Gage Data Acquisition Device, Micro-Measurements, Vishay Precision Group, Malvern (PA), USA). Force levels were computed from signals recorded with the force sensor as the RMS value over 6 sec.

3.2.3 Experimental results

When analysing filtered EMG signals from all muscles, it appeared that mostly LP, TP and QL muscles were activated.

For illustration, Fig. 3.7 shows the time evolution of EMG amplitudes of right RA, LD and LP of one subject during LIC tests for three different loads. The EMG amplitudes for RA are high because they are not normalised with a MVC test dedicated to the RA). LD had weak EMG amplitudes, i.e. less than 10%MVC, whereas LP had moderate EMG amplitudes.

This observation was rather logical due to the fact that the Sorensen test was a prone back extension exercise. Weak activation of abdominal muscles and LD was also obtained for trunk extension posture previously [155]. So, only extensor muscles of the trunk, i.e. back muscles, should have been activated; so were LP, TP and QL.

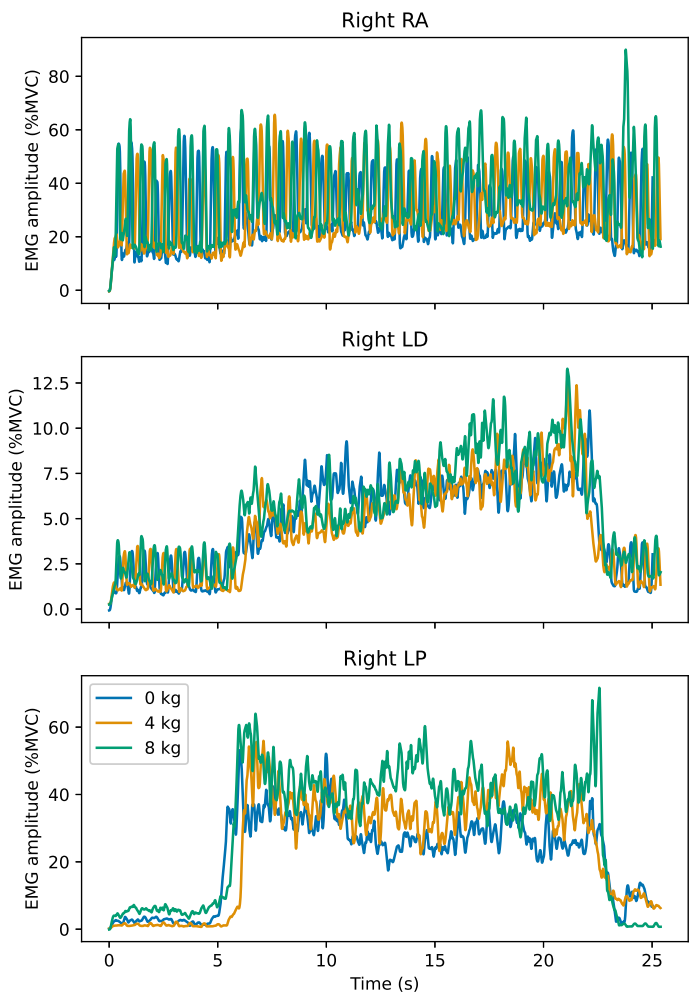


Fig. 3.7 Time evolution of the EMG amplitudes (%MVC) for (Top) right RA, (Middle) right LD and (Bottom) right LP during LIC tests for three loads.

4

Multibody approach

In this chapter, we will introduce the main model used in the thesis which is associated to the experiment described in the previous chapter. Prior to the model, the basic principles of multibody approach are presented: its formalism and the inverse dynamics method which is particularly used in biomechanics and so in this thesis. After the model, the simplest form of the hybrid approach, i.e. the EMG-based deterministic distribution of a global equivalent force into the muscle fascicles of the model, is presented as well as the optimisation-based approach to which the hybrid one is compared.

4.1 Multibody dynamics

Multibody dynamics refers to the kinematic, static and/or dynamic study of any polyarticulated system(s), denoted as **MBS**. These systems are composed of a succession of bodies connecting to each other via joints (Fig. 4.1). **MBS** motion is described thanks to the equations of Newton for body translations and of Euler for body rotations. Applications of multibody dynamics are numerous: human body, robots, vehicles or mechanisms. Furthermore, different analyses can be used in multibody approach to characterise a **MBS** such as its motion time evolution when submitting it to known forces and torques (direct dynamics) or, on the contrary, the computation of forces and torques given the positions, velocities and accelerations of the **MBS** (inverse dynamics). The latter will be further described here below.

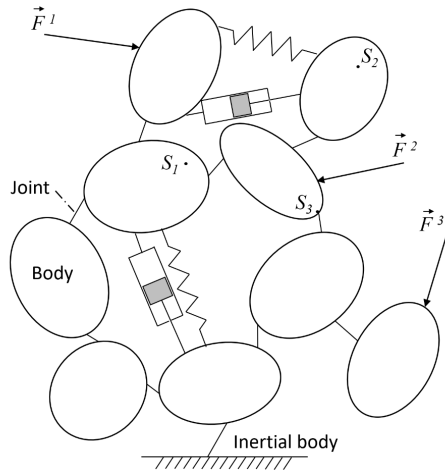


Fig. 4.1 A **MBS** is a set of bodies connected by joints, subject to possible motion constraints, internal and external forces and/or torques.

4.1.1 Formalism

As said previously, **MBS** are composed of bodies and joints, alternatively connected in the kinematic chains. Bodies can be rigid or flexible and joints, usually considered massless, represent any device able to confer **DoF** between bodies. To describe the motion of **MBS**, several types of generalised coordinates, denoted $\mathbf{q}$, can be used

- absolute, i.e. each body configuration of the **MBS** is described with respect to the inertial frame;
- relative, i.e. each body configuration is described via the joint internal motion relatively to the previous body in the kinematic chain;
- natural, i.e. each body configuration is described via the absolute position of specific material points.

In our formalism at the root of the homemade **ROBOTRAN** multibody [158], rigid bodies and relative generalised coordinates are used. In case of a tree-like structure, i.e. without any kinematic constraints, described by relative joint coordinates, a **MBS** possesses as many **DoF** as generalised coordinates.

In this thesis, we will resort to closed-loop **MBS** (Fig. 4.1). Such **MBS** contain one or several loops of bodies in their structure, leading to motion constraints between the generalised coordinates. To illustrate this, in the

case of the human body, a closed-loop configuration exists in the elbow joint in which the humerus, radius and ulnar bones would be the first branch, composed of a succession of bodies, i.e. bones, and joints, i.e. the elbow. The second branch of the loop would be the brachialis muscle inserting on the humerus and the ulna. In addition to the gravity force, an example of external force would be a dumbbell held in the hand.

Closed-loop MBS are a specific case of constrained MBS, in which loop constraints are holonomic, i.e. algebraic, assembly constraints. For these systems, equations of motion must be solved conjointly with loop constraints, i.e. solve

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{c}(\mathbf{q}, \dot{\mathbf{q}}, \mathbf{frc}, \mathbf{trq}) = \mathbf{Q}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{J}^T(\mathbf{q})\boldsymbol{\lambda} \quad (4.1)$$

subject to

$$\mathbf{h}(\mathbf{q}) = \mathbf{0} \quad ; \quad \dot{\mathbf{h}}(\mathbf{q}, \dot{\mathbf{q}}) = \mathbf{J}(\mathbf{q})\dot{\mathbf{q}} = \mathbf{0} \quad ; \quad \ddot{\mathbf{h}}(\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}}) = \mathbf{0} \quad (4.2)$$

in which

- $\mathbf{q}$, $\dot{\mathbf{q}}$ and $\ddot{\mathbf{q}}$ stand for the generalised relative positions, velocities and accelerations;
- $\mathbf{M}(\mathbf{q})$ represents the symmetric, positive-definite mass matrix of the MBS;
- $\mathbf{c}(\mathbf{q}, \dot{\mathbf{q}})$ gathers the centripetal, Coriolis, gyroscopic terms as well as the external forces $\mathbf{frc}(\mathbf{q}, \dot{\mathbf{q}})$ and torques $\mathbf{trq}(\mathbf{q}, \dot{\mathbf{q}})$;
- $\mathbf{Q}(\mathbf{q}, \dot{\mathbf{q}})$, deliberately isolated in the right-hand-side of Eq. 4.1, denotes the efforts within the joints (actuators or passive elements);
- $\mathbf{J}(\mathbf{q}) \triangleq \frac{\partial \mathbf{h}}{\partial \dot{\mathbf{q}}}$ denotes the constraint Jacobian matrix;
- $\boldsymbol{\lambda}$ are the Lagrange multipliers associated with the constraints $\mathbf{h}(\mathbf{q})$.

If there are m independent algebraic constraints and n generalised coordinates, the number of DoF of the MBS is $n - m$.

4.1.2 Coordinate Partitioning Technique

Equations 4.1 and 4.2 form a system of differential algebraic equations, which can be difficult to solve directly due to the presence of algebraic equations. A practical way to solve them is to transform equations 4.1

and 4.2 into a reduced set of ordinary differential equations, using the Coordinate Partitioning Technique [159].

First, the generalised coordinates must be split into $n - m$ independent and m dependent generalised coordinates. The independent generalised coordinates will correspond to the $n - m$ DoF of the system.

Formally, the vector of the generalised coordinates is partitioned as

$$\mathbf{q} = \begin{pmatrix} \mathbf{q}_u \\ \mathbf{q}_v \end{pmatrix} \quad (4.3)$$

where $\mathbf{q}_u$ are the $n - m$ the independent and $\mathbf{q}_v$ the m dependent generalised coordinates respectively.

By applying the same partitioning to all vectors and matrices in Eq. 4.1 and 4.2, we can express the positions, velocities and accelerations of the dependent generalised coordinates as functions of the independent generalised coordinates, i.e. Eq. 4.2 lead to

$$\mathbf{q}_v = f_1(\mathbf{q}_u) \quad ; \quad \dot{\mathbf{q}}_v = f_2(\mathbf{q}, \dot{\mathbf{q}}_u) \quad ; \quad \ddot{\mathbf{q}}_v = f_3(\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}}_u) \quad (4.4)$$

where f_1 (non-linear system), f_2 and f_3 (linear system) are functions relating dependent to independent generalised variables (for more details, see [159]).

Similarly, Equation 4.1 becomes a set of $n - m$ ordinary differential equations:

$$\mathbf{M}_r(\mathbf{q}_u) \ddot{\mathbf{q}}_u + \mathbf{c}_r(\mathbf{q}_u, \dot{\mathbf{q}}_u, \mathbf{frc}, \mathbf{trq}) = \mathbf{Q}_r(\mathbf{q}_u, \dot{\mathbf{q}}_u) \quad (4.5)$$

in which the subscript r indicates the reduced form. In Equation 4.5, the dependent variables and the Lagrange multipliers λ have been properly eliminated by the Coordinate Partitioning reduction.

In a similar way, one can introduce the motion of the so-called driven generalised coordinates into the MBS. Driven generalised coordinates describe joints whose trajectory is fully imposed over the simulation. They are denoted as $\mathbf{q}_d$.

A second partitioning is required to process these driven coordinates in Equation 4.5. The vector of the independent generalised coordinates is split in a similar way to Equation 4.3:

$$\mathbf{q}_u = \begin{pmatrix} \mathbf{q}_{uu} \\ \mathbf{q}_d \end{pmatrix} \quad (4.6)$$

where $\mathbf{q}_{uu}$ denotes the first set of independent coordinates.

Equation 4.5 reads

$$\begin{pmatrix} \mathbf{M}_r^{uu} & \mathbf{M}_r^{ud} \\ \mathbf{M}_r^{du} & \mathbf{M}_r^{dd} \end{pmatrix} \begin{pmatrix} \ddot{\mathbf{q}}_{uu} \\ \ddot{\mathbf{q}}_d \end{pmatrix} + \begin{pmatrix} \mathbf{c}_r^u \\ \mathbf{c}_r^d \end{pmatrix} = \begin{pmatrix} \mathbf{Q}_r^u \\ \mathbf{Q}_r^d \end{pmatrix} \quad (4.7)$$

The first line of the last equation can be solved with respect to $\ddot{\mathbf{q}}_{uu}$, what gives the final form of a direct dynamics problem of the MBS

$$\mathbf{M}_r^{uu} \ddot{\mathbf{q}}_{uu} + \mathbf{c}_r^u = \mathbf{Q}_r^u - \mathbf{M}_r^{ud} \ddot{\mathbf{q}}_d \quad (4.8)$$

The second line of Equation 4.7 can be used, when appropriate, to compute efforts $\mathbf{Q}_r^d$ associated with the driven generalised coordinates

$$\mathbf{Q}_r^d = \mathbf{M}_r^{du} \ddot{\mathbf{q}}_{uu} + \mathbf{M}_r^{dd} \ddot{\mathbf{q}}_d + \mathbf{c}_r^d \quad (4.9)$$

They are very useful in biomechanics since these efforts may represent net torques or internal forces inside the joints, i.e. biological markers that can give relevant information for biomechanical analyses. In fact, these efforts, $\mathbf{Q}_r^d$, computed within a direct dynamics formulation (Eq. 4.7) are nothing more than a local inverse dynamics (see next section) of an MBS whose motion is not fully imposed.

4.1.3 Inverse dynamics

In biomechanics, we are usually interested in knowing muscle forces and/or joint forces and torques given the measured kinematics generally obtained from an upfront experiment. In this case, inverse dynamics is the numerical method which suits better for computing joints efforts $\mathbf{Q}$. In case of an unconstrained MBS, inverse dynamics reads

$$\mathbf{Q} = \boldsymbol{\phi}(\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}}, \mathbf{frc}, \mathbf{trq}) \quad (4.10)$$

where $\mathbf{Q}$ are now the unknowns, and $\boldsymbol{\phi}$ represent the MBS model dynamics, whose computation evaluation has only an $\mathcal{O}(n)$ complexity because inverse dynamics does not require the explicit computation of the mass matrix $\mathbf{M}(\mathbf{q})$ ($\mathcal{O}(n^2)$ complexity). This advantage is obviously lost if one computes $\boldsymbol{\phi}$ through its direct formulation

$$\boldsymbol{\phi} = \mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{c}(\mathbf{q}, \dot{\mathbf{q}}, \mathbf{frc}, \mathbf{trq})$$

Similarly to the equations of motion derived for the direct dynamics computation for closed-loop MBS, for the inverse dynamics, Equation 4.1 becomes

$$\mathbf{Q} = \boldsymbol{\phi}(\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}}, \mathbf{frc}, \mathbf{trq}) - \mathbf{J}^T(\mathbf{q})\boldsymbol{\lambda} \quad (4.11)$$

subject to the constraints

$$\mathbf{h}(\mathbf{q}) = \mathbf{0} \quad ; \quad \dot{\mathbf{h}}(\mathbf{q}, \dot{\mathbf{q}}) = \mathbf{J}(\mathbf{q})\dot{\mathbf{q}} = \mathbf{0} \quad ; \quad \ddot{\mathbf{h}}(\mathbf{q}, \dot{\mathbf{q}}, \ddot{\mathbf{q}}) = \mathbf{0} \quad (4.12)$$

Using the same reduction technique as for direct dynamics, in case of inverse dynamics, Equation 4.5 becomes

$$\mathbf{Q}_r = \boldsymbol{\phi}_r(\mathbf{q}_u, \dot{\mathbf{q}}_u, \ddot{\mathbf{q}}_u) \quad (4.13)$$

In case of an overactuated problem, i.e. muscle redundancy, Equation 4.13 can be written as

$$\mathbf{B}(\mathbf{q}_u)\mathbf{Q}_{over} = \boldsymbol{\phi}_r(\mathbf{q}_u, \dot{\mathbf{q}}_u, \ddot{\mathbf{q}}_u) \quad (4.14)$$

in which the n_{over} ($> (n - m)$) efforts $\mathbf{Q}_{over}$ are the unknowns; $\mathbf{B}$ being a $(n - m) \times n_{over}$ rectangular matrix. The number of solutions being infinite, additional criteria must be used to solve it.

The inverse dynamics problem only requires the kinematics, i.e. positions, velocities and accelerations, of the independent generalised coordinates as input for the computation. as for direct dynamics, possible driven variables can be introduced for specific reasons, e.g. locking a joint, requiring an equivalent re-partitioning of Equation 4.13:

$$\mathbf{Q}_r^d = \boldsymbol{\phi}_r^d(\mathbf{q}_u, \dot{\mathbf{q}}_u, \ddot{\mathbf{q}}_u) \quad (4.15)$$

In our formalism, the driven joints efforts (Eq. 4.15) are computed at the same time than the joint efforts (Eq. 4.13). In biomechanics applications, it means that muscle forces and internal joint efforts are obtained simultaneously in ROBOTRAN. This is not always guaranteed in other multibody software such as OpenSim.

Particular case: Static configuration In the case of a static configuration, i.e. null velocities and accelerations, only the positions of the independent generalised coordinates, $\mathbf{q}_u$ are necessary for the computations. For the purpose of this thesis, two numerical methods are used in ROBOTRAN to

compute muscle forces and intervertebral efforts.

The first numerical method is the classical inverse dynamics (Eq. 4.14) and is used by the optimisation-based computations (Section 4.4). The only input is the position of the lowermost S1/L5 lumbar IVD. The unknowns in this computation are the muscle forces which are resolved thanks to the optimisation process, to take the muscle redundancy problem into account (Eq. 4.14).

The second numerical method is an inverse equilibrium, also called an equilibrium with exchange of variables. This second method is used for the computation of a global equivalent force for the hybrid approach in its simplified version (see Section 4.3). Normally, the equilibrium is used to find the equilibrium positions of a MBS when its velocities and accelerations are put to 0. Starting from Equation 4.5, the equilibrium usually consists in finding $\mathbf{q}_u^*$ such that

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

However, it is possible to perform an exchange of variables in order to compute other parameters p of the system, e.g. a mass or a force:

$$\begin{aligned} \{\mathbf{q}_u^*\} &\leftrightarrow \mathbf{x}^* \{p\} \\ \mathbf{c}_r(\mathbf{x}^*, \mathbf{0}, \mathbf{frc}, \mathbf{trq}) &= \mathbf{Q}_r(\mathbf{x}^*, \mathbf{0}) \end{aligned} \quad (4.16)$$

For our case, the only independent generalised coordinate, i.e. the position of the lowermost lumbar IVD, being known, the exchange variable is assigned to a global equivalent force which is split in a deterministic way among the muscle fascicles.

Finally, the formalism presented makes it easy to identify contributions from static and dynamic sources, i.e.

$$\mathbf{Q}_r = \boldsymbol{\phi}_r^{dyn}(\mathbf{q}_u, \dot{\mathbf{q}}_u, \ddot{\mathbf{q}}_u) + \boldsymbol{\phi}_r^{stat}(\mathbf{q}_u) \quad (4.17)$$

and

$$\mathbf{Q}_r^d = \boldsymbol{\phi}_r^{d,dyn}(\mathbf{q}_u, \dot{\mathbf{q}}_u, \ddot{\mathbf{q}}_u) + \boldsymbol{\phi}_r^{d,stat}(\mathbf{q}_u) \quad (4.18)$$

4.2 Model of the Sorensen test

The 3D musculoskeletal model was developed with the ROBOTRAN software [158]. It was based on a fully articulated thoracolumbar spine and rib

cage model published previously for a 25-year-old man (height 175 cm and mass 78 kg) [105] available in the OpenSim software [103].

It is composed of 29 bodies:

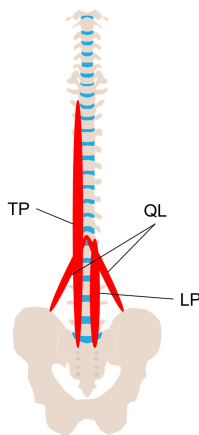
- The pelvis;
- The sacrum;
- The 5 lumbar vertebrae;
- The 12 thoracic vertebrae;
- The 2 twelfth ribs;
- The 2 arms, the 2 forearms and the 2 hands;
- The head;
- And an additional 29th body added to the ninth thoracic vertebra in order to add a vertical load to the subject as realised during the experiment (see Chapter 3)

The model includes 84 (42×2 sides) muscle fascicles of **LP**, **TP** and **QL** muscle groups¹. The three muscles **LP**, **TP** and **QL** are schematically represented beside Table 4.1 where their fascicles are detailed. Only these three muscles were considered based on the upfront experiment (see Chapter 3 for more details). Also only muscle fascicles which had insertion points on each side of the S1/L5 joint, involving the only independent generalised coordinate, were considered in the model.

In terms of **MBS** modelling, each muscle fascicle was incorporated in the model via a closed-loop of bodies in which the fascicle was bi-articulated with spherical joints and allowed a translation characterising the muscle fascicle action (see Fig. 4.2). This way of proceeding allowed to make the joint force Q_i (N) associated with the muscle translation q_i (m) appear as unknown in the inverse dynamic model. Closed loops were solved using the Coordinate Partitioning technique (Eq. 4.3) prior to solve the model in terms of muscle forces.

Dynamic parameters of bodies, i.e. mass m and center of mass (**COM**), were also based on the model developed by Bruno et al. [105]. Inertia was not considered as we studied only static configurations. The model

¹For the sake of brevity, the term "muscle" will be used in the rest of the manuscript to refer to either a muscle or a muscle group.



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

Table 4.1 Muscle fascicles included in the model and their schematic representation. L stands for lumbar vertebra, T for thoracic vertebra and R for rib. I1, I2 and I3 refer to different insertion points on the pelvis. R12 stands for the twelfth rib and each subscript refers to a different insertion point on this rib.

could be scaled to different subjects. For that purpose, a global scaling was used for masses, COM locations and dimensions of bodies. For the mass, a scale factor was defined as $SF_{m,i} = \frac{m_{subject,i}}{m_{ref}}$ where $m_{subject,i}$ was the body mass of the i^{th} subject of the study and m_{ref} the reference body mass, i.e. 78 kg. Similarly, for the COM location and dimensions of a body, a height scale factor was defined as $SF_{h,i} = \frac{h_{subject,i}}{h_{ref}}$ where $h_{subject,i}$ was the height (in meter) of the i^{th} subject of the study and h_{ref} the reference height, i.e. 1.75 m. The global scaling process consisted in multiplying reference values for masses, COM locations and dimensions of bodies from the model of Bruno et al. [105] by the appropriate scale factor. This global scaling technique was similar to the global scaling performed in the OpenSim software.

Unknowns in the model for both hybrid and optimisation-based approaches were muscle forces, i.e. joint forces $\mathbf{Q}$ (see Fig. 4.2), and were solved by inverse equilibrium (Eq. 4.16) for the hybrid approach and by inverse dynamics (Eq. 4.13) for the optimisation-based approach (see Section 4.1.3 for more details about the numerical methods). The underlying hypothesis for this computation was that muscles in the model acted as pure

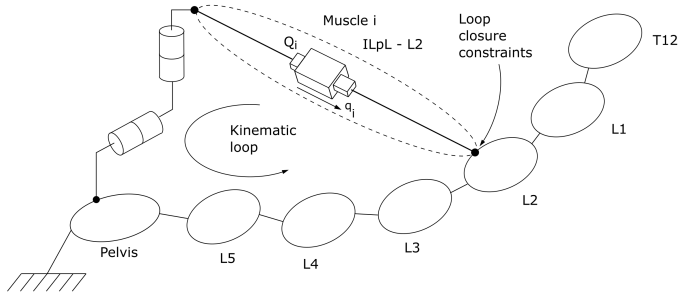


Fig. 4.2 A kinematic loop generated is by each muscle fascicle (ILpL - L2 here for example). Each fascicle is bi-articulated (on the pelvis via two rotations and on L2 via a spherical joints) and allows a translation characterising the muscle fascicle action. Loop closure is necessary at the insertion point of the fascicle (spherical joint on L2 here) to solve the kinematic constraints. There are 84 kinematic loops in total in the model.

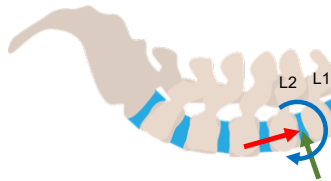


Fig. 4.3 Intervertebral resultant efforts – compressive force in red, shear force in green and bending torque in blue – acting from L2 on L1 vertebral segment. Drawn arrows define positive values. Adapted from [19].

force transmitters. This hypothesis was realistic as it was a static configuration. There were therefore no force-length relationship or force-velocity relationship which would need the use of a Hill-type model.

As the main results of the study, intervertebral efforts, i.e. the compressive force N , the shear force T and the bending torque M_b , were computed thanks to driven joint coordinates $\mathbf{q}_d$ located within IVD. These coordinates, being assigned to a constant value, enable to compute the associated constraint forces, N and T , and torque, M_b in particular at the four superior lumbar (Lx) IVD levels (Fig. 4.3).

To summarise, the model was developed under the following assumptions

- Static configuration (Sorensen test), i.e. null velocities and accelera-

tions $\dot{\mathbf{q}} = \ddot{\mathbf{q}} = 0$;

- Muscle fascicles modelled as pure force transmitters, i.e. no dependencies of length or velocity;
- Bilateral (left and right) muscles were assumed to produce identical forces, so there is neither lateral shear, lateral bending nor axial torques;
- The only external force was the gravity acting in the anterior-posterior direction pointing forward;
- Bodies had no need of inertia as there was no motion;
- The spine and its surrounding intrinsic muscles (multifidus and other deep back muscles defined in Fig. 1.4) were assimilated to a beam whose intervertebral efforts – compression and shear forces and bending torque – are computed according to the beam theory through constrained joint forces and torques. As a consequence, intervertebral efforts could be under- or overestimated as the surrounding muscles' actions, e.g. intertransversarii laterales and mediales, interspinales and multifidus, were not taken into account in the model;
- The posterior contacts that a vertebra may have with its two adjacent vertebrae have not been modeled. All intervertebral efforts are considered at the level of the IVD;
- Intervertebral efforts were translated into equivalent stresses considering a perfectly cylindrical IVD, for all the lumbar levels, with a cross-section area of 18 cm^2 [102, 160] whose neutral fibre is located in the middle of the disc.

4.3 Hybrid approach

4.3.1 Overview of the process

The process, whose global overview is illustrated in Fig. 4.4, is composed of three main parts: the MBS model, the hybrid approach and the optimisation-based approach. The central part of the process is the MBS model which is used by the two computation processes. The goal is to confront the muscles forces F_{musc} and the intervertebral efforts N , T and M_b , for compressive, shear forces and bending torque respectively, computed with a hybrid

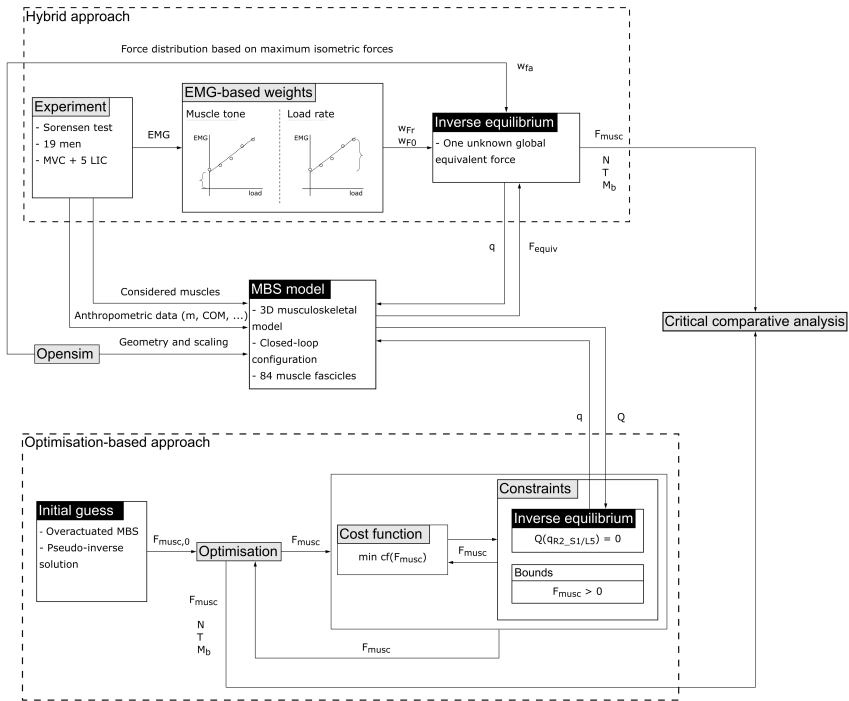


Fig. 4.4 Overview of the process involving the musculoskeletal MBS model and the two considered approaches and computations: hybrid and optimisation-based. Black boxes with blank titles refer to a computation directly issuing from the MBS model.

approach, with these obtained with a purely mathematical optimisation-based approach.

Deliberately, the process started with experimental measurements. The latter helped to develop the MBS model and to define EMG-based weights w_{F0} and w_{Fr} (defined later in Section 4.3). Based on the two latter, muscles forces and intervertebral efforts for the hybrid approach were computed.

On the contrary, the computation of the muscle forces and intervertebral efforts by numerical optimisation was carried out regardless of the hybrid approach, on the basis of cost functions classically found in the literature (see Table 4.2).

The results of both approaches could thus be compared and confronted

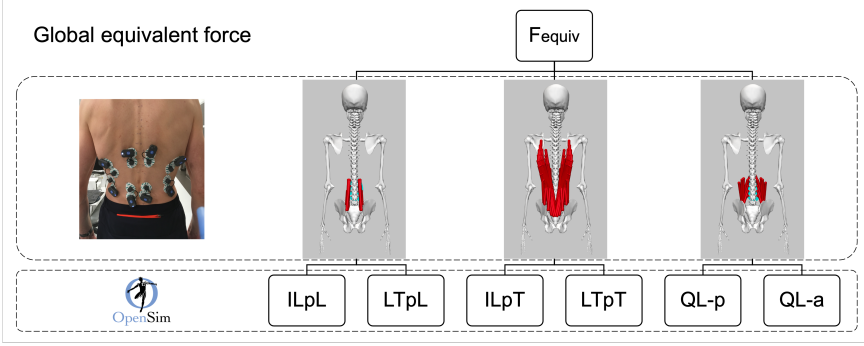


Fig. 4.5 Distribution of the global equivalent force into muscles and muscle fascicles. The first level refers to the distribution between **LP**, **TP** and **QL** muscles thanks to **EMG** signals. The second level refers to the distribution between each fascicle inside each muscle thanks to the values of maximum isometric force provided by the OpenSim model [105]. For details about muscle fascicles, refer to Table 4.1.

in a critical and objective way.

4.3.2 EMG-based weights

The hybrid approach was fully based on **EMG** amplitudes from the experiment. To take the physiological – and not mathematical – muscle redundancy into account, all joint forces implemented in the muscle fascicles depended on one unique global equivalent force (F_{equiv} in Fig. 4.4 and in Eq. 4.19). This unique global equivalent force was distributed among muscles and then among fascicles according to Fig. 4.5. For this, the idea is to (i) distribute that unique global equivalent force among the three muscles **LP**, **TP** and **QL** using EMG-based weights and (ii) distribute the force inside each muscle among all fascicles of this muscle in accordance with data from the model of Bruno et al. [105].

Mathematically, the distribution of the unique global equivalent force illustrated in Fig. 4.5 could be written as

$$F_{fa} = w_{fa} * w_{EMG} * F_{equiv} \quad (4.19)$$

where F_{fa} is the force in a muscle fascicle, w_{EMG} the weight based on **EMG** amplitudes and w_{fa} the weight for the fascicle inside a given muscle.

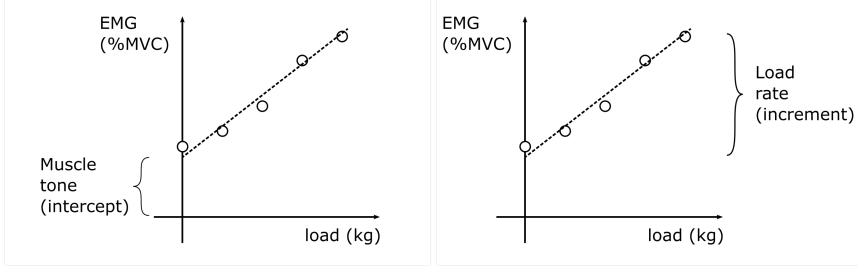


Fig. 4.6 EMG-based weights are based (Left) on the muscle tone, i.e. 0-kg **EMG** amplitude and (Right) on the load rate, i.e. increment of a non-negative linear regression of the **EMG** amplitudes for **LIC**.

However, Eq. 4.19 has been made more complex to take into account the muscle tone, i.e. when there was no load in the experiment, and the load rate, i.e. the muscle response to a load increase. In other words, the distribution illustrated in Fig. 4.5 was performed twice, once for the muscle tone and once for the load rate. Consequently, the EMG-based weights w_{EMG} in Eq. 4.19 were split into two weights:

- w_{F0} the weight for the muscle tone (see Fig. 4.6 Left);
- w_{Fr} the weight for the load rate (see Fig. 4.6 Right).

To define these two weights, a non-negative linear regression was performed between **EMG** amplitudes and loads for each muscle. On one hand, the y-axis intercepts (%**MVC**) of the fit were used (see Fig. 4.6 Left) to compute the coefficients w_{F0} for each muscle, i.e.

$$w_{F0,m} = \frac{intercept_m}{\sum_m intercept_m} \quad (4.20)$$

where m indicates the muscle (**LP**, **TP** or **QL**). On the other hand, increments (%**MVC**) of the fits were used as illustrated on the right in Fig. 4.6 to compute the coefficients w_{Fr} :

$$w_{Fr,m} = \frac{increment_m}{\sum_m increment_m} \quad (4.21)$$

where m indicates the muscle (**LP**, **TP** or **QL**).

The last weight to be defined was the weight $w_{fa,i}$ associated to the distribution of the force between the fascicles i of the same muscle. For the i^{th} fascicle of a given muscle, the weight $w_{fa,i}$ was considered as the ratio

between the maximum isometric force for the i^{th} fascicle in the given muscle divided by the sum of the maximum isometric forces of the fascicles of the given muscle, i.e.

$$w_{fa,i} = \frac{F_{max,i}}{\sum_i F_{max,i}} \text{ for a muscle}$$

where the maximum isometric forces $F_{max,i}$ are taken from the model of Bruno et al. [105].

Finally, based on Eq. 4.19, the force in each fascicle is defined as

$$F_{fa,i} = w_{F0} * w_{fa,i} * F_{equiv,0} + w_{Fr} * w_{fa,i} * F_{equiv,r} \quad (4.22)$$

$$\text{with } F_{equiv} = F_{equiv,0} + F_{equiv,r} \quad (4.23)$$

where $F_{equiv,0}$ is the global equivalent force for the muscle tone, i.e. for an unloaded configuration in the Sorensen test posture, and $F_{equiv,r}$ the global equivalent force for the load rate.

4.3.3 Global equivalent force

To compute the global equivalent force, for both muscle tone and load rate, inverse equilibrium, which is nothing else than inverse dynamics with $\dot{\mathbf{q}} = \ddot{\mathbf{q}} = 0$ in Eq. 4.13, was used as illustrated in Fig. 4.4. The case of muscle tone must be computed prior to solve the case of load rate, i.e. Eq. 4.22 reduces to $F_{fa} = w_{F0} * w_{fa} * F_{equiv,0}$. Once the global equivalent force associated with the muscle tone $F_{equiv,0}$ was found, the one associated with the load rate $F_{equiv,r}$ could be computed for all loads, i.e. inverse equilibrium equation with respect to the global equivalent force (Eq. 4.16) via inverse dynamics formulation (Eq. 4.13).

Similarly, associated intervertebral efforts were obtained via Equation 4.15, i.e. compute $\{N_j, T_j, M_{b,j}\}$ with $j = 1 \dots 4$ for the four upper lumbar levels.

4.4 Optimisation-based approach

The optimisation-based computation was the extreme opposite of the above EMG-based computation, to which it will be confronted (hereafter). This purely mathematical approach was deliberately chosen to be fully independent of EMG input. This approach needed therefore to define cost functions to solve the redundancy problem and to select an appropriate optimisation process.

Number	Objective	Equation	Abbreviation
1	Minimise the sum of squared muscle forces [18]	$\min \sum F_{musc}^2$	F^2
2	Minimise the sum of squared muscle stresses [18]	$\min \sum \left(\frac{F_{musc}}{PCSA} \right)^2$	s^2
3	Minimise the sum of cubed muscle stresses [18]	$\min \sum \left(\frac{F_{musc}}{PCSA} \right)^3$	s^3
4	Minimise the largest relative muscle force [112, 113]	$\min \max \left(\frac{F_{musc}}{F_{max}} \right)$	mM

Table 4.2 Cost functions used in the optimisation process. F_{musc} is the muscle force, F_{max} the maximum isometric force and PCSA the physiological cross-sectional area of each muscle fascicle.

4.4.1 Cost functions

Four cost functions were studied (see Table 4.2). All of them had a similar objective: to favour the smallest muscle force. Minimising muscle stresses (s^2 and s^3 in Table 4.2) tends to maximising endurance [111].

PCSA and maximum isometric forces derived from the model of Bruno et al. [105]. Their values needed to be adjusted for each subject. PCSA and maximum isometric forces being linearly related through the MMS, i.e. $F_{max} = MMS * PCSA$, only an adjustment method for PCSA values was necessary. For that purpose, PCSA of ES, hypothesised as LP and TP muscles together, and PCSA of the QL muscle were computed based on anthropometric measurements of all subjects via regression equations suggested by Marras et al. [161]. Regression equations for ES and QL were respectively [161]:

$$PCSA_{ES,Sx} = 52.18 - 11.69 \frac{h_{Sx}}{BM_{Sx}} \quad (4.24)$$

$$PCSA_{QL,Sx} = -0.387 + 0.041 * H_{Sx} * BM_{Sx} \quad (4.25)$$

where Sx denotes the x^{th} subject, h_{Sx} his height in cm, H_{Sx} his height in m

and BM_{Sx} his body mass in kg. $PCSA$ as defined in Eq. 4.24 and 4.25 were associated with the whole muscle and not for each fascicle. So, the $PCSA$ of each fascicle could be estimated as

$$PCSA_{fa_i, Sx} = \frac{PCSA_{fa_i}}{\sum_i PCSA_{fa_i}} * PCSA_{Sx}$$

where $PCSA_{fa_i}$ is the $PCSA$ of the i^{th} fascicle of the muscle and whose value comes from [105]. Note that for a LP or TP fascicle, $\sum_i PCSA_{fa_i}$ was the sum of $PCSA$ of both LP and TP fascicles. Finally, maximum isometric muscle forces were computed assuming a MMS value of 100 N/cm^2 [105].

4.4.2 Optimisation process

A trust-region algorithm for constrained optimisation was used to solve the muscle redundancy problem. The process could end (i) if both maximum absolute value of the Lagrangian gradient and constraint violation were smaller than the above tolerance or (ii) if the radius of the trust region was smaller than the tolerance, set to 10^{-3}

The pseudo-inverse solution was used as the algorithm initial guess. For that solution, the MBS inverse equilibrium (Eq. 4.14) was overactuated by considering all the muscles forces Q_{over} as unknowns of the problem. Equation 4.14 recalls

$$\begin{bmatrix} \mathbf{B}(\mathbf{q}_u) \\ [(n - m) * n_{over}] \end{bmatrix} \begin{bmatrix} \mathbf{Q}_{over} \\ [n_{over} * 1] \end{bmatrix} = \begin{bmatrix} \boldsymbol{\phi}(\mathbf{q}_u, \dot{\mathbf{q}}_u, \ddot{\mathbf{q}}_u) \\ [n_{over} * 1] \end{bmatrix}$$

with n_{over} the number of muscles in the model and $n - m$ the number of degree of freedom of the model which was equal to 1 in this case (the only independent generalised coordinate being associated to the rotation around the medio-lateral axis between the first sacral vertebra (S1) and the fifth lumbar vertebra (L5)). This corresponded to the sole motion in the model that muscles must master to satisfy the Sorensen test posture.

Two constraints were implemented. For that part of the optimisation process, the only independent generalised coordinate of the MBS model was the rotation of the S1/L5 lumbar disc around the medio-lateral axis. The associated joint torque was constrained to be zero, because it was assumed that forces were only produced by the muscles. The second constraint imposed muscles forces to be positive, which the above pseudo-inverse solution obviously did not guarantee. No upper constraint bounds were

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put for muscle forces.

5

Sensitivity analysis of the optimisation process

It was not the purpose of this thesis to develop a new optimisation process to solve the muscle redundancy problem but rather to include [EMG](#) into the muscle force estimation and embedded into the optimisation process. Therefore, this chapter aims at performing a sensitivity analysis on the chosen optimisation algorithm and its parameters. The objective is to eliminate any uncertainty that could arise from the optimisation process to ensure the robustness of the estimated muscle forces.

5.1 Choice of the optimisation method

As already presented in Section [4.4](#), the chosen algorithm for the optimisation process was a trust-region algorithm for constrained optimisation. The latter was implemented using the Python function [\[162\]](#)

```
scipy.optimize.minimize(method='trust-constr')
```

The 'trust-constr' method was chosen because it is one of the method recommended for constrained minimisation. The 'BFGS' (for Broyden, Fletcher, Goldfarb and Shanno) algorithm which is a quasi-Newton method is implemented by default in the 'minimize' function. However, it is not

designed for constrained problems, other than bound constraints. Therefore, the ‘trust-constr’ method was used for the optimisation process.

This algorithm is an implementation of the Byrd and Omojokun’s trust region algorithm for solving nonlinear equality constrained optimisation problems [163, 164]. Without going into details of the algorithm, it requires the cost function and equality constraints to be twice continuously differentiable. Furthermore, the cost function should be scalar, while equality constraints could return a scalar or a vector whose dimensions are smaller or equal to the number of unknowns of the problem. In the case of the gradient vector and the Hessian matrix of the cost function are not provided analytically to the algorithm, they are estimated using a two-point finite difference with an absolute step size.

Once the method has been selected, its parameters must be carefully chosen to ensure a good and reliable convergence. The tolerance and the initial guess were two of the most critical parameters. Therefore, these two parameters were modified and their impacts studied for the present study. Besides, as the mM cost function (Table 4.2) was not differentiable, it was approximated by an estimated function to study the differences induced by the algorithm.

5.2 Method for the sensitivity analysis

The model used for this sensitivity analysis was the model of the Sorensen test posture as described in Section 4.2 with the physiological parameters of the generic model of Bruno et al. [105]. Only two cost functions among these presented before were tested for the present analysis: s^3 and mM (Table 4.2). The sensitivity analysis is carried out on three optimisation parameters: the tolerance, the initial guess and a new cost function, approximation of mM.

For the two cost functions, different tolerances for the process termination threshold were tested. To make these numerical methods tangible, let us see what they represent from a numerical and physical points of view:

1. From a numerical point of view, the optimisation process terminates successfully if:
 - (a) The maximum absolute value of the Lagrangian gradient of the cost function and the constraint violation are smaller than the specified tolerance (called *gtol* for the analysis) or;
 - (b) The radius of the trust region is smaller than the specified tolerance (called *xtol* for the analysis), meaning that the solution lies

within in the required accuracy.

2. From a physical point of view, the first condition represents the slope of the cost function, i.e. N^3/N for s^3 and N^{-1} for mM . For both cost functions, units of the constraint violation are Nm as the constraint is associated to the bending torque inside the lowermost lumbar IVD. The units of the second condition are N as this termination condition reflects how close to an acceptable solution the computation is.

Three analyses were carried out to verify the relevancy of the tolerances. Firstly, both $gtol$ and $xtol$ were modified at the same time from 10^{-1} to 10^{-8} with a 0.1-step factor. Secondly, the $gtol$ tolerance was fixed to 10^{-3} while the $xtol$ tolerance varied on the same range of tolerances as previously. Thirdly, the $xtol$ was fixed while the $gtol$ tolerance varied.

The numbers of iterations of the optimisation process for each tolerance and each cost function were reported for comparison. Furthermore, the evolution of optimised forces in several muscle fascicles as function of the tolerance value was analysed. The chosen muscle fascicles (two per muscle) were those with the greatest theoretical maximum isometric forces, i.e. ILpL-L4 and LTpL-L5 for LP, ILpT-R11 and LTpT-T9 for TP and QL-p I2L4 and QL-a I3T12 for QL.

For the initial guess, one deterministic and ten randomised vectors were tested. The deterministic initial guess was composed of the muscle forces found with the pseudo-inverse solution (Eq. 4.14), as described in Section 4.4. The ten randomised vectors were built around muscle forces found with the pseudo-inverse solution as follows

For the i^{th} muscle fascicle,

$$\begin{aligned} &\text{if } F_{musc,i}|_{PI} < 0 \\ &\quad F_{0,i} = X \times 10 \\ &\text{else } F_{0,i} = F_{musc,i}|_{PI} \pm X \times \frac{F_{musc,i}|_{PI}}{2} \end{aligned}$$

where PI indicates the value found with the pseudo-inverse solution, F_0 , the initial guess and X is a random value between 0 and 1. In other words, the initial value for a muscle fascicle was in a 50% range on both sides of the muscle force found with the pseudo-inverse solution.

For mM , as the function was not twice continuously differentiable, an approximation of the function, which respects this condition, was implemented as follows,

5 | Sensitivity analysis of the optimisation process

Given $r_i \triangleq \frac{F_{muscle,i}}{F_{max,i}}$, the ratio of the force of the i^{th} muscle fascicle over its maximum isometric force and given $\epsilon > 0$, a parameter representing the quality of the approximation, the new cost function to minimise was [165]

$$\min r_{max} + \epsilon \ln \left(\sum_i \exp \left(\frac{r_i - r_{max}}{\epsilon} \right) \right) \quad (5.1)$$

where r_{max} is the maximum value of the ratios r_i . This approximation of **mM** is called *logsumexp*. Twenty evenly-spaced values between 10^{-6} and 1 were tested for the ϵ parameter. The root-mean-square deviations (**RMSD**) of muscle forces found by *logsumexp* and the muscle forces found with **mM** are reported in Section 5.3.3 for each value of the ϵ parameter. **RMSD** was defined as

$$RMSD = \sqrt{\frac{1}{n} \sum_{i=1}^n (LSE|_{\epsilon_i} - mM)^2} \quad (5.2)$$

where n is the number of muscles, $LSE|_{\epsilon_i}$ the muscle forces found by the *logsumexp* cost function for the i^{th} value of ϵ and mM the muscle forces found by **mM**.

To compare the results, both muscles forces and intervertebral efforts are compared in Section 5.3 as a function of the optimisation parameters of interest.

5.3 Choice of the optimisation parameters

5.3.1 Impact of the tolerance

For the different tolerances and for each cost function, the numbers of iterations of the optimisation process are indicated in Table 5.1. Firstly, it can be seen that the optimisation processes with **mM** did not converge for tolerances greater than 10^{-5} when both tolerances varied, even if the maximum number of iterations for the optimisation process was increased from 1000 to 2000 iterations. Thereby, when only one of the two tolerances varied, it was decided to not explore values further than 10^{-5} .

When both tolerances were modified at the same time, the number of iterations increased when the tolerances decreased (Table 5.1). In particular, the number of iterations increased drastically for **mM**. When looking at the optimised muscle forces (Fig. 5.1), they are constant for s^3 whatever the tolerance value was. However, for **mM**, the optimised forces stabilised for a tolerance of 10^{-3} . Interestingly, this corresponds to the change in the

	Both varied		<i>xtol</i> varied		<i>gtol</i> varied	
	s^3	mM	s^3	mM	s^3	mM
10^{-1}	170	3	158	166	180	3
10^{-2}	163	14	168	276	211	14
10^{-3}	176	292	176	292	176	292
10^{-4}	209	483	202	445	211	402
10^{-5}	250	975	226	/	188	521
10^{-6}	322	/	-	-	-	-
10^{-7}	375	/	-	-	-	-
10^{-8}	387	/	-	-	-	-

Table 5.1 Number of iterations for each cost function and each tolerance. Numbers in **bold** indicates that the optimisation process because of the first termination condition. A '/' indicates that the optimisation process did not converge. A '-' indicates that the test was not performed.

termination condition. Indeed, for the two largest tolerances, the optimisation process for mM stopped because of the first termination condition, i.e. the infinity norm of the Lagrangian gradient and the violation constraint were smaller than the tolerance (see **bold** numbers in Table 5.1). Once the termination condition was related to the radius of the trust region, the value of optimised forces seemed to stabilise for all the muscle fascicles (Fig. 5.1).

When the *gtol* tolerance was fixed, decreasing the *xtol* tolerance resulted in more iterations for the optimisation process for both cost functions. The optimisation process with mM did not converge for 10^{-5} . This time, all the optimisation processes stopped because of the radius of the trust region. It can be noticed that the number of iterations for s^3 is similar in this case to the number of iterations when both tolerances were modified. Optimised forces for s^3 were constant for all the tested tolerances (Fig. 5.2). For mM, optimised forces were constant on average (Fig. 5.2), the variability being likely associated to the non-differentiable behaviour of the mM cost function.

Finally, when the *xtol* tolerance was fixed, decreasing the *gtol* tolerance had different consequences depending on the cost functions. For s^3 , the number of iterations varied but was not correlated with the decrease of the *gtol* tolerance (Table 5.1). Besides, it can be noticed that the optimisation processes stopped because of the radius size of the trust region. Furthermore, optimised forces were also constant (Fig. 5.3) as observed previously for the two other analyses on the tolerances. On the contrary, for mM, the

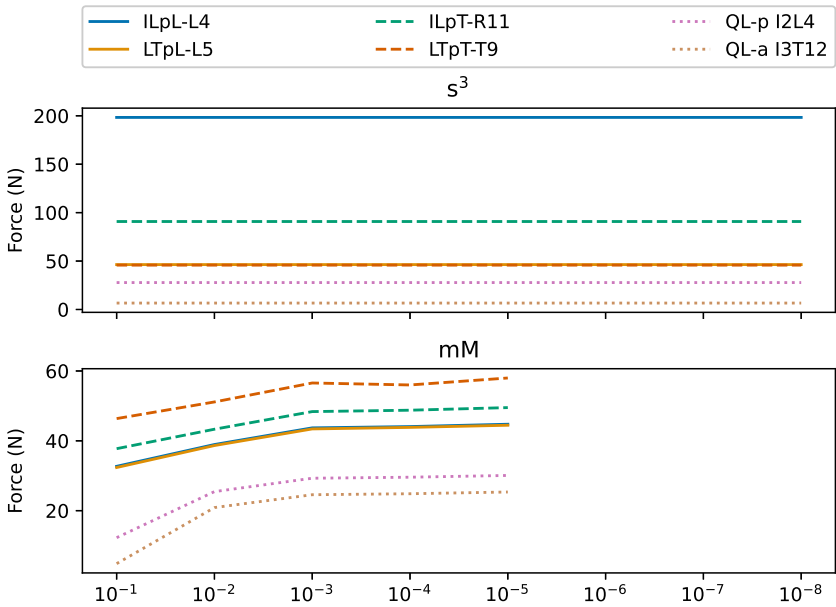


Fig. 5.1 Optimised forces in muscle fascicles with both cost functions when both tolerances were modified at the same time.

decrease of the *gtol* was associated with an increase of the number of iterations. For the two largest tolerances, the numbers of iterations were very small as the optimisation processes stopped because of the first termination condition. Once the termination condition was associated with the radius size of the trust region, the number of iterations increased more. Finally, optimised forces seemed to reach a constant value for a tolerance of at least 10^{-3} (Fig. 5.2) as highlighted when the two tolerances were modified.

In conclusion, this detailed analysis of tolerances ensures that the desired results (the forces) converge to a relatively stable value. Although we can never rule out the possibility of local optima, this analysis is reassuring compared to using deterministic optimisation for the purposes of this thesis.

5.3.2 Impact of the initial guess

As for the tolerances, whatever the initial guess given to the optimisation process (the pseudo-inverse solution or one of the ten randomised vectors), muscle forces found for the cost function s^3 were all identical which is

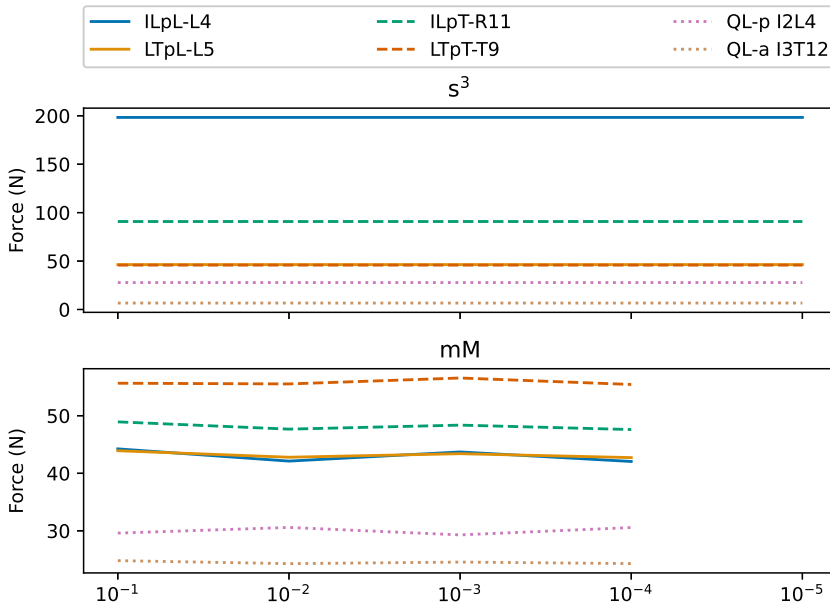


Fig. 5.2 Optimised forces in muscle fascicles with both cost functions when $gtol$ was fixed and $xtol$ varied.

reassuring in terms of the robustness of the solution¹. However, for the cost function mM , solutions found were all different. For illustration, muscle forces found starting from the pseudo-inverse solution compared to the mean and standard deviation of muscle forces found with the ten randomised initial guesses are presented in Fig. 5.4 for LP . Although the means of muscles forces for the ten randomised initial guesses were close to the muscle forces of the pseudo-inverse solution (except for the $LTpL$ fascicle inserting at L1), standard deviations were high for some muscle fascicles, e.g. $ILpL$ inserting at the second lumbar vertebra or $LTpL$ inserting at the first lumbar vertebra for LP .

Nevertheless, all muscle forces (solutions found with pseudo-inverse solution and means of the ten randomised initial guesses) were in the same order of magnitude even for the muscle fascicles for which the standard deviation was high. Only the force exerted by the $LTpL$ -L1 fascicle was

¹but which in no way demonstrates that the optimum is global, given the type of method used

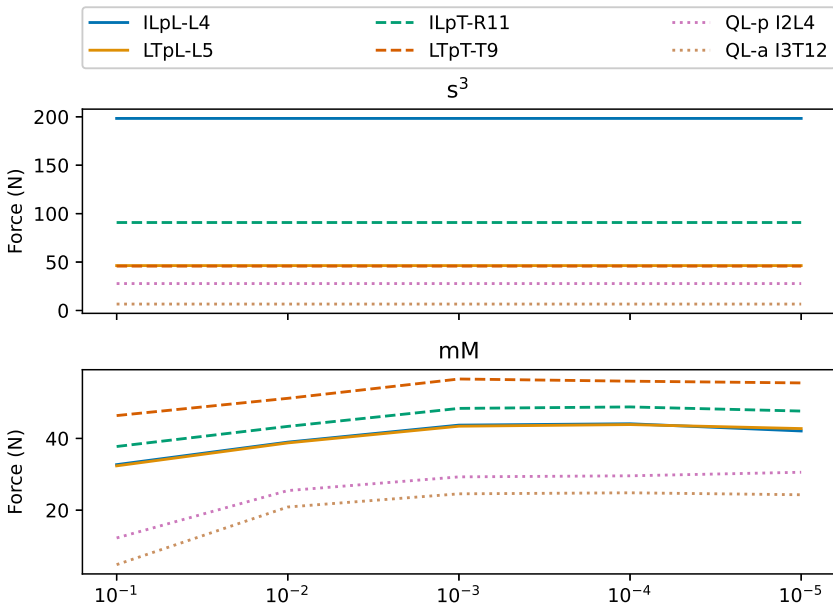


Fig. 5.3 Optimised forces in muscle fascicles with both cost functions when $xtol$ was fixed and $gtol$ varied.

different according to the initial guess. Furthermore, when looking at intervertebral efforts (Fig. 5.5 for compressive forces), one could see that the impact of the different muscle forces was weak: standard deviations were relatively small. In conclusion, for mM , although there was a high variability in muscle forces, their effects on intervertebral efforts were hardly felt. No explanation could be found for this result.

Finally, as when studying the impact of the tolerances, all the optimisation processes executed for the analysis of an initial guess stopped following the radius of the trust region being smaller than the tolerance.

5.3.3 Impact of the *logsumexp* cost function

Among the twenty values tested for the ϵ parameter (Eq. 5.1), two of them did not converge because the maximum number of iterations, i.e. 1000, was reached. $RMSD$ between mM and the approximate *logsumexp* cost function with different values of ϵ are presented in Fig. 5.6. As expected, the smaller the ϵ value, the smaller the $RMSD$. However, it might be noticed that

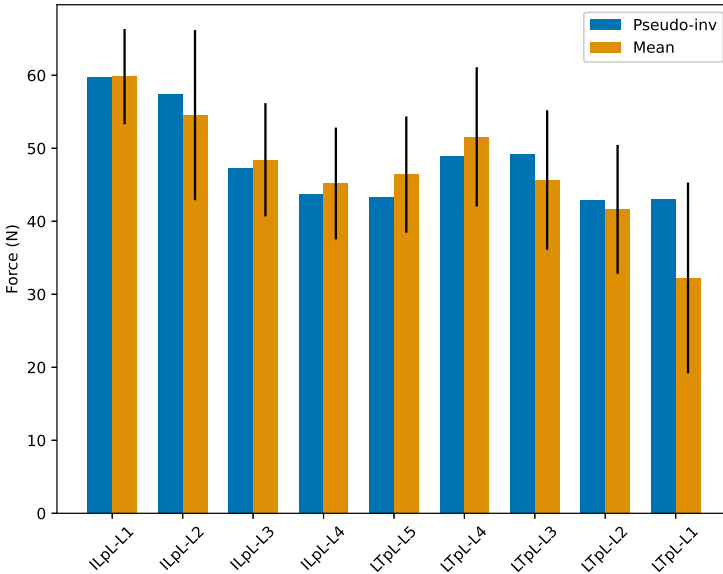


Fig. 5.4 Forces in LP muscle fascicles found by the optimisation process for the cost function mM for different initial guesses. Muscle forces found starting from the pseudo-inverse unique solution are represented in comparison with mean and standard deviation (black vertical bars) of muscle forces found with the ten randomised initial guesses.

for some smaller values, $RMSD$ increased. It meant that the optimisation process found a solution for the constrained problem, but this solution was not the closest approximation to mM . Nevertheless, $RMSD$ was relatively small, i.e. from 4 to 12 N depending on ϵ .

Also, the smaller the ϵ value, the greater the number of iterations of the optimisation process required to solve the problem (see boxed graph in Fig. 5.6). The value of ϵ should therefore be chosen according to the quality of the approximation and the time allowed for the optimisation process.

As a last comment about the approximate cost function, all the optimisation processes stopped in this case because of the first termination condition, i.e. the infinity norm of the Lagrangian gradient and the violation constraint were smaller than the tolerance (except for the smallest ϵ value for which the optimisation process stopped because of the radius of the trust region).

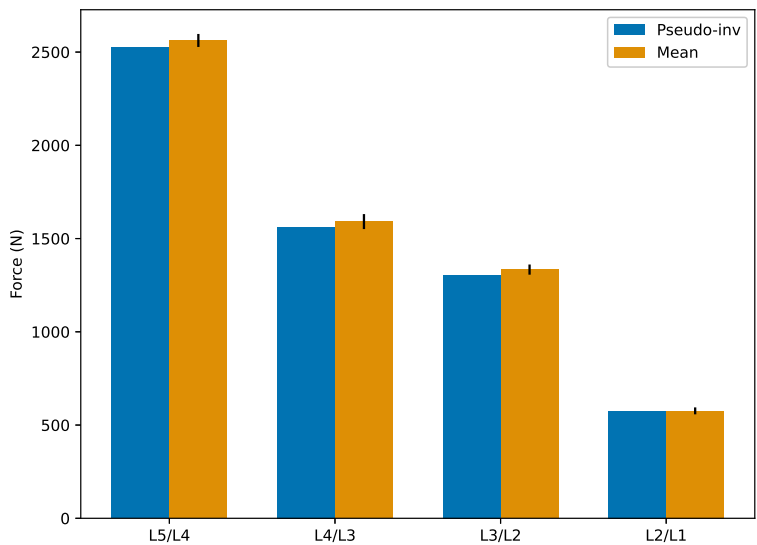


Fig. 5.5 Compressive forces at each lumbar IVD found by the optimisation process with the cost function mM starting from the pseudo-inverse solution and mean and standard deviation (black vertical bars) of compressive forces starting from the ten randomised initial guesses.

To analyse further the approximate function, a fixed ϵ value, i.e. the local minimum in Fig. 5.6, $\epsilon = 0.05456$ although sensitive, was chosen to implement the ten randomised initial guesses, identical to the ones implemented for the previous analysis. As for mM , ten different solutions were obtained, but again all in the same order of magnitude and with a relatively small impact on intervertebral efforts (not showed). The function approximation was consequently not suitable to ensure to have the same solution for different initial guesses. This point highlights the numerous local minima that the problem had.

5.4 Conclusion of the sensitivity analysis

This sensitivity analysis about the optimisation process and its parameters allowed us to dispel doubt about some optimisation uncertainties.

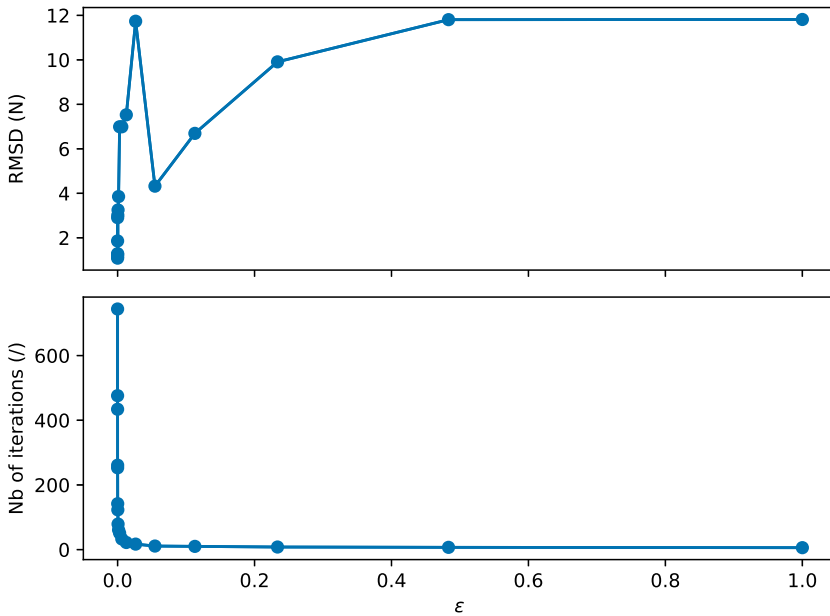


Fig. 5.6 (Top) RMSD (Eq. 5.2) and (Bottom) number of iterations performed by the optimisation process for the approximate *logsumexp* cost function as function of the ϵ parameter.

A trust-region algorithm for constrained optimisation was considered for finding muscle forces for the trunk muscles. The tolerance for termination conditions was set up at 10^{-3} as the latter was a good compromise between the precision of computed muscle forces and an affordable number of iterations (and so CPU time) for the optimisation process.

For the initial guess, the solution found with the pseudo-inverse algorithm was considered as a good start despite the variety of muscle forces found for **mM**.

Finally, an approximation of **mM**, i.e. *logsumexp* was implemented to respect the condition of the cost function to be twice continuously differentiable. This approximation was function of an ϵ parameter which reflected its quality. Twenty values of this parameter were tested among with eighteen values converged to a solution. Despite a good approximation, it was not possible, for a fixed ϵ value, to find an identical solution starting from ten randomised initial guesses. This is why it was decided to keep **mM** cost

5 | Sensitivity analysis of the optimisation process

function in the frame of this thesis as it converged all the time to a solution despite not being differentiable.

PART III

Towards the hybrid approach: results and discussion

6

Predefined static postures in weights devices: a pilot study

This chapter summarises the first experiment about the quantification of forces of back and abdominal muscles for predefined static postures in weights devices. The main objective of this pilot study was to calibrate **EMG** signals with forces computed thanks to an equivalent **MBS** model. Calibration is an essential step before studying any kind of postures or motions and computing muscle forces via optimisation. Calibration had been performed for all trunk motions, i.e. flexion, extension, lateral bending and axial rotation, thanks to four predefined static postures in spine reinforcement devices. In this chapter, the hybrid approach is not used. The purpose of the study was rather focused on the construction of look-up tables relating **EMG** amplitudes and forces.

6.1 Experiment

For this experiment, six young male subjects (age: 23.0 ± 2.2 years; mass: 70.3 ± 8.4 kg; height: 1.76 ± 0.05 m) were recruited. The experiment was realised at the 'Médecine Physique' department of the 'Cliniques Universitaires Saint-Luc' (Brussels, Belgium) and consisted in explanations, warm-up, muscle palpation and segments' measurement, **EMG** placement and **LIC** tests for all trunk motions.

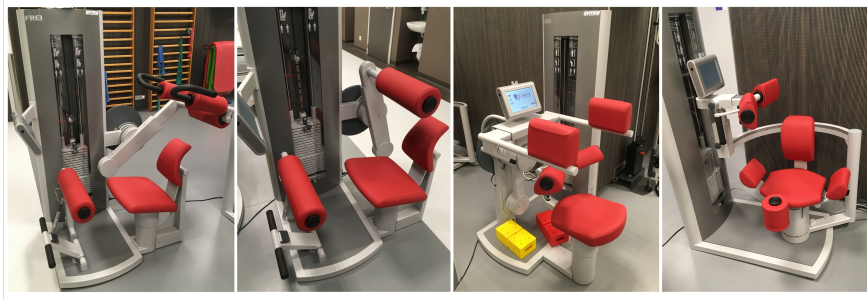


Fig. 6.1 Spine reinforcement devices for flexion, extension, lateral bending and axial rotation from left to right respectively.

As for the experiment explained in Chapter 3, each subject had to meet the eligibility criteria, i.e. lower limbs of equal length, balanced pelvis and spine alignment in the frontal plane and lack of concave/convex scoliosis curves. The placement of EMG electrodes for studied muscles was identical to the ones described in Chapter 3.

Spine reinforcement devices (FREI medical GmbH, Kirchzarten, Germany) were used to perform isometric contractions (Fig. 6.1). For each device, a static posture was established as follows

- Flexion test: 10° forward flexion;
- Extension test: backward extension at 20° forward flexion;
- Lateral bending test: 15° lateral bending on both right and left sides;
- Axial rotation test: 15° axial rotation on both right and left sides.

For each configuration, four different loads (from 5 to 20 kg with a 5-kg step) were applied randomly. For each LIC, the subject was asked to move into the desired position. Then, after a 3-second stabilisation, the subject had to keep the posture during 10 seconds. The rest duration was set at 30 seconds between each load and several minutes between each device, i.e. the time needed to shift to the next device. A familiarisation time was allowed to each subject and for each device prior to the recording of EMG signals.

During the experiment, several segment's measurements were taken in order to build the corresponding MBS model:

- In standing position,

- In the anteroposterior direction,
 - * Abdomen width at the L3 level;
 - * Thorax width at the T7 level, i.e. the distance between the T7 spinous process and the xiphoid process;
 - * Upper arm width (with the arms along the body).
- In the mediolateral direction,
 - * Distance between the two superior posterior iliac spines;
 - * Abdomen width (with the calliper laid on the iliac crest);
 - * Thorax width at the T7 level (with the calliper at the level of the posterior axillary folds);
 - * Width between shoulders.
- In the vertical direction,
 - * Abdomen height, i.e. distance between L5 and T12 vertebrae;
 - * Thorax height, i.e. distance between T12 and T1 vertebrae.
- In supine position,
 - Vertical distance between the navel and the pubic symphysis;
 - Vertical distance between the navel and the sixth rib;
 - Vertical distance between the navel and the lowest point of the rib cage.
- In prone position,
 - Vertical distance between the line connecting the two superior posterior iliac spines and the L5 vertebra.

Additionally, for the flexion and extension devices, the lever arms (Fig. 6.2) used to apply the external load were reported for each subject.

6.2 Models

In total, for each subject, six variants of the **MBS** model were developed to compute muscle forces for the different trunk motions, i.e. one model for flexion, one for extension, two for lateral bending (one for each side) and two for axial rotation as well. All the models were composed of 22 bodies: the head, two upper arms, two forearm and hand segments and 17 trunk segments, each associated to a lumbar or thoracic vertebra. Dimensions



Fig. 6.2 Lever arms (white arrows) reported for the (Left) flexion and (Middle) extension devices. (Right) Calibration for the forces to apply in the model – example on the lateral bending device.

and body parameters were taken from anthropometric data published by Winter [166] and from experimental measurements previously cited. For all the models, natural lumbar and thoracic spine curvatures were taken into account thanks to values found in the literature. The total curvatures were assumed to be 50° [167, 168] and 30° [168, 169] for the lumbar lordosis and the thoracic kyphosis respectively, evenly spread between each segment. The whole spine was also tilted forward by an angle of 30° which was the image of the sacral slope [168]. The spine curvatures were not adapted for each subject while the other geometric data was.

The models differed by the simulated motion, i.e. the plane in which the trunk rotated, and by the modelled muscle. Each model stood for one-DoF posture, which was represented by one elementary rotation at the S1/L5 level in the associated plane. The rotation angle of the S1/L5 level was not directly equal to the posture angle in the device, but rather a proportion of it, based on the distribution of the motion among the lumbar IVD proposed by Rozumalski et al. [52]. To avoid the muscle redundancy problem and as the weights devices were assumed to isolate the action of the most agonist muscle, the strategy was to model the muscle as a point-to-point force transmitter whose application points corresponded to origin and insertion points of the muscle. This strategy was close to single-equivalent-muscle models proposed by Dreischarf et al. [16].

The modelled muscle for each weights device was:

- Bilateral RA for flexion;

- Bilateral **LP** for extension;
- Ipsilateral, i.e. on the same side than the one of the motion, **QL** for lateral bending and;
- Contralateral, i.e. on the opposite side than the one of the motion, **OE** for axial rotation.

Muscle forces were computed by inverse equilibrium (Eq. 4.16) as described in Section 4.1.3. For each posture, the weights device was not modelled but its resultant force was applied at the application point reported during the experiment and according to its direction. For flexion, the application point was at the elbow point and the force pointed towards the posterior direction and upwards. For extension, it was at the spinous process of the T2 vertebra, pointing towards the anterior direction and upwards. For both lateral bending and axial rotations, the application point was at the shoulder level. But the force direction was upwards and towards the center of the human body for lateral bending and towards the back and towards the median line for axial rotation. The resultant force of each weights device and for each load was experimentally measured with an external force sensor during a calibration phase prior to the experiment with subjects (Fig. 6.2 right). Indeed, the force was not equal to the load weight, i.e. mass times the gravity constant, due to friction and pulleys present in the device.

To build up the look-up tables calibrating **EMG** amplitudes with muscle force, linear regressions have been performed between **RMS** values of **EMG** amplitudes and loads and between muscle forces computed with models and loads.

6.3 Results and discussion

Results are showed in Fig. 6.3 for the left axial rotation. We observed that **EMG** amplitudes were linearly dependent on the loads, while muscle forces showed a good linear relationship with loads. Mathematically, this meant that there was a linear relationship between **EMG** amplitudes and trunk muscle forces (also visible for other trunk motions and muscles but not showed). No direct comparison of muscles forces could be made with the literature but obtained values seemed physically realistic, i.e. muscle forces normalised by the gravity constant were equivalent to loads that could be lifted (Fig. 6.3 bottom). For the lateral bending, a first quantification was also obtained with realistic values for almost all subjects. Indeed,

for two subjects, the model and its associated linear regression predicted unrealistic negative forces for loads smaller than 2.5 kg. For asymmetric tasks, i.e. lateral bending and axial rotation, while EMG amplitudes were not identical for the right and left sides, models could not distinguish the difference between both. Therefore, muscle forces obtained in the right-side QL for the right lateral bending were exactly identical to the ones obtained for the left-side QL for the left lateral bending, and similarly for the axial rotations. It was due to the symmetrical nature of the model.

For flexion and extension, unrealistic muscle forces were obtained. For flexion, muscle forces were negative for loads smaller than 20 kg while for extension, muscle forces were miscellaneous. For one subject, muscle forces were negative for all loads. Furthermore, for all subjects, the order of magnitude for muscle forces did not make any sense as the latter were all above 50000 N in absolute values. It was due to the combined effects of the small lever arms of LP in the model, i.e. ≈ 4 cm, and the almost parallel alignment of LP with the spine. These unrealistic results for both flexion and extension suggested that the muscle insertion points were not precisely located.

Experimental results could already reflect the different muscle strategies that could appear (see Fig. 6.3 on the left), even with weights devices assumed to activate the most agonist muscle. In particular, for the subject 6, the slope of the linear regression was not as high as for the other subjects. After investigation, it appeared that this subject had a better motor control, due to its formation and profession as a physiotherapist. Furthermore, he probably performed the task differently than the other subjects and used both QL and LD. There was clearly a combined activation of several muscles for some static postures although we thought that the spine reinforcement devices would isolate the activation of the most agonist muscle of each motion.

This pilot study showed that it was plausible to calibrate EMG amplitudes of trunk muscles with associated forces computed through a model. It was also a preliminary quantification of trunk muscles in the frame of this thesis. However, two limitations appeared. Firstly, it was necessary to refine the insertion points location of the muscles. Secondly, although the weights devices were selected in order to highlight the activation of one main muscle strategy, it experimentally appeared that several muscles were activated, inducing different strategies between subjects.

Although very instructive, this pilot study exhibited "paradoxically constructive" limitations. The latter led to the conclusions that to pursue

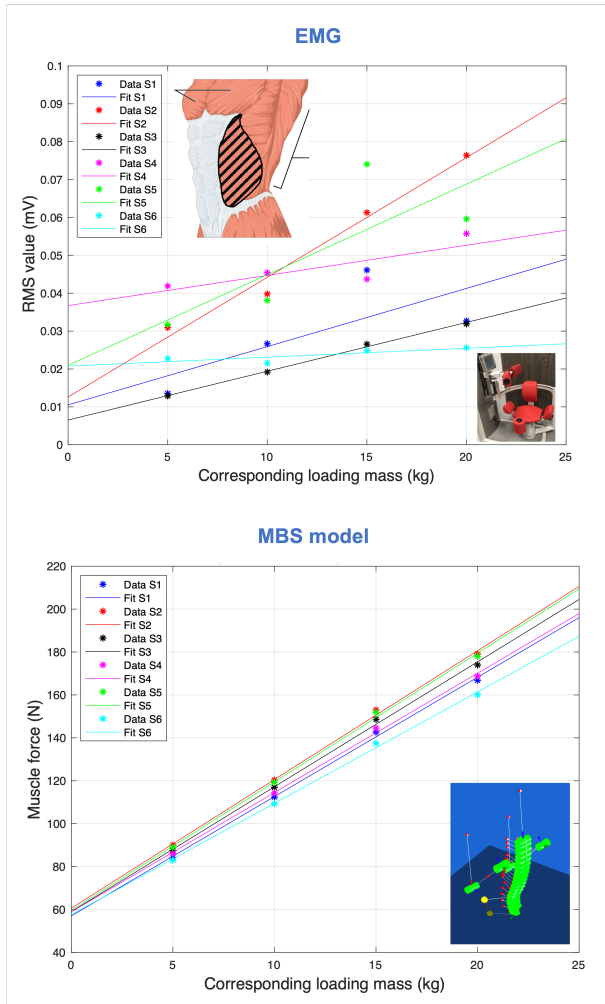


Fig. 6.3 (Top) **EMG** amplitudes (in mV) and (Bottom) muscle forces (in N) as function of the loads (in kg) in the left **OE** for the isometric axial rotation towards the right for the 6 subjects.

the development of the hybrid approach, it was necessary to select a static posture which recruited only agonist muscles with a relatively high intensity but without using any complex weights device. Such an exercise could be the Sorensen test as described in Section 3.2. Furthermore, to improve the accuracy of our model, it was decided to extract the musculoskeletal model

of the trunk developed by Bruno et al. [105] from the OpenSim software and to model the muscle fascicles in a more detailed way rather than the equivalent-muscle model used in this study.

Associated publications

The results of this chapter were presented at

- the 9th ECCOMAS Thematic Conference on Multibody Dynamics 2019 [170];
- the 44^{ème} Congrès de la Société de Biomécanique 2019 [171], the abstract of which has been published in [172].

7

Hybrid approach vs optimisation-based approach

In this chapter, the first version of the hybrid ([Hybrid-1](#)) approach, i.e. an EMG-based deterministic distribution of a global equivalent force, is compared with pure optimisation-based computations. This comparison is made in two steps. First, the hybrid approach is applied on the sample of 19 subjects to highlight the different muscle strategies. Second, is it applied to a subject who deliberately produces three different muscle strategies. This second step aimed to take into account the muscle redundancy problem thanks to the use of [EMG](#).

7.1 Application to the sample of 19 subjects

7.1.1 Experiment and model

For this first application, the experiment and the model were explained in Section [3.2](#) and Section [4.2](#) respectively. The hybrid and the optimisation-based approaches were explained in Section [4.3](#) and [4.4](#).

Hybrid and optimisation-based approaches were compared based on predicted muscle forces and intervertebral efforts. To eliminate confounding factors, normalised efforts were used, i.e. forces or torques divided by the subject's body mass.

Furthermore, to carry out a more in-depth comparison, forces measured during MVC tests were implemented in the MBS model for the optimisation-based approach only. The goal was to confront cost functions to a maximum contraction measured experimentally. For that purpose, the mass equivalent to the force from MVC tests for each subject was added to the 'load' body in the MBS model used to load the subject. Concerning muscle forces computed in that case, the difference between the force of one particular muscle fascicle and the maximum isometric force associated with this fascicle was computed as follows

$$\Delta = \frac{F_{musc} - F_{max}}{F_{max}} * 100 \quad (\%)$$

A positive Δ stands for an increase with respect to the maximum isometric force while a negative Δ stood for a decrease. A threshold, arbitrarily put at 50%, was used for the difference exceeding, i.e. a positive difference, to eliminate non relevant cost functions.

7.1.2 Muscles forces and intervertebral efforts in response to external load

Muscle forces Whatever the approach used, i.e. the hybrid computation and even the different criteria within the optimisation-based computation, muscle strategies were globally coherent but revealed locally significant differences that can be instructive from a physiological point of view (see Fig. 7.1 for the case of no external load). Firstly, the hybrid approach distinguished itself with a greater recruitment of muscle fascicles from all the muscles (LP, TP and QL) while the optimisation-based solutions seemed to favour one muscle and/or disregarded another. Overall, it was clear and somewhat reassuring that LP were the most activated muscles for both hybrid and optimisation-based solutions and for all loads applied to the subjects, except for F^2 which recruited LP and TP muscle fascicles in a similar way. One could observe that there were three trends about variability in muscle forces (see the error bars in Fig. 7.1):

- For F^2 , s^2 and s^3 , there was no variability at all. The same muscle strategy was observed in all subjects. This result was normal as the scaling did not bring any change in the optimisation computation.
- On the contrary to the other cost functions, mM showed some variability due to the stopping criterion in the numerical computation. There was therefore a subject dependency but which barely impacted

the average of muscle forces, and consequently their quality.

- The greatest variability was obviously observed for the EMG solution, inherently due to the experimental variability.

Finally, for optimisation-based solutions, the greatest maximum isometric force in a fascicle was, the greatest muscle force in this fascicle was.

About LP, there was a bigger recruitment of the ILpL-L4, muscle fascicle joining the pelvis and the fourth lumbar vertebra, for EMG, s^2 and s^3 . For the hybrid solution, this greater muscle force in ILpL-L4 was the consequence that (i) LP were the most activated muscles in the Sorensen test posture for 15-sec isometric contractions and (ii) the force distribution inside LP muscles favoured the ILpL-L4 fascicle as its maximum isometric force was the largest. The same reason also explained why the solutions s^2 and s^3 had a greater muscle force in this fascicle as well. F^2 and mM solutions showed relatively low and evenly-distributed forces in LP muscle fascicles compared to the three other solutions.

Forces in TP muscle fascicles were the second greatest forces but strategies differed between solutions. The hybrid solution aimed at showing that muscle forces were lower compared to the optimisation-based solutions. For the optimisation-based solutions, F^2 differed by recruiting TP muscle fascicles more, specially fascicles of ILpT, going from the pelvis to the fifth to ninth rib but inserting first on the second lumbar vertebra. s^2 and s^3 showed the greatest muscle forces in TP overall for the ILpT fascicles going to the tenth, eleventh and twelfth ribs. Similarly to LP muscles, it was due to a greater maximum isometric force available in these fascicles. For the other fascicles in the optimisation-based methods, the comment made previously still held: the greatest maximum isometric force in a fascicle was, the greatest the muscle force in this fascicle was.

QL was the most interesting muscle to look at, in terms of comparison between the hybrid solution and the mathematical ones. Indeed, EMG revealed strategies recruiting the QL the most, specially the posterior fibers connected the pelvis with the four lumbar vertebrae with a mean normalised muscle force equal to 1.2 N/kg, which was greater than all normalised muscle forces for TP muscles for the hybrid solution. Compared to the latter, mathematical solutions completely missed the QL except the mM solution which activated all the QL fascicles but with far less magnitude than the hybrid approach (at least half smaller on average). F^2 , s^2 and s^3 solutions activated a bit the lowest posterior QL fascicles.

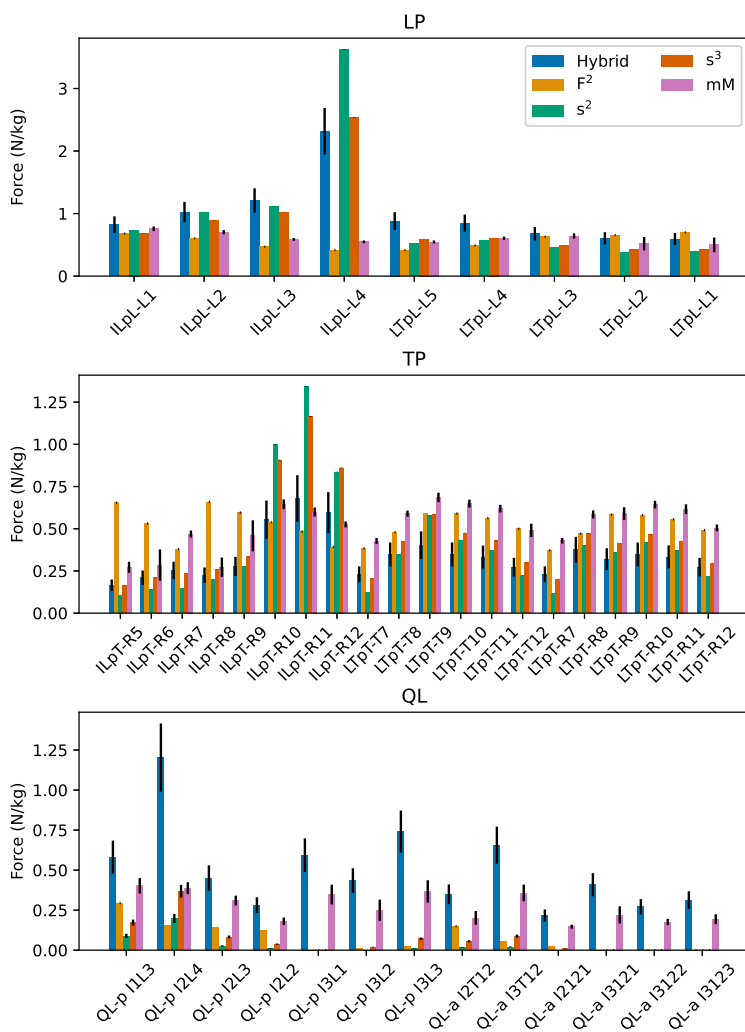


Fig. 7.1 Normalised muscle forces (in N/kg) computed by each method for **LP**, **TP** and **QL** with no external load. The standard deviation is illustrated thanks to the error bar.

Finally for muscle forces, adding loads increased muscle forces globally whatever the method used without modifications in the pattern of forces in muscle fascicles with respect to the pattern obtained for the unloaded case.

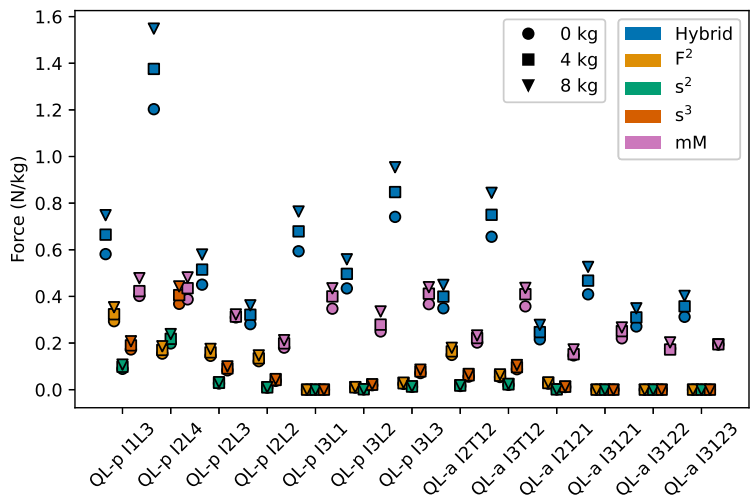


Fig. 7.2 Normalised muscle forces (in N/kg) computed by each method for **QL** for 0-kg, 4-kg and 8-kg loads. Each point, representing the mean of normalised muscle forces, stands for a specific external load (symbol) and for a given method (color).

Besides, this result still held for the **QL** (Fig. 7.2) whose the recruitment was revealed by the hybrid approach mostly.

Intervertebral efforts Intervertebral efforts, obtained by constraining the associated motion via driven variables (Eq. 4.15) in the inverse equilibrium solution, were directly a consequence of the muscle forces computed previously. Firstly, regarding the variability, the comments made for muscle forces still held. However, once could observe a noticeable variability (Fig. 7.3) between subjects for bending moments for F^2 , s^2 and s^3 , simply due to the different lever arms from the scaling. Secondly, the greater recruitment of muscle fascicles for the hybrid solution resulted in greater compressive forces for this solution for all lumbar **IVD** levels as well as a greater shear force for the L5/L4 and L4/L3 levels (Fig. 7.3, Fig. 7.4 and Table 7.1). However, the hybrid approach showed lower shear forces for the two superior lumbar levels and reduced bending moments (in absolute value) for the same levels as for the L4/L3 level as well. This result held

when adding loads to subjects (see Fig. 7.4 and Table 7.2 and 7.3 for more details).

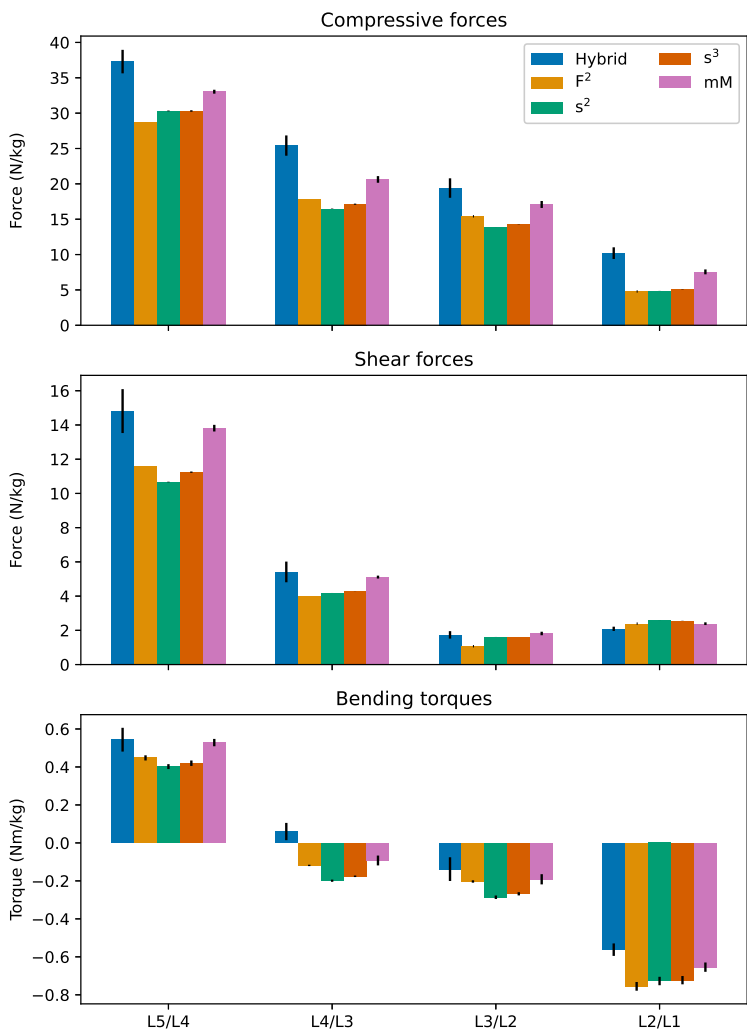


Fig. 7.3 Normalised compressive and shear forces (in N/kg) and bending moment (in Nm/kg) obtained by each method for the four superior lumbar IVD levels with no external load. The standard deviation is illustrated given by the error bar.

Looking at optimisation-based solutions, mM stood out being the closest

Load (kg)	EMG	F ²	s ²	s ³	mM
Level L5/L4					
0	37.3±1.7	28.7±0.0	30.3±0.1	30.3±0.1	33.0±0.3
2	39.4±1.8	30.2±0.2	31.9±0.1	31.8±0.2	34.8±0.3
4	41.4±1.9	31.6±0.3	33.4±0.3	33.4±0.3	36.3±0.4
6	43.5±2.1	33.1±0.4	34.9±0.4	34.9±0.4	37.9±0.5
8	45.5±2.3	34.5±0.6	36.4±0.6	36.4±0.6	39.7±0.7
Level L4/L3					
0	25.4±1.4	17.8±0.0	16.5±0.0	17.1±0.1	20.6±0.5
2	26.9±1.5	18.7±0.1	17.3±0.1	18.0±0.1	21.9±0.3
4	28.4±1.6	19.7±0.2	18.2±0.2	18.9±0.2	22.6±0.5
6	29.9±1.8	20.6±0.3	19.0±0.2	19.7±0.3	23.6±0.4
8	31.3±2.0	21.5±0.4	19.8±0.3	20.6±0.3	24.5±0.7
Level L3/L2					
0	19.4±1.4	15.4±0.0	13.8±0.0	14.2±0.0	17.1±0.5
2	20.5±1.5	16.2±0.1	14.5±0.1	15.0±0.1	18.1±0.2
4	21.6±1.5	17.0±0.2	15.2±0.1	15.7±0.1	18.8±0.4
6	22.7±1.7	17.8±0.2	15.9±0.2	16.4±0.2	19.6±0.4
8	23.8±1.8	18.6±0.3	16.7±0.3	17.2±0.3	20.4±0.6
Level L2/L1					
0	10.2±0.8	4.8±0.0	4.9±0.0	5.1±0.0	7.6±0.4
2	10.9±0.9	5.1±0.0	5.1±0.0	5.4±0.0	8.0±0.2
4	11.5±0.9	5.3±0.1	5.4±0.1	5.6±0.1	8.3±0.3
6	12.2±1.0	5.6±0.1	5.7±0.1	5.9±0.1	8.6±0.3
8	12.8±1.1	5.8±0.1	5.9±0.1	6.2±0.1	8.9±0.6

Table 7.1 Compressive forces (mean±sd) in N/kg obtained by each method for the four superior lumbar levels and for all the loads.

to the hybrid solution, specially for bending moments and shear forces – except at the lowest lumbar level – while F^2 , s^2 and s^3 were relatively close to each other but the furthest from the hybrid solution.

In view of Tables 7.1, 7.2 and 7.3, the lowest lumbar level L5/L4 showed the greatest intervertebral efforts. For superior lumbar levels, magnitude of compressive forces decreased as well as the magnitude of shear forces except for the uppermost lumbar level, i.e. L2/L1. For bending moments, the greatest magnitudes were obtained at the highest lumbar levels followed closely by the lowest lumbar level. L4/L3 and L3/L2 levels showed

Load (kg)	EMG	F ²	s ²	s ³	mM
Level L5/L4					
0	14.8±1.3	11.6±0.0	10.7±0.0	11.2±0.0	13.8±0.2
2	15.7±1.4	12.2±0.1	11.2±0.1	11.9±0.1	14.6±0.1
4	16.5±1.4	12.9±0.1	11.8±0.1	12.5±0.1	15.2±0.2
6	17.4±1.5	13.5±0.2	12.4±0.2	13.1±0.2	15.9±0.2
8	18.3±1.6	14.1±0.3	13.0±0.2	13.7±0.2	16.7±0.4
Level L4/L3					
0	5.4±0.6	4.0±0.0	4.2±0.0	4.3±0.0	5.1±0.1
2	5.8±0.6	4.3±0.0	4.5±0.0	4.5±0.0	5.4±0.1
4	6.1±0.7	4.5±0.1	4.7±0.1	4.8±0.1	5.7±0.1
6	6.5±0.7	4.8±0.1	5.0±0.1	5.1±0.1	6.0±0.1
8	6.8±0.7	5.0±0.1	5.2±0.1	5.3±0.1	6.3±0.2
Level L3/L2					
0	1.7±0.2	1.1±0.0	1.6±0.0	1.6±0.0	1.8±0.1
2	1.9±0.2	1.2±0.0	1.8±0.0	1.8±0.0	2.0±0.1
4	2.0±0.2	1.3±0.0	1.9±0.0	1.9±0.0	2.1±0.1
6	2.2±0.3	1.4±0.0	2.1±0.0	2.1±0.0	2.3±0.1
8	2.3±0.3	1.5±0.1	2.2±0.1	2.2±0.1	2.5±0.1
Level L2/L1					
0	2.1±0.1	2.4±0.0	2.6±0.0	2.6±0.0	2.4±0.1
2	2.3±0.1	2.6±0.0	2.8±0.0	2.8±0.0	2.6±0.1
4	2.4±0.1	2.8±0.0	3.0±0.0	3.0±0.0	2.8±0.1
6	2.6±0.2	3.0±0.1	3.2±0.1	3.2±0.1	3.0±0.1
8	2.8±0.2	3.2±0.1	3.4±0.1	3.4±0.1	3.3±0.2

Table 7.2 Shear forces (mean±sd) in N/kg obtained by each method for the four superior lumbar levels and for all the loads.

relatively low magnitudes, the hybrid solution getting the lowest.

Bending moments were positive for the L5/L4 level and negative for other lumbar levels except for the L4/L3 level obtained by the hybrid solution. Bending moments reflected the behaviour of internal fibres of IVD. Positive bending moments indicated IVD in which posterior internal fibres were in compression and anterior internal fibres in traction; negative bending moments indicated the opposite behaviour.

Adding loads logically resulted in greater magnitudes in all intervertebral efforts (Table 7.1, 7.2 and 7.3) without modifying either the efforts

Load (kg)	EMG	F ²	s ²	s ³	mM
Level L5/L4					
0	0.5±0.1	0.5±0.0	0.4±0.0	0.4±0.0	0.5±0.0
2	0.6±0.1	0.5±0.0	0.4±0.0	0.4±0.0	0.6±0.0
4	0.6±0.1	0.5±0.0	0.4±0.0	0.5±0.0	0.6±0.0
6	0.6±0.1	0.5±0.0	0.5±0.0	0.5±0.0	0.6±0.0
8	0.7±0.1	0.5±0.0	0.5±0.0	0.5±0.0	0.6±0.0
Level L4/L3					
0	0.1±0.1	-0.1±0.00	-0.2±0.0	-0.2±0.0	-0.1±0.0
2	0.1±0.1	-0.1±0.0	-0.2±0.0	-0.2±0.0	-0.1±0.0
4	0.1±0.1	-0.1±0.0	-0.2±0.0	-0.2±0.0	-0.1±0.0
6	0.1±0.1	-0.1±0.0	-0.2±0.0	-0.2±0.0	-0.1±0.0
8	0.1±0.1	-0.1±0.0	-0.2±0.0	-0.2±0.0	-0.1±0.0
Level L3/L2					
0	-0.1±0.1	-0.2±0.0	-0.3±0.0	-0.3±0.0	-0.2±0.0
2	-0.1±0.1	-0.2±0.0	-0.3±0.0	-0.3±0.0	-0.2±0.0
4	-0.1±0.1	-0.2±0.0	-0.3±0.0	-0.3±0.0	-0.2±0.0
6	-0.1±0.1	-0.2±0.0	-0.3±0.0	-0.3±0.0	-0.2±0.0
8	-0.1±0.1	-0.2±0.0	-0.3±0.0	-0.3±0.0	-0.2±0.0
Level L2/L1					
0	-0.6±0.0	-0.8±0.0	-0.7±0.0	-0.7±0.0	-0.7±0.0
2	-0.6±0.0	-0.8±0.0	-0.8±0.0	-0.8±0.0	-0.7±0.0
4	-0.6±0.0	-0.8±0.0	-0.8±0.0	-0.8±0.0	-0.7±0.0
6	-0.6±0.0	-0.9±0.0	-0.8±0.0	-0.8±0.0	-0.7±0.0
8	-0.6±0.0	-0.9±0.0	-0.9±0.0	-0.8±0.0	-0.8±0.0

Table 7.3 Bending moments (mean±sd) in Nm/kg obtained by each method for the four superior lumbar levels and for all the loads.

distribution between IVD levels or the behaviour described here before (Fig. 7.4).

Finally, when comparing intervertebral efforts between themselves, the greatest intervertebral was the bending moment with a maximum equivalent stress of 2.30 MPa (by the hybrid approach, for 8 kg and for L5/L4 level) by referring to the beam theory and considering a circular IVD of 18 cm² [102, 160] with a neutral fibre located in the middle of the disc. For the same case, equivalent stresses associated to compressive and shear forces were 1.87 MPa and 0.75 MPa respectively. Equivalent stresses are used for

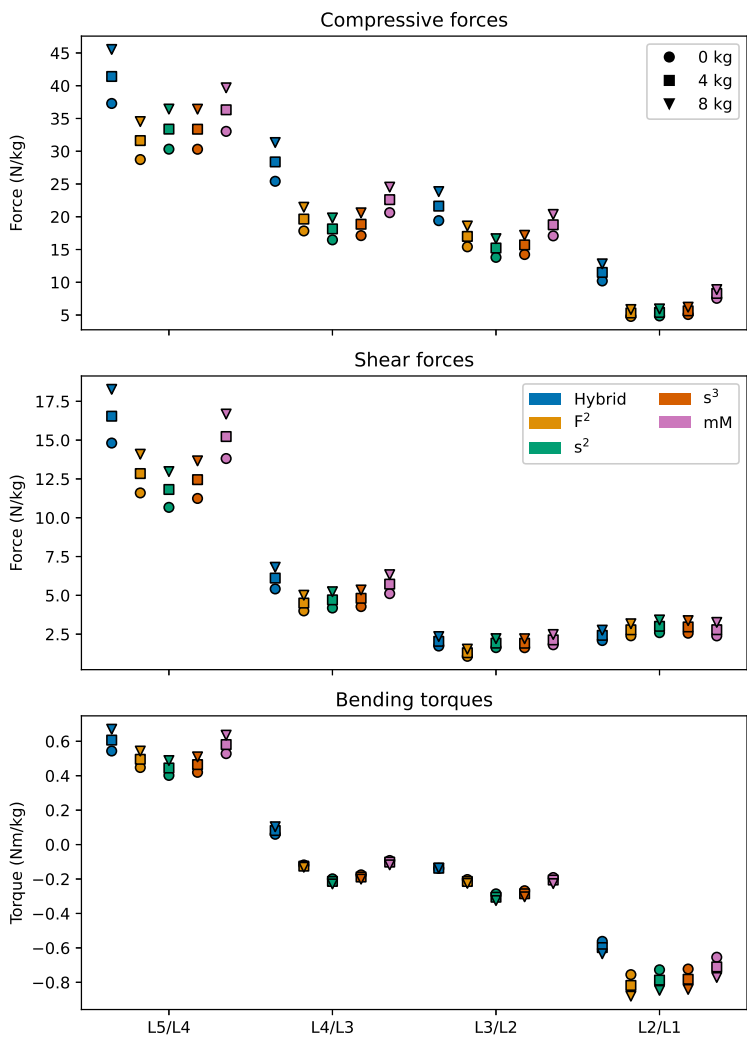


Fig. 7.4 Normalised compressive and shear forces (in N/kg) and bending moment (in Nm/kg) obtained by each method for the four superior lumbar IVD levels for 0-kg, 4-kg and 8-kg loads. Each point, representing the mean of normalised forces or torques, stands for a specific external load (symbol) and for a given method (color).

comparison with the literature, which is detailed in Section 7.1.4.

7.1.3 Muscles forces in response to maximum loading

Maximum loading could be simulated only for F^2 , s^2 , s^3 and mM by adding in the model a virtual load equivalent to the force measured during MVC tests. Maximum isometric forces could not be simulated using the hybrid approach. When looking at muscle forces in this case (see Fig. 7.5 and Tables 7.4, 7.5 and 7.6), muscle forces exceeded for F^2 and s^2 . The largest exceeding was for F^2 for the ILpT-R5. F^2 exceeded the maximum isometric force four times while only once for s^2 . s^3 did not exceed the 50%-limit but was close for one LP fascicle (ILpL-L4) with a 44%-difference. The highest difference for mM was 27% for two TP fascicles. QL was the only muscle for which maximum isometric forces were not reached.

Muscle fascicle	F^2	s^2	s^3	mM
ILpL-L1	+8%	+18%	+9%	+26%
ILpL-L2	-23%	+30%	+15%	-2%
ILpL-L3	-49%	+20%	+10%	-33%
ILpL-L4	-76%	+105%	+44%	-68%
LTpL-L5	-39%	-23%	-12%	-16%
LTpL-L4	-24%	-12%	-6%	-0%
LTpL-L3	+22%	-10%	-5%	+24%
LTpL-L2	+43%	-16%	-8%	+4%
LTpL-L1	+55%	-13%	-6%	-24%

Table 7.4 Difference (%) w.r.t. the maximum isometric force obtained by each method (except the hybrid approach) for LP in case of maximum loading. Values in **bold** indicate a 50% exceeding of maximum isometric force.

7.1.4 Relevancy of the hybrid approach

Hybrid approach similar to mM but more physiologically correct In this study, the hybrid approach gave muscle forces more representative of the measured EMG signals compared to muscle forces obtained by optimisation. In terms of muscle recruitment, it looked like mM . Furthermore, both hybrid approach and mM gave similar magnitudes for intervertebral efforts. However, mM tended to underestimate LP recruitment, i.e. if LP and TP were only considered, the hybrid approach would be closer to muscle forces obtained with s^3 in terms of magnitude. In other words, the muscle forces with the hybrid approach had the magnitude of those with s^3 , but the pat-

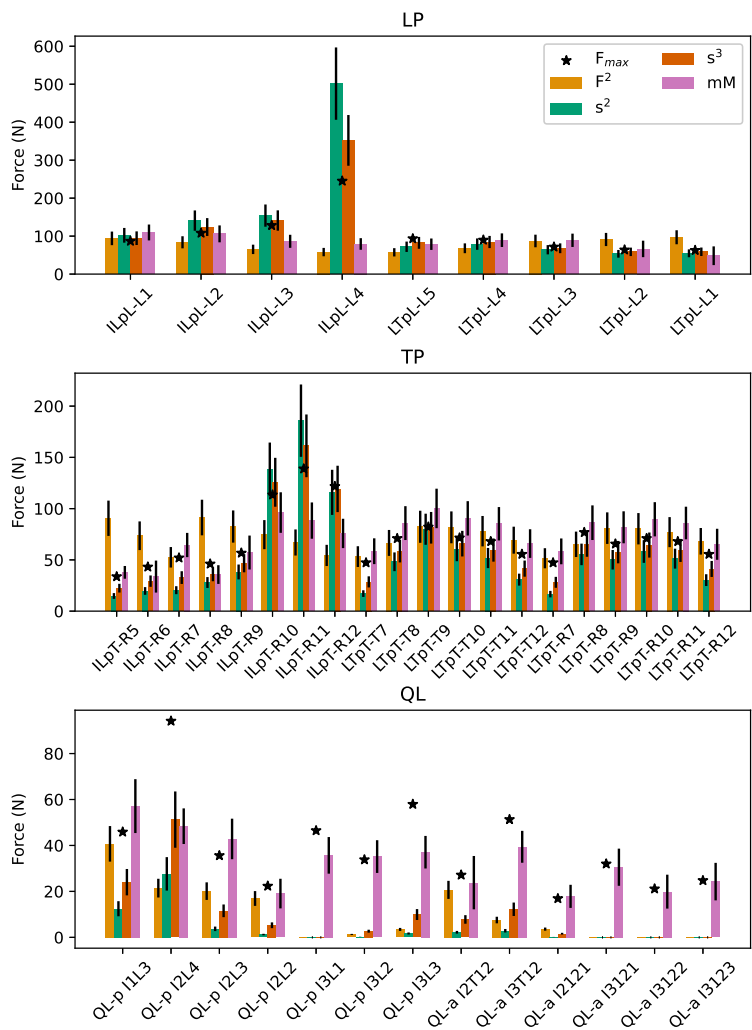


Fig. 7.5 Muscle forces (N) in the three muscles (LP, TP and QL) in response to maximum load. Maximum isometric forces for each fascicle are indicated with a black star. Standard deviations are indicated with a black vertical error bar.

tern of those with mM . Besides, all the optimisation-based solutions, except mM , failed to recruit QL. This ability of an hybrid approach to recruit all the

Muscle fascicle	F ²	s ²	s ³	mM
ILpT-R5	+169%	-56%	-33%	+14%
ILpT-R6	+70%	-54%	-32%	-23%
ILpT-R7	+1%	-61%	-37%	+24%
ILpT-R8	+98%	-39%	-22%	-23%
ILpT-R9	+45%	-33%	-18%	+1%
ILpT-R10	-34%	+22%	+11%	-15%
ILpT-R11	-52%	+34%	+16%	-36%
ILpT-R12	-55%	-5%	-2%	-38%
LTpT-T7	+13%	-64%	-40%	+23%
LTpT-T8	-6%	-32%	-17%	+21%
LTpT-T9	+0%	-3%	-1%	+22%
LTpT-T10	+14%	-16%	-8%	+26%
LTpT-T11	+15%	-24%	-12%	+25%
LTpT-T12	+25%	-45%	-25%	+19%
LTpT-R7	+9%	-65%	-41%	+23%
LTpT-R8	-15%	-28%	-15%	+12%
LTpT-R9	+23%	-24%	-12%	+25%
LTpT-R10	+13%	-18%	-9%	+26%
LTpT-R11	+13%	-25%	-13%	+26%
LTpT-R12	+23%	-46%	-26%	+17%

Table 7.5 Difference (%) w.r.t. the maximum isometric force obtained by each method (except the hybrid approach) for **TP** in case of maximum loading. Values in **bold** indicate a 50% exceeding of maximum isometric force.

muscle fascicles had already been highlighted for an upright configuration [128].

Other cost functions have been proposed previously but were not considered in this work for different reasons. Linear cost functions, i.e. minimising the sum of muscle forces, relative muscle forces or muscle stresses [18, 113] have been eliminated from the study because they produced non physiological solutions in terms of muscle recruitment. Indeed, for the Sorensen test posture, muscle forces obtained for linear cost functions were greater than maximum isometric forces (not showed). Cost functions involving relative muscle forces such as $\min \sum \left(\frac{F_{musc}}{F_{max}} \right)^p$ where p is a number greater than 2 were also excluded as they produced exactly the same results as $\min \sum \left(\frac{F_{musc}}{PCSA} \right)^p$ because **PCSA** and maximum isometric forces were directly

Muscle fascicle	F ²	s ²	s ³	mM
QL-p-I1-L3	-11%	-73%	-48%	+24%
QL-p-I2-L4	-77%	-71%	-46%	-48%
QL-p-I2-L3	-43%	-90%	-68%	+20%
QL-p-I2-L2	-24%	-95%	-77%	-16%
QL-p-I3-L1	-100%	-100%	-100%	-23%
QL-p-I3-L2	-96%	-99%	-92%	+5%
QL-p-I3-L3	-94%	-97%	-83%	-36%
QL-a-I2-T12	-24%	-92%	-71%	-14%
QL-a-I3-T12	-85%	-94%	-76%	-23%
QL-a-I2-R12 ¹	-79%	-99%	-91%	+6%
QL-a-I3-R12 ¹	-100%	-100%	-100%	-4%
QL-a-I3-R12 ²	-100%	-100%	-100%	-7%
QL-a-I3-R12 ³	-100%	-100%	-100%	-2%

Table 7.6 Difference (%) w.r.t. the maximum isometric force obtained by each method (except the hybrid approach) for **QL** in case of maximum loading. I1, I2 and I3 refer to different insertion points on the pelvis. R12 stands for the twelfth rib and each subscript refers to a different insertion point on this rib.

linked via [MMS](#).

EMG: relevant but underused input As a reminder, using **EMG** as input for musculoskeletal models was not new. As explained in Section 2.1.3, Granata and Marras proposed EMG-assisted modelling techniques to make the most of **EMG** signals [121]. Their equations were used afterwards and compared with other EMG-assisted optimisation approaches [120, 127, 131, 132, 133]. All these studies focused on load-lifting tasks which were mainly dynamic. In this perspective, muscle forces computations depend on muscle length and velocity, which are difficult to estimate and are often subject-dependent.

In our work, a static configuration was preferred to eliminate dependencies to muscle dynamics. This allowed the hybrid approach to be defined as a relatively simple weighting technique between the normalised **EMG** signals of the three muscles considered to find a unique global equivalent force. Despite these simplifications, intervertebral efforts computed with the hybrid approach were close to intervertebral efforts computed by optimisation. The hybrid approach had the advantage to reflect actual muscle

strategies better. This result had been previously obtained [127] but did not make consensus for EMG-assisted optimisation [120, 128]. Results obviously depend on how EMG signals are considered in the force definition. Beyond comparison between different hybrid approaches and their (simple or more complex) relationship to muscle forces, we could ask ourselves why EMG which show promising result as input for musculoskeletal models is still not enough considered for that purpose [173].

Impact of the QL An important finding of this work was the more intense recruitment of the most anterior QL fibres highlighted by the hybrid approach which was consistent with the smallest bending moments observed for the most superior lumbar levels. It seemed to be aimed at preventing the spine from buckling by surrounding it with more muscle forces. As compensation, recruiting the QL more resulted in greater compressive forces but reduced shear forces. This finding was logical physiologically as the spine is supposed to resist to compression forces and less to shear forces and bending moments. This result underlines the fact that the hybrid approach produces muscles forces which are guesses supposed to be closer to the physiological solution.

Towards validation Due to the Sorensen test posture, IVD were loaded in the posteroanterior direction, i.e. aligned with the gravity. Shear forces were obviously higher than those obtained for a standing configuration. The greatest shear force was obtained at the L5/L4 level for the 8-kg load computed with the hybrid approach with a mean shear stress of 0.75 MPa. A value of 1.1 MPa had been obtained previously in-vivo at the same level but for a standing position and bending forward [102]. According to Gallagher and Marras [174], our shear stress corresponded to an excessive shear for the lumbar spine. Indeed, a limit of 1000N shear is supposed to be acceptable for a relatively low frequency, less than 100 cycles a day [174]. The greatest shear for our study was around 1350 N, which is above the 1000N limit. However, it only occurred once during the experiment and should not appear regularly as the Sorensen test posture is not a daily task at all. It only insisted on the challenging loading in the anteroposterior direction. A one-occurrence shear limit was set up to 2000N which is far more than the shear value obtained in our study [174].

Compressive forces were obviously smaller than for a standing position but still intense with the greatest compressive force also at the L5/L4 level for the hybrid approach and for 8 kg with a value of 1.87 MPa.

Ghezelbash et al. developed subject-specific regression equations to estimate compressive and shear forces at L5/L4 level for lifting tasks in an upright standing position [146]. Using their equations, normalised compressive and shear forces were 41.9 N/kg and 6.1 N/kg respectively, for the 8-kg load. These values were lower than those obtained with the hybrid approach. However, it might be explained by the standing position used to develop the regression equations.

Intervertebral efforts were consequently in the range of reported values but direct comparisons were limited as our study was the first, to the best of authors' knowledge, to compute muscle forces and intervertebral efforts in the Sorensen test posture. Finally, it must be noticed that actual intervertebral efforts in the Sorensen test posture should differ from values obtained in our study for the following reason. We have assumed that the intervertebral efforts produced by the model are those generated by the spine and the surrounding muscles, as an image of a wire cable surrounded by its sheath. It is likely that hip extensors have a non-negligible action on intervertebral efforts as well [153, 175, 176]. To find the real intervertebral efforts, it would be necessary to separate the contribution of the different elements.

Clinically speaking, it was reassuring to find that L5/L4 disc level was the most stressed. Indeed, this disc level is the first one to dysfunction in the human body with spinal degeneration [177, 178], pain and functional handicap. High risk of adjacent segment disease for this level compared to other lumbar disc levels has also been reported previously [14].

7.1.5 Sensitivity of the model

One of the greatest frustrations when computing intervertebral efforts with a musculoskeletal model is to not be able to validate the model with in-vivo measurements as the latter are almost unavailable (except for specific cases [142]). Validating the model is therefore nearly impossible. However, it is possible to study the sensitivity of the model and look at which parameters would imply the greatest modifications.

Considering intervertebral efforts as output in our study, it can be said in view of the results (see muscle forces in Fig. 7.1 and intervertebral efforts in Fig. 7.3 for F^2 , s^2 and s^3) that the model was not very sensitive to both mass and height modifications. Indeed, variability in muscle forces and intervertebral efforts for F^2 , s^2 and s^3 was very limited and the only input parameters that were modified were the anthropometric features of the

subjects. Nevertheless, lumbar spinal loads were showed to be linearly dependent in body height and body weight [179]. The homogeneity of the sample of subjects for the experiment in this study might also explained the non-sensitivity of the model to anthropometric measurements.

However, the muscle recruitment strategies (obtained by hybrid approach and optimisation) seemed to produce varied intervertebral efforts with a notable difference between methods which recruited the QL muscle and these which did not. The model was therefore sensitive to recruited muscles and forces they produced; this aspect will be discussed later in Section 10.2. More precisely, adding recruitment of a muscle such as QL produced different intervertebral efforts. On the contrary, despite different magnitudes in fascicles of a same muscle, one observed similar intervertebral efforts. For example, while s^2 and s^3 produced different magnitudes in fascicles of LP and TP muscles, intervertebral efforts were relatively identical for both methods.

It is therefore useful to study what happens if a muscle is omitted from the model or if its effect is overestimated compared to the other muscles. More broadly, in the case of a spine surgery such as a spinal fusion [43, 44], what would be the limit for each intervertebral effort at each IVD with respect to muscle recruitment? This finding is not insignificant as long as, from a muscular point of view, the participation of QL in the movements of the back represents a real question on behalf of the physiotherapists; question to which the present study tries to answer, in the most quantitative way possible.

7.2 Application to one subject producing three different muscle strategies

In the previous section, the hybrid approach showed promising results to highlight individual muscle strategies compared to optimisation-based solutions. However, the muscle redundancy problem can also occur when one subject produces different muscle strategies for the same static configuration. In order to distinguish the different muscle strategies and their subsequent intervertebral efforts, one additional experiment was necessary. The latter aimed at showing how EMG was relevant as input for the quantification of muscle forces, especially in the perspective of solving the muscle redundancy problem. Whereas the mathematical optimisation cannot see any differences in muscle forces, it is expected that the solutions from the

hybrid approach reflect the muscle strategies better.

7.2.1 Experiment

One 28-year-old male subject (175cm, 90kg) participated voluntarily in the experiment. The subject was an experienced physiotherapist with a good motor control. He was therefore able to produce three different muscle strategies in the Sorensen test posture as described in Section 3.2. Lying on a table in a prone position with the trunk unsupported, he had to maintain the trunk aligned with the lower limbs while being

1. the most relaxed as possible (relaxed configuration),
2. contracted at maximum (stressed configuration) and,
3. in an intermediate level between relaxed and stressed (mid configuration).

Each configuration was performed twice: unloaded and loaded with an external mass equal to 20% of the body mass. EMG signals were recorded bilaterally on the same muscles as described in Section 3.2. The identical EMG processing was also used for rectification, filtering and smoothing. To normalise the EMG signals of back muscles, i.e. LP, TP, QL and LD, MVC in the Sorensen test posture was performed as described in Section 3.2.

However, this time, to better normalise EMG signals of abdominal muscles, i.e. RA and OE, the McQuade test was carried out [180, 181]. For that, the subject was placed in a supine position with the knees bent at 90°, feet not fixed but flat on the ground and arms across the chest. At the signal, the subject had to curl up until the inferior angle of the scapula was visible. The examiner ensured a manual resistance required to produce a MVC. Each McQuade test lasted 6 seconds and occurred three times with one-minute rest between each repetition.

MVC tests in both Sorensen test and McQuade test postures were performed before the different configurations. The EMG amplitudes of the MVC tests in the Sorensen test posture were used to normalise the EMG amplitudes of back muscles (LP, TP, QL and LD), whereas those in the McQuade test posture were used for the abdominal muscles (RA and OE).

The analysis of the EMG signals showed varied muscle strategies as expected. A first experimental result was the non- or weak activation of abdominal muscles, even for the most stressed configuration. This behavior for the abdominal muscles was verified thanks to an additional test performed during the same experiment. During this test, the subject, unloaded,

was asked to produce the stressed configuration but while recruiting the abdominal muscles. Even for this test, the abdominal muscles were weakly activated. Therefore, the abdominal muscles were not considered further in the analysis.

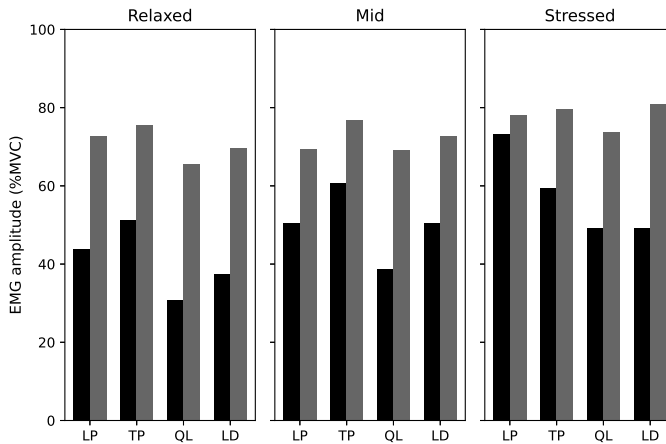


Fig. 7.6 EMG amplitudes (%MVC) of LP, TP, QL and LD for the three configurations and for both unloaded (in black) and loaded (in grey) cases.

When looking at EMG amplitudes of back muscles (Fig. 7.6), loaded cases produced obviously highest EMG amplitudes. As expected for the Sorensen test, paravertebral muscles, i.e. LP and TP were the most activated. For the relaxed and mid configurations, TP were the most activated for both unloaded and loaded cases. For the stressed configuration, LP were the most activated for the unloaded case.

LD was not expected to be as active, particularly for the loaded cases. LD produced the highest EMG amplitude among all EMG amplitudes for the loaded stressed configuration. This high activation of LD in this study may come from the bilateral role of this muscle in helping for hyperextension of the trunk [60].

When looking at the unloaded case only (black bars in Fig 7.6), one could see the increase in the EMG amplitudes, especially for LP and QL, from the relaxed to the stressed configuration. LP had EMG amplitudes of 44%MVC, 51%MVC and 73%MVC and QL 31%MVC, 39%MVC and 49%MVC for the relaxed, mid and stressed configurations respectively. TP and LD had an increase in EMG amplitudes but only between the first two

conditions. When adding load, there is no similar increase: all the muscles showed a high EMG amplitude (grey bars in Fig 7.6). The difference in EMG amplitudes for each muscle between the configurations were 11%MVC or less.

7.2.2 Model

As a consequence of the experiment and of the recorded EMG signals, the model described in Section 4.2 was slightly modified to include the LD. For that purpose, one¹ fascicle of the muscle was added bilaterally to the model, linking the pelvis to the eleventh rib. So, the new-built model was composed of 31 (29 existing + both eleventh ribs) bodies and 86 (2×42 existing + 2×1 LD) muscle fascicles (and kinematic constraints).

For the hybrid approach, the same methodology as the one described in Section 4.3 was used to design the EMG-based weights, but modified to include the LD. The force in each fascicle was still a function of the global equivalent force modulated by the EMG-based weight and the weight associated to the distribution among the fascicles. Although there was only one LD muscle fascicle, it did not change this distribution: the fascicle was assigned the weight as if the other fascicles of the LD muscle were modelled.

For the optimisation-based approach, only two cost functions were considered: s^3 and mM. It was a direct consequence of the work carried out on the sample of 19 subjects when modelling an artificial maximum loading. Compared to the optimisation-based computations described in Section 4.4, it was necessary to add the scaling of the PCSA of the LD. As performed similarly for paravertebral muscles and QL, the regression equations developed by Marras et al. [161] were used:

$$PCSA_{LD} = -0.387 + 0.041 * H * BM \quad (7.1)$$

where H is the height of the subject in m and BM his body mass in kg. The rest of the methodology was completely identical.

7.2.3 Results and discussion

Logically, the optimisation-based solutions could not distinguish the different muscle strategies. Indeed, solutions found using s^3 and mM were identical for the three configurations as there was no EMG input for the

¹The other fascicles of the LD did not attach on both sides of the S1/L5 IVD, but above. Therefore, there were not modelled as it would only rigidify an already rigid segment.

optimisation-based approach. So, in Fig. 7.7, the bars associated to the optimisation-based solutions should be represented three times but would be strictly identical. On the contrary, the hybrid approach was able to find three different solutions, one for each configuration.

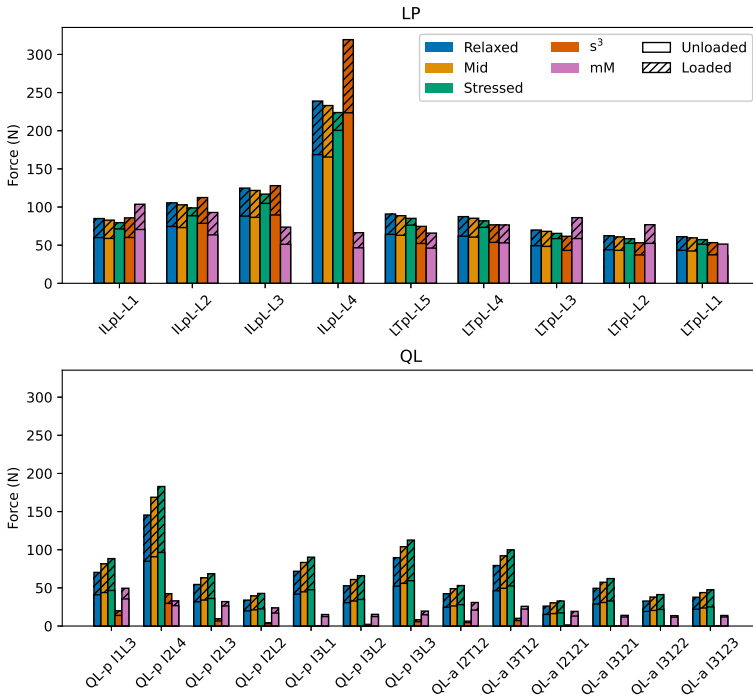


Fig. 7.7 Forces in the muscle fascicles of **LP** (Top) and **QL** (Bottom) for the three configurations, the two optimisation-based solutions and for both unloaded (non-hatched bars) and loaded (hatched bars) cases.

Muscle forces When looking closer, it can be seen that the highest forces were found in **LP**, and especially for s^3 which favoured the muscle fascicle with the greatest maximum isometric force as seen in the previous section. Another main finding was the highest recruitment of QL with the hybrid approach compared to the optimisation-based solutions. For **LP**, **TP** and **LD**, hybrid and optimisation-based approaches were closer.

When considering only the unloaded case, muscle forces for the relaxed and mid configurations were relatively close while they were higher for the

stressed configuration, in particular for **LP**. However, for the loaded case, the stressed configuration did not produce the highest muscle forces for **LP** while it did for **QL**. Furthermore, for the latter, muscle forces for the loaded case were almost twice greater than these for the unloaded case. These findings were a direct consequence of how **EMG** input was considered in the hybrid approach. Indeed, when looking back at Fig. 7.6, one could see the low difference of **LP EMG** amplitudes between the unloaded and loaded cases. For the stressed configuration, this difference was less than 10%**MVC** which resulted in a force difference of about 25 N for **ILpL-L4** muscle fascicle. The same fascicle had a difference almost three times bigger for the relaxed configuration. For the **QL**, the difference in **EMG** amplitudes was bigger and could consequently be seen in muscle forces. This increase in **EMG** amplitude when adding load was reflected by the load rate explained in Section 4.3. A small increase meant a low load rate and therefore a small difference in muscle forces. To solve this low load rate, it would probably be more appropriate to perform the non-linear regression on more points, and so adding more loads to the subject during the experiment. The hybrid approach is consequently dependent on the number of loads adding to the subject.

Intervertebral efforts As already said, the muscle forces being intermediate results in the quantification of the impacts on the spine, it was necessary to look at the intervertebral efforts. Being somehow the image of what happened at the level of muscles, there was also, as expected, no difference for the optimisation-based solutions for the different configurations (Fig. 7.8). For the hybrid approach, unlike muscle forces, the stressed configuration produced the greatest compressive forces for both unloaded and loaded cases. For shear forces and bending torques, it was only verified for the loaded case and for the lowermost lumbar levels. For the unloaded case, the three configurations produced relatively similar shear forces. Interestingly, although it was difficult to highlight an increase in **EMG** amplitudes from the relaxed to the stressed configuration for the loaded case (Fig. 7.6), it was highlighted for intervertebral efforts, even if the differences were low; from 4 to 13% difference between the relax and stressed configurations and from 2 to 5% difference between the mid and stressed configurations. Consequently, the hybrid approach seems to be an appropriate method to solve the muscle redundancy problem and to highlight differences in muscle forces and intervertebral efforts for different recorded muscle strategies.

Besides, the behaviors of intervertebral efforts underlined in the previ-

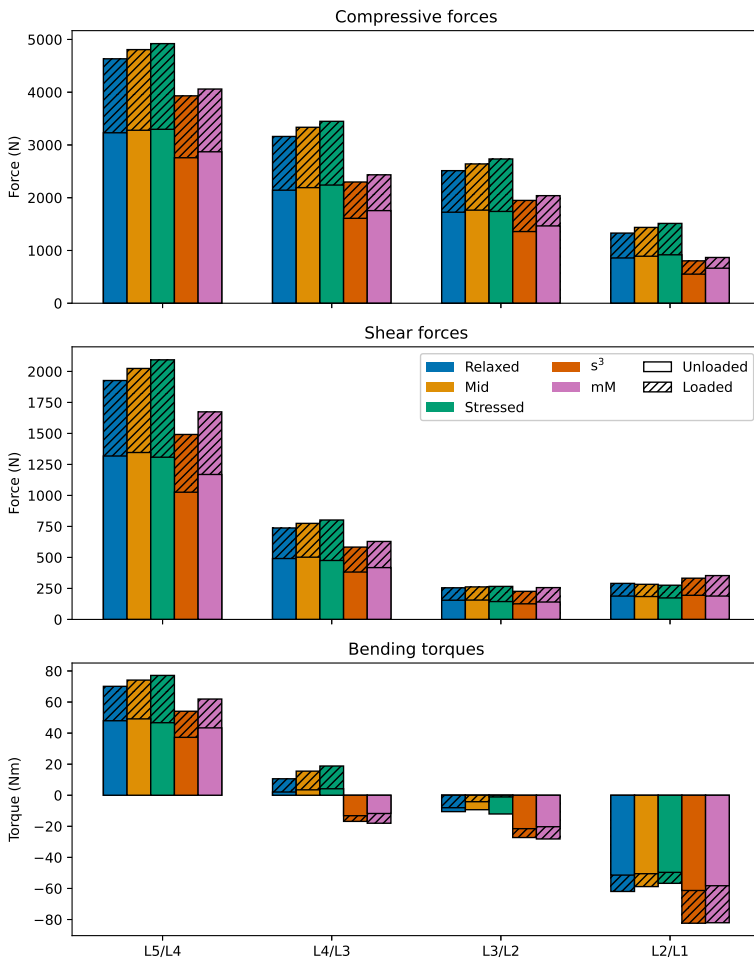


Fig. 7.8 Intervertebral efforts in the four superior lumbar discs corresponding to (Top) compressive and (Middle) shear forces and (Bottom) bending torques. Non-hatched and hatched bars represented unloaded and loaded cases respectively.

ous section for the sample of 19 subjects still held. First, both compressive and shear forces were greater for the lowermost lumbar levels, i.e. coming closer to the pelvis. Secondly, the most extreme lumbar levels showed the highest bending torques. Thirdly, adding load shifted all the intervertebral efforts towards greater values except for bending torques at the intermedi-

ate **IVD** levels. For intermediate levels, bending torques could have positive or negative small values due to the shift of the line of the bending torque. The zero-point crossing for the hybrid approach was around L3/L2 level, identified by the smaller bending torques for the loaded case compared to the unloaded case (Fig. 7.8).

Equivalent stresses Furthermore, the bending torques were associated to the greatest equivalent stresses. Indeed, the normal stress associated to the bending torque for the L5/L4 level was 3.58 MPa for the loaded case in the stressed configuration while it was 2.73 MPa and 1.16 MPa for the normal stress associated to the compressive force and the tangent stress associated to the shear force respectively. Again, the lowermost lumbar level, i.e. L5/L4 **IVD** level was the most subject to constraints [177, 178]. However, it must be noticed that both optimisation-based solutions produced their greatest stress at the uppermost lumbar **IVD**, i.e. at L2/L1 level. **s³** and **mM** had a normal stress associated to the bending torque of -3.82 MPa and -3.81 MPa respectively. Negative values indicated that the resultant bending torque at the L2/L1 level had to act 'alone' against the gravity. It could be represented as a cantilever beam. This behavior was less marked for the hybrid approach thanks to the greatest forces observed in the **QL**. This finding will be further explored later in the manuscript in Section 10.2.

Comparison with the sample of 19 subjects Due to the heavier external load, intervertebral efforts reached higher stresses compared to the sample of 19 subjects. In particular, for the shear force, the 2000N-limit mentioned previously was exceeded. As a reminder, Gallagher and Marras reported a shear limit of 2000N for a one-occurrence activity [174]. Whereas this value was not reached at all for the sample of 19 subjects, it was clearly exceeded in this case due to the heavier load, i.e. more than twice heavier than the higher external load for the sample of 19 subjects. Besides, the subject expressed a certain muscle fatigue (not evaluated) after the mid and stressed configurations in the loaded case. For the relaxed configuration for the loaded case, the shear force was slightly under the 2000N-limit.

LD To end this study, one can analyse the sensitivity of the model when adding the **LD**. To do so, a graphical comparison of the compressive force obtained with a model which included the **LD** and a model which did not is showed in Fig. 7.9. There was no clear difference between the two graphs, what could mean that the model was very little sensitive to the addition of

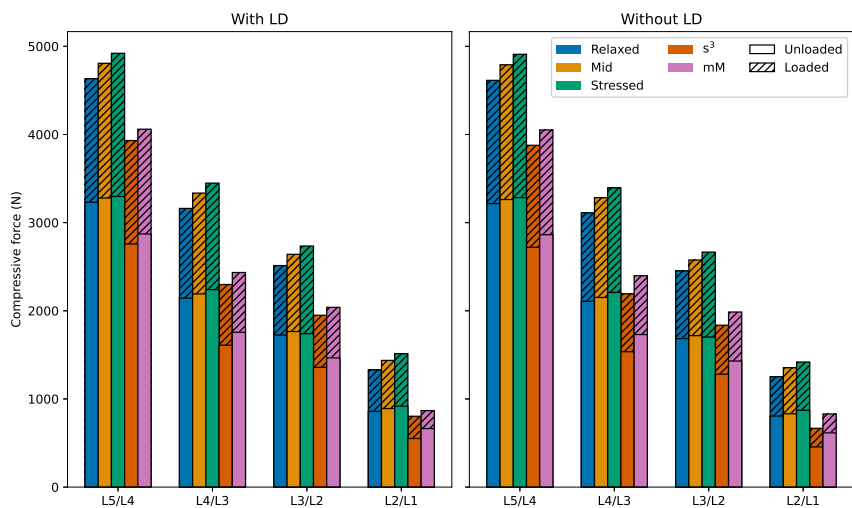


Fig. 7.9 Compressive forces computed with the model (Left) including the LD and (Right) not including the LD. Non-hatched and hatched bars represented unloaded and loaded cases respectively.

the LD muscle. However, this finding must be nuanced as only one fascicle of this muscle was modelled. It implied that its action was very moderate compared to the action of the 42 other fascicles (for one side). That being said, the linear regression used to scale the PCSA (Eq. 7.1) overestimated the PCSA compared to the PCSA of LP, TP and QL, for which the scaled PCSA were slightly underestimated. So, LD had virtually a higher capacity of generating higher force for both hybrid and optimisation-based approaches. Even so, its action had few impact on intervertebral efforts. Therefore, it does not need to be taken into account in the model of the Sorensen test posture.

7.3 Discussion

In this chapter, we explored the Hybrid-1 approach, i.e. an EMG-based deterministic distribution of a global equivalent force into muscle fascicles of the model. This approach was applied to a sample of 19 subjects, to highlight its capacity to distinguish the different muscle strategies, and to a subject who produces three different muscle strategies in the same configuration, to take into account the muscle redundancy problem.

In both applications, the hybrid approach showed potential compared with the optimisation-based approach. For the first application, the hybrid approach aimed at reproducing intervertebral efforts obtained by purely mathematical solutions. However, it had the advantage to be more representative of the muscle strategies experimentally recorded. For the second application, the hybrid approach was able to distinguish three different muscle strategies whereas optimisation-based solutions could not. These two applications illustrated that the hybrid approach, and consequently **EMG** or any experimental input related to the muscle force, was relevant to quantify muscle forces.

For the first application, it could also be added that the hybrid approach recruited all the muscle fascicles while some were disregarded by purely mathematical solutions. This finding had been previously highlighted for trunk muscles, but for an upright configuration [128]. Our results brought new perspective for trunk muscles in a prone position and the ability of our hybrid approach to find plausible solutions.

Despite the care given to experimental acquisitions and to the development of the model, there exist limitations in this work which can be spilt into two levels: those related to the **EMG**-based methodological approach and those related to the model itself.

About the **EMG**-based method, the hybrid approach was limited to superficial muscles. For the lumbar spine, there were only three muscle groups that could be measured during the experiment of the first application: **LP**, **TP** and **QL**. The latter was considered in the literature as a deep muscle but whose muscle activity might be measured under conditions [182] that we respected meticulously during the experiments. This point will be further explained in Section 10.3. On the contrary, the multifidus was not considered in this work despite studies measuring its activity superficially [32, 183, 184]. The reason was a too close distance with the L5 vertebra on palpation. As explained before, adding back muscles should impact intervertebral efforts. Using intramuscular **EMG** during the experiment to record muscle activity of deeper muscles such as multifidus among others might help to refine the model and to study its sensitivity to these muscles. One should be aware that adding intramuscular **EMG** to the protocol would most certainly make the experiment longer and more complex. In the case of hybrid approaches, surface **EMG** should be preferred to remain attractive with respect to the mathematical approaches.

For a prone back extension exercise, only **EMG** signals from back muscles were sufficiently intense to be used in the hybrid approach. However, this

method strongly depends on the studied motion. Considering trunk flexion, abdominal muscles, i.e. [RA](#) and [OE](#), should be considered as well as back muscles due to co-contraction [130]. Trunk flexion and extension are mainly visible in the lumbar spine. For lateral bending or axial rotation, the thoracic spine cannot be considered as a rigid body anymore. It would imply to add lateral muscles such as transverse abdominis or [LD](#). We showed in the second application that the model was not very sensitive to the addition of [LD](#). It might not be valid for lateral bending or axial rotation. However, our study, deliberately focused on a simple static configuration allowing to identify quite precisely the main muscles acting in it, yet revealed the interest of making the most of [EMG](#) signals in order to highlight via a [MBS](#) model the discrepancies between physiologically hybrid and optimisation-based approaches.

The model and its data had some limitations as well. Despite the intention to limit the number of muscle parameters, it was necessary to estimate [PCSA](#) and maximum isometric forces of each muscle fascicle. [PCSA](#) were approximated with linear regressions based on trunk measurements [161]. They were based on spinal levels located by palpation, the latter having a poor inter-examiner reliability [185]. The hypothesis of using linear regressions for [PCSA](#) was strong and could not exclude propagation errors due to misplacement of markers on spinal levels. Furthermore, maximum isometric forces were assumed to be linearly dependent to [PCSA](#) via [MMS](#). This second hypothesis could also generate errors as the [MMS](#) value could vary in trunk muscles [97] and could even not be a constant for all the muscles involved in the model [98]. Finally, as already said for the second application, the [PCSA](#) scaling might over- or underestimate the actual [PCSA](#), as already discussed for the [LD](#). In the frame of enlarging a generic model to a subject-specific model, one potential solution is to continue the development of open-access database for physiological muscle parameter such as [PCSA](#) [7, 8].

Another limitation was the scaling method applied in this study. The scaling was assumed to be applied globally on each body mass and dimensions. It took the assumption that all the subjects had similar proportions and mass distribution, what might be not sufficiently precise. In the frame of developing a subject-specific model for surgical planning, this limitation could lead to erroneous muscle and intervertebral efforts. To encounter this limitation, the use of radiography [42, 43, 44, 47] and compute-tomography scans [45, 46] to adapt generic models showed promising results.

Finally, in this [Hybrid-1](#) approach, there was no play for the [EMG](#)-

based weights and so on the values of muscle forces. To overcome this deterministic behavior, one step further would be to embed [EMG](#) into the optimisation process itself. This will be the second version of the hybrid ([Hybrid-2](#)) approach covered in the next chapter.

Associated publications

The results of Section [7.1](#) were presented successively at

- the 45^{ème} Congrès de la Société de Biomécanique 2020 [[186](#)], the abstract of which has been published in [[187](#)];
- the 46^{ème} Congrès de la Société de Biomécanique 2021 [[188](#)], the abstract of which has been published in [[189](#)];
- the 10th ECCOMAS Thematic Conference on Multibody Dynamics 2021 [[190](#)];
- the 47^{ème} Congrès de la Société de Biomécanique 2022 [[191](#)], the abstract of which has been published in [[192](#)].

The results of Section [7.1](#) have been fully published as a scientific contribution in Multibody System Dynamics [[193](#)].

The results of Section [7.2](#) were presented at

- the 11th ECCOMAS Thematic Conference on Multibody Dynamics 2023 [[194](#)];
- the 48^{ème} Congrès de la Société de Biomécanique 2023 [[195](#)] whose the abstract has been published in [[196](#)].

8

EMG-assisted optimisation

This chapter focuses on the [Hybrid-2](#) approach, for which [EMG](#) is embedded into the optimisation process itself. The idea behind [Hybrid-2](#) is to guide the optimisation to a more physiological result among the infinite possibilities of solutions of the muscle redundancy problem. Embedding [EMG](#) into the optimisation allows us to give more margin of choice for the solutions of muscle forces compared to [Hybrid-1](#) presented in the previous chapter.

As for the previous chapter, the present hybrid approach is tested on two samples: one composed of 19 subjects and one composed of one subject who produces three different muscle strategies. All the configurations were performed in the Sorensen test posture.

8.1 How to embed EMG?

Two ways of embedding [EMG](#) into the optimisation process have been developed in the frame of this thesis. The first way was to consider [EMG](#) as an additional constraint function to the optimisation process. The second was to modify cost functions to include [EMG](#) in the latter directly. From an implementation point of view, the question was: should [EMG](#) be included at the level of cost functions or at the level of constraint functions?

On the basis of the results of Section 7.1, only two cost functions were taken on for this work: s^3 and [mM](#). Indeed, the [Hybrid-1](#) approach gave

muscle forces close to s^3 in terms of magnitude but close to mM in terms of muscle recruitment. By embedding EMG in the optimisation process, we made the distinct following assumptions in terms of muscle forces:

- For s^3 , a muscle recruitment physiologically closer to the experiment, i.e. more recruitment of QL;
- For mM , an increase in force magnitudes of LP and therefore a decrease in these of TP and QL.

In our approach, let is point out that there is no objective to identify the most suitable cost function, but rather to analyse the impact of embedding EMG into the optimisation process.

8.1.1 EMG as constraint functions

The first approach was to consider EMG as additional constraints. In fact, it meant to add two constraint functions, in case of the three muscles LP, TP and QL. The principle underlying the EMG-based constraint functions was to express each muscle with respect to another (Fig. 8.1). In the optimisation process, muscle fascicles were gathered into muscle so that the sum of forces in one muscle was expressed as a function of the sum of forces in another muscle.

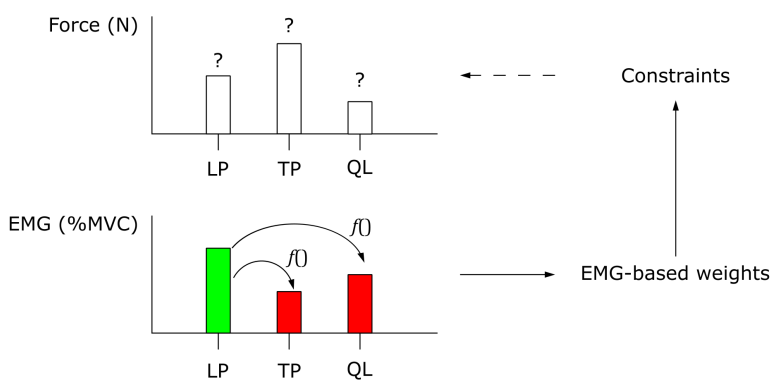


Fig. 8.1 EMG as constraint functions.

Considering LP as the reference muscle¹ for the constraint functions (Fig. 8.1), one could identify the two following constraint functions for a

¹This arbitrary choice is based on the greater muscle forces expected in this muscle.

given external load denoted m_j

$$\sum_i F_{musc,i}|_{LP,m_j} - w_{LP/TP} * \sum_i F_{musc,i}|_{TP,m_j} = 0 \quad (8.1)$$

$$\sum_i F_{musc,i}|_{LP,m_j} - w_{LP/QL} * \sum_i F_{musc,i}|_{QL,m_j} = 0 \quad (8.2)$$

with the weights $w_{LP/TP}$ and $w_{LP/QL}$ given respectively by

$$w_{LP/TP} = \frac{EMG_{LP,m_j}}{EMG_{TP,m_j}} \quad (8.3)$$

$$w_{LP/QL} = \frac{EMG_{LP,m_j}}{EMG_{QL,m_j}} \quad (8.4)$$

where EMG_{X,m_j} is the EMG amplitude of the X muscle for the external load m_j . As a reminder, EMG amplitudes were normalised EMG amplitudes, which allows to compare muscles between themselves. Besides, it must be noticed that EMG amplitudes in Equations 8.3 and 8.4 were not the EMG amplitudes experimentally measured. They were their estimates using a non-negative linear regression as performed in Section 4.3.

8.1.2 EMG included in the cost functions

The second approach consisted in modifying existing cost functions s^3 and mM to include EMG-based weights in them. Each cost function was therefore subdivided into three parts, one for each muscle to take into account the associated EMG input.

For s^3 , the modified cost function for the mass j was given by

$$\begin{aligned} \min \quad & (w_{LP,m_j} \sum_i \left(\frac{F_{musc,i}|_{LP,m_j}}{PCSA_i|_{LP,m_j}} \right)^3 + \\ & w_{TP,m_j} \sum_i \left(\frac{F_{musc,i}|_{TP,m_j}}{PCSA_i|_{TP,m_j}} \right)^3 + \\ & w_{QL,m_j} \sum_i \left(\frac{F_{musc,i}|_{QL,m_j}}{PCSA_i|_{QL,m_j}} \right)^3) \end{aligned} \quad (8.5)$$

and for mM

$$\min \max \{\Gamma_i\} \text{ for } i = 1 \dots 42 \quad (8.6)$$

$$\text{with } \Gamma_i = w_{EMG,m_j} \frac{F_{musc,i}}{F_{max,i}}$$

where i refers to the muscle fascicle and the EMG-based weight w_{EMG,m_j} is identical for all fascicles of a same muscle m_j .

In both Equations 8.5 and 8.6, the EMG-based weights were given by the following relationships

$$w_{LP,m_j} = \frac{EMG_{LP,m_j}}{EMG_{LP,m_j} + EMG_{TP,m_j} + EMG_{QL,m_j}} \quad (8.7)$$

$$w_{TP,m_j} = \frac{EMG_{TP,m_j}}{EMG_{LP,m_j} + EMG_{TP,m_j} + EMG_{QL,m_j}} \quad (8.8)$$

$$w_{QL,m_j} = \frac{EMG_{QL,m_j}}{EMG_{LP,m_j} + EMG_{TP,m_j} + EMG_{QL,m_j}} \quad (8.9)$$

where EMG_{X,m_j} is the EMG amplitude of the X muscle for the external load m_j . These relationships were similar Equations 4.20 and 4.21 described in Section 4.3 for the muscle tone and load rate factors. Again, EMG amplitudes were normalised EMG amplitudes, which allows to compare muscles between themselves. Moreover, EMG amplitudes from Equations 8.7 to Eq. 8.9 were not the EMG amplitudes experimentally measured. They were their estimates using a non-negative linear regression as performed in Section 4.3.

8.2 Application to the sample of 19 subjects

8.2.1 Muscle forces

When applying the two approaches of EMG-assisted optimisation on the 19-subject sample, we could notice (Fig. 8.2 for three fascicles and Fig. A.1 and A.2 for all the fascicles) than results of muscle forces matched with the two assumptions made at the beginning of this chapter, in comparison with the purely mathematical computation.

Indeed, for s^3 (Fig. 8.2 top), there was clearly a greater recruitment of the QL for one of the two approaches, i.e. EMG embedded as constraints. Furthermore, whereas there was no variability for s^3 with the purely mathe-

mathematical computation, there was a clear variability for the two approaches proposed to embed **EMG**, due to the variety of muscle strategies recorded experimentally from one subject to another. Finally, the increase in force for **QL** was for all fascicles (Fig. A.1).

For **mM** (Fig. 8.2 bottom), there was an increase of magnitude for **LP** for the same approach, going hand in hand with a decrease of magnitude for **TP** but not in **QL**. When looking at all the muscle fascicles (Fig. A.2, the decrease in force for **TP** was observed for almost all the fascicles. The increase for **QL** was valied for all the fascicles.

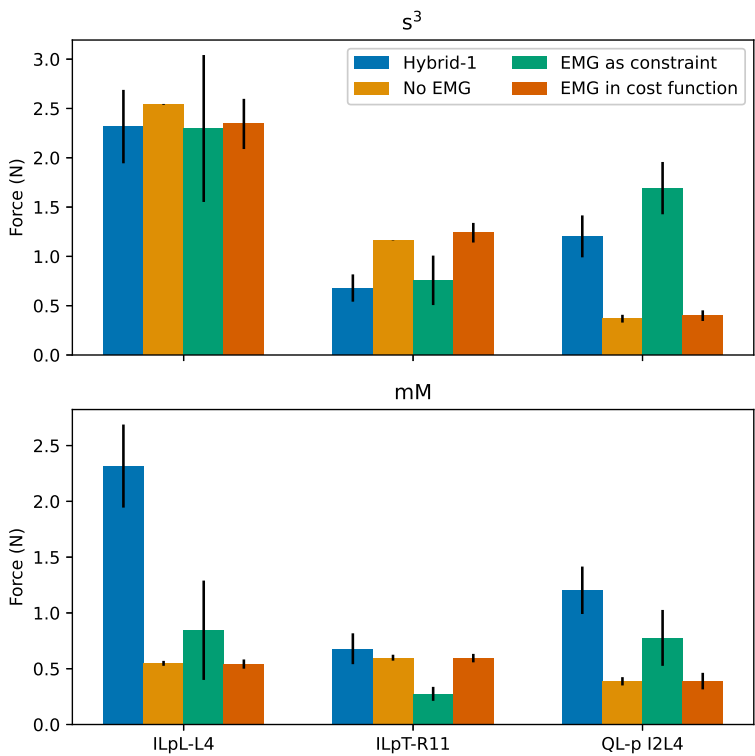


Fig. 8.2 Normalised muscle forces (N/kg) for **ILpL-L4**, **ILpT-R11** and **QL-p I2L4** obtained by the **Hybrid-1** approach, by the pure mathematical computation, by adding **EMG** as constraints and by modifying the cost function (cF) for the unloaded configuration. (Top) s^3 and (Bottom) **mM** cost functions.

When looking closer at both approaches, one could remark that making cost functions more complex to embed EMG information did not bring any relevant results, compared to the purely mathematical solutions. This finding joined the conclusion made recently for cost functions in gait modelling: the simpler the cost function was, the better the correlation between measured EMG signals and muscle activations obtained by optimisation [113].

On the other hand, considering EMG at the level of constraint functions in the optimisation process clearly helped to reflect the muscle strategies recorded experimentally (Fig. 8.2). Besides, muscle forces obtained with this approach were, for some fascicles, close to these obtained by the Hybrid-1 approach.

For s^3 , there was a high increase in QL recruitment, even higher than the magnitudes observed for mM (see the fascicle QL-p-I2L4 in Fig. 8.2). In particular, this latter fascicle was associated with a high maximum isometric force, which could explain why this fascicle had the greatest force among all fascicles in QL. LP was still the more activated muscle, with the greatest normalised muscle forces. TP however had lower muscle forces. Some TP fascicles produced almost zero force (see fascicles ILpT-R7 and LTpT-T7 and -R7 in Fig. A.1). These fascicles were not specifically associated to a smaller maximum isometric force and did not have similar insertion points. Despite that they all inserted at the T7 vertebra or the seventh rib, the first insertion points were either on the L4 or L2 vertebra. There was no real physiological explanation for this result.

For mM, there was an overall increase in LP muscle forces, but not as high as the Hybrid-1 approach (Fig. 8.2). There was also an increase in QL muscle forces, which was not intended. Indeed, as for the pure mathematical solution given with mM, some QL fascicles produced more forces than some LP fascicles, what was not completely coherent with the experiment. This pointed out that mM still aimed at activating all muscle fascicles relatively evenly. Adding EMG as constraints helped to balance a bit between muscles but not that much. TP ‘suffered’ the most with a decrease of most of its fascicles (Fig. A.2).

8.2.2 Intervertebral efforts

Intervertebral efforts obtained with the two approaches for embedding EMG were globally between those obtained by the Hybrid-1 approach (upper bound) and those obtained with the purely optimisation solution (lower bound) (Fig. 8.3).

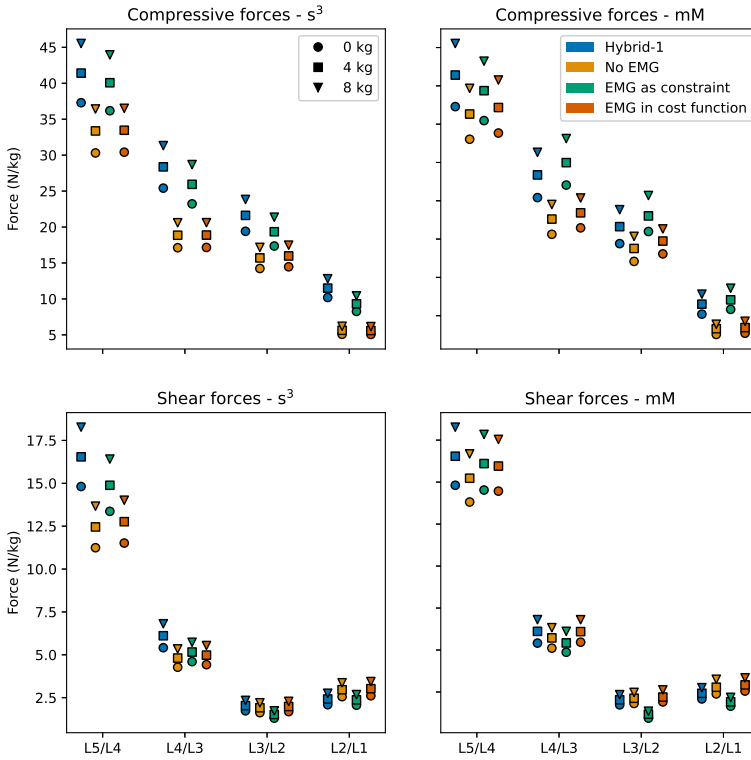


Fig. 8.3 Normalised (Top) compressive and (Bottom) shear forces (in N/kg) obtained by embedding **EMG** in (Left) s^3 and (Right) mM cost functions for 0-kg, 4-kg and 8-kg loads.

Besides, when **EMG** was taken as constraints, values of intervertebral efforts were closer to these obtained by the **Hybrid-1** approach (blue and green points in Fig. 8.3). Conversely, intervertebral efforts obtained with and without modifying the cost functions were also close to each other (orange and red points in Fig. 8.3). In Chapter 7, we concluded that the **Hybrid-1** approach was the closest to mM considering intervertebral efforts. When embedding **EMG** as constraints, there were less differences between intervertebral efforts of the two cost functions. Interestingly, adding **EMG** as constraint functions produced the lowest shear forces for the uppermost lumbar disc levels for both cost functions. The only common point between the two cost functions for this configuration was the reduced magnitudes

of TP muscle fascicles, compared to the other approaches. Reduced muscle forces in TP seemed to produce less shear forces² in the uppermost lumbar levels.

8.2.3 Conclusion

In conclusion, we have developed a Hybrid-2 approach which consists in embedding EMG into the optimisation process. Two approaches were investigated for that purpose: adding EMG as constraint functions and including EMG in cost functions.

For the application on the sample of 19 subjects illustrating the diversity of muscle strategies, adding EMG as constraint functions produced muscle forces more representative compared to the recorded EMG signals. The results were even more representative than the Hybrid-1 approach, in particular for the s^3 cost function. On the contrary, modifying the existing cost functions gave muscle forces relatively close to those obtained by purely mathematical computations.

With respect to intervertebral efforts, adding EMG as constraint functions was also more auspicious since the differences of intervertebral efforts between s^3 and mM was reduced. Previously, the Hybrid-1 approach was only similar to mM. The Hybrid-2 approach with EMG added as constraint functions looked like a good improvement of Hybrid-1 while letting the optimisation process more margin of solutions for muscle forces.

8.3 Application to one subject producing three different muscle strategies

The same strategy of embedding EMG was applied on the sample composed of a subject in the Sorensen test posture who produces three different muscle strategies, i.e. relaxed, mid and stressed (as in Section 7.2.1). As a reminder of the previous chapter, only the Hybrid-1 approach had been able to distinguish the three different strategies, whereas the optimisation-based solutions obviously had not. The Hybrid-2 approach should also distinguish the three different muscle strategies while letting the optimisation process find the solution.

²As a reminder, in the Sorensen test posture, shear forces were aligned with the gravity force.

8.3.1 Muscle forces

By looking at the muscle forces, it was clear that the Hybrid-2 approach was able to distinguish the different muscle strategies (green and red bars in Fig. 8.4) while the optimisation-based solutions were identical for the three configurations (orange bar in Fig. 8.4). Furthermore, as for the sample of 19 subjects (Section 8.2), introducing EMG as constraint had more influence than modifying the cost functions. For s^3 , it decreased the muscle force in LP fascicles while it increased the muscle force in QL fascicle (this result is showed in Fig. 8.4 for the strongest LP fascicle and the strongest QL fascicle). For mM, it increased muscle forces in both LP and QL (Fig. 8.4 right) whereas muscle forces in TP fascicles were smaller (not showed).

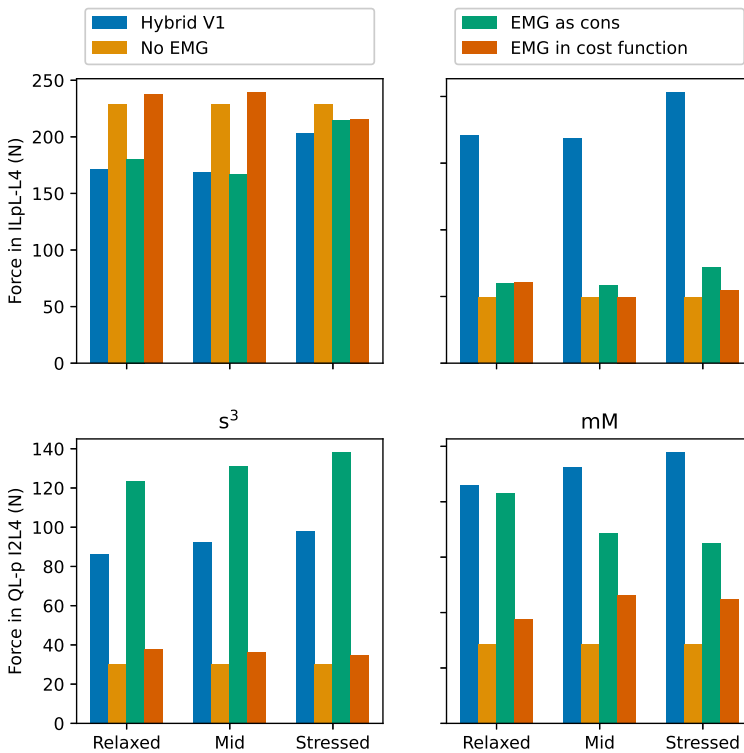


Fig. 8.4 Muscle forces in (Top) ILpL-L4 and (Bottom) QL-p I2L4 (in N) obtained by embedding EMG in (Left) s^3 and (Right) mM cost functions for the unloaded case.

When adding load (not showed), the muscle forces obtained with the [Hybrid-2](#) approach were also distinguished according to the configuration used by the subject. Muscle forces were globally obviously greater compared to the unloaded case but not for all the muscle fascicles and all the configurations. As mentioned in the previous chapter, this was due to the use of only two masses during the experiment (Section 7.2). It meant that the EMG-based weights were based on the non-negative linear regression performed on only two masses. Thus, a small difference between the two loading conditions for a muscle, even the strongest, resulted in small differences in the model for the hybrid approach. A minimum of three or four loading conditions should have been appropriate to use our hybrid approach in its best configuration.

8.3.2 Intervertebral efforts

Concerning the intervertebral efforts in this case (Fig. 8.5), we observed almost the same behavior as for the sample of 19 subjects. Intervertebral efforts obtained for the [Hybrid-2](#) approach were between those obtained with the purely mathematical solution and those obtained by the [Hybrid-1](#) approach. Also, intervertebral efforts obtained by embedding EMG as constraint functions and these obtained by [Hybrid-1](#) approach were relatively close and the highest. Conversely, the purely optimisation approach and modifying the cost function with EMG input were even closer and the lowest. However, this behavior was only observed for s^3 .

For [mM](#), modifying the cost function with EMG was closer to the solutions based on the hybrid approach rather than the purely mathematical solution (see on the right in Fig. 8.5).

Besides, for both cost functions, adding EMG as constraints functions resulted in the lowest shear forces for the two uppermost lumbar levels compared to the other methods. Finally, one could notice that there were few differences between the three different muscle strategies. This was due to the Sorensen test posture which generated already high magnitude for intervertebral efforts even for the most relax configuration. However, for the compressive forces at the L4/L3 level for [mM](#) (top right graph in Fig. 8.5), the hybrid approach with EMG embedded as constraints showed a clearer difference between the three muscle strategies.

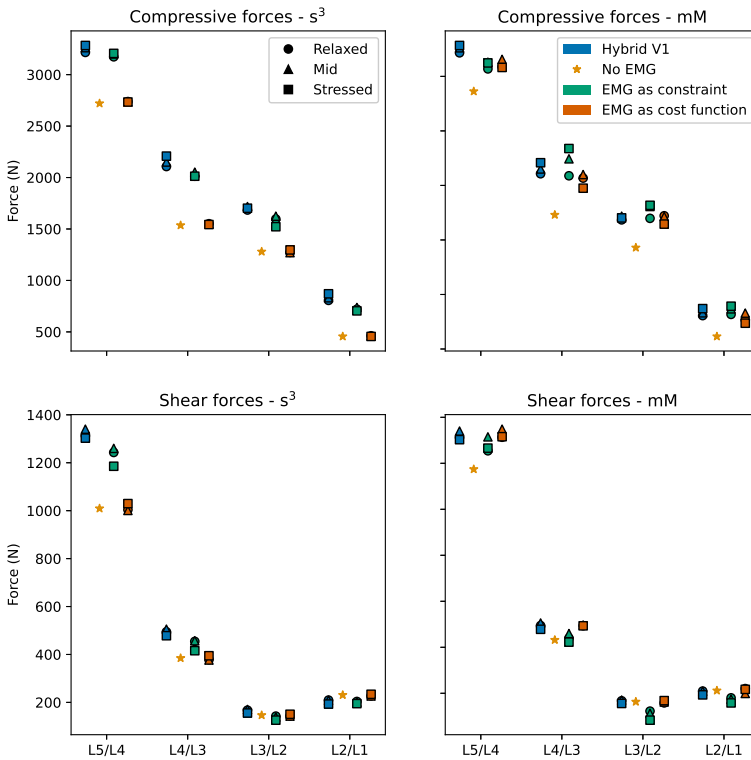


Fig. 8.5 (Top) Compressive and (Bottom) shear forces (in N) obtained by embedding EMG in (Left) s^3 and (Right) mM cost functions for the unloaded case.

8.3.3 If modelling LD

If the LD was added to the model, as in Section 7.2, muscle forces for the approach using EMG as constraint functions were not realistic. Indeed, muscle forces in LD were the greatest of all muscle forces whose values were above 600N. One could assume two origins for these unrealistic values.

Firstly, the LD may have been favoured with the EMG-based weights, due to its moderate action and the use of only two loading conditions to build up the EMG-based weights. Secondly, as already mentioned, the equation used to scale the PCSA (Eq. 7.1) overestimated the computed PCSA with respect to its initial value. Although it had few impact when using the Hybrid-1 approach, it definitely had an influence for the Hybrid-2

approach.

More generally, it would mean that if hybrid approaches had the potential to estimate muscle forces as showed in this thesis, they must be accompanied with an improvement of subject-specific model and parameters [7].

8.4 Conclusion about the EMG-assisted optimisation

Our hybrid approach with EMG embedded in the optimisation process paves the way for innovative results. Indeed, the Hybrid-1 approach showed that EMG could lead to muscle forces that were more representative of the experimental EMG signals while obtaining intervertebral efforts that were similar to the ones obtained with purely mathematical approaches. However, the Hybrid-1 approach was deterministic and had no margin on muscle forces. Thanks to the Hybrid-2 approach, the margin on muscle forces was let to the optimisation process to find the most suitable solution but while considering experimental input. For that purpose, two trails had been explored: adding EMG as constraints and modifying the existing cost function to include EMG.

The first trail gave even better results than the Hybrid-1 approach in terms of both muscle forces and intervertebral efforts. About muscle forces, for s^3 , a larger activation of the QL had been obtained which led to an overall activation of all modelled muscle fascicles, as observed previously for the EMG-assisted optimisation approach [128]. For mM, there was a decrease in TP going hand in hand with a slight increase in LP but it generated an increase in QL as well.

Embedding EMG as constraint functions produced intervertebral efforts that were closer to the Hybrid-1 approach. It also decreased the difference in intervertebral efforts between s^3 and mM whereas the Hybrid-1 approach was only similar to mM. Finally, this trail generated the lowest shear forces among all methods for the two uppermost IVD levels. This feature of the Hybrid-2 approach with EMG as constraint functions might be of interest in modelling scoliotic spines given the greater lateral shear forces that appear to the asymmetry of the posture.

The second trail, i.e. modifying the cost functions with EMG input, did not modify clearly muscle forces and intervertebral efforts compared to the purely mathematical solutions, for both s^3 and mM.

In conclusion, the Hybrid-2 approach is a relevant improvement of the Hybrid-1 approach for static configurations. Regarding the Hybrid-2

approach, **EMG** should be embedded as the level of constraint functions in the optimisation process to ensure a relevant influence of its addition.

9

Towards dynamic movements

In this chapter, the first version of the hybrid approach will be applied on a simple low-dynamics motion, i.e. a slow flexion/extension of the trunk. The aim was twofold: (i) to investigate whether the hybrid approach, studied in the previous chapters for a static configuration, could be applied to dynamic configurations, i.e. motions and (ii) to compare intervertebral efforts obtained in the flexion/extension motion with those obtained in-vivo and available in the previously mentioned open-access database, Orthoload¹.

In this chapter, a new experiment will be described although roughly analogous to the main experiment described in Chapter 3. It implied that the model changed for the static configuration in the Sorensen test posture. Also, a model dedicated to the dynamic motion was developed.

It should be noted that in this chapter, the term ‘dynamics’ refers to slow dynamics from an experimental point of view. From a modelling point of view, it means that the dynamic motion can be assimilated to a succession of quasi-equilibrium configurations.

9.1 Orthoload

Orthoload is an open-access database developed by Julius Wolff Institut from Charité Universitätsmedizin Berlin (Germany). It gathers internal

¹<https://orthoload.com/database/> (consulted on September 21st, 2023)

or joint efforts measured in-vivo thanks to implanted prostheses. Today available data is for the hip, the knee, the shoulder, the vertebral body and the internal spinal fixator. In this study, the device of interest is the vertebral body replacement, which consists in replacing a spinal functional unit, i.e. a vertebra with its two adjacent IVD. Up to now, six patients were implanted [142] to replace a vertebra which suffered from a vertebral body compression (type A of the classification proposed by Magerl et al. [197]). Once implanted, the patients were proposed to perform an important amount of various daily tasks during which the in-vivo efforts inside the replaced vertebra were recorded in real-time.

The implant used to record internal efforts is represented in Fig. 9.1 and was described by Graichen et al. [198]. To develop their instrumented implant, they modified the SYNEX implant (Stractec Medical, Oberdorf, Switzerland) to integrate a build-in circuit. The latter is able to receive external power, to compute intervertebral efforts, temperature and to send these measurements back to a receiver.

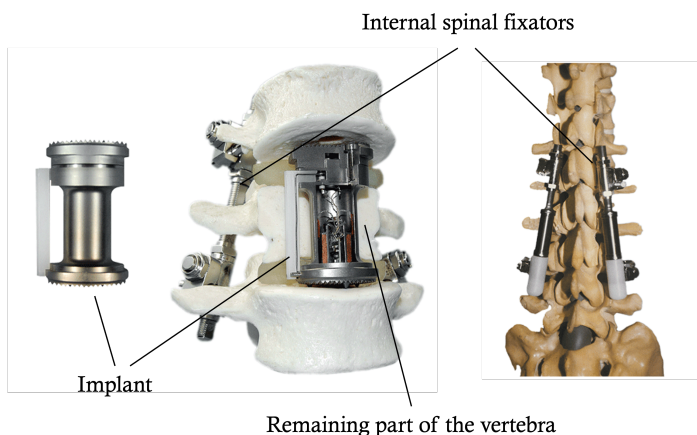


Fig. 9.1 (Left) Instrumented implant; (Middle) Its positioning in the spine to replace a functional unit; (Right) Internal spinal fixators placed dorsally to stabilise the spine. The remaining part of the vertebra can be seen in the middle figure. Adapted from [199].

As illustrated in the block-diagram in Fig. 9.2, the primary circuit is embedded in the implant. It receives by induction the necessary electrical power which is used to compute the forces and the torques according to the three axes as well as the implant temperature. These measurements are

made thanks to six strain gauges and one thermal resistor. Once the efforts and the temperature are measured, calibrated and modulated to be sent by radio transmission (using 150 MHz frequency) to the secondary circuit.

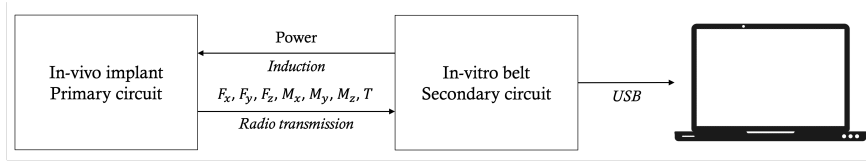


Fig. 9.2 block diagram of the implant operation during motion acquisition. *Italic* information is for communication methods.

In the case of the vertebral body implant, the secondary circuit is in-vitro on a belt placed around the trunk of the patient at the level of the implant. This circuit powers the primary circuit, receives the measurements and transmits them to the computer. It also regulates the magnetic field, produced by the power oscillator, and synchronises data previously modulated. Finally, the computer displays the measurements in real-time.

From a mechanical point of view, the vertebral body implant takes most of applied efforts but not all of them. Indeed, when looking at Fig. 9.1, one can see the internal spinal fixators. The latter are placed during the surgery to stabilise the spine dorsally. Furthermore, it can also be seen that the whole vertebra is not removed. Indeed, the most posterior part is kept. No particular reason was given for this choice [142]. It was only said that a fraction of the functional unit was removed [200]. Anyway, the retention of a fraction of the bone and IVD means that the implant does not absorb all the forces and torques. Part is supported by the internal spinal fixator and part by the remaining vertebra [142]. One study estimated by FEM simulation the percentages of axial load supported by each structure and it concluded that the implant, the spinal fixators and the remaining part of the vertebra supported respectively 82%, 12% and 6% of the axial load for a replacement of the L4 vertebra [200]. This distribution will be taken into account for our MBS comparison with the intervertebral efforts coming from the Orthoload database.

To achieve that comparison, it has been decided to study the trunk flexion/extension with a straight back when standing upright as reported in the database. This choice allowed us to recover intervertebral efforts for three trials in the Orthoload database: two performed by the subject called "wp1" and one performed by the subject called "wp2".

9.2 Experiment

The experiment was very analogous to the one described in Chapter 3 but differed at the same time as it was composed of two parts: a static configuration with three loads – called the Sorensen test case in this chapter – and a dynamic configuration with three trials – the Orthoload case.

9.2.1 Subjects

To avoid confounding factors when comparing intervertebral efforts, it was decided that our subjects had to match their anthropometric features with these from the subjects recovered in the Orthoload database. These features are summarised in Table 9.1 for all the subjects. Our subjects were with a good self-reported health, were recreationally active [201] and never had back pathology.

	This study		Orthoload	
Subject	S1	S2	wp1	wp2
Height (m)	1.77	1.82	1.68	1.69
Mass (kg)	68	63	66	74
Age (years)	60	59	62	71
Sex	M	M	M	M
Fractured vertebra	/	/	L1	L1

Table 9.1 Anthropometric features for the dynamic study. Data for the Orthoload subjects are from [142].

9.2.2 Experimental procedure

The experiment took place in a single session including explanations, warm-up, muscle palpation, placement of electrodes, MVC tests for back muscles, MVC test for abdominal muscles, LIC tasks and dynamic trunk flexion/extension.

The warm-up, the muscle palpation, the placement of electrodes and the MVC tests for back muscles were strictly identical to the ones described in Section. 3.2. As a reminder, EMG signals were recorded on back (LP, TP, QL and LD) and abdominal (RA and OE) muscles. To normalise the EMG signals of abdominal muscles, the McQuade test (Fig. 9.3) was carried out [180, 181]. For that, the subject was placed in a supine position with the knees bent at 90°, feet not fixed but flat on the ground and arms across the

chest. At the signal, the subject had to curl up until the inferior angle of the scapula was visible. The examiner ensured a manual resistance required to produce a **MVC**. Each McQuade test lasted 6 seconds and occurred three times with one-minute rest between each repetition.



Fig. 9.3 McQuade test.

As for **MVC** tests, **LIC** tests were performed in two positions as well: in the Sorensen test posture and in the McQuade test posture. For both positions, there were three trials associated with three different loads: 0%, 5% and 10% of body mass. Each **LIC** task lasted 15 sec with 1-min rest in-between.

Finally, three dynamic trials were realised, i.e. trunk flexion/extension motion with a straight back (Fig. 9.4). To limit the influence of the muscle dynamics and to keep the assumption of the muscles acting as pure force transmitters², the subjects were asked to perform the trunk flexion/extension as slow as possible. Standing upright, they had to bend forwards as far as possible while keeping their back straight and come back to their starting position. Their hands had to remain along their thighs throughout the movement. Two dynamic trials were performed with this condition. Afterwards, a last trial was performed but this time, the subjects could use a self-selected speed. The instruction was to use a spontaneous speed. This trial had for purpose to mimic as best the trials recorded in the Orthoload database. Besides, it allowed to validate if the hybrid approach was able to deal with dynamic motions while it had been developed only for static configurations. The two slow-speed motions will be averaged in the following analysis.

During the dynamic trials, a stick with two visual sensors was attached to the subjects in order to compute the flexion angle with respect to the

²and so to apply the **Hybrid-1** approach to compute muscle forces and intervertebral efforts

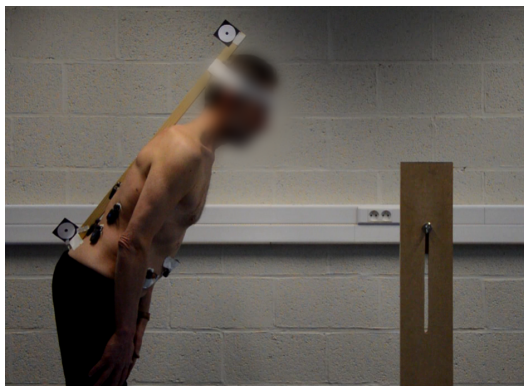


Fig. 9.4 Snapshot of the trunk/extension motion with the straight back.

vertical direction (Fig. 9.4). The stick was attached to the lumbar region and the head. All the dynamic trial have been filmed for visual analysis. The flexion angle with respect to the vertical direction was computed in post-process when analysing the images with the Tracker³ software (Open Source Physics). It was assumed that the lower sensor did not move during the whole motion. During the flexion/extension motion, the subjects received auditory feedback to keep them inform whether they had their straight back.

9.2.3 Experimental results

During the experiment and after the analysis of the EMG signals during the dynamic trials, it rapidly appeared that abdominal muscles could be reasonably neglected in the model as their EMG amplitudes did not exceed 10% of their respective MVC (see Fig. 9.5). For OE, the EMG amplitude did exceed 10% but it was not consistent for all dynamic trials. Indeed, it only appeared for one slow trial and the fast trial. It may suggest a compensatory behavior in order to stabilise and maintain the trunk in its more flexed position. In S1 (not showed), there was less activation for abdominal muscles compared to S2 (Fig. 9.5). It was therefore decided not to model the abdominal muscles.

In conclusion, in the model, only back muscles – LP, TP, QL and LD – were considered.

³<https://physlets.org/tracker/index.html>, consulted on December 12th 2023.

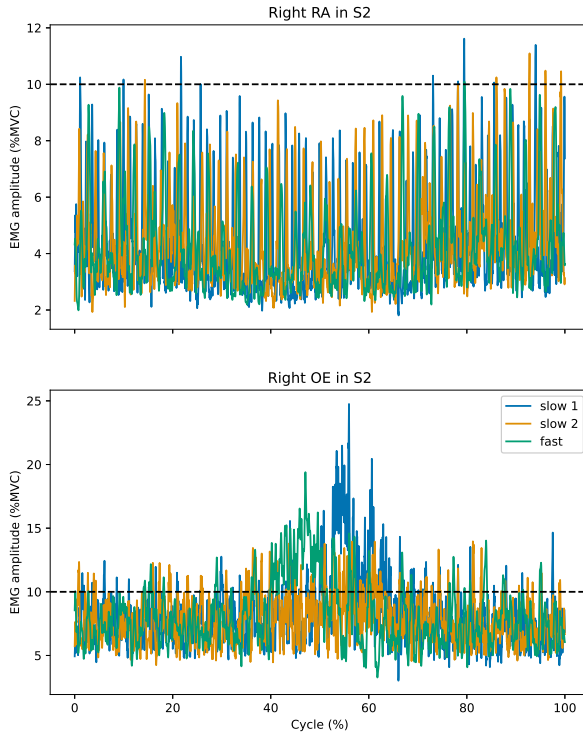


Fig. 9.5 Time evolution of the filtered and smoothed EMG signals (%MVC) of (Top) right RA and (Bottom) right OE in S2 for dynamic trials.

9.3 Model

As in Section. 7.2.2, the model described in Section 4.2 was slightly modified to include the LD. So, the new-built model was composed of 31 (29 existing + both eleventh ribs) bodies and 86 muscle fascicles (and kinematic constraints). Again, to study the sensitivity of the model to the addition of LD, a model without the LD fascicle was also developed. Each of them was used for both static and dynamic models whose numerical analyses were performed differently.

For the static model (Sorensen test case), the hybrid approach was ap-

plied as described in Section 4.3. A non-negative linear regression was performed on the three **EMG** amplitudes as function of the three loads, and that for each back muscle. This step had for purpose to compute the EMG-based weights as defined by Eq. 4.20 and 4.21. These weights were afterwards used in Eq. 4.22 to compute the force associated to each muscle fascicle. This force was function of the global equivalent force spread out into the muscle fascicles, which was solved using the inverse equilibrium numerical analysis. Intervertebral efforts were computed by constraining the motion of the associated **IVD** levels.

For the dynamic model (Orthoload case), the hybrid approach had to be modified due to the time evolution. Indeed, with time, the position changed but the **EMG** amplitudes did as well. It was therefore necessary to modify the equation of the EMG-based weights as well as the equation of the force in the muscle fascicle. For the latter, it was simply assumed that the global equivalent force in the case of a dynamic motion was no more split into a part associated to the muscle tone and another part associated to the increase of load. In other words, we came back to Equation 4.19 but which became for the dynamic case

$$F_{fa}^d = w_{fa} * w_{EMG}^d * F_{equiv}^d \quad (9.1)$$

where the exponent d indicated the dynamic case. The weight w_{fa} did not change as it described the distribution of the force between the fascicles of a same muscle and was based on maximum isometric forces reported by Bruno et al. [105]. The EMG-based weight for the dynamic case w_{EMG}^d were given by

$$w_{EMG}^d = \frac{EMG_{m,t_j}^d}{\sum_i EMG_{m_i,t_j}^d} \quad (9.2)$$

For a given muscle m and for a given time t_j of the dynamic motion, its EMG-based weight is its **EMG** amplitude divided by the sum of **EMG** amplitudes of all the muscles. The **EMG** amplitude in this specific case is the mean of the right and left **EMG** amplitudes of the muscle. Finally, the global equivalent force associated to the dynamic case F_{equiv}^d was computed for each time t_j with the inverse equilibrium numerical analysis. The dynamic motion was consequently considered as a succession of quasi-equilibrium configurations for which velocities and accelerations were assumed to be equal to 0. It was assumed that the whole trunk flexion/extension happened at the S1/L5 level, where the only independent coordinate of the **MBS** was.

9.4 Results

9.4.1 Static model - Sorensen test case

For **EMG** in the Sorensen test case, non-negative linear regressions were performed on **EMG** amplitudes as a function of loads. The regressions are illustrated in Fig. 9.6 for the four back muscles of the model. Immediately, the different muscle strategies between the two subjects appeared. Firstly, S2 tended to produce higher **EMG** amplitudes than S1. The difference was around 20% for **LP** and **TP** but it was the most important for **QL** with more than 30% of difference. Secondly, although both subjects presented similar slopes for three of the four muscles (and consequently a constant difference), there was a higher increase of **EMG** amplitudes in **LD** for heavier load in S2 compared to S1. So, whereas there was less than a 10%-difference between **EMG** amplitudes for the unloaded case, the difference reached 30% for the heavier load.

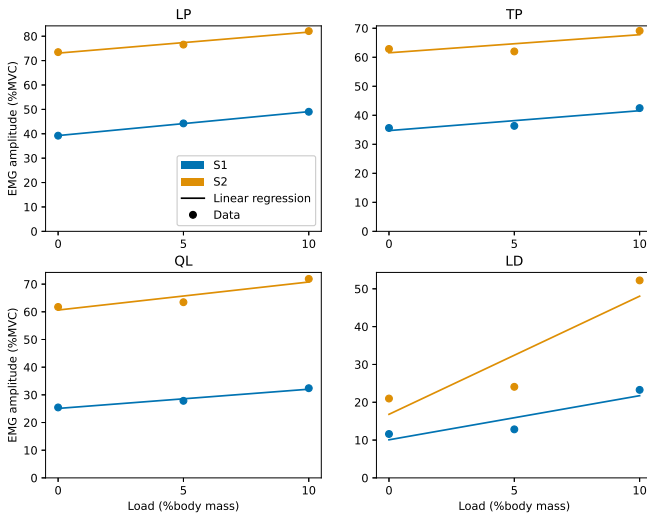


Fig. 9.6 Non-negative linear regressions of **EMG** amplitudes (%MVC) as function of the load (%body mass). Data are the mean of the right and left **EMG** amplitudes for each muscle.

Muscle forces obtained with the models (including or not the **LD** muscle) are showed for the most loaded case in Fig. 9.7. It appeared that **LD** had few

impact on muscle forces obtained for **LP**, **TP** and **QL**. Indeed, the maximum difference obtained between forces computed with or without including **LD** was 2.4% for both subjects. It has been therefore decided to present only results of the Sorensen test case computed with the model not including the **LD**.

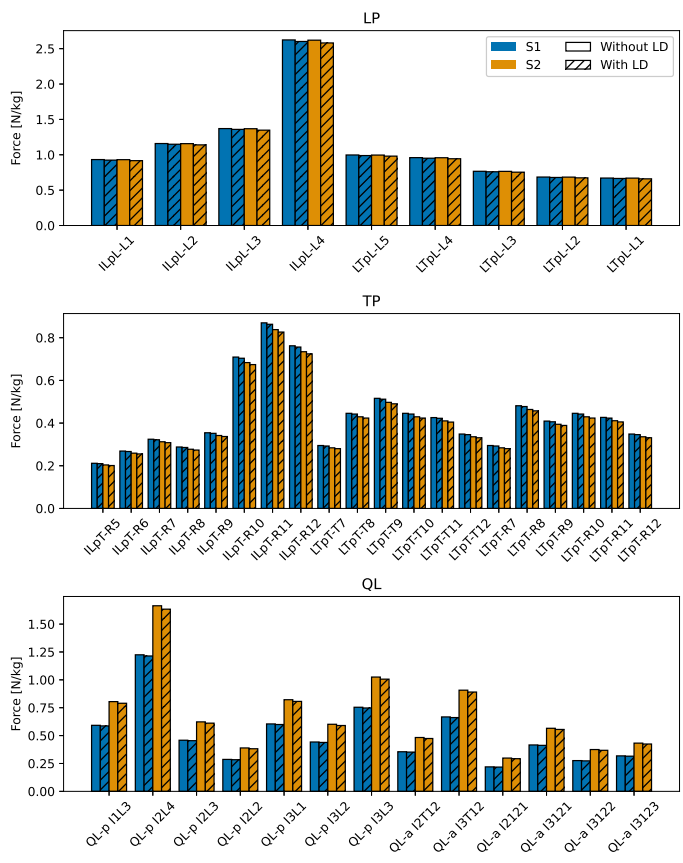


Fig. 9.7 Normalised forces in muscles fascicles for the most loaded case (10% of the body mass) computed by the model with and without **LD** for the Sorensen test case.

The differences observed with non-negative linear regressions in Fig. 9.6 were also visible in computed muscle forces (Fig. 9.7). Forces in fascicles of **LP** and **TP** were relatively close. The ‘positive’ shift in **EMG** amplitudes for S2 for these two muscles disappeared with the ratio of **EMG** amplitudes

in Eq. 4.20 and 4.21. However, for QL, there was a difference in terms of forces as there was a difference in terms of EMG amplitudes. S2 had greater forces in QL fascicles with a difference of 26% compared to S1. There was consequently different muscle strategies used by S1 and S2. Finally, for a full report despite that the LD is not considered in the following results, when LD was added to the model, there was almost twice more force in its fascicle in S2, i.e. 0.51 N/kg, compared to S1, i.e. 0.28 N/kg, which indicated also different muscle strategies for this muscle between both subjects.

Finally, let us mention that normalised muscle forces obtained with the Hybrid-1 approach for the two subjects in the Sorensen test case were relatively close to normalised muscle forces obtained for the sample of 19 subjects.

Subject	Load (%body mass)	L5/L4	L4/L3	L3/L2	L2/L1
Compressive forces (N/kg)					
S1	0	35.89	23.86	18.58	9.26
	5	39.34	26.23	20.40	10.25
	10	42.78	28.60	22.22	11.25
S2	0	38.11	26.12	20.06	10.60
	5	42.03	28.96	22.21	11.88
	10	45.96	31.80	24.36	13.15
Shear forces (N/kg)					
S1	0	14.28	5.27	1.74	2.20
	5	15.71	5.86	2.01	2.52
	10	17.14	6.44	2.28	2.84
S2	0	15.40	5.68	1.82	2.11
	5	17.09	6.36	2.11	2.42
	10	18.78	7.05	2.41	2.73
Bending torques (Nm/kg)					
S1	0	0.52	0.02	-0.15	-0.58
	5	0.57	0.03	-0.15	-0.62
	10	0.62	0.04	-0.16	-0.66
S2	0	0.58	0.08	-0.11	-0.55
	5	0.64	0.11	-0.11	-0.58
	10	0.70	0.13	-0.10	-0.61

Table 9.2 Normalised intervertebral forces (N/kg) and torques (Nm/kg) for both subjects for the Sorensen test case.

Intervertebral efforts associated with obtained muscle forces for the

Sorensen test case are showed in Fig. 9.8 with values available in Table 9.2. As for the previous studies, the greater compressive and shear forces were obtained for the lowermost lumbar levels, i.e. the closest to the pelvis. Similarly, the greater bending torques were obtained for the most extreme IVD levels, i.e. L5/L4 and L2/L1 while the intermediate levels were associated with low bending torques. Adding load produced greater forces and torques (Table 9.2).

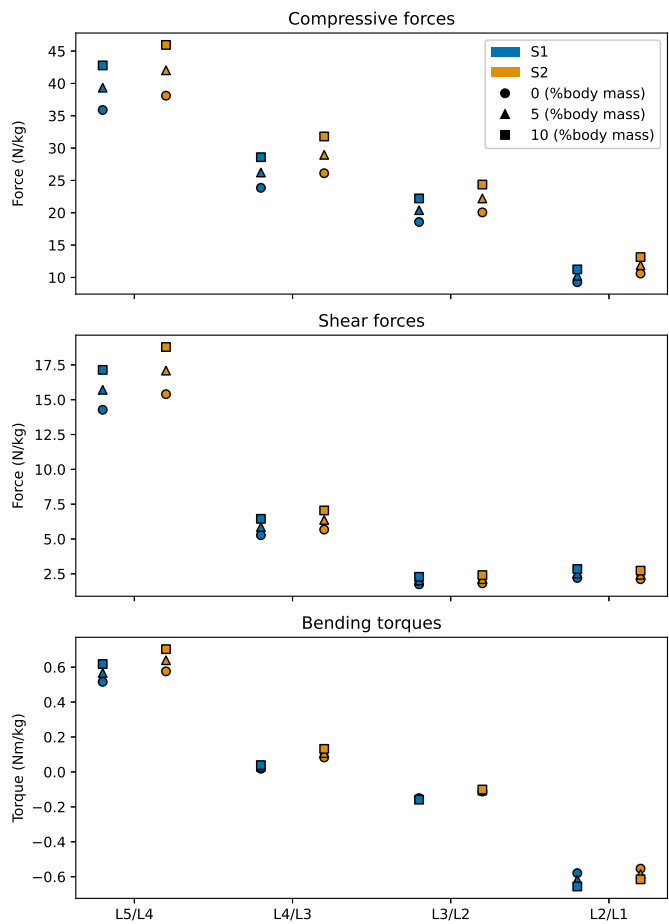


Fig. 9.8 Normalised intervertebral efforts for both subjects and all loading conditions for the Sorensen tes case (model without LD).

In terms of equivalent stresses (Table 9.3), the greatest stress was also

associated with the bending torques with 2.07 MPa obtained for S1 at L2/L1 level. The normal stresses associated with bending showed the greatest values for the extreme lumbar levels while the normal stresses associated with compressive forces were the greatest for intermediate lumbar levels. Overall, the most constrained IVD level was the L5/L4 level.

Subject	L5/L4	L4/L3	L3/L2	L2/L1
Normal stress (compressive)				
S1	1.62	1.08	0.84	0.42
S2	1.61	1.11	0.85	0.46
Tangent stress (shear)				
S1	0.65	0.24	0.09	0.11
S2	0.66	0.25	0.08	0.10
Normal stress (bending)				
S1	1.95	0.12	0.50	2.07
S2	2.06	0.39	0.33	1.80

Table 9.3 Maximum equivalent stresses (MPa) for both subjects for the Sorensen test case.

9.4.2 Dynamic model - Orthoload case

In this section, the model (including LD again) with the hybrid approach will be tested for dynamic motions with slow-speed and normal-speed (called fast for the rest of the section) trunk flexion/extension. The model and the hybrid approach will be compared at four levels: (i) the flexion angle reached, (ii) a sensitivity analysis and a validation of the model and the hybrid approach, (iii) intervertebral efforts at the different lumbar levels, and (iv) the comparison with intervertebral efforts from Orthoload database.

Flexion angle First of all, it was interesting to estimate flexion angles, with respect to the vertical direction, reached by our subjects during the experiment. This analysis did not require the use of the model and the hybrid approach: it was only the result of the videos analysis. For subjects from the Orthoload database, the flexion angles were also obtained by videos analysis with the Tracker software.

Globally, all the curves presented a certain homogeneity with the maximum flexion angles reached at mid cycle (Fig. 9.9). S1 from this study

Motion	S1	S2	Trial	wp1	wp2
Slow	78.9	56.4	1	55.6	66.3
Fast	85.8	59.6	2	51.7	/

Table 9.4 Maximum flexion angles (degrees) reached by our subjects (S1 and S2) and subjects from Orthoload database (wp1 and wp2).

reached the greatest flexion angle (Table 9.4) while S2 reached an angle similar to the angles reached by the Orthoload subjects.

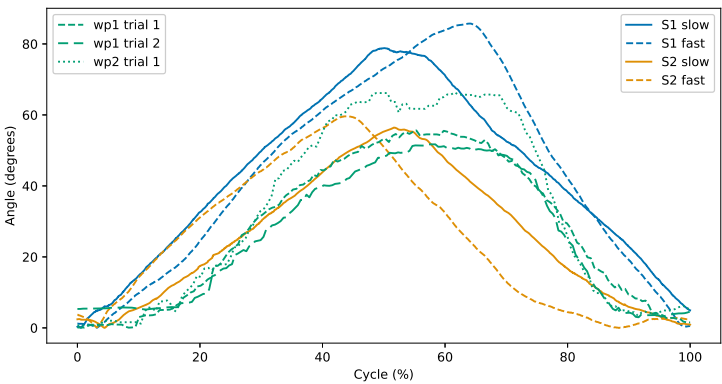


Fig. 9.9 Flexion angles (degrees) of all subjects as function of the motion cycle.

By looking closer, one could see that for slow-speed motions, both S1 and S2 reached their maximum flexion angles around 50% of the cycle. However, when increasing their execution speed, S1 reached his peak later, i.e. greater flexion period, while S2 reached his peak earlier, i.e. shorter flexion period. Besides, for Orthoload subjects, maximum flexion angles were plateaux, i.e. flattened curves.

Sensitivity and validation of the dynamic model To characterise the model, it had been decided to perform a sensitivity analysis with respect to the muscle included into the model. For that purpose, the compressive forces were used as indicators. Results are showed in Fig. 9.10. Compressive forces were normalised by the compressive force in the upright standing position, assumed to be the force at the beginning of the motion. Finally, for

the comparison, compressive forces were taken at the L2/L1 lumbar level as it was the level corresponding to the implant level in Orthoload subjects.

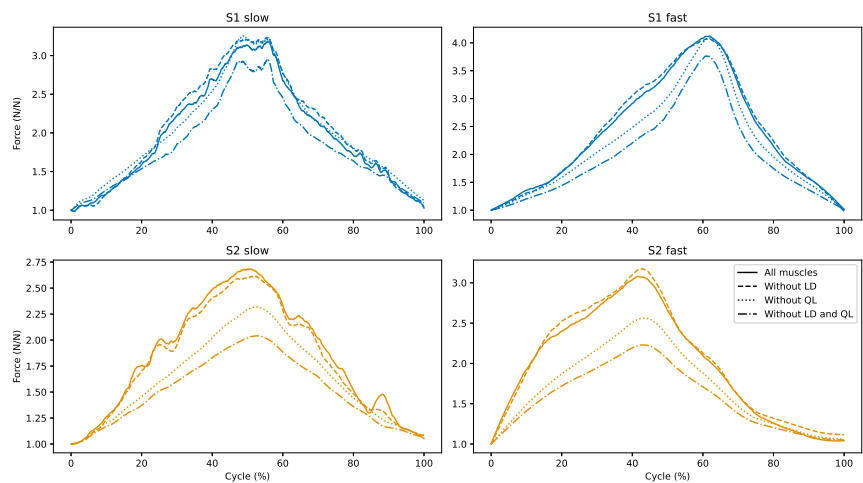


Fig. 9.10 Normalised compressive forces (N/N) as function of the muscles involved or not in the model for both subjects for the Orthoload case.

Considering the results of Fig. 9.10 and the values of normalised compressive forces in Table 9.5, the same conclusion could be drawn with respect to the presence of LD in the model. It had few impact on compressive forces so its action could be neglected for further analysis. The reason of this weak impact was the way of modelling the LD: it had only one fascicle included in the model. Moreover, its action was limited due to the distribution of the global equivalent force used in the Hybrid-1 approach.

Quite the opposite, if the QL was removed from the model, compres-

	All muscles	no LD	no QL	no LD no QL
S1 slow	3.2	3.2	3.3	2.9
S1 fast	4.1	4.1	4.1	3.8
S2 slow	2.7	2.6	2.3	2.0
S2 fast	3.1	3.2	2.6	2.2

Table 9.5 Peak normalised compressive forces (N/N) at the L2/L1 level for the Orthoload case as function of muscles involved into the model.

sive forces could be really underestimated, especially for S2 who showed greater EMG amplitudes for this muscle. QL is an important stabiliser of

the trunk [182]. Therefore, its action could not be neglected.

In conclusion, similarly to the static model, it had been decided to remove the LD muscle from the model for the rest of the analysis.

Once the impact of the muscles included into the model has been studied, the model could be validated. For that purpose, one could notice in Fig. 9.10 that on both sides of the mid-cycle, the compressive forces seemed to be their own image, in particular for the slow-speed trial (Fig. 9.10 left). Compressive forces and flexion angles should therefore be linearly correlated.

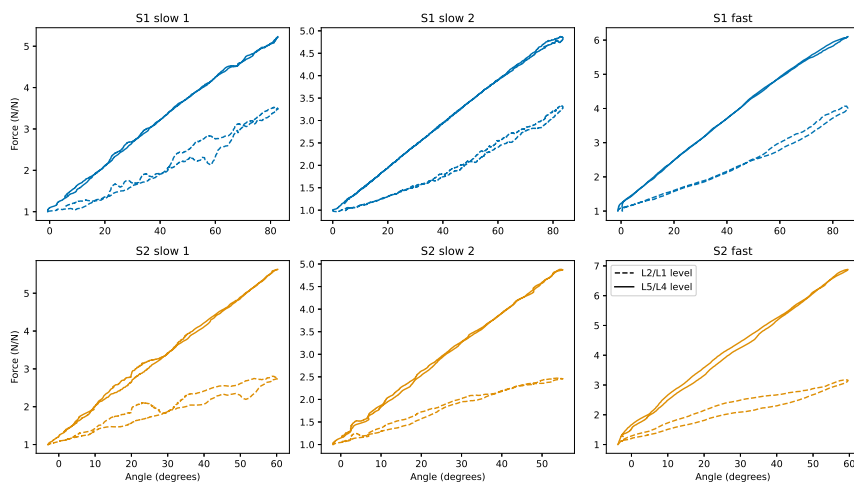


Fig. 9.11 Normalised compressive forces (N/N) at L5/L4 and L2/L1 levels as function of the flexion angle for each dynamic trial and both subjects (model without LD).

Compressive forces at the most extreme IVD levels, i.e. L5/L4 and L2/L1, seemed to be linearly related to the flexion angles (Fig. 9.11). Two levels were studied to validate this relationship. The only difference between the two levels was the greater slope and greater values observed for the L5/L4 level.

The linear relationship between the compressive force and the flexion angle had been already reported in the literature [16, 102, 109, 134, 146, 202]. Takahashi et al. proposed a linear regression based on in-vitro pressure measurements and disc areas measured by medical imaging radiography [202].

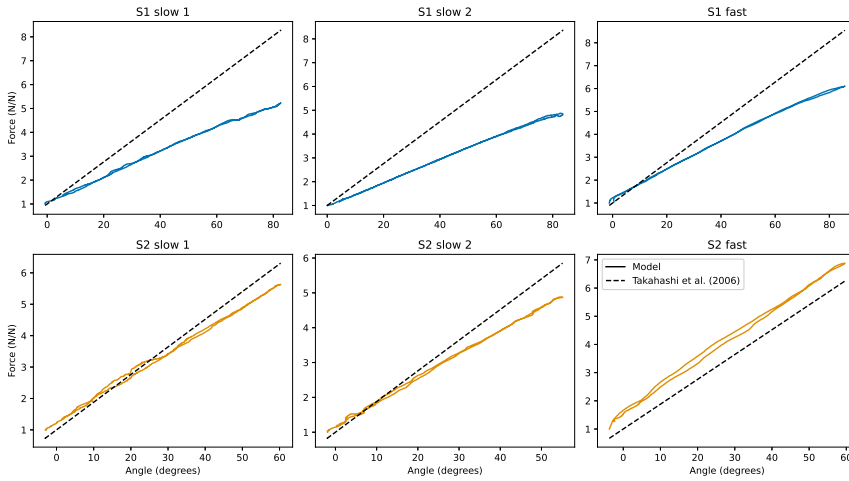


Fig. 9.12 Normalised compressive forces (N/N) at L5/L4 level from the model and from the literature [202] as function of the flexion angle for each dynamic trial and both subjects (model without LD).

They concluded that the compressive force had to be

$$force\ (N) = 64.281 * angle\ (degrees) + 730.166 \quad (9.3)$$

In Fig. 9.12, the linear regression obtained by Takahashi et al. seemed to fit relatively well the compressive forces obtained by our model, especially for S2. For S1, computed compressive forces were lower than these obtained by the linear regression. This finding was an indirect validation of the model and the hybrid approach for a dynamic case.

Besides, if there existed a linear correlation between compressive forces and flexion angles, then their maximum values should be correlated temporally (with respect to the motion cycle). For the fast dynamics, S1 had a peak of flexion angle delayed with respect to 50% cycle (Fig. 9.9). The same delay could be observed in Fig. 9.10 (Top right) for compressive forces. Similarly, S1 had an early peak flexion angle (Fig. 9.9) which was also visible for compressive forces (Fig. 9.10 bottom right).

Intervertebral efforts for the Orthoload case Extreme (maximum for positive or minimum for negative values) intervertebral efforts obtained for the dynamic motions, both slow and fast conditions, were greater than

maximum intervertebral efforts obtained for the Sorensen test case for both subjects (Table 9.6). Values for slow and fast conditions were relatively close for each intervertebral effort and each IVD level. Besides, the dynamic motions did not change the trend of the efforts among the IVD levels. Compressive and shear forces were still the highest for the lowermost lumbar levels and the bending torques were the highest for the two extreme levels. Furthermore, when comparing intervertebral efforts between themselves with their equivalent stresses (not showed), the bending torques generated the greatest stresses in three of the four lumbar levels. Only in the L3/L2 level, the greatest stress was provoked by the compressive force.

Subject	Motion	L5/L4	L4/L3	L3/L2	L2/L1
Compressive forces (N/kg)					
S1	Slow	50.66	39.26	28.72	17.82
	Fast	50.84	40.04	28.77	18.49
S2	Slow	39.82	27.28	22.38	10.84
	Fast	40.68	28.45	22.93	11.59
Shear forces (N/kg)					
S1	Slow	26.00	6.02	0.93	1.31
	Fast	25.44	5.91	1.04	1.37
S2	Slow	21.17	4.89	1.03	1.32
	Fast	21.13	4.84	1.10	1.42
Bending torques (Nm/kg)					
S1	Slow	1.02	0.63	0.27	-0.49
	Fast	1.00	0.64	0.26	-0.47
S2	Slow	0.84	0.28	0.10	-0.55
	Fast	0.84	0.31	0.10	-0.53

Table 9.6 Normalised intervertebral forces (N/kg) and torques (Nm/kg) for both subjects for the Orthoload case (model without LD).

Comparison with Orthoload subjects The last comparison for the dynamic model and the hybrid approach was the comparison with intervertebral efforts from the Orthoload database for the L2/L1 level. They are showed in absolute values (forces in N and torques in Nm) in Fig. 9.13 for the subjects from this study and the Orthoload subjects.

First, it can be observed that the bending torque measured by the implant was so small compared to bending torques from this study that it

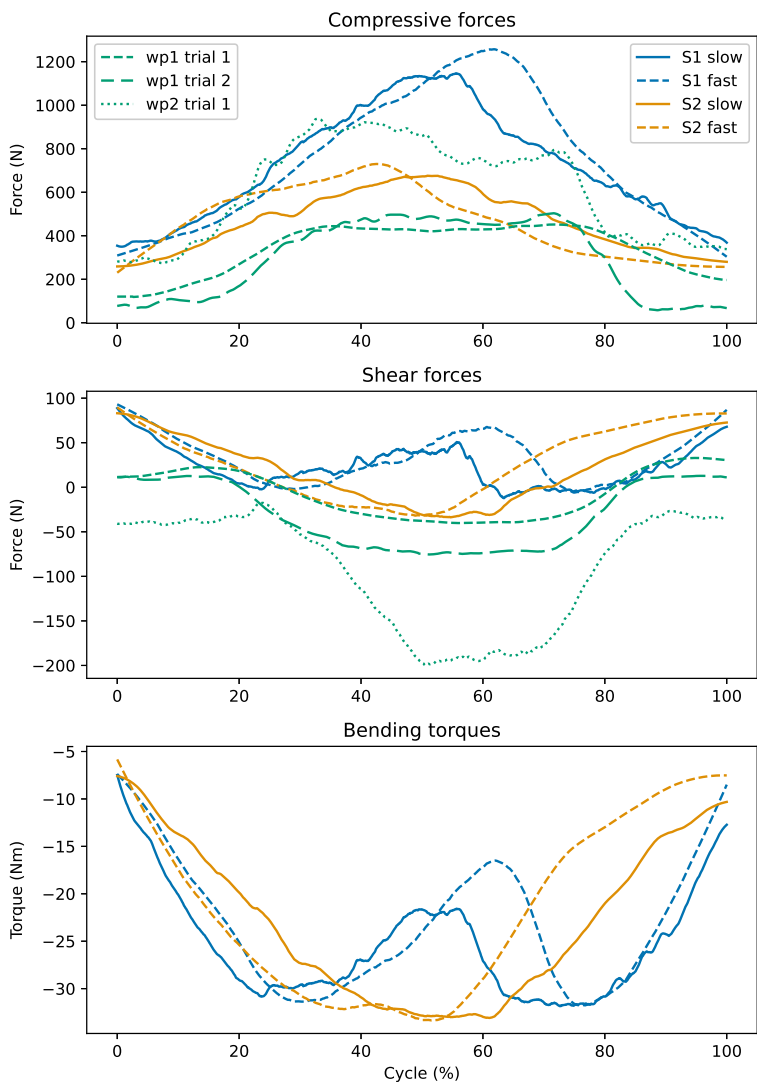


Fig. 9.13 Intervertebral efforts for all subjects and all trials (model without LD) – Comparison between this study (S1, S2) and Orthoload (wp1, wp2).

was not displayed on the graph. This difference between this study and Orthoload was not surprising as Dreischarf et al. [142], themselves, warned that the prosthesis, replacing a vertebral functional unit, and the internal

spinal fixators resulted in the fusion of several neighbouring segments. Consequently, it results in the creation of a mechanical bridge that takes up some of the efforts, and the bending torque in particular, which therefore did not act on the measuring device as it would between healthy segments.

Secondly, shear forces in Orthoload subjects were opposite to shear forces obtained with the model and the hybrid approach.

Compressive and shear forces computed by equivalent models of healthy subjects and those measured in Orthoload patients were in a similar range of magnitudes. One of the Orthoload subjects (wp1) showed the smallest compressive forces. From all subjects, S1 produced the greatest compressive forces. Furthermore, they looked similar in their curve shapes. For compressive forces, this result was relatively obvious with a bell-shaped curve, in particular for healthy subjects. There was a plateau for each curve of Orthoload subjects. Compressive forces were linearly correlated to the flexion angles whose the curves were also similar.

For shear forces, a closer look revealed almost a different strategy for each subject but with a common global trend. For healthy subjects, maximum shear forces were reached at the beginning and at the end of the cycle, i.e. in standing position. However, for S1, instead of decreasing until a minimum value (as for S2 with a negative value), it oscillated to reach a local maximum. This behavior was also visible for the bending torque. No physiological explanation has been found for this behaviour. For Orthoload subjects, the maximum values were obtained at the maximum flexion angle.

Besides, for the bending torques, S2 had its maximum value at mid cycle. The same could have been observed for S1 without the oscillatory behavior, which gave a – double – maximum value around 20-30% and 70-80% of the motion cycle.

Finally, if compressive and shear forces were normalised with respect to their values in upright standing position (Fig. 9.14), it could be noticed that the difference of force between minimum and maximum values were much higher for Orthoload subjects. For compressives forces, the Orthoload subject wp1 reached a maximum 100% greater than S1, which produced the highest compressive force between our two subjects. For shear forces, whereas there was a variation slightly greater to 1 unit (N/N) for our subjects, it was greater than 9 units for Orthoload subjects.

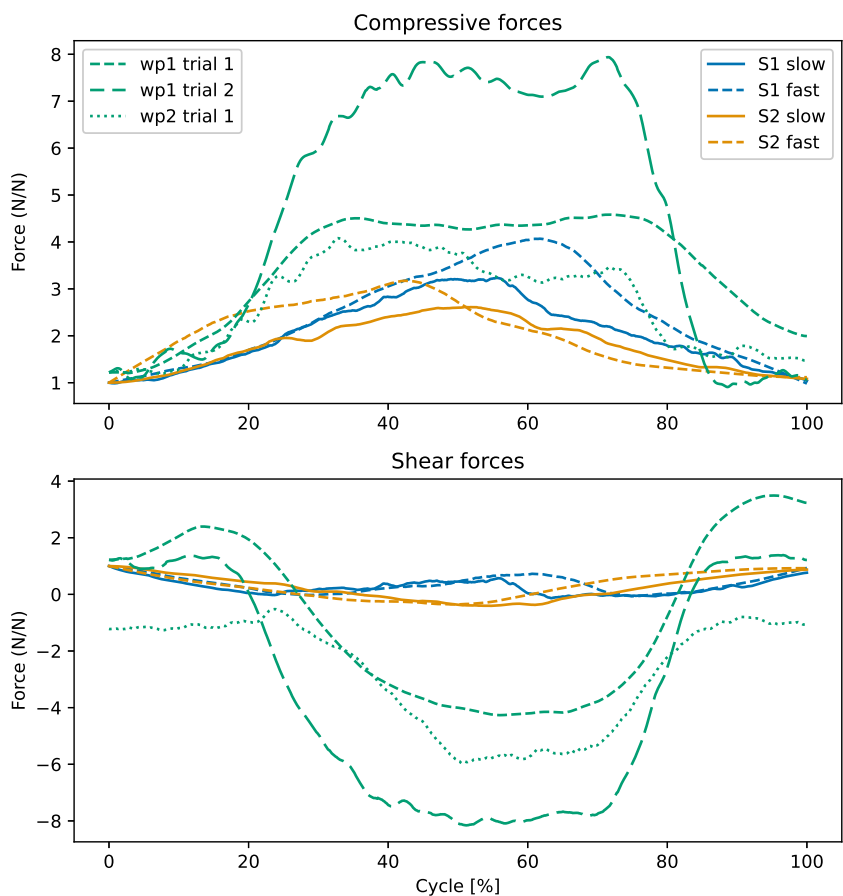


Fig. 9.14 Normalised compressive and shear forces (N/N) for all subjects and all trials (model without LD).

9.5 Discussion

9.5.1 Muscle strategies in the Sorensen test and Orthoload cases

For both static and dynamic configurations, it was obvious that both subjects did not produce the same muscle strategies (Fig. 9.6). There were two main differences. First, S1 produced smaller EMG amplitudes which could be explained by a better motor control, training and knowledge of his back muscles. Secondly, S2 generated higher EMG amplitudes of QL for heavier

load which suggested a strategy involving more this muscle, compared to S1. The impact of QL had been already pointed out in Section 7.1. Recruiting the QL more resulted in greater compressive forces, which happened in this study as well (Fig. 9.10). The variability of QL had already been reported in the literature in terms of muscle activity [203, 204] and anatomically [33]. This point will be further discussed in Chapter 10.

Finally, it has been showed that LD had few influence on LP, TP and QL muscle forces for the Sorensen test case (Fig. 9.7) and on compressive forces for the Orthoload case (Fig. 9.10). As already mentioned previously in this manuscript, only one LD fascicle was added to the model, which could explain the weak influence of taking this muscle in consideration. Besides, its action was limited due to the distribution of the global equivalent force used in the Hybrid-1 approach. Furthermore, LD acts more in hyperextension of the trunk [60] or in back extension with heavy load (as in Section 7.2.1). However, it could be observed for S2 (Fig. 9.11 and 9.12) that compressive forces at the beginning and the end of the motion were associated with negative flexion angles. There were two possible explanations: an inaccuracy in the computation of the flexion angle, which is discussed in the limitations below, or an actual trunk hyperextension, which could be consequently correlated with a greater activation of LD by S2 compared to S1.

9.5.2 Towards the validation of the hybrid approach for dynamic configurations

As a reminder, the objective of this study was to confront the hybrid approach developed in static configurations to a dynamic case. An indirect validation of the hybrid approach was proposed by modelling the trunk flexion/extension of two subjects to compare computed intervertebral efforts with these from the Orthoload database.

Before any comparison with the Orthoload database, we found a linear correlation between compressive forces and the flexion angle. Since this correlation exists in the literature as well [16, 102, 109, 134, 146, 202], it was a first validation of the model and the method, i.e. the Hybrid-1 approach. It meant that even with a relatively simple deterministic distribution of a global equivalent force into all the muscle fascicles, the method was accurate enough to compute representative compressive forces. The execution speed of the motion (slow or fast) did not influence the relationship between compressive forces and flexion angles. It would suggest that considering the back muscles as pure force transmitters, i.e. no muscle-length or muscle-

velocity relationships, could be sufficient for dynamic configurations.

Also, there was a similar trend for intervertebral efforts between static and dynamic cases. Regarding their equivalent stresses (Table 9.3 for the Sorensen test case), the bending torques were the greatest among all intervertebral efforts, in particular for the two extreme IVD levels, while the compressive forces were the highest for the L3/L2 level. The only difference was for the L4/L3 level for which compressive forces were the highest in the static case but the bending torques were the highest in the dynamic case. Besides, for both static and dynamic cases, the L5/L4 level was the most stressed level as it has been showed in Chapters 7 and 8 as well as in the literature [177, 178]. Again, it would tend to promote the use of the hybrid approach for dynamic configurations.

Talking about the comparison with the Orthoload database itself, there was some agreement between our subjects and Orthoload subjects, especially for compressive forces. The orders of magnitude were similar. The only difference was the presence of a plateau on the Orthoload curves, not present in all implanted patients [142]. The plateau was actually explained by the motion performed by the Orthoload subjects. One explanation could be that they did not keep their back exactly straight. Indeed, the flexion could be split in two, with a sagittal rotation of the pelvis first followed by a slight rotation at the level of another spine segment. Globally the back was straight – the instructions given by Orthoload staff were not too restricted [142] – but the real strategy produced a plateau on compressive forces. Dreischarf et al. corroborated this explanation in their study [142]. They also proposed other origins (not detailed here) to explain the plateau. They mainly concluded that the correlation between compressive forces and the flexion angle in implanted patients was dependent on the morphology and how the motion was performed.

In term of differences, it must be mentioned that our subjects were in good self-reported health while the Orthoload subjects were in post-operative follow-up. Therefore, it might impact the compressive forces that could be reached. Indeed, a post-operation subject should produce less muscle forces and so less compressive forces, which was admitted by Dreischarf et al. when they analysed data for the Orthoload subjects [142]. While this reduced compressive force was verified for wp1 (Fig. 9.13), it was not observed for wp2. The latter even showed greater compressive forces than S2. For our healthy subjects, compressive forces at the L2/L1 level were slightly lower compared to compressive forces available in the literature [47]. Our subject produced compressive forces between 250 and

350N and between 450 and 810N for a 0-degree and 30-degree flexion respectively, whereas compressive forces 500 and 1000N for the same angles were obtained with subject-specific models [47].

For shear forces, the comparison with Orthoload subjects was more laborious. In particular, shear forces were more intense in the Orthoload subjects and they were oriented in the opposite direction⁴. These differences might be explained by the implant itself. Indeed, forces measured by the implant were for a vertebral functional unit while the forces computed with the hybrid approach were associated with one disc level. If compressive forces could be compared, shear forces could not. It would require to compare the mean of shear forces at both L2/L1 and L1/T12 levels. Also, the Orthoload subjects could have developed an adaptive strategy following their injury, operation and rehabilitation which could result in a different distribution of intervertebral efforts.

9.5.3 Limitations and future work

The main limitation of this study was that dynamic trials were not exactly modelled as dynamic motions. They were rather considered as a succession of quasi-static equilibrium configurations for which the muscles could be assimilated to pure force transmitters. The [Hybrid-1](#) approach was used accordingly to compute muscle forces and intervertebral efforts. However, to study real dynamic configurations, it would be necessary to adapt the model and the method.

Firstly, the muscle dynamics should be taken into account in the [Hybrid-1](#) approach. For that purpose, equations such as these proposed by Granata and Marras [121] presented in Section 2.1.3 could be implemented. The impact of the muscle dynamics could thereby be analysed to estimate whether or not it should be considered in the muscle's model.

Secondly, it would be necessary to add inertia properties to all the bodies of the model. The latter were useless in this study as the model was solved as a quasi-static motion, i.e. using a succession of equilibrium conditions. It would allow to distinguish the intervertebral efforts induced by the gravity – associated with statics – and these associated with dynamics.

Another limitation was that it was assumed in the model that the sagittal rotation of the whole trunk was at the S1/L5 level, i.e. a pure rotation of the pelvis. However, it is very likely that it was a too strong hypothesis, i.e. the motion was not only induced at that level but also in the hips. It could

⁴It is not a problem of frames convention.

likely explain the difference of pattern showed by S1 for shear forces and bending torques, compared to S2. For future experimental acquisitions, it would be necessary to control this variability, either by training the subjects before recording the motion, or by using more accurate system for kinematic acquisitions such as motion capture system or inertia measurement units.

10

Analysis of the quadratus lumborum

10.1 Introduction and objectives

The QL has been regularly pointed out in this thesis for different reasons. First, it was unexpectedly activated during the Sorensen test, acting not only as a stabiliser but also as a major muscle involved in the motion – in both static and dynamic configurations. Besides, the QL was the muscle for which there was a larger diversity of muscle strategies. This muscle was ignored by all the cost functions, except mM, for the purely optimisation-based solutions. Finally, it had influence on intervertebral efforts compared to muscle strategies using only LP and TP. For all these reasons, it had been decided to go further into the analysis of the QL.

The QL (Fig. 1.5) is a deep back muscle which inserts mainly from the iliac crest to the twelfth rib and the transverse processes of the lumbar vertebrae [33, 60, 205]. It is divided into three layers: anterior, middle and posterior layers. From an anatomical point of view, QL lies mainly under the ES. This muscle is subject to high intra- and inter-individual variability in terms of number of fascicles, their prevalence and their sizes [33]. QL mainly acts during the lateral trunk flexion [203, 206, 207, 208, 209, 210]. In the literature, the QL action is qualified from modest [33] to moderate in trunk extension [206, 208, 210] as well as an important stabiliser of the trunk [182].

It has a weak activation for trunk flexion [203, 206] and axial rotation [206]. Finally, it had an accessory respiratory role [207]. Intramuscular EMG on the QL also showed that its posterior fibres, i.e. QL-p, were more active in lateral trunk flexion and trunk extension whereas the anterior fibres, QL-a, were more active only in lateral flexion [208]. The same method also highlighted that the QL was involved in the movement of the pelvis [211].

From all the studies previously cited, they were all based on EMG signals except two using the resonance magnetic imaging [209, 210]. All the results were therefore based on indirect experimental measurements which did not allow any quantification of muscle force (in N). To the best of authors' knowledge, there is no study which focused on the quantification of the forces of the QL and its influence on intervertebral efforts. The first objective of this chapter is to go deeper into the second part. Therefore, the impact of the QL on intervertebral efforts with respect to LP and TP is studied thanks a parametric study, i.e. the QL action was defined as a varying parameter in the model.

Secondly, to reinforce the interest in the hybrid approach and the use of EMG, it is necessary to validate the use of surface EMG for the QL as it is a deep back muscle. Indeed, there is only one study which reported the use of surface EMG for recording the QL activity [182] while the other used intramuscular EMG. Consequently, we designed an experimental protocol to validate the use of surface EMG for the QL.

Finally, it has been reported in the literature that the QL had important inter-individual and intra-individual anatomical variations [33]. Our third study will show that there also exists a sex-related variability for the action of the QL for LIC in the Sorensen test posture.

Each study required a dedicated experiment and/or model which is described in each section.

10.2 Impact of the QL on the lumbar spine

10.2.1 Parametric model

The model used for the present parametric study is the model described in Section 4.2, i.e. the model with LP, TP and QL muscles. The model was adapted for a 25-year-old male subject (height 1.75m and mass 78 kg). A strategy similar to the Hybrid-1 approach was used to solve the muscle redundancy problem, i.e. a global equivalent force spread out into the muscle fascicles. The more recent Hybrid-2 approach was not used as it has

been developed after this study.

The parameter of the model was the activated QL fraction from 0 to 40% with a 5% step of the global equivalent force. The upper bound was chosen arbitrarily. The combined action of LP and TP varied from 100 to 60% respectively. Contrary to the hybrid approach in which LP and TP were separated into two different muscles, each one with their respective EMG amplitudes, they were gathered into one muscle group for this study. In other words, there were no EMG-based weights in this study. The weight was the fraction from 0 to 1 describing the activation of the QL, which impacted directly the activation of the group LP-TP. Inside each group, the distribution between muscle fascicles was based on the maximum isometric forces reported by Bruno et al. [105].

Intervertebral efforts were computed to evaluate the impact of the QL.

10.2.2 Results and discussion

Results are presented in Fig. 10.1. For a given fraction of QL, compressive forces increased when coming closer to the pelvis, as mentioned in the previous studies of this thesis. It was due to the greater number of muscle fascicles in the lower lumbar region, which induced a larger force resultant.

Shear forces were mainly found at the L5/L4 level, which was the most stressed IVD level [177, 178]. For a 18-cm² disc area, equivalent shear stresses for the L5/L4 level varied from 0.48 to 0.80 MPa, which were in the range of the values reported in the literature for a flexed posture while standing [102, 134]. Shear forces were intense compared to the limits reported in the literature. Indeed, Gallagher and Marras suggested a value of 1000N shear for a task executed at a frequency less than 100 cycles a day [174]. This 1000N limit value was exceeded even without involving the QL. Intense shear forces were more due to the Sorensen posture itself than the activation of the QL. That being said, a 40% fraction of the QL resulted in a shear value of 1500N, reflecting a 50% difference compared to the value without QL involved.

Bending torques were the largest for the extreme IVD levels. They reflected how internal fibres in each IVD were loaded. For a QL fraction below 20%, the positive bending torque at the L5/L4 level indicated that IVD posterior fibres were in compression while anterior fibres were in traction (Fig. 10.2 top). The negative bending torque at all the other levels revealed the opposite trend (Fig. 10.2 bottom). Indeed, most of the muscles acted at the L5/L4 level, pulling L4 upwards (in the posterior direction).

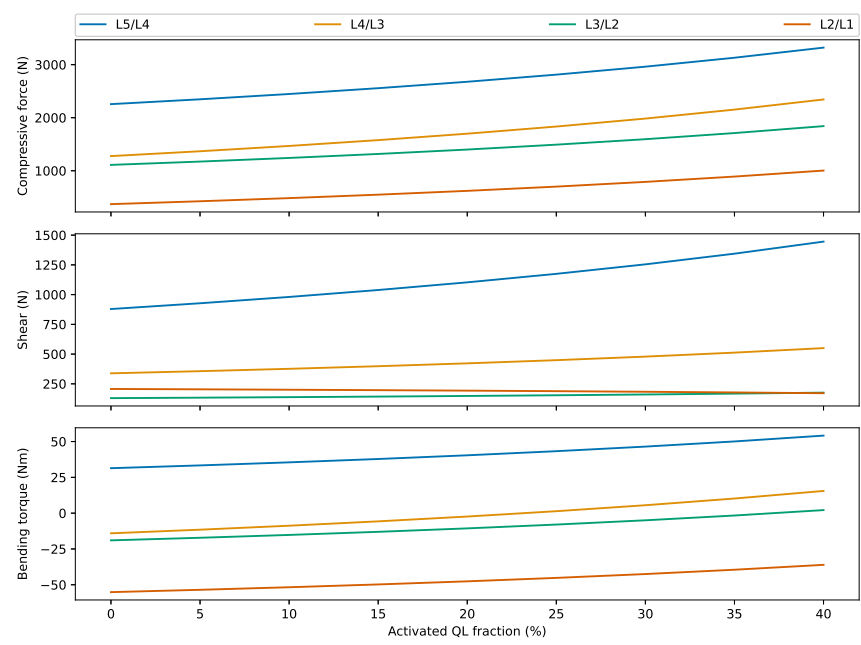


Fig. 10.1 Intervertebral efforts as function of the fraction of the activated QL.

Above L5/L4 level, each segment might be seen as a cantilevered beam.

For an increasing activated QL fraction, compressive forces increased in all IVD levels due to the larger force resultant from the QL muscle, itself due to the shorter levels arms of QL fascicles. Shear forces increased in all IVD levels, except in L2/L1. This was thought to be for the same reason as the compressive forces. Finally, bending torques decreased in cantilevered segments which resulted in less forces in IVD internal fibres. The positive bending torque at L5/L4 level was reinforced, involving greater forces in IVD fibres. For intermediate IVD levels, the bending torques crossed the zero-value at some QL fraction which indicated a modification of the role of the internal fibres.

In conclusion, the QL has an impact on on intervertebral efforts in the lumbar spine.

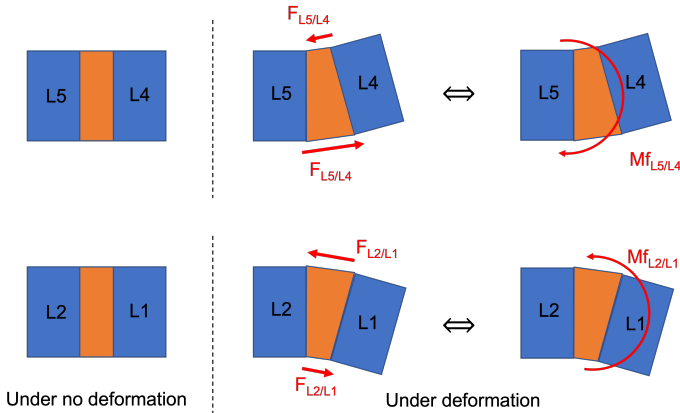


Fig. 10.2 Schematic of the role of the internal fibres of a **IVD** for the (Top) uppermost and (Bottom) lowermost lumbar levels. The resulting bending torque is represented on the right.

10.3 Surface EMG of the QL

The objective of this second study was to identify if surface **EMG** was sensible to measure the **EMG** amplitude of **QL**. In this study, following the work carried out by McGill et al. [182], surface **EMG** was used to see if we could distinguish **QL** from **LP** and **TP** muscles during four isometric contraction tests in young male subjects. Among the four configurations, two were chosen for their potential to recruit **LP** and **TP** more intensely and two for their potential to recruit the **QL** more intensely.

10.3.1 Experiment

Twenty-six participants (morphological features available in Table 10.1) voluntarily took part in the study after signing an informed consent. Inclusion criteria were being a male, being aged between 20 and 30 years old, between 1.70 and 1.90m tall, with a body mass index (**BMI**) smaller than 27 kg/m² and having nor past or current back pain neither history of back pathology, i.e. scoliosis or spondylolysis. Participants were free to stop the experiment at any time without any justification. This study was authorised by the Hospital Ethics Committee of the Université catholique de Louvain¹.

Males were chosen for this study to ensure a homogeneous sample and

¹Belgian agreement number: B403201523492

because precision and sensitivity of the palpation were showed optimised for them [212].

Feature	Males (n=26)
Age (years)	22.9±1.8
Height (m)	1.80±0.06
Mass (kg)	71.96±7.03
BMI (kg/m ²)	22.20±1.90

Table 10.1 Anthropometric data (mean±sd) of subjects.

The experiment took place in a single session including explanations, muscle localisation, warm-up, placement of electrodes, MVC tests and four specific isometric contractions.

The muscle localisation, the warm-up, the placement of electrodes and MVC tests in the Sorensen test posture were all identical to the ones presented in Section 3.2. The only difference was the selection of the muscles whose EMG activity was recorded. Only LP, TP and QL were anatomically localised for EMG acquisition. In fact, only LP and TP among the back muscles could produce crosstalk with respect to QL.

The eligibility criteria (equal lengths of lower limbs, frontal balance of the pelvis and the spine and lack of gibbosity) were also identical to the ones presented in Section 3.2. None subject was excluded after the verification of eligibility criteria.

Isometric contractions consisted in four different exercises, two (A and B in Fig. 10.3) supposed to activate more intensely LP and TP and two (C and D in Fig. 10.3) supposed to activate more intensely QL. A first test was carried out in the Sorensen test posture (Sorensen) as for the MVC tests but without linking the subject to the metal rod. Sensory feedback was set up for the Sorensen test via a rod placed at the level of the subject’s shoulder blades. By keeping contact with this rod, the subject could maintain the horizontality. The second test was the Ito test (Ito). The subject was lying (not attached) in a prone position on the table with his arms across the chest and hands on opposite shoulders. At the signal, the subject had to raise his head and trunk while keeping the arms crossed and the feet in contact with the table. The third test was the lateral Sorensen (LS) test. For that test, the subject was lying on his left side with the trunk unsupported, the lower limbs being secured as for the Sorensen test. Sensory feedback was implemented with a rod placed at the level of the left shoulder to help the subject to maintain the horizontality. The fourth and last test was the QL

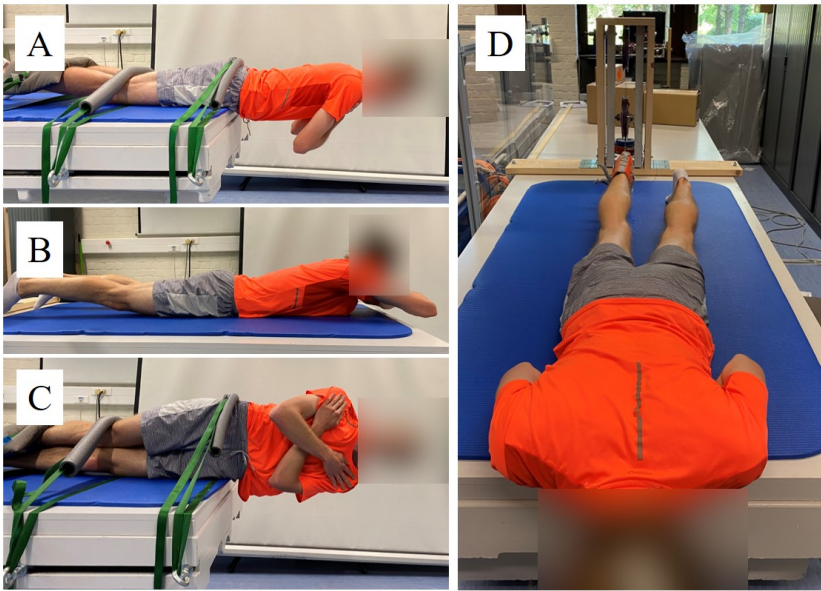


Fig. 10.3 Isometric contraction tasks (A) Sorensen test, (B) Ito test, (C) Lateral Sorensen test, (D) Functional test of the QL.

functional test (FT). This test was inspired by the test movement proposed by Kendall et al. [60] for the QL, i.e. the subject must do a lateral elevation of the pelvis while the examiner pulls on his leg to create a slight extension. To standardise the test, a two-pulley device (Fig. 10.4) was mounted to apply an external horizontal load to the leg to create a slight extension. The external load was equal to 10% of the body mass. The lateral elevation of the pelvis was explained to the subject during 5 min prior to the test. For all isometric contraction tests, the contraction time was 15 sec and auditory feedback was given by counting the time with a neutral but loud voice.

Statistical analysis was performed using SigmaPlot (version 13.0, San Jose, CA, USA). Outcomes were EMG amplitudes of each muscle and test, factors of the statistical analysis. As the sample did not follow a normal distribution, two repeated measures one-way analysis of variance (ANOVA) on ranks were carried out, one by factor (test and muscle). Statistical significance was considered for a $p\text{-value} < 0.05$.

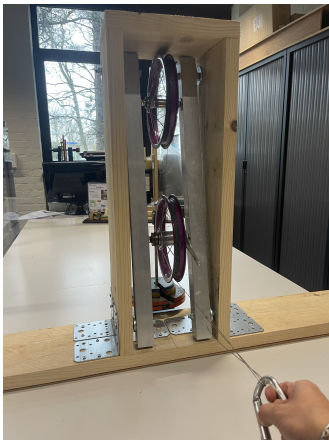


Fig. 10.4 Two-pulley device used during FT to apply the external horizontal load.

10.3.2 Results

EMG amplitudes are showed for all tests and for all muscles in Fig. 10.5.

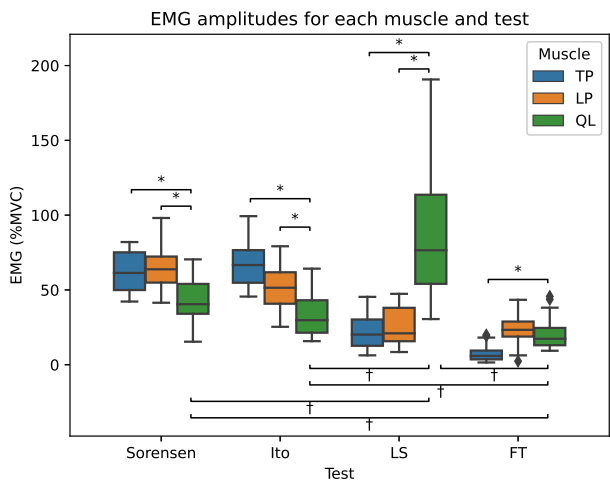


Fig. 10.5 EMG amplitudes (%MVC) for all tests and for all muscles. Diamonds denote outliers. * indicates significant p-values for the factor 'Test' and † significant values for the factor 'Muscle'.

As expected, Sorensen and Ito activated more intensely **LP** and **TP**. **LS** produced the greatest **EMG** amplitudes for the **QL**. However, more surprisingly, **FT** did not produce the highest intensity in the **QL** but in the **LP**.

Test	Median (Q25, Q75) (%MVC)			P-values	
	TP	LP	QL	LP vs QL	TP vs QL
Sorensen	61.4 (49.9, 75.1)	63.8 (54.9, 72.3)	40.5 (34.1, 54.0)	<0.001	<0.001
Ito	66.6 (54.8, 76.6)	51.5 (40.8, 61.8)	29.8 (21.4, 43.1)	<0.001	<0.001
LS	20.2 (12.7, 30.3)	21.0 (15.7, 38.0)	76.5 (54.1, 113.6)	<0.001	<0.001
FT	5.9 (3.6, 9.5)	23.3 (18.8, 28.9)	17.4 (13.0, 24.6)	0.055	<0.001

Table 10.2 Median and interquartile range (%MVC) of **EMG** amplitudes for **TP**, **LP** and **QL** and for the four tests and p-values of the repeated measures one-way **ANOVA** on ranks for the factor 'Test'. Significant p-values are in **bold**.

From Table 10.2, **EMG** amplitudes of **QL** were significantly different from **EMG** amplitudes of **LP** for all tests (p-values<0.001) except **FT** (p-value=0.055) and significantly different from **EMG** amplitudes of **TP** for all tests (p-values<0.001).

Sorensen vs Ito	0.057	Ito vs LS	0.014
Sorensen vs LS	0.003	Ito vs FT	<0.001
Sorensen vs FT	<0.001	LS vs FT	<0.001

Table 10.3 P-values of the repeated measures one-way **ANOVA** on ranks for the factor 'Muscle' for the **QL**. Significant p-values are in **bold**.

Considering **QL** only (Table 10.3), its **EMG** amplitudes were significantly different for each pair of tests except for the pair Sorensen vs Ito (p-value=0.057).

In Fig. 10.6, **EMG** amplitudes of **QL** are showed as function of **EMG** amplitudes of **LP** and **TP** respectively. **LS** clearly distinguished itself with high **EMG** amplitudes of the **QL** compared to both **LP** and **TP** and compared to the other tests (upper left of both graphs in Fig. 10.6). At the opposite, **EMG** amplitudes for Sorensen and Ito were mixing themselves but were

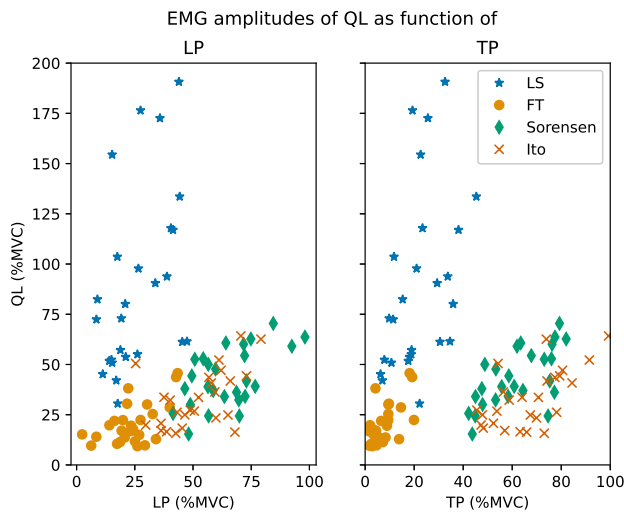


Fig. 10.6 EMG amplitudes (%MVC) of the QL as function of EMG amplitudes of (on the left) LP and (on the right) TP for the four isometric contraction tests. Each point represents a subject.

characterised by higher EMG amplitudes of LP or TP with lower EMG amplitudes of QL. Finally, for the FT, while there was a clear distinction in EMG amplitudes for TP compared to Sorensen and Ito (very low EMG amplitudes of TP in FT), the difference was less clear for LP.

10.3.3 Discussion

McGill et al. stated that deep muscle activity such as QL, psoas major and abdominal wall could be measured with surface EMG if (i) electrodes were appropriately placed and (ii) a magnitude of error was recognised, i.e. the obtained EMG amplitude is correct within a range of 15% on either side of the value [182], especially for spine modelling and clinical applications. In this study, appropriately placed surface EMG were used to measure EMG amplitudes of LP, TP and QL during four isometric contraction tests to differentiate QL from both LP and TP.

Surprisingly, FT was not good enough to differentiate QL from LP while LS was. This result was relatively opposite to what can be found in reference books for physiotherapy training: FT gave the most satisfactory differentiation while lateral trunk flexion recruited other muscles, ES and

abdominal muscles among others, than QL [60, 213]. It was believed here that LS produced much more intense EMG amplitudes than lateral trunk flexion when standing or side-lying bridge. These higher EMG amplitudes in QL gave significant differences with LP and TP for LS. Furthermore, FT does not differentiate always QL from paravertebral muscles, especially if the leg is slightly hyperextended, which could have happened in this study. Varied muscle strategies could also occur in subjects despite the training and the selected test. Finally, greater EMG amplitudes of QL were observed for heavier loads of 34 kg [203]. The smaller load used in this study, i.e. 7.2 kg on average, might explain the non-differentiation between QL and LP.

Except for FT in which no significant difference was observed between LP and QL, QL could be differentiate from TP and LP in all other tests. If for Sorensen and Ito, this differentiation could be hard to be measured with surface EMG, LS showed significant differences between QL and LP or TP. This result strongly encouraged that surface EMG could be used to differentiate QL as soon as electrodes were placed correctly [182] and that the most superficial muscles, LP here, were inhibited [203]. Considering QL, only the pair Sorensen-Ito did not produce significant differences which was somewhat reassuring as QL was expected to be far less active for these two tests.

Finally, measuring QL with surface EMG seemed reasonable as results for QL for LS were consistent with results when using intramuscular EMG to measure QL activity for the side-bridge test or for lateral trunk flexion exercises [203, 206, 207, 208]. EMG amplitudes could not directly be compared but trends were similar.

10.3.4 Limitations and future work

This study had several limitations, the main one being the non-direct comparison with intramuscular EMG for validation. This study was part of a larger project of quantifying back muscle forces using non-invasive measurements combined with musculoskeletal models. Intramuscular EMG were therefore excluded.

Crosstalk A possible crosstalk from adjacent muscles could not be excluded when recording QL activity with surface EMG. Crosstalk could not be quantified with non-invasive measurements [72] but could be reduced with a small electrode size, an inter-electrode distance smaller than 20mm, a low skin impedance and a proper electrode localisation [72, 73, 75].

Crosstalk is also reduced when **EMG** signals are normalised by isometric **MVC** [72]. This study ensured to implement all these recommendations, in particular the proper localisation of the **QL** electrode. For back muscles, a previous study showed that activity of lumbar multifidus could not be isolated from activity of **ES** with surface **EMG** [214] but these two muscles work conjointly in back extension. **QL** not being aligned with **LP** and **TP**, its activity is believed to be less sensitive to crosstalk of **LP** and **TP**.

Joint action of OE Activity of the **OE** was not measured during the experiment while its activation had been demonstrated in lateral trunk flexion [206, 208, 215]. Therefore, it cannot be stated that **EMG** amplitudes of **QL** in this study did not reflect partially **EMG** amplitudes of ipsilateral **OE**. An additional study should be necessary. The latter should implement an additional **MVC** test specific for **OE**.

Specific MVC test for QL For **LS**, **EMG** amplitudes of **QL** were far above the 100%**MVC** value. It could be simply explained by the fact that **MVC** values used for normalisation were measured in the Sorensen test posture for all the muscles. Using **LS** for standardisation of **QL** would have been more appropriate [65, 216] but having a single **MVC** position was voluntarily chosen. Despite the **EMG** amplitudes of **QL** exceeding 100%**MVC**, it did not change the conclusion than part of **QL** activation could be captured with surface **EMG** carefully located on the muscle and admitting a certain error. Using surface **EMG** for **QL** could bring high perspectives in modelling the trunk and for the development of musculoskeletal models based on **EMG** signals.

Sample composition Finally, the sample was composed only of male participants with normal **BMI** selected to facilitate the palpation of muscles. Future studies should include female subjects and participants with various **BMI**.

10.3.5 Conclusion

In conclusion, using surface **EMG** is limited for deeper muscles. However, for the **QL** which is a deep back muscle, it is believed that well-located electrodes might reflect partially the activity of this muscle.

In this study, it was showed the **QL** could be differentiated from **LP** and **TP** with well-located electrodes and by selecting intense isometric

contractions which could isolate the action of the QL with respect to LP and TP. LS was the test which could isolate QL the best.

This result brings perspectives of using surface EMG not only for superficial back muscles to develop musculoskeletal models and hybrid approaches which could reflect varied muscle strategies better.

10.4 Evidence of sex-related differences

10.4.1 Introduction

As previously mentioned in Section 3.2, the Sorensen test is commonly used to assess isometric endurance of the extensor muscles of the trunk. Until now, many studies have focused on the fatigability of the trunk muscles during the Sorensen test, highlighting a sex difference in the time-to-exhaustion values, i.e. endurance. Indeed, the holding time of the posture was significantly longer in women than men [153, 154, 156, 217].

A lot of researchers have looked into this result and suggested many hypotheses that could explain this sex-related difference. Hunter et al. proposed the idea of a difference in mechanical blood flow restriction provoked by muscle contraction [218]. Russ et al. mentioned the use of different substrates (more glycolytic pathways for men and more fat-oxidative pathways for women) [219]; this could be due to a dissimilarity in muscle composition (ratio between fibres I and fibres IIa,b, ratio between these fibres and fat cells within the muscle) and/or in muscle size between male and female [220]. Moreover, Kouzaki et al. submitted the hypothesis of neuromuscular activation patterns such as variable load sharing, i.e. when fatigue occurs, one or several muscles that were until then deactivated relay muscles previously activated [221]. Many more hypotheses concerning morphological differences between sexes were also proposed, such as the relative lower weight of the trunk [175] and lumbopelvic area size and shape variations such as greater lumbar lordosis and larger pelvis in women [153]. However, few studies [76, 100, 156] have observed whether there was a sex-related difference in the strength exerted by the trunk muscles in the Sorensen test posture and even less if there was a sex-related difference in the EMG amplitudes of the trunk muscles.

Yet, strength and motor control are very important components for the treatment as well as for the prevention of many musculoskeletal disorders including chronic LBP [222]. Therefore, the aim of this study was to investigate sex-related differences in EMG amplitudes of the back muscles – LP

and **QL** – in a young and healthy population while holding the Sorensen test posture via short-period **LIC**.

10.4.2 Materials and methods

Thirty-six participants (morphological features available in Table 10.4) voluntarily participated in this study after signing an informed consent. Inclusion criteria for this experiment were being aged between 18 and 30 years old, having a standard **BMI** (between 18.5 and 24.9 kg/m²), no past or current back pain or history of back pathology, i.e. scoliosis or spondylolysis, and being between 1.70 and 1.90m tall for males and between 1.60 and 1.80m for females. This study was authorised by the Hospital Ethics Committee of the Université catholique de Louvain².

	Females (n=17)	Males (n=19)	Two-sided t-test p-value
Age (years)	21.94±2.68	22.63±2.06	0.39
Height (m)	1.68±0.05	1.80±0.06	<0.001
Mass (kg)	63.59±5.98	73.79±8.92	<0.001
BMI (kg/m ²)	22.47±1.75	22.68±2.41	0.77
TLL ratio (-)	0.97±0.07	1.04±0.06	0.0014

Table 10.4 Anthropometric data (mean±sd) of subjects. **TLL ratio** is the normalised ratio between the trunk height and the lower limb length. Significant p-values are in **bold**.

The experiment took place in a single session on the university campus, including explanations, warm-up, muscle localisation, placement of electrodes and various tests. Two groups were formed based on sex. Both groups performed different tasks: three **MVC** tests followed by five **LIC** tasks. Materials and methods used for this study were exactly identical to the ones described in Section 3.2.

Statistical analysis was performed using the R software (version 4.0.2, Vienna, Austria). Two groups – males and females – were considered for the statistical analysis.

The primary outcome was sex-related differences in **EMG** amplitudes during **LIC** tests. So, a mixed two-way **ANOVA** was carried out to investigate the effect of sex (male vs female), load (0-8 kg) and their interaction on **EMG** amplitudes. Data used for the mixed two-way **ANOVA** are summed

²Belgian agreement number: B403201523492

(right + left) normalised **EMG** amplitudes divided by 2 for both muscles. Homoscedasticity, normality of the residuals and normality of the random effect were verified prior to the two-way **ANOVA**.

As loads for **LIC** tests were absolute masses and not relative to subject bodyweight, an additional one-way **ANOVA** was carried out on normalised loads, i.e. load divided by subject body mass, to indicate any influence of the sex factor.

The secondary outcome was the normalised force (in N/kg) measured during **MVC** tests. It was calculated as the mean of force amplitudes measured by the force sensor during **MVC** tests divided by the subject body mass. A Welch two-sample t-test — the two samples did not have the same variance — was carried out to compare the means of the relative force obtained in each group. A Pearson correlation test was carried out to study the correlation between the normalised force and the normalised ratio between the trunk height and the lower limb length (**TLL ratio**) in each group.

Statistical significance for all tests was considered for a $p\text{-value} < 0.05$.

10.4.3 Results

Males had significantly greater height ($p\text{-value} < 0.001$), mass ($p\text{-value} < 0.001$) and **TLL ratio** ($p\text{-value} = 0.0014$) compared to females (Table 10.4). Males had a **TLL ratio** greater than 1 while females had a **TLL ratio** smaller than 1. It meant that males had on average a greater trunk while females had longer lower limbs. **BMI** and age were not significantly different.

Load (kg)	LP		QL	
	M	F	M	F
0	55.32±15.18	57.86±13.16	43.81±14.18	56.14±17.24
2	59.21±17.34	62.32±13.04	46.94±17.83	58.48±15.97
4	59.57±15.42	65.96±14.54	49.64±16.12	62.98±17.21
6	65.81±17.34	73.55±14.06	54.91±19.36	70.70±16.36
8	68.19±15.78	76.32±14.75	56.82±18.68	74.83±17.29

Table 10.5 Summed **EMG** amplitude (%**MVC**) (mean±sd) of **LP** and **QL** muscles for the different loads.

Summed **EMG** amplitude are showed in Table 10.5 for both **LP** and **QL** muscles, for both sexes and for each load. From Table 10.6, no sex-related difference was found in the **LP**. However, for the **QL** muscle, a sex-related difference was found for the highest load (8 kg) with a significantly ($p = 0.02$) greater **EMG** amplitude of **QL** for females compared to males.

		LP		QL	
Sex factor					
	0kg	1.0		0.22	
	2kg	1.0		0.26	
	4kg	0.83		0.17	
	6kg	0.80		0.06	
	8kg	0.80		0.02	
Load factor		M	F	M	F
	0kg vs 2kg	0.15	0.09	0.42	0.50
	0kg vs 4kg	0.09	<0.001	0.03	0.01
	0kg vs 6kg	<0.001	<0.001	<0.001	<0.001
	0kg vs 8kg	<0.001	<0.001	<0.001	<0.001
	2kg vs 4kg	1.0	0.28	0.49	0.22
	2kg vs 6kg	0.005	<0.001	<0.001	<0.001
	2kg vs 8kg	<0.001	<0.001	<0.001	<0.001
	4kg vs 6kg	0.001	<0.001	0.06	0.003
	4kg vs 8kg	<0.001	<0.001	0.003	<0.001
	6kg vs 8kg	0.80	0.80	0.50	0.26
Interaction sex/load factor		0.04		0.13	

Table 10.6 P-values of the mixed two-way ANOVA test. For the load factor, each p-value refers to the statistical difference between the two loads given in the first row. Significant p-values are in **bold**.

Distributions of both LP and QL muscles as a function of the load and the sex are showed in Fig. 10.7 to illustrate the sex-related difference.

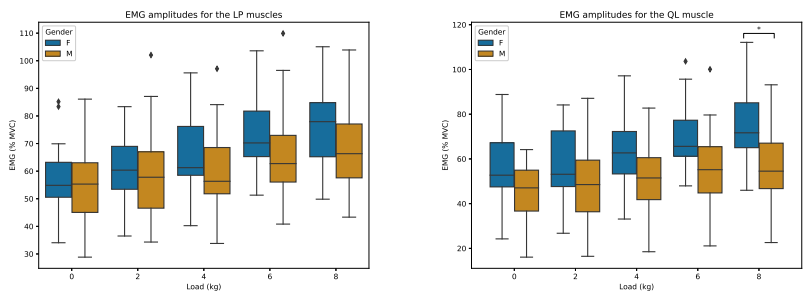


Fig. 10.7 EMG amplitudes (%MVC) of the (Left) LP and (Right) QL for both sexes and for all LIC tests. Diamonds denote outliers. * indicates significant p-values for the sex factor.

From Table 10.6, for LP, there was a significant difference between 0-kg and 6-kg loads, between 0-kg and 8-kg loads, between 2-kg and 6-kg loads, between 2-kg and 8-kg loads, between 4-kg and 6-kg loads and between 4-kg and 8-kg loads for both sexes. The only difference between males and females for LP was an additional significant difference between 0-kg and 4-kg load.

For QL (Table 10.6), for both males and females, there was a significant difference between 0-kg and 4-kg loads, between 0-kg and 6-kg loads, between 0-kg and 8-kg loads, between 2-kg and 6-kg loads, between 2-kg and 8-kg loads and between 4-kg and 8-kg loads. Again, females showed an additional significant difference compared to males between 4-kg and 6-kg loads.

Overall, for both LP and QL muscles and for both sexes, there was a significant difference between low loads (2 and 4 kg) and high loads (6 and 8 kg) but also between no external load (0 kg) and high loads.

Load (kg)	M	F	p-value
0	0±0	0±0	1.0
2	2.74±0.28	3.17±0.30	0.13
4	5.48±0.56	6.34±0.60	<0.001
6	8.23±0.84	9.51±0.89	<0.001
8	10.97±1.12	12.69±1.19	<0.001

Table 10.7 Normalised loads (%body mass) (mean±sd) and associated p-value for the sex factor. Significant p-values are in **bold**.

From Table 10.7, there were significant sex-related differences in normalised loads for 4-kg, 6-kg and 8-kg loads with p-values of 8.09e-4, 1.86e-7 and 1.6e-12, respectively. No significant differences were found for 0-kg (p-value=1) and 2-kg (p-value=0.14) loads.

Normalised loads being significantly greater for females, sex-related differences in EMG amplitudes might be truncated. However, if EMG amplitudes were drawn as a function of normalised loads and if a non-negative linear regression was made for each sex, the sex-related difference for QL (Fig. 10.8 right) was still noticeable. LP showed similar EMG amplitudes for both sexes (Fig. 10.8 left), highlighting the non-existent sex-related difference found before.

There was a significant influence of the interaction between the sex factor and the load factor for the LP muscles (p=0.04) but not for the QL muscle (p=0.13) (Table 10.6).

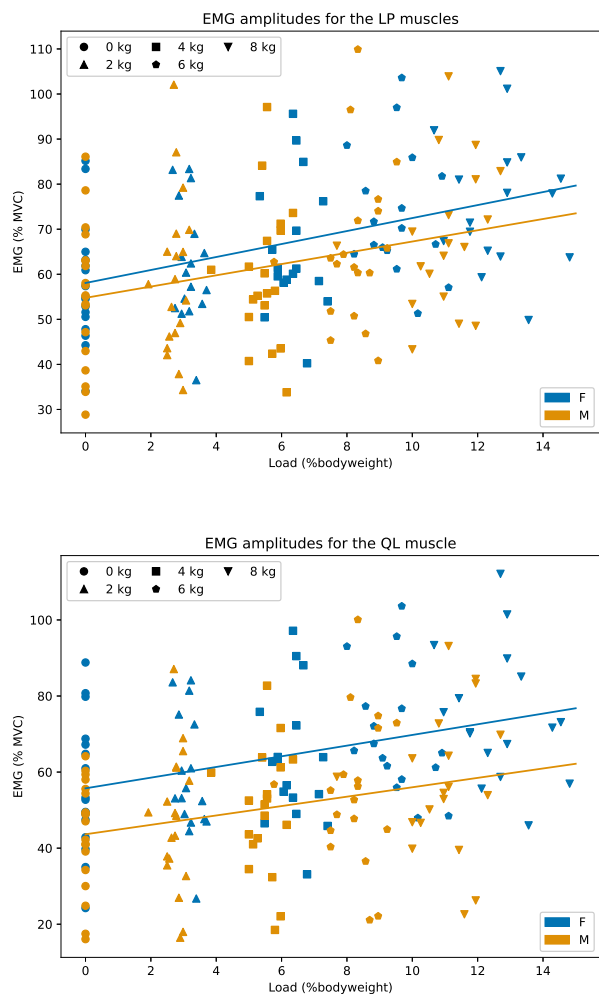


Fig. 10.8 EMG amplitudes (%MVC) of the (Left) LP and (Right) QL for both sexes and for all LIC tests as a function of normalised loads.

Finally, males showed a significantly ($p<0.001$) greater normalised force (4.25 ± 1.37 N/kg) than females (2.60 ± 0.78 N/kg). No correlation was found between the normalised force and the TLL ratio, 0.25 for females ($p\text{-value}=0.33$) and -0.16 for males ($p\text{-value}=0.51$).

10.4.4 Discussion

EMG amplitudes of LP and QL muscles in the Sorensen test posture

It was previously reported that the EMG amplitude for LP during the Sorensen test rarely exceeded 40-50%MVC [153]. EMG amplitudes in this study with no load were slightly higher with 55%MVC for males and 58%MVC for females. EMG amplitude differences might be explained by the studied motion: in this study, contractions were only measured over 15 sec compared to the 240-sec duration of the Sorensen test. Higher amplitudes (57%MVC) were reported but for a full extension of the trunk starting from a fully flexed posture [155]. The present motion could be considered as a force exertion and major active movement of the trunk [61, 223]. To find that the LP muscles were the most activated muscles was not surprising. A more unforeseen result was moderate QL EMG amplitudes especially since the QL is a deeper trunk muscle. It was still considered as a global stabilizer of the spine and so participated in the motion [223].

Influence of the sex It was showed that females had greater QL EMG amplitudes for the highest load (8kg) compared to males. It would suggest a sex-related strategy difference for higher load with females using the QL muscle more than males. Furthermore, considering normalised loads, greater QL EMG amplitudes were also visible for females for all loads and not only for the highest. This trend to use the QL muscle more might result from the sexual dimorphism of the pelvis; the latter being wider and smaller in females while narrower and taller in males [224]. Furthermore, it has been showed that females have a slight anterior pelvic tilt during gait while males have a more neutral pelvic position [224]. A greater anterior pelvic tilt in females was also possible in the Sorensen test posture. The latter would induce a greater lumbar lordosis and shorten the lever arms of both LP and QL muscles. Muscle activations would therefore be more intense in females. It might be important to take into consideration this variation of motor control strategy between sexes for rehabilitation of musculoskeletal disorders during motor control exercises of the trunk and glutes.

Males developed a significantly greater normalised force than females as already showed in the literature with subjects in the same posture [76, 100, 156], in standing posture [154] and in a semi- or fully-seated posture [225, 226]. Normalised force values in this study were however smaller for both sexes than those obtained previously in the same posture, possibly explained by protocol differences in MVC tests [156] or the use of a Roman

chair, allowing more force in the extensors with the initial spine flexion [76, 100]. About the sex-related difference in the trunk strength, differences in composition of thoracic and lumbar muscles in young males and females might explain the greater normalised force in males, also related to greater PCSA in males [161]. Mannion et al. showed that firstly, males had larger muscle fibres than females and secondly, females had a greater presence of type I fibres while males had a similar proportion of type I and II fibres [220]. Type I fibres being related to the muscle endurance, their greater presence in females might explain the smaller normalised force that they can develop. This result would still hold with respect to the greatest paravertebral muscle endurance observed in females due to the greater presence of type I fibres [153]. Males generated therefore a greater normalised force due to a higher presence of type II fibres which are more related to muscle force. This difference in fibre presence was also related to a different use of glycolysis in males and females [219].

It was hypothesised also that a greater endurance in females reflected a proportionally lighter trunk compared to males [153, 175]. This study showed no correlation between the TLL ratio and the normalised force in both sexes. The TLL ratio reflected the location of the centre of mass. A smaller ratio consequently meant that the trunk centre of mass was closer to the pelvis and resulted in a smaller torque to counteract – torque induced by the trunk weight. A smaller torque meant less resistance and so could allow to develop more forces but it was not the case as females developed less relative force. If sex-related morphological differences existed to explain the relative force, they would not result from the segment lengths but more from segment masses [175, 179]. However, women did show a significantly greater TLL ratio compared to men. Although this ratio was not related to the relative force and consequently to the muscle strength, it is believed that this ratio could have an effect on the muscle endurance.

Influence of the load For both LP and QL, significant differences in EMG amplitudes were highlighted between low (2 and 4 kg) and high (6 and 8 kg) loads and between no external load and high loads. This result might be of interest for therapeutic rehabilitation concerning the loading strategy for both reducing pain and injury recovery. Previous studies showed the importance of starting with low-loading exercises before high-loading namely to improve muscle endurance [227, 228]. The importance of the Sorensen test had also been highlighted before high-loading deadlift [229]. However, it has also been reported that the LP muscles did not show the

greater extent in muscle activity when increasing load or exercise intensity; the greater extent being noticed more in the gluteus maximus [100]. In the present work, increases in muscle activity were obtained for both **LP** and **QL** muscles with a similar extent, suggesting a potential interest of reinforcement of the trunk extensors. **LIC** in the Sorensen test posture could therefore be considered in the training process as mid-loading exercises.

Limitations and Perspectives Assessing the fatigue of the trunk extensor muscles with the Sorensen test has already been discussed previously in the literature, especially in regards to the moderate to high muscle contraction of the hip extensors [175, 176, 222]. Contraction of the hip extensors is inevitable for the Sorensen test. In the present study, no **EMG** signals were measured for the hip extensors but subjects pointed their hamstring muscles as the muscle in which they felt the most muscle work. The Ito test was proposed as an alternative for the Sorensen test to diminish the muscle contraction of the hip extensors [176]. However, the Sorensen test posture was found easier to standardise for both **MVC** and **LIC** tests. Another alternative to limit the hip contribution would be to position the subject in a semi-seated posture [225].

Assessing the muscle activity of deep muscles with surface **EMG** is discussed as it is always uncertain if the recorded activity is only from the muscle itself or is a combination of activities from several muscles, i.e. crosstalk. Particularly, the **QL** muscle is posteriorly covered by insertions of several abdominal muscles but also by paraspinal muscles. Using intramuscular **EMG** was therefore preferred for assessing its activity for various tasks [203, 206, 230]. Yet, the **QL** muscle is more superficial at the L3 level and previous work evaluated the possibility of recording partially its activity, especially for intense isometric contractions and **MVC** [182, 207]. It was also showed in the previous section with a dedicated study (see Section 10.3). Sex-related differences highlighted in this study for **QL EMG** amplitudes tended also to suggest that surface **EMG** might reflect partially **QL** activity. Future work could focus on the validity of surface **EMG** for recording the **QL** muscle activity, for various or specific tasks.

Conclusion To conclude, the present study demonstrated that there were sex-related and load-related differences for isometric contractions in the Sorensen test posture. First, for **LIC**, females did show a sex-related strategy for which greater **EMG** amplitudes of the **QL** muscle were obtained compared to males. **LP** muscles were the most activated muscles but did not

show a significant difference between both sexes. Sexual dimorphism in the pelvic area was a possible explanation for differences in motor control strategies.

Secondly, both **LP** and **QL** muscles saw their amplitudes increasing with loads with a similar extent. Clinically, the difference between low- and high-loading tasks might be of interest for rehabilitation and injury recovery. Also, the more important **QL** involvement should be considered to avoid muscle overload during rehabilitation (muscle soreness, contracture or fatigability). Indeed, most of the rehabilitation protocols for **LBP**, patellofemoral or iliotibial band syndromes use motor control exercises, as for example the single-leg deadlift exercise, which involves a good balance in both sagittal and frontal planes recruiting the **LP**, the **QL** and the glutes.

Finally, a sex-related difference in the normalised force developed by subjects during **MVC** tests was found. Males developed a greater normalised force than females which was consistent with the greater muscle endurance in females reported in the literature. Previous evidence suggested differences in morphology and in muscle composition.

10.5 Conclusions of the analysis of the QL

This chapter focused on the analysis of the **QL** at different levels.

Related to the development of the model and the hybrid approach, the parametric study reflecting the fraction of **QL** involved in the muscle strategy showed that this muscle could have an important impact on the lumbar spine. Particularly, its recruitment resulted in greater compressive and shear forces at almost all **IVD** levels. Bending torques were also impacted but it depended on the **IVD** level. This finding suggests that **QL** has to be considered into the model of the trunk when studying back extension exercises.

From an experimental point of view, it was necessary to validate that surface **EMG** could be used for the **QL** despite its anatomical position, i.e. a deep back muscle. Without this verification, it was not possible to use the **QL** in the hybrid approach, which is based on surface **EMG** signals, without ensuring the validity of the recorded signals. Our results showed that the **QL** could be distinguished from **LP** and **TP** with well-located electrodes. Therefore, the use of surface **EMG** could be considered to record the **QL** activity in a modelling perspective.

Finally, also from an experimental point of view, it was showed that **EMG** amplitudes of **QL** were different in males and in females for **LIC** in the

Sorensen test posture. Sexual dimorphism in the pelvic area was a possible explanation for differences in motor control strategies. This finding had no direct consequence in the frame of this thesis. However, it is an important perspective for future work in the frame of the development of subject-specific musculoskeletal models. To the best of the authors' knowledge, there only exists one trunk model dedicated to female subjects [231]. Bruno et al. modified their generic model [105] to adapt it for female subjects but their study focused on the causes of vertebral fracture in the thoracic spine, in particular in elderly women [231]. Considering the higher prevalence of AIS in young female subjects, it seemed important that future work focused on the development of female subject-specific model and the determination of physiological parameters for muscle quantification.

Associated publications

The results of Section 10.2 were presented at the 28th Congress of the International Society of Biomechanics 2021 [232].

The results of Section 10.4 have been published as a scientific publication [77] with co-authors other than the thesis supervisors.

PART IV

Discussion and perspectives

11

Conclusions

In this thesis, we have presented and discussed the developments achieved in this thesis to quantify back and abdominal muscle forces through [MBS](#) models and non-invasive experimental measurements. This final chapter takes a step back to recall the main developments and their contributions, to discuss them and put them in perspective for future work. Finally, an overall conclusion is drawn.

11.1 Contributions

As introduced in Part [I](#), the objectives of this thesis were associated with three distinct developments whose contributions are summarised in this section.

11.1.1 Experimental development

With regard to the quantification of muscle forces, the principle of the hybrid approach is to rely on experimental measurements, [EMG](#), to guide the optimisation process towards the most physiological solution. For this purpose, it is essential to have well thought out the design of an experiment and protocol to extract [EMG](#) signals that are sufficiently reliable to be used.

To ensure a good understanding of the quantification of back muscle forces, it was decided to limit the study to a static configuration. The

Sorensen test posture was identified as a suitable exercise as it generated intense contractions, and thus intense EMG signals, of back muscles and it required the modelling of the gravity force only. Standardisation had been ensured to allow intra- and inter-individual comparisons. It was translated into the implementation of MVC tests for EMG normalisation and the addition of visual, auditory and sensory feedback given to the subjects during the experiment.

A good quality of EMG signals comes with a good localisation of the muscles and a good EMG processing. For that reason, a particular care was dedicated to the study of a suitable filter and its associated cut-off frequency. The muscles' location and the placement of electrodes were also carefully handled.

The design of the experiment and the protocol being carefully developed, it had been extended and applied in this thesis to similar experiments taking into account few adjustments: different muscle strategies for a same static configuration, dynamic motions and comparison of different static configurations.

Finally, in order to reveal the uncertainty of using surface EMG on the QL, which is a deep back muscle, a specific study was carried out to verify that the QL could be distinguished from LP and TP.

11.1.2 Multibody modelling

A MBS musculoskeletal model of the trunk associated with the Sorensen test experiment had been developed. It derived from the fully articulated thoracolumbar spine and rib cage model proposed by Bruno et al. [105]. Our model assumed that muscles were assimilated to pure force transmitters. Muscle forces and resulting intervertebral efforts were computed via inverse equilibrium and constrained motions.

A parametric study of the activation of the QL showed that this muscle had an important impact on intervertebral efforts in the lumbar spine. This study strengthened the proposition for including the QL fascicles in the model.

11.1.3 Methodology: hybrid approach

The thesis also focused on the development of the hybrid approach for static configurations to involve the EMG signals as input in the model. This part distinguishes two complementary approaches:

- An EMG-based deterministic distribution of a global equivalent force into muscle fascicles of the model;
- An EMG-assisted optimisation in which EMG signals are added to the optimisation process.

The features of the two hybrid approaches are summarised in Table 11.1 in comparison with purely optimisation-based solutions.

	Optimisation-based	Hybrid-1	Hybrid-2
EMG as model input	No	Yes	Yes
Optimal solution (overactuation)	Yes	No	Yes
Muscle strategy differentiation	No	Yes	Yes

Table 11.1 Comparison of hybrid approaches with optimisation-based solutions.

The first approach showed promising results for static configurations to distinguish muscle strategies and to solve the muscle redundancy problem. Intervertebral efforts obtained for the lumbar spine were similar to these reported in the literature as well as these obtained with purely optimisation-based computations implementing cost functions specific to back muscles. This first approach was also tested on a dynamic case to study its limitations. Despite it was developed for static configurations, intervertebral efforts, especially compressive forces, were in good agreement with these from an open-access database. In other words, the Hybrid-1 approach, based on a relatively simple deterministic distribution of a global equivalent force into muscle fascicles, was accurate enough to compute muscle forces and intervertebral efforts. However, it could not allow a margin of play for computed muscle forces, i.e. the found solution was not the result of an optimisation computation (Table 11.1).

To give a room for calculated muscle forces, the Hybrid-2 approach was developed to embed EMG signals into the optimisation process itself. This approach was closer to the ones of EMG-assisted optimisation reported in the literature. Only two cost functions, i.e. s^3 and mM, were kept for this development. EMG signals have been implemented at two levels: (i) as constraint functions of the optimisation process and (ii) directly as a part of the cost functions. EMG embedded as constraint functions gave

the most promising results, in particular for s^3 . It aimed to recruit the QL more compared to the purely optimisation-based solution. The Hybrid-2 approach however gave unrealistic muscle forces when LD was added to the model and when too few loads were used during the experiment.

11.2 Discussion and future perspectives

Although the hybrid approach proposed in this thesis showed promising results, it had to be put into perspective given the complexity of the human body and its potential use.

Hybrid approach First of all, the hybrid approach was proposed as an alternative to purely mathematical approaches which often depend on multiple (physiological) parameters that are not always available or difficult to estimate [126]. In this thesis, by assuming that muscle acted as pure force transmitters, for static (or quasi-static) configurations, most of the physiological parameters could be ignored. However, it was still necessary to estimate two parameters: PCSA and maximum isometric forces for each muscle fascicle. For PCSA, linear regressions based on anthropometric measurements developed by Marras et al. [161] were used to estimate these of ES, QL and LD. Inside each muscle, PCSA was spread out into the muscle fascicles on the basis of the maximum isometric forces reported by the generic model of Bruno et al. [105]. Once PCSA were estimated for each muscle fascicle, maximum isometric forces were finally obtained from PCSA by a simple linear relationship via the MMS factor. To carry out these steps, it was successively assumed that (i) linear regressions proposed by Marras et al. [161] were accurate enough, (ii) ES was only composed of LP and TP, (iii) the composition of the muscle was uniform among all subjects and (iv) there existed a linear relationship between maximum isometric forces and PCSA via a (hypothesised) constant.

Although results, both muscle forces and intervertebral efforts, obtained in this thesis seemed realistic, there were some limitations. When modelling LD, its associated PCSA was overestimated which results in large unrealistic forces in this muscle. Moreover, as already mentioned previously, assuming MMS to be a constant through all the trunk muscles was showed to be limited as it may vary in trunk muscles from 40 to 130 N/cm² as function of age and back pain [97] and could even not be a constant for all the muscles involved in the model [98]. To overcome this limitation, future work could consist in adding PCSA measurements during the experimental protocol.

Ultrasonography may be added for real-time measurements of **PCSA** of the trunk muscles [94]. It would require a sensitivity study to estimate the number of measurements per muscle. Indeed, one measurement could not be sufficient as it would only reflect locally the **PCSA** of the muscle. On the other hand, it would not be possible to take as many measurements as the number of fascicles in each muscle. In the case of **QL**, it would not be possible to perform **PCSA** measurements due to its deeper localisation.

Estimating maximum isometric forces for each fascicle experimentally does not seem like a realistic approach as it is impossible to isolate the action of one fascicle.

Regarding the **MMS**, its value must be appropriate to the population sample since it is subject-dependent [97]. Finally, for static configurations, a constant **MMS** value may be sufficient. Indeed, Buchanan showed variant **MMS**, but for elbow flexion/extension [98]. Therefore, for dynamic configurations, it would be necessary to evaluate if **MMS** varies or not. The associated study would be based on the comparison between experimental values of trunk forces or torques in a specific motion and predicted values found with the model, as performed in [97].

To conclude about the hybrid approach and the physiological parameters to estimate, results of the hybrid approach for slow-dynamic configurations suggested that a succession of quasi-equilibrium configurations may be sufficient to quantify trunk muscle forces and intervertebral efforts. In other words, muscle dynamics (force-length and force-velocity relationships) could be neglected for trunk muscles if the motion is performed relatively slowly. If this finding came to be confirmed, it would mean that there is mainly one muscle parameter, i.e. **PCSA**, to estimate when using the hybrid approach developed in this thesis. This finding may prevent the avoidance of modelling due to the large number of parameters to be estimates and may encourage people from wider fields, e.g. clinical staff, to use **MBS** musculoskeletal models for the trunk, either for teaching or rehabilitation purposes.

Multibody model Although very correlated to the previous point, the **MBS** model used in this thesis had its own limitations. Firstly, it was hypothesised that the only independent generalised coordinate was located at the S1/L5 level, i.e. the whole "free motion" happened at only one disc level. This hypothesis had one important consequence for modelling: only muscle fascicles inserting above and below the S1/L5 level were modelled. It implied that the middle fibres of the **QL** were not taken into account and

that the **LD** was composed of only one muscle. The latter brought important impact and large unrealistic forces at some point in the thesis. Furthermore, by imposing the motion at the S1/L5 level, it was impossible to estimate intervertebral efforts for this **IVD**. Yet, it has been showed that this **IVD** was also subject to the large loads in the lumbar spine [233, 234]. Besides, if assuming that the motion happened at the S1/L5 level only had few impact for static configurations, it was not true for dynamic motions. Indeed, it was likely that our subjects in the Orthoload case experiment used their hips to achieve larger trunk flexion. It would impact the muscle forces and intervertebral efforts.

To scale the model to each subject of this thesis, a global scaling was applied to the mass and dimensions of each body composing the **MBS** model. However, it is likely that the proportions were not identical for all the subjects. To reach better subject-specific models, several ways are possible. Firstly, geometry and kinematics could be measured during the experiment thank to motion-capture system. Secondly, medical imaging could be imported into a generic model to scale the latter appropriately, e.g. [44]. Thirdly, open-access database with anatomical references could be used. Besides, open-access database would be required for sharing values of physiological parameters such as **PCSA** or **MMS**. Lerchl et al. insisted in their review for database for comparison of computed and in-vivo spinal loads, values for **IVD** intrinsic parameters and muscle parameters as well [7].

Experimental improvements Despite the care given to experimental acquisitions, there were still limitations. Static configurations in this thesis focused on a prone extension exercise, i.e. the Sorensen test, to quantify back muscles forces. For this configuration, surface **EMG** were limited to **LP**, **TP**, **QL** and **LD**. Given that the hybrid approach is only based on **EMG** signals recorded superficially, some muscle could be added to better mimic muscle strategies. First, the multifidus whose activity were previously recorded superficially [32, 183, 184], could be added as well although we considered it was too close to the L5 vertebra on palpation. Therefore, it could not be measured according to our opinion. Danneels et al. proved however a good reliability for surface **EMG** on the multifidus, what could give perspectives for future work. Also, considering the Sorensen test posture, it has been showed that there was a significant contribution of hip extensors [175, 176]. Furthermore, when adding load in an extension exercise similar to the Sorensen test, it has been showed that the gluteus maximus had a greater extent in **EMG** amplitudes compared to paravertebral muscles [100]. It

would suggest a significant contribution of hip extensors in the Sorensen test posture, which could be recorded with surface [EMG](#) as well. It would however require to modify the model to include the muscles, but overall to define the motion differently as in the current model as hip extensors do not insert on both sides of the S1/L5 level.

To use the hybrid approach as done in this thesis, it was required to load the subject with several masses during the experiment. Results showed that at least three different loads – including the unloaded case – should be used. Furthermore, added masses had to be defined in percentages of the body mass. Finally, they should be between 0 and 20% of the body mass. Indeed, the latter was already constraining and may change the muscle strategies that subjects could naturally produce.

Clinical perspectives In addition to advancing the knowledge of trunk muscle force estimation, the models and results of this thesis could likely benefit both physiotherapists and surgeons. More specifically, physiotherapists could use the models to highlight the influence of a particular muscle for a patient. This would likely allow them to adapt their treatment. The models could also be directly used as tools in the rehabilitation process. In the case of scoliosis, when the deformation of the spine is too great and physiotherapy cannot prevent its development, surgery is required. In this context, this thesis should contribute to the improvement of developing an objective planning tool for spine surgery.

About the quantification of muscles itself, it would require to develop open-access database with physiological parameters obtained in-vivo. In particular, in the perspective of the hybrid approach proposed in this thesis, [PCSA](#) should be measured in subjects with [AIS](#). Barba et al. quantified trunk muscles forces and their resulting intervertebral forces in 132 [AIS](#) patients [235]. They obtained muscle forces similar to in-vivo [EMG](#) measurements: greater muscle forces in the convex side of the scoliosis. They pointed out the lack of in-vivo data for comparison but also the lack of data for physiological parameters. For [PCSA](#), they scaled values obtained with EOS imaging on the basis of available data of healthy children [167] and using a distribution similar to the one performed in this thesis for the [PCSA](#) scaling. For muscles again, it would also necessary to develop their geometry [8]. Scaling a generic model to the [AIS](#) morphology is not complicated but the scaling could bring geometrical conflicts between muscles and bone structures [235]. Each scoliosis being unique, one could imagine a guideline to solve geometrical conflicts, e.g. to take into account the curved shape of

the muscle as proposed by Hwang et al. [131, 132].

In the introduction to this thesis, it was said that geometry was relatively well known in the literature. Indeed, as pointed in Chapter 2, there exist many generic models for the spine which can be easily scaled to match the morphology of a scoliotic patient. However, most (if not all) of these models were developed for male subject. Yet, there is a higher prevalence of female scoliotic patients [24]. To the best of author's knowledge, there exists only one female model [231] but developed for healthy and old females. Like the work carried out by Barba et al. [235], it would necessary to develop models – intentionally put in the plural form because they may not all have the same purpose – specific to female subjects and to validate them for female scoliotic patients.

Secondly, there exist several types of scoliosis classified according to their Lenke type [236]. Geometrically, deformations can occur in the thoracic spine, the lumbar spine or in both. It may bring modifications to the model presented in this thesis. Indeed, as our model was assumed to have symmetry in muscles, only the segment of the muscle fascicle located on the lowermost part of the lumbar spine was modelled. The other segments were assumed to have the same muscle forces and were not modelled as they would only bring rigidity to a body already considered as rigid. However, in AIS patients, the concave side of the scoliosis usually means a decrease in muscle activation [235]. Therefore, it would impact the geometry of the muscles and modify their forces. Finally, as the model is not symmetrical anymore, lateral shear forces and frontal torques will appear at each IVD level. The lateral shear force is an important biological marker for scoliosis. In our model, both are already implemented and operational.

Finally, once all these aspects are mastered, there would still be necessary to identify configurations for which muscle forces and intervertebral efforts are quantified in order to plan a spine surgery. It is more likely that both static and (slow) dynamic configurations would be needed; the first ones for calibration of the EMG signals¹ and the second ones to keep in step with reality. Indeed, operated patients hope to recover a regular lifestyle with daily tasks. Tasks such as walking or load-lifting may be of interest and are already investigated in the literature [56, 57, 102]. At that point, all the medical staff, such as physiotherapists and surgeons, should be included into the loop to take into account their experience, their work routine and the time they may allow for the experiment.

¹For scoliotic patients, MVC may not be used as it is not always appropriate for patients with pain [237]. In this case, sub-MVC should be considered.

11.3 Overall conclusion

In this thesis, we have presented a hybrid approach to quantify back and abdominal muscle forces and resulting intervertebral efforts. Hybrid approaches are not new and have the advantage to take into account the real muscle strategies unlike the solutions found by mathematical approaches. The proposed hybrid approach relies on an upfront experiment to make the most of **EMG** as input of a **MBS** model.

In the frame of the hybrid approach, it was necessary to develop a well thought experiment and protocol to ensure a good quality of **EMG** signals. The Sorensen test was identified as a suitable static configuration to quantify back muscle forces. Indeed, this test generated intense **EMG** signals of back muscles and did not involve the modelling of any external device. The Sorensen test posture was used for several experiments to illustrate the variety of muscle strategies among a sample of 19 subjects and the variety of muscle strategies produced by one subject.

A **MBS** model associated with the experiment was developed to compute muscle forces and intervertebral efforts. The model includes **LP**, **TP** and **QL** fascicles. It was showed that the latter had an important impact of the intervertebral efforts when its activation increased. The model was adapted to the different experiments and population samples and could have an additional muscle modelled, i.e. **LD**.

The hybrid approach was developed in two steps. First, it was proposed to use a relatively simple **EMG**-based distribution of a global equivalent force spread out among all the muscle fascicles included in the model. This **Hybrid-1** approach showed promising results with intervertebral efforts close to those obtained with purely optimisation-based solutions. More specifically, the **Hybrid-1** approach was able to reflect the different muscle strategies recorded in the subjects. Also, it was able to solve the muscle redundancy problem for one subject showing three different muscle strategies for the same static configuration.

In a second time, it was proposed to embed **EMG** signals into the optimisation process itself. Unlike the **Hybrid-1** approach, it allowed the optimisation process to have room for manoeuvre on muscle forces. **EMG** were added as constraint functions and injected in the classical cost functions. Embedding **EMG** as constraint functions showed the better results with muscle strategies improved compared to the **Hybrid-1** approach. Intervertebral efforts were close to the ones reported in the literature taken into account the Sorensen test posture, which generated greater anteroposterior

shear forces.

A simple dynamic configuration, i.e. a trunk flexion/extension with a straight back, was also investigated. The objective was twofold: compare intervertebral efforts with in-vivo data and confront the [Hybrid-1](#) approach, designed for static configurations, to a dynamic motion. Compressive forces were similar to the in-vivo measurements and there was a linear correlation between compressive forces and flexion angles. This findings showed that the hybrid approach could be used for dynamic motions, even performed at a normal speed, without considering the muscle dynamics. It also increased interest in the proposed hybrid approach.

Finally, additional experimental studies were presented to validate the use of surface [EMG](#) on the [QL](#), i.e. a deep back muscle, and to highlight sex-related differences in [QL](#) activation in the Soresen test configuration.

This thesis had for objective to improve the estimation of back muscle forces and intervertebral efforts thanks to an innovative hybrid approach which relies on the determination of few physiological parameters. Estimation of muscle forces was identified as one of the last challenges to the development of realistic spine models and to the development of an objective planning tool for spine surgery.

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PART V

Appendices



Hybrid-2 approach: EMG-assisted optimisation

A.1 Additional graphs for muscle forces

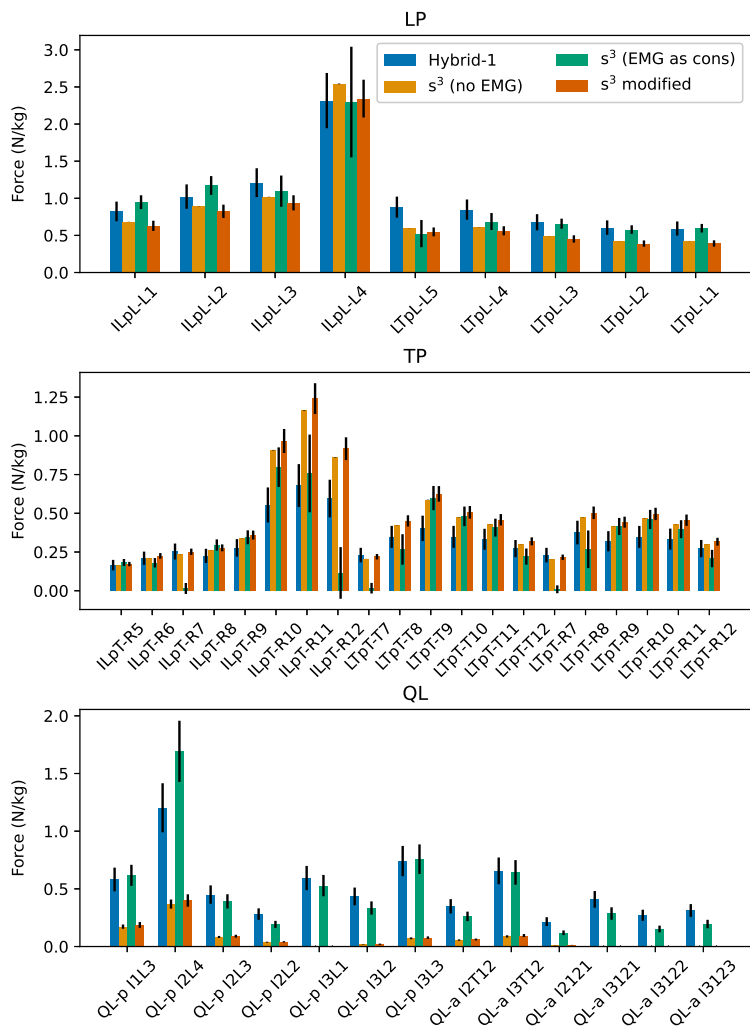


Fig. A.1 Normalised muscle forces (in N/kg) for **LP**, **TP** and **QL** obtained by the **Hybrid-1** approach, by the pure mathematical computation with s^3 , by adding **EMG** as constraints and by modifying the cost function for the unloaded configuration.

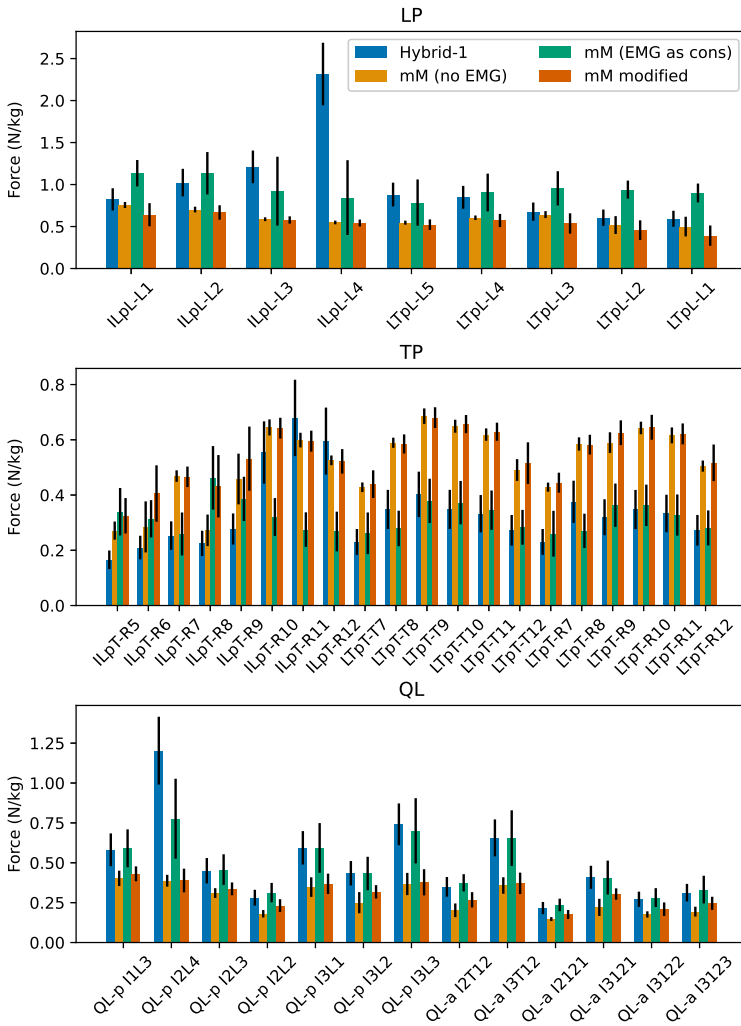


Fig. A.2 Normalised muscle forces (in N/kg) for **LP**, **TP** and **QL** obtained by the **Hybrid-1** approach, by the pure mathematical computation with **mM**, by adding **EMG** as constraints and by modifying the cost function for the unloaded configuration.