Construction of musculoskeletal systems for anatomical simulation

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RÉSUMÉ

L'usage d'humains virtuels s'est démocratisé à de nombreuses activités ces dernières années. Au-delà de la chirurgie virtuelle, les corps virtuels sont de plus en plus utilisés pour concevoir des dispositifs médicaux, des véhicules et des outils de notre quotidien plus généralement. Ils se sont avérés être également d'extraordinaires supports à l'apprentissage de l'anatomie. De récents films (*Avatar, Le seigneur des anneaux, etc*) ont démontré que l'anatomie et la biomécanique peuvent être utilisées pour concevoir des personnages d'une grande qualité. Cependant, reproduire le comportement des structures anatomiques demeure une tâche complexe, et de nombreuses connaissances variées sont nécéssaires à la mise en place de simulation de qualité de nos organes. Ceci fait de la modélisation pour la simulation d'humains une problématique non résolue, une tâche fastidieuse, mais également un sujet de recherche fascinant.

À travers ces travaux de thèse, nous abordons cette problématique de la construction de systèmes musculo-squeletiques pour ces domaines variés : *animation, biomécanique et aide à l'apprentissage*. Notre objectif est de simplifier le processus entier de création en le rendant plus intuitif et plus rapide. Notre approche consiste à pallier à chacune des difficultés, à savoir : *la représentation et la manipulation de connaissances anatomiques, la modélisation géométrique et la simulation efficace de systèmes musculosquelettiques* grâce à trois principales contributions introduites durant ces travaux de recherche.

Notre première contribution se focalise sur la construction biomécanique d'un modèle hybride du rachis lombaire. Dans ces travaux, nous montrons que les approches hybrides combinant des systèmes de corps rigides et des modèles éléments finis permettent d'obtenir des simulations en temps intéractifs, précises, et respectant les principes de l'anatomie et de la mécanique.

Notre seconde contribution s'intéresse aux problématiques liées à la complexité des connaissances anatomiques, physiologiques et fonctionnelles. En se basant sur une ontologie de l'anatomie et une ontologie inédite de la physiologie humaine, nous introduisons un pipeline pour la construction automatique de modèles simulant les fonctions de nos organes. Celles-ci permettent d'exploiter les connaissances anatomiques complexes via des requêtes simples. Les sorties de ces requêtes sont utilisées pour créer des modèles simulables retranscrivant les aspects fonctionnels tels qu'ils ont été formalisés et décrits par les anatomistes.

Enfin, notre troisième contribution : le *transfert d'anatomie*, permet d'adapter les modèles géométriques et mécaniques à la morphologie de patients spécifiques. Cette nouvelle méthode de recalage permet de reconstruire automatiquement l'anatomie interne d'un personnage défini par sa peau en transférant les organes d'un personnage de référence. Elle permet de pallier à la nécessité de re-construire ces géometries pour chaque nouvelle simulation, et contribue ainsi à accélérer la mise en place de simulations spécifiques à une grande variété d'individus de morphologie différente.

ABSTRACT

The use of virtual humans has spread in various activities in recent years. Beyond virtual surgery, virtual bodies are increasingly used to design medical devices, vehicles, and daily life hardware more generally. They also turn out to be extraordinary supports to learn anatomy. Recent movies (Avatar, Lord of the Rings, etc) demonstrated that anatomy and biomechanics can be used to design high-quality characters. However, reproducing the behavior of anatomical structures remains a complex task, and a great amount and variety of knowledge is necessary for setting up high quality simulations. This makes the modeling of human body for simulation purposes an open problem, a tedious task, but also a fascinating research subject.

Through this PhD, we address the problem of the construction of biomechanical models of the musculoskeletal systems for several domains : *animation, biomechanics and teaching*. Our goal is to simplify the entire process of model design by making it more intuitive and faster. Our approach is to address each difficulty : *the representation and use of anatomical knowledge, the geometrical modeling and the efficient simulation of the musculoskeletal system* thanks to three novel contributions introduced during these research works.

Our first contribution focuses on the biomechanical construction of a hybrid model of lumbar spine. In this work, we show that hybrid approaches that combine both rigid body systems and finite element models allow interactive simulations, accurate, while respecting the principles of anatomy and mechanics.

Our second contribution addresses the problem of the complexity of anatomical, physiological and functional knowledge. Based on a novel ontology of anatomical functions of the human body, we introduce a novel pipeline to automatically build models that simulate physiological functions of our bodies. The ontology allows us to extract detailed knowledge using simple queries. The outputs of these queries are used to set up simulation models of the functional aspects as they were formalized and described by anatomists.

Finally our third contribution, the anatomy transfer, allows the mapping of available geometrical and mechanical models to the morphology of any specific individual. This novel registration method enables the automatic construction of the internal anatomy of any character defined by his skin, by transferring organs from a reference character. It allows to overcome the need to re-construct these geometries for each new simulation, and it contributes to accelerate the simulations setup for a range of people with different morphologies.

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PLAN

CHAPTER

1 INTRODUCTION

Open problems in modeling and simulation of musculoskeletal system

The modeling and simulation of musculoskeletal system has become an area subject to great interest in recent years. A high level of anatomical precision has proven to be necessary in many Computer Graphics applications, from visualizing the internal anatomy for education purposes, to anatomical simulation for feature films, ergonomics, medical, or biomechanical applications (e.g. optimizing muscle energy).

In the context of this work, the modeling of musculoskeletal systems describes the process for representing systems using specific tools, with the aim at re-transcribing criteria and characteristics that define the system. These criteria are mainly of two types :

– From anatomical type such as the external shape of the structures composing the system, or the shapes of elements constituting these structures.
– From physiological type as the biomechanical behavior of each structure, the interactions between these structures, and the system interactions with the external environment.

The specifics tools for modeling are theoretical tools (*Mathematics, Computer science, Mechanic, . . ., etc*) and software that implement theses theoretical tools (*SOFA, Opensim, ABBACUS, Maya, Blender, . . ., etc*).

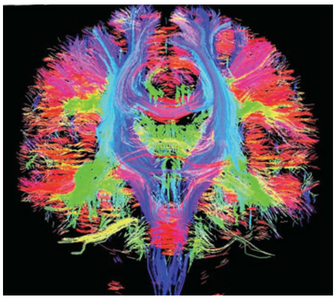
The simulation of anatomical system describes the evaluation in time of the different states of the system in specified conditions. It first aim at reproducing the real behavior of each system component in given conditions in order to predict other states.

The modeling and simulation of anatomical system play a major role in many fields.

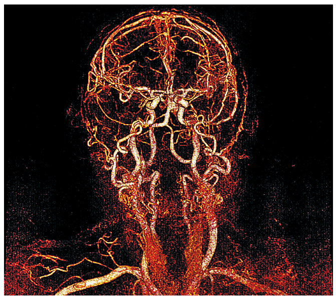
In the **field of health**, they become indispensable to the good practice of numerous medical activities. Modeling for visualization as well as simulation contributes to an accurate **diagnosis** of known diseases. They allow a better understanding of illness as well as a precise characterization of responsible agents [RAC*13]. A good illustration is the possibilities offered by reconstruction methods allowing to characterize causes of dysfunction such as lung nodule

(Fig.1.1a) or tumor. This type of data reconstruction and visualization are widely used to improve surgical interventions planning. Reconstruction techniques as tractography (Fig.1.1b) and angiography (Fig.1.1c) represent the only way to get in-vivo overview of nerve fibers path and an overview of circulatory system. If modeling for visualization proved essential for medical practices, it is also true for simulation which is increasingly integrated in the surgical process.

(a) *Detection of lung nodule*



(b) *Tractography.*



(c) *Angiography*

Numerous works such as [RSC*07, Pay12] demonstrate how **simulation can assist surgery**. This assistance translates into the possibility to predict the state of a system after a specific procedure to optimize it, and its effects on organs [SJK13, SKM*13, CRS*14]. The purpose is to reduce risks on patient. This way of acting is commonly used for cranio-maxillofacial [KGKG98] or spine [ALP*00] surgeries.

Modeling and simulation are also used in the **field of sport and rehabilitation**. It mainly aims at analyzing performance in order to better understand it and to improve it [LKH*02, Yea98, MLT*08] as illustrated in Fig.1.2.

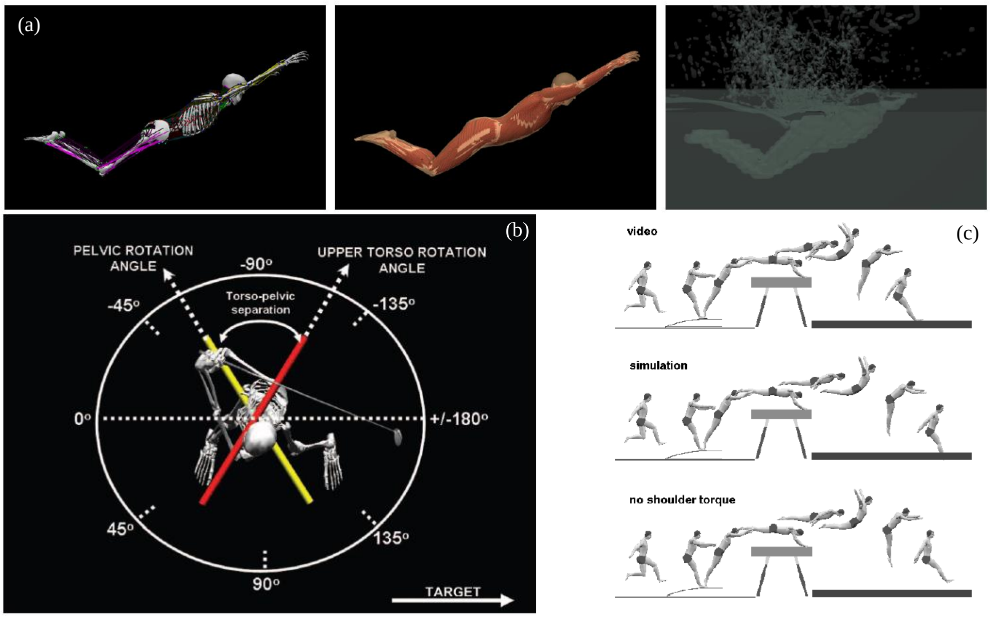


FIGURE 1.2 – *Simulation in the sport field : (a) Simulation of the swimming [Si13]. (b) Simulation of golf movement [MLT*08]. (c) Simulation of gymnastic movements [Yea98]*

To enhance **the design of device** that interact with our body (cf Fig.1.3), engineer are increasingly relying on simulation (*e.g : medical devices : prosthesis or orthosis, transport* *device : air bag, chair, car belt, and commonly objects : shoes, gloves, ...*). This results from the improvement of modeling and simulation tools that are becoming more accessible and accurate. Thus, they allow reduction in the cost and time on the design. These benefits are also due to the decreased number of testing phases with real prototypes on human since they are performed with simulation.

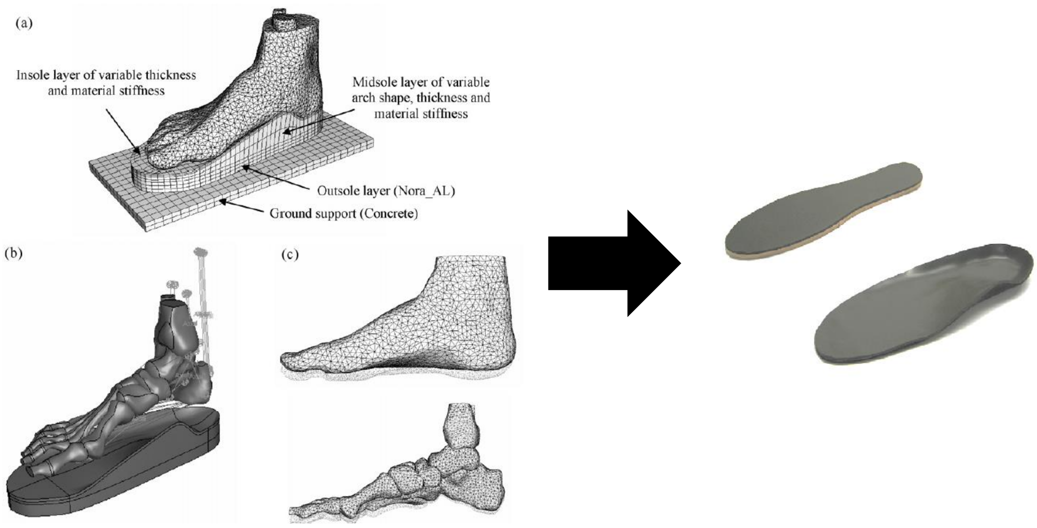


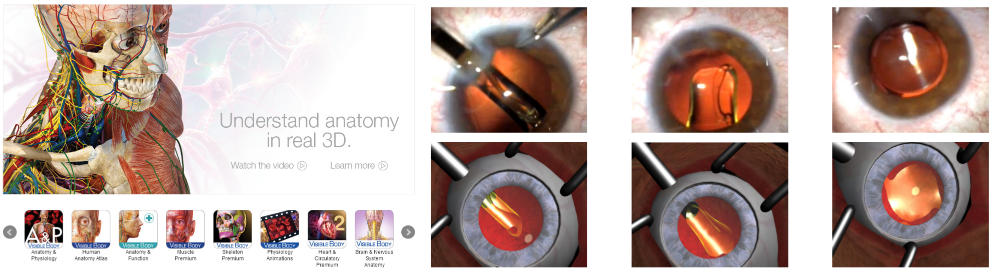
FIGURE 1.3 – *Simulation to assist the design of daily life tools [CZ08].*

Simulation-based **learning** is becoming widely established within medical education. Since training in the real-life context is not always possible for reasons of safety, costs, or didactics, alternative ways are needed to achieve clinical excellence. Simulation-based learning offers obvious benefits to novices learning invasive procedural skills, especially in a trend of decreasing clinical exposure. In the ways of learning anatomy with modeling and simulation tools, we mainly distinguish two types of applications.

On the one hand, we find tools that enable the visualization of the geometry of organs within the body [bio, inn, zyg, vis, hea] (cf Fig.1.4a) and how these organs work in synergy with their adjacent structures through animation or simulation [esc, ani]. These tools complementary to classical teaching of anatomy contribute to a better learning since they allow to observe phenomenon that usually required in-vivo or in-vitro experiments. Similarly, the evolution of technological devices such as Kinect have pushed forward these learning approaches, and allowing to learn from his own body. This concept called embodiment have proved its benefits on our way of learning anatomy [KBSC14a, MFB*13] and currently represent a field of interest in research [BUD*14].

On the other hand, there are tools that enable to practice medical and surgical procedures [CDA99, DCC*13] as shown in Fig.1.4b. Some of them include haptic devices coupled with simulation software to allow to virtually interact with organ [DCC*13]. The visual and force feedback contribute to re-transcribe feeling of real intervention [SVT*11]. The use of these means represents good complementary tools to existing learning methods [PR13] instead of replacing them.

The **movie and animation** industry does also not escape from the use of virtual human. While the goal and the constraints of their use are not the same as those described above, the approaches, and the required tools remain identical. The aim of simulation in these areas is to propose the most realistic animation, with credible deformations and photo-realistic texture



(a) *The use of 3D model to assist the learning process [bio].*

(b) *Using simulation to practice surgical gesture [DCC*13].*

and rendering [SKP08]. The purpose is to make people believe that what they are watching is real and this is the main validation of these simulations [imm11, the08]. They are mainly used for special effects and for humanoid character [the11, the03a, ava09]. Recent years witnessed huge improvements in anatomically-based simulation, especially in terms of computational efficiency [PMS12] and realism [LST09].

FIGURE 1.5 – *Simulation in the movie and animation field to produce visual effect [imm11].*

Achieving high quality simulation of musculoskeletal system usually required three main stages.

The first step consists in learning anatomy and physiology of the system to simulate. At the end of this step, all system components, their structure, their functionality and their involvement in the phenomenon to reproduce are known and formalized. This set of complex anatomical knowledge represents the basis of the process, and is used to be associated to the medical field and therefore not within everyone's reach.

The second step aims at representing each system entity. The choices of representation depend on the goal of the simulation. These representations includes geometrical modeling, mathematical and biomechanical modeling of the internal behavior of each entity, and the modeling of the interactions between system elements, and between the system and the external environment. For instance, some simulations require *Finite Element* (FE) meshes with specific type of nodes, while other are more efficient with rigid body systems. The cost of setting up a 3D anatomical model for a given character is high since it required geometrical modeling skills, mathematical and biomechanical skills in addition to anatomical knowledge. This task is very time consuming and tedious, as it requires modeling of the bones, organs, muscles, as well as connective and fat tissues. With real humans, it is possible to take advantage of 3D imaging, such as MRI [BAGD07]. However, this route is difficult or even impossible for fictional characters, ranging from Popeye to Avatar's Na'vi. Highly realistic animations showing muscles or tendons deforming the skin typically require precise anatomical models and simulation of the internal anatomy. Moreover, the control of the fat distribution is important for achieving the associated secondary dynamics effects. The current tools available for artists to model anatomical deformations [MM13] as well as early academic work [WVG97, SPCM97] extensively rely on user input, essentially amounting to setting up the musculature from scratch.

At the end of the modeling stage, begins the third step whose purpose is to simulate the system. The simulation aim at reproducing phenomena initially targeted while respecting the constraints specific to each domain. Achieving this stage, required skills in Computer Science, Mathematics and Mechanics to efficiently implement tools for simulation and visualization.

Even if the use of virtual musculoskeletal systems are required in numerous fields, in different ways, and for various purposes, the improvement of modeling and simulation tools did not prevent the process of construction of these system to be tedious and complex. Achieving high quality simulation remains hard because of all of the knowledge and skills required at each stage of the process. All these skills are rarely within everyone's reach.

Contributions

In this research work, we address the question of creating simulation models while fitting needs of various area. Our goal is to simplify the process of modeling, with the purpose to make it faster, less tedious and ideally more accessible to the greater number of people. To reach these goals, we worked on three main contributions.

Hybrid spine model :

We introduce a novel 3D dynamic model of the lumbar spine that combines both FEM and multibody systems. We show that accurate simulations can be obtained with easier modeling and faster computation times than using the traditional finite element method. This work illustrates that hybrid modeling and simulation fit medical needs without penalizing precision. Special care is paid to the construction of the model. Each vertebra is modeled using a rigid body. The position of the rigid body, the inertia matrix and the mass of the vertebra are computed based on the geometrical shape. Each rigid body controls a surface used for the contacts that occur between two vertebrae on the spinal process during the extension. The facet joints are modeled using 6D elastic joints that allow restricted relative translations and rotations. The ligaments are modeled using non-linear springs attached to the vertebrae. Each spine unit contains six ligaments : the anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL), the ligamentum flavum (FL), the transverse ligament (TL), the interspinous ligament (IL) and the supraspinous ligament (SL). The most complex object is the disc, modeled as a heterogeneous finite element model for two of its components : the nucleus and the annulus. The two bony plates are considered undeformable and are constrained in displacement en velocity to the rigid bodies of the vertebrae. These plates allow to connect the finite element model of the disc to the rigid bodies of the vertebrae and then allow to propagate load and force in the spine through the disc. The model is validated against the experimental and theoretical results of the literature on flexion, extension, lateral bending as well as axial rotation.

Ontology based modeling :

This contribution shows that structured anatomical knowledge that can be used to automatically build models for the simulation and the visualization of functions of our organism. We introduce a novel pipeline for the construction of biomechanical simulations by combining generic anatomical knowledge with specific data. Based on functional descriptors supplied by the user, the list of the involved anatomical entities (currently bones and muscles) is generated using formal knowledge stored in ontologies, as well as a physical model based on reference geometry and mechanical parameters.

This simulation-ready model can then be registered to subject-specific geometry with MRI data as input to perform customized simulations. The user can provide additional specific geometry, such as a simulation mesh, to assemble with the reference geometry. Subject-specific information can also be used to individualize each functional model. The model can then be visualized and animated.

This pipeline dramatically eases the creation of biomechanical models. We detail an example of a musculoskeletal simulation of knee flexion and hip flexion and abduction, based on rigid bones and the Hill muscle model, with subject-specific 3D meshes non-rigidly attached to the simulated bones.

Anatomy transfer

Starting from a model is built for simulation using the two previous contributions, this contributions allows to reconstruct anatomy and simulation meshes for a set of character. We propose the first semi-automatic method for creating anatomical structures, such as bones, muscles, viscera and fat tissues. This is done by transferring a reference anatomical model from an input template to an arbitrary target character, only defined by its boundary representation (skin) instead of MRI data of the contribution in Sec.1. We call this process "Anatomy Transfer". The purpose is to quickly get a wide range of model ready for simulation. The fat distribution of the target character needs to be specified. We can either infer this information from MRI data, or allow the users to express their creative intent through a new editing tool. The rest of our method runs automatically : it first transfers the bones to the target character, while maintaining their structure as much as possible. The bone layer, along with the target skin eroded using the fat thickness information, are then used to define a volume where we map the internal anatomy of the source model using harmonic (Laplacian) deformation. This way, we are able to quickly generate anatomical models for a large range of target characters, while maintaining anatomical constraints.

CHAPTER

2

STATE OF THE ART

2.1 INTRODUCTION TO THE ANATOMY AND THE PHYSIOLOGY OF THE MUSCULOSKELETAL SYSTEM

This section reviews the anatomy and physiology of the musculoskeletal system. If it does not go further on the way that work each articular system of our organism, it focuses on the description of each component of the musculoskeletal system.

The human musculoskelatal system is the set of organ system that gives to vertebrates the ability to move hence the name *locomotor system*. It is composed by the skeletal system, the joint system, and the muscular. These sub-systems include both soft and hard tissues.

The main hard tissues are the *bones* which compose the skeletal system and allow to support the body. Soft tissues are mainly the *ligaments* and the *cartilaginous tissues* which are parts of the joint system, the *muscles* whose role is to drive the body mobility, and the *inter-vertebral discs* which aim to limit excessive motion in the spine. Musculoskeletal soft tissues generate motion, protection of vital organs and gives his shape to our body.

The remainder of this section is as follows. The first sub-section (Sec.2.1.1) introduce anatomical terms such as the main anatomical directions, planes, sections, and the main movements existing in anatomy. The purpose is to define words that will be found several time within the manuscript to ease the description of anatomical systems. Since this thesis focuses on the modeling and the simulation of the musculoskeletal system, sections 2.1.2 to 2.1.5 summarize important characteristics that define each component of the musculoskeletal system. They describe in detail each type of these structures, their shape, their composition, their mechanical behavior, and their role and implication on the functions of the musculoskeletal system.

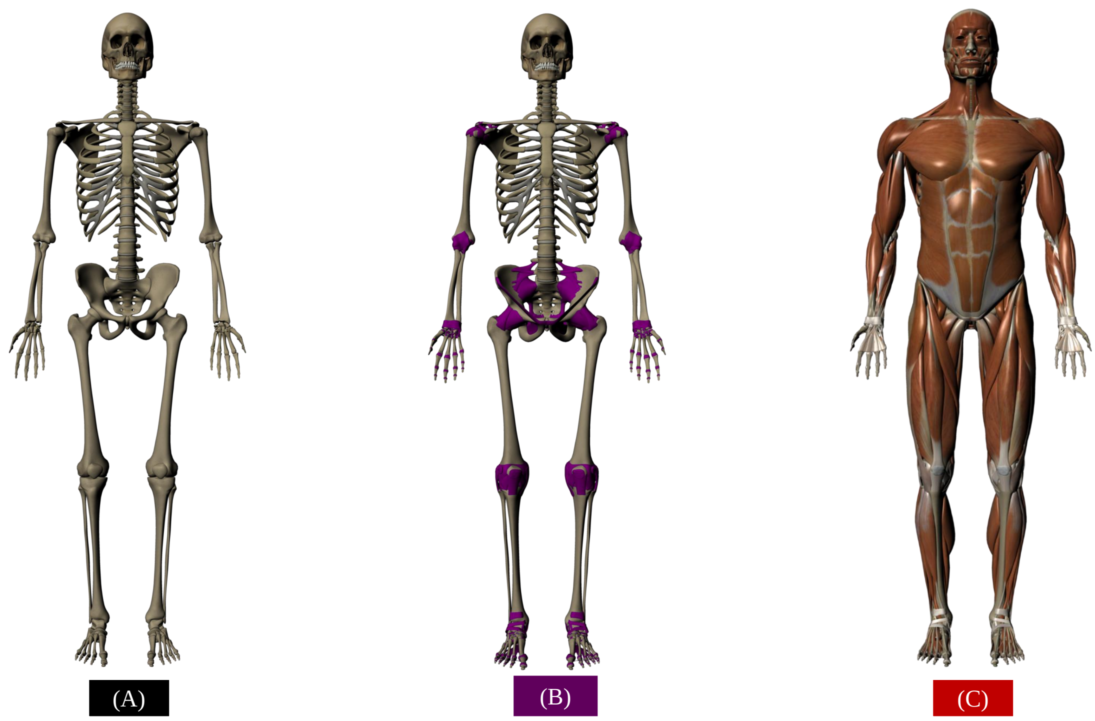


FIGURE 2.1 – *Musculoskeletal System : (A) Skeletal system, (B) Joint system (ligaments and connetive tissues), (C) Muscles system*

2.1.1 Anatomy terminology

To prevent confusion and to standardize the way of describing anatomy and physiology, anatomical terms have been introduced. These terms allow to specify orientations and directions when discussing human anatomy, and then ease communication and description of ideas. These terms are preferred to express relative position and orientation of organs or body parts, mainly because they do not rely on the orientation of the viewer. In our description of these terms, we focus on four categories related to the musculoskeletal system description : directions, plane, depths, and movements. Most of the definitions below and images are provided and adapted from [MO09, Kap07, RS, Jon].

Anatomical directions :

The directions terms specify the locations of structures in the body or the direction and orientation in relation to other structures. They are universal and cover directions away, toward and within the body. These terms that express directions or relative positions are illustrated in the Fig.2.2 and are :

– *Anterior* : Towards the front (the eyes are anterior to the brain) (*ventral*).
– *Posterior* : Toward the back (the pharynx is posterior to the oral cavity) (*dorsal*).
– *Inferior* : Refers the fact that a part is below another or towards the feet (*caudal*).
– *Superior* : Represents the fact that a part is above another or closer to head (*cranial*).
– *Medial* : Relates to the imaginary midline dividing the body into equal right and left halves (*i.e the nose is medial to the eyes*).
– *Lateral* : Means towards the side with respect to the imaginary midline (*i.e the ears are lateral to the eyes*).
– *Proximal* : Describes a part that is closer to the trunk of the body or closer to another specified point of reference than another part (*i.e the elbow is proximal to the wrist*).
– *Distal* : Refers the fact that a particular body part is farther from the trunk or farther from another specified point of reference than another part (*i.e fingers are distal to the wrist*).

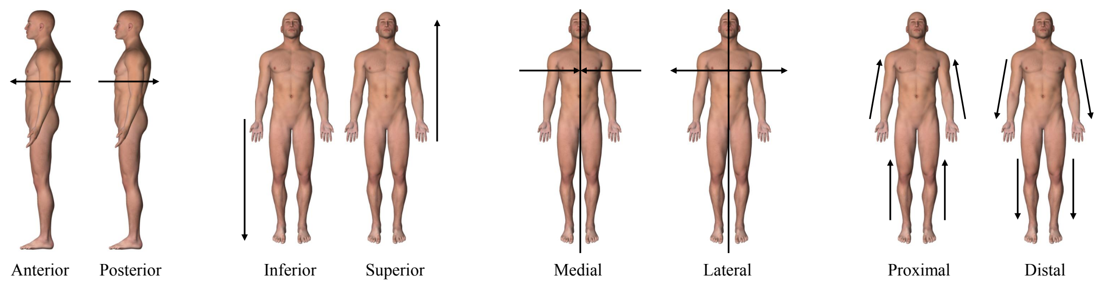


FIGURE 2.2 – *Anatomical main directions*

Anatomical planes :

Anatomical planes are the planes which present a two-dimensional section of the body or the body parts (cf Fig.2.3). Our anatomy can be sectioned into three different anatomical planes to ease its description by referring to the point of view that might be considered. These three planes are illustrated in Fig.2.3 and described below.

– *Sagittal plane* : It is the plan which divides the body into left and right sections.

- *Midsagittal (median) plane* : It is the plan which divides the body into equal halves at midline, it is a specific type of sagittal plane.
- *Frontal (coronal) plane* : It is the plan which divides the body into anterior and posterior sections.
- *Transverse (horizontal) plane* : It is the plan which divides the body into superior and inferior sections.

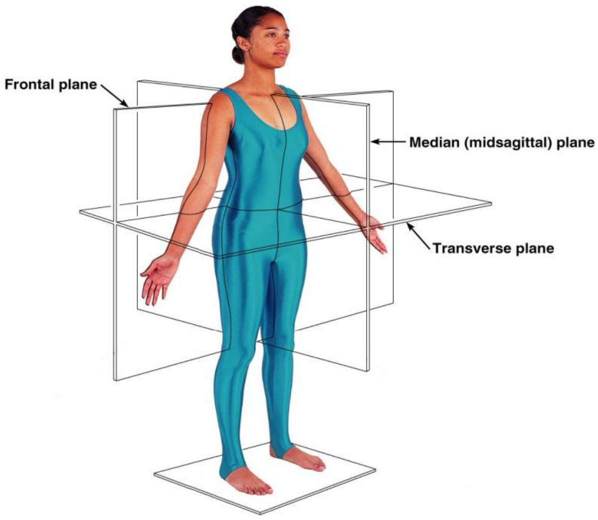


FIGURE 2.3 – *Anatomical planes.*

Anatomical depths :

There are three specific terms to refer the depth relative to the surface while describing anatomy (cf Fig.2.4).

- *Superficial* : It refers to an entity situated near the surface. *Peripheral* also means outward or near the surface.
- *Deep* : It is used to describe parts that are located more internally.
- *Intermediate* : It is used to describes a component located in the middle between the superficial and the deep layers.

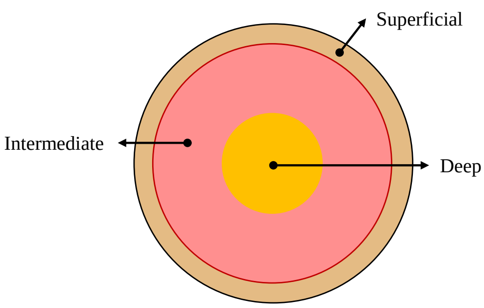


FIGURE 2.4 – *Anatomical depths.*

Anatomical main movements :

Specific terminology is also used to describe joint movement as well. These movements are illustrated in Fig.2.5 and described down below.

- *Flexion* : refers to a movement that decreases the angle between two body parts (*i.e flexion at the elbow is decreasing the angle between the ulna and the humerus*).
- *Extension* : refers to a movement that increases the angle between two body parts (*i.e extension at the elbow is increases the angle between the ulna and the humerus*).
- *Abduction* : it is a movement away from the midline. This movement refer to a rotation around an axis perpendicular to the frontal plane.
- *Adduction* : it is a movement towards the midline. This movement is the same type of movement as the *abduction*, but in the opposite direction.
- *Medial rotation* : it is a rotating movement towards the midline.
- *Lateral rotation* : it is a rotating movement away from the midline.
- *Axial rotation* : it is a rotating movement around a given axis (*i.e axial rotation of the vertebra C6 around the vertebra C7*).
- *Pronation* : it moves the palm of the hand so that it is facing posteriorly (*i.e your forearms are pronated when typing on a keyboard*).
- *Supination* : it moves the palm of the hand so that it is facing anteriorly (*i.e your hands are supinated when holding a bowl of soup*).
- *Dorsiflexion* : refers to extension at the ankle, so that the foot points more superiorly.
- *Plantarflexion* : refers flexion at the ankle, so that the foot points more inferiorly.

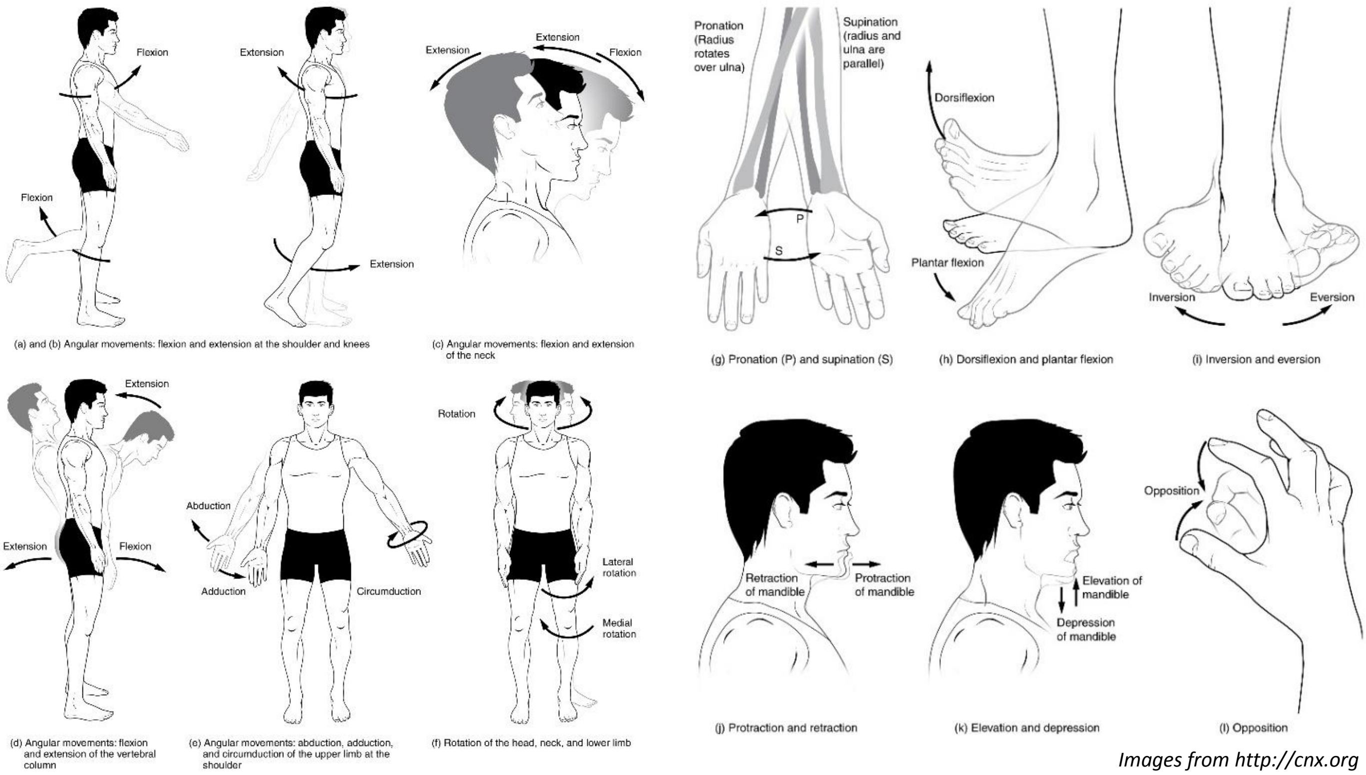


FIGURE 2.5 – *Anatomical main movements.*

2.1.2 Skeletal system

Bones are the main component of the skeletal system (cf Fig.2.6.B) whose the purpose is to serve as structural support and protection for internal organs, to provide rigid kinematic links and muscle attachment sites, and to facilitate muscle actuation and body movement. It is a specialized connective tissue which has unique structural and mechanical properties that allows it to carry out these tasks.

Microscopically, the fundamental structural unit of bone is the osteon, which is composed of concentric layers of a mineralized matrix surrounding a central canal containing blood vessels and nerve fibers (cf Fig.2.6.A). The mineral portion of bone contains calcium and phosphate, and thus serve as reservoir for these two components [NF01].

Macroscopically, each bone is composed of two major parts : the outer shell which is made of *cortical bone* (*or compact bone*), and the *cancellous bone* (*or trabecular bone*) inside (cf Fig.2.6.A). The cortical bone is dense, very stiff and forms the shaft of the bone. Its structure is similar to that of the ivory. The cancellous bone consists of loosely knit structures (thin plates) that remind a honeycomb. This tissue is arranged in concentric lacunae containing lamelae but does not contain haversian canal (*canal which runs longitudinally through the bone and contain blood vessels and nerves which supply the bone*). Trabecular bone is softer than the cortical bone. It gives supporting strength to the ends of the weight-bearing bone.

Numerous bones have a cavity that contains the *bone marrow* as illustrated in Fig.2.6.A. Bone marrow is a viscous tissue that resides in the confines of bones and houses the essential pluripotent precursor cells for the living organism. It is the major source of cells which have the role of renewing the elements in the blood, and which contribute to the regeneration of tissues such as bone, cartilage, muscle, adipose, tendon, ligament [GA08].

The bone is an anisotropic material, its mechanical behavior is highly dependent on its geometry [NF01]. These geometry criteria include the length, the cross-sectional area and the distribution of bone tissue around the neutral axis. Considering these parameters, we distinguish 4 types of bone : the flat bones, the short bones, the long bones and the irregular bones as shown in cf Fig.2.6.C.

2.1.3 Joint system (Ligaments and connective tissues)

If bones contribute to our body movements, this is mostly thanks to the joints. Joints represent the area where two bones are attached for the purpose of permitting body parts to move.

According to their type of motion and the shape of bones, we distinguish 7 groups of joints : *immovable joints* {0 DOF} (e.g, skull), *ball-and-socket joint* {3 DOF} (e.g, hip), *hinge joint* {1 DOF} (e.g, elbow), *condyloid joint* {2 DOF} (e.g, fingers), *gliding joint* {various DOFs depending on the joint} (e.g, wrist), *saddle joint* {2 DOF} (e.g, thumb), *pivot joint* {1 DOF} (e.g, neck) (cf Fig.2.7).

Joints are composed of bones and connectives tissues. We find 3 types of joints within the human body regarding their compositions : *fibrous joint*, *cartilaginous joint*, and *synovial joint*. Each joints of the skeletal system are usually connected, stabilized and surrounded by 3 principal dense connective tissues : ligaments, tendons and joint capsules (cf Fig.2.8). These passive structures contribute to guide joint motions, to prevent excessive motion while augmenting the mechanical stability of these articular systems. They have the function to passively modulate force transmitted from one structure to another during locomotion.

The ligaments and capsules connect pairs of bones, they act as static restraints. Tendons attach bones to muscles, and transmit load from muscle to bone in order to initiate joint motion, or to maintain body posture [NF01].

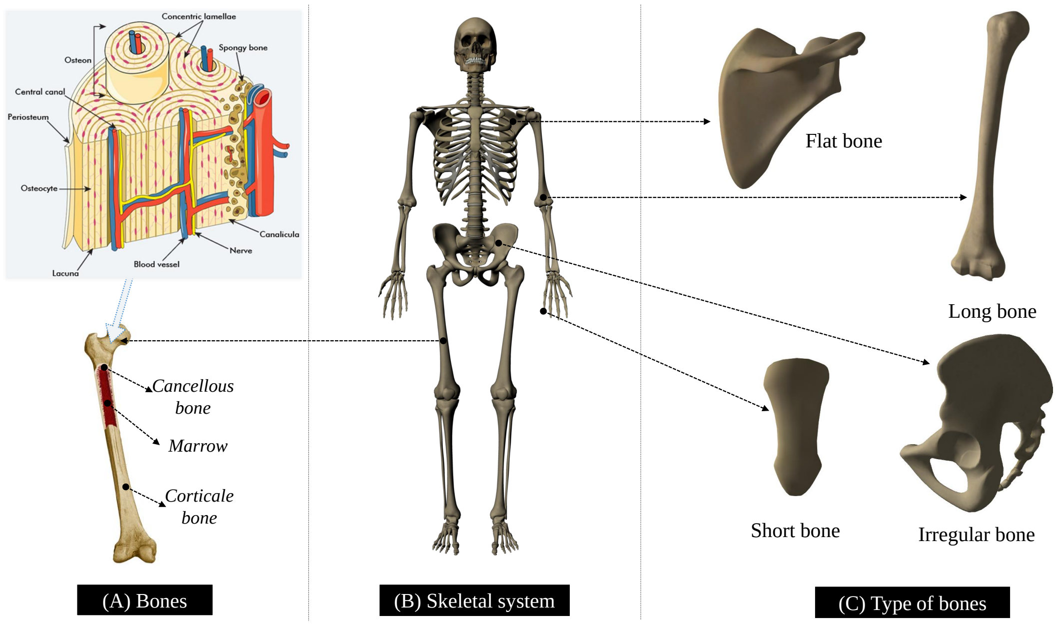


FIGURE 2.6 – *Skeletal system description*

These connectives tissues are composed essentially of collagen and elastin with various proportions (cf Fig.2.8). The collagen limits the extensibility of these components during motion, and the elastin helps to make them softer. One major difference between ligament and tendon is their fiber arrangement (cf Fig.2.8). Tendon are mostly composed of parallel arrays of collagen fibers closely packed together while some within ligament are less well organized.

This specific composition provides them with non-linear elasticity as shown in Fig.2.9. The stress-strain curve contains three main parts : the toe region, the linear region and the yield and failure region. Most physiological activity occurs in the toe and the linear regions. This elastic behavior is modeled from experimental measurements on isolated fibers of ligament, tendon, and other connective tissues.

2.1.4 Muscles

Muscles are the active tissues within the body that generate forces to drive motion. The muscular system consists in 3 types of muscle depending on their physiological functions : the cardiac muscle (*composing the heart*), the smooth muscles (*these muscles surround the hollow internal organs*), and the skeletal muscles which are attached to the skeleton via tendons and which are the driving force of the body mobility.

All of these classes of muscles are controlled by the autonomic nervous system and contract unconsciously except the skeletal muscles. Skeletal muscle contractions are controlled by the somatic nervous system, and most of the time, contractions are done consciously. These voluntary contractions (chemical signals) are transformed into force transmitted to the underlying skeleton. Thus, the main role of muscles is to provide strength and protection to the skeleton by distributing loads and absorbing shock. Muscles are usually divided in 3 parts : the *origin* (attachment of muscle to the stationary bone), the *belly* and the *insertion* (attachment of muscle to the movable bone) (cf Fig.2.11). They enable bone to move at the joints (dynamic work) and provide the maintenance of the body posture against force (static work). Anatomically, they characterize body shape, and they represent the most abundant tissue in our organism, accoun-

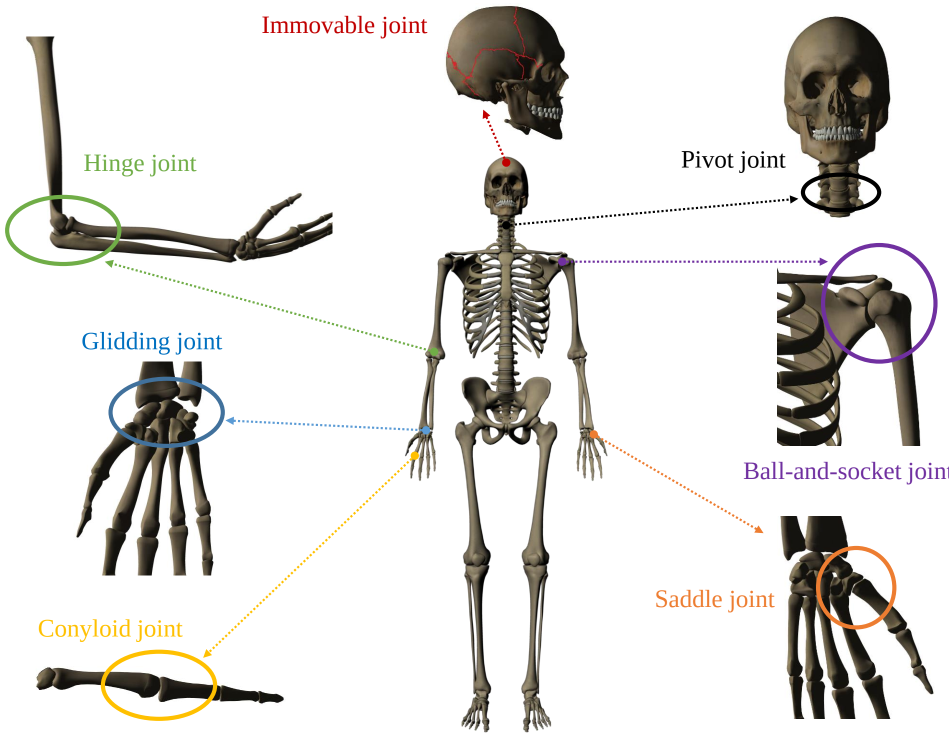


FIGURE 2.7 – *Examples of joint type*

ting for 40% to 45% of the total body weight. There are more than 430 skeletal muscles in the human body.

The structural unit of skeletal muscle is the *muscle fiber*. The muscle fiber is a long cylindrical cell with hundreds of nuclei. It consist of *myofibrils*, which are invested by delicate plasma membrane called the *sarcolemma*. The myofibrils is made up of several *sarcomeres* (cf Fig.2.10) that contain numerous filaments : *the actin* (thin filaments), *the myosin* (thick filaments), *the titin* (elastic filaments), and *the nebulin* (stiff filaments). Actin and myosin represent the contractile part of the myofibrils which are the basic units of the contraction. During a muscle contraction, the lengths of these filaments remain constant and slide past each other to increase their overlap, producing an overall shortening effect in the muscle [NF01, Lee10] as illustrated in Fig.2.10. Regarding the length change and force level, we find two type of contraction :

- The *isotonic* contraction, in which the muscle length changes while producing force (shortening for *concentric contractions*, and lengthening for *eccentric* contractions).
- The *isometric* contraction, in which the muscles length remain unchanged while producing force.

Each muscle is composed of bundles of fibers (*the fascicles*) surrounded by a fascia of fibrous connective tissue called the *epimusium* (cf Fig.2.10). Epimusium connect the muscle with the tendon. The fascicles are separated by a layer of connective tissue known as the *perimysium*. The muscles fascicles form the contractile component and the tendon form the series elastic component of the motor units of the musculoskeletal system. All the produced forces are transmitted to the bones via the tendons. Tendon connect muscle to bone either at a narrow area or over a wide and flattened area : *the aponeurosis*. Within muscle, the *sarcomere* represent the functional unit of the contractile system. All the events that take place in one sarcormere are

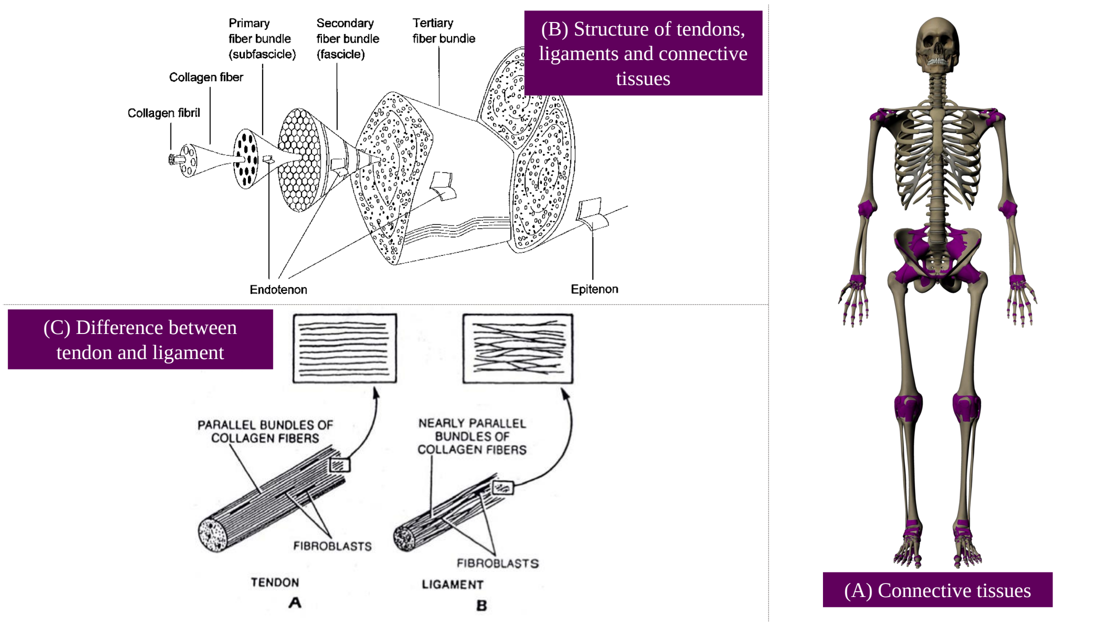


FIGURE 2.8 – *Ligaments, tendons and connective tissues description. (Image (B) and (C) extract from [NF01])*

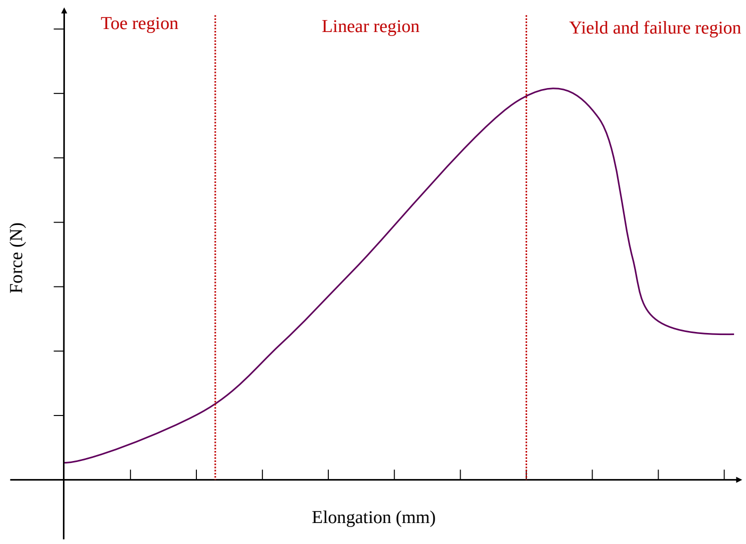


FIGURE 2.9 – *Ligament and tendon mechanical behavior.*

duplicated in the others [NF01]. The functional unit is the motor unit, and includes a single motor neuron and the muscles fibers. Each motor unit may contain less than a dozen muscle fibers. In a large muscles (*e.g : the gastrocnemius*) that perform coarse movements, motor unit contain between 1000 to 2000 muscle fibers.

In the process of understanding the skeletal muscle physiology, it is indispensable to observe its architecture. The muscle architecture describes the internal arrangement of fascicles within this complex organ, it involves the angular orientation of the fibers bundles with respect to the aponeurosis, this orientation is known as the *pennation angle*. More precisely, the pennation angle represents the angle between the muscle tendinous attachment and the longi-

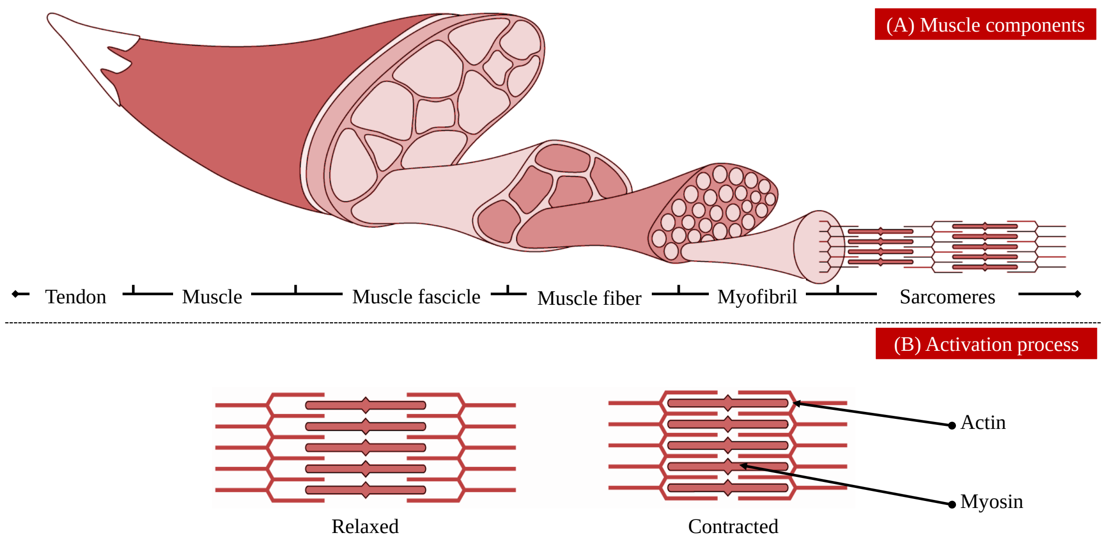


FIGURE 2.10 – *Musculoskeletal system composition. (Image extracted from [Lee10])*

tudinal axis of the muscle fascicle [NTH01]. Depending on to the pennation angle, we find a wide variety of muscle type, all shown and detailed in Fig.2.11. This angle influences the force intensity transmitted to the bones [Zaj89].

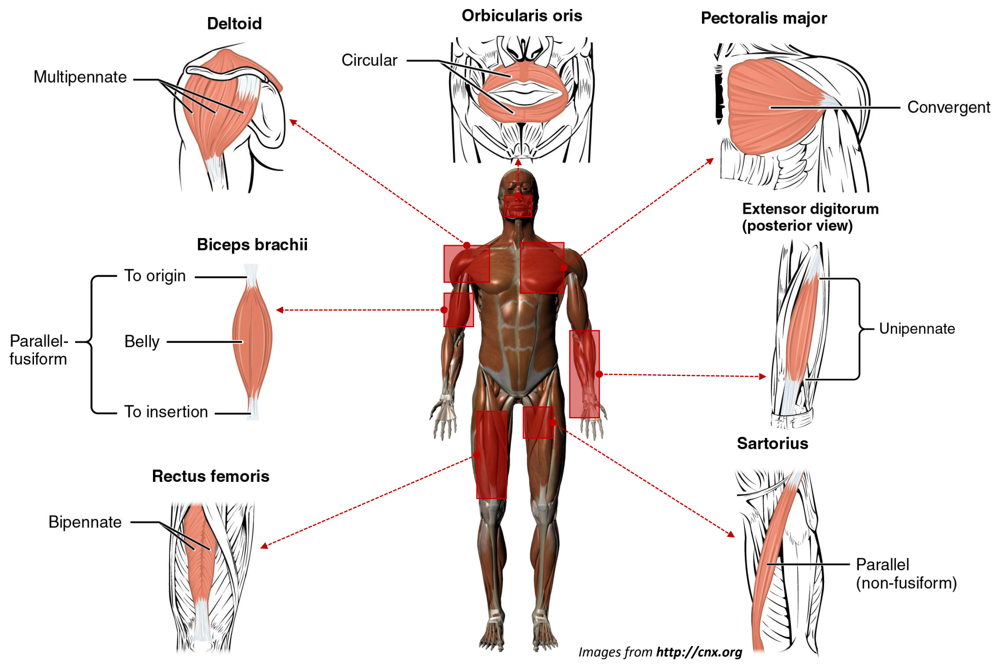


FIGURE 2.11 – *Musculoskeletal system and muscle architecture*

Concerning the mechanical behavior of muscle, the force components within a fiber are modeled from the measurement of isolated muscle fibers, which directly reflect the non-linear properties due to the sliding filaments [Hil38, Zaj89, Sch93, The03b]. Two fundamental functional properties : the force-length and the force-velocity laws (cf Fig.2.12) are established and frequently considered into a variety of biomechanical models to study muscle function. Pennation effects scale the forces generated by a fiber [Zaj89]. Aging affects muscle properties and muscles forces [The03b]. When a whole muscle is stretched or shortened to different length (force length property represent in red and blue in Fig.2.12), the muscle generates a non-linear elastic force consistent with the sliding filament theory of muscle contraction [Lee10]. With no activation, only the passive restorative force against increased stretching is produced. With activation, the sum of both active and passive force are generated. The force-velocity property of muscle describes the relationship between the velocity at which muscle shortens and the amount of force it produces (cf Fig.2.12). When there is no load on the muscle, the maximum velocity of shortening is experienced, and as the external load increases, the velocity of shortening decreases.

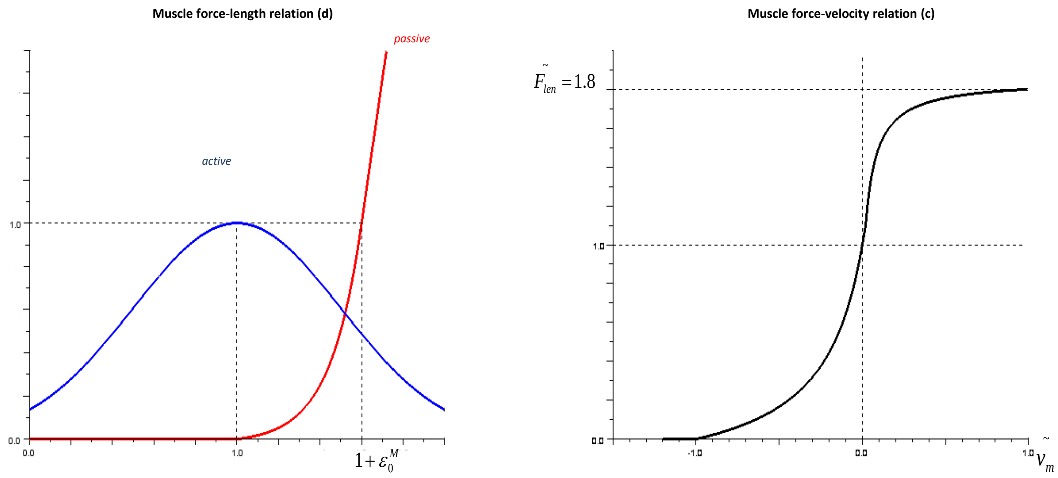


FIGURE 2.12 – *Muscle fiber stress-strain curve*

2.1.5 Inter-vertebral discs

In addition to muscles, ligaments, tendons, and cartilaginous tissues, there are other complex soft tissues which do not fit to any of the cited categories : *the inter-vertebral discs*.

Inter-vertebral discs are composed of four parts : the *annulus fibrosus*, the *nucleus pulposus* and the two *bony endplates* that link it to the vertebrae (see Fig.2.13). They are designed to bear and distribute loads, and to restrain excessive motion of the vertebrae. They are well suited to this dual role thanks to their location between the two vertebrae, and also because of the unique composition of their inner and outer structures.

The inner portion of the discs, the nucleus pulposus, is a gelatinous mass located in a position that varies from one segment of the spine to another. On the cervical segment, the nucleus is really small and is located on the middle portion of the disc. On the thoracic segment, the nucleus is bigger than on the cervical part, but it remains located in the middle portion. The nucleus in the lumbar spine is found on the posterior part of the disc, it is gradually larger as we get closer to the sacrum. The nucleus consists of almost 90% water in young individual, this water content is higher at birth and decreases to 70% as the disc degenerates with age. The rest of the nucleus consist of proteoglycan and type II collagen. This composition enables the nucleus to absorb compressive forces.

The nucleus is surrounded by the annulus fibrosus which is composed of several layers of fibrocartilage which are essentially made of collagen fiber (40% of type II collagen, 60% of type I collagen). The crisscross arrangement of the coarse collagen fiber bundles within the fibrocartilage allows the annulus fibrosus to withstand high bending and torsional loads. As the disc ages, the collagen undergoes irreducible cross-linking and the amount of type I collagen increase, replacing type II collagen. This change contributes to weaken the disc in the same way as the amount of water decreases as the age rises.

Above and below the annulus and nucleus, there are the bony endplates, composed by hyaline cartilage. Bony endplates links the disc with the vertebral body.

Inter-vertebral discs exhibit viscoelastic properties and hysteresis. Age contribute to degenerate the discs and lead them to be less viscoelastic in consequence to change in each part of this organ (*i.e the reduction in the amount of water within annulus and nucleus*).

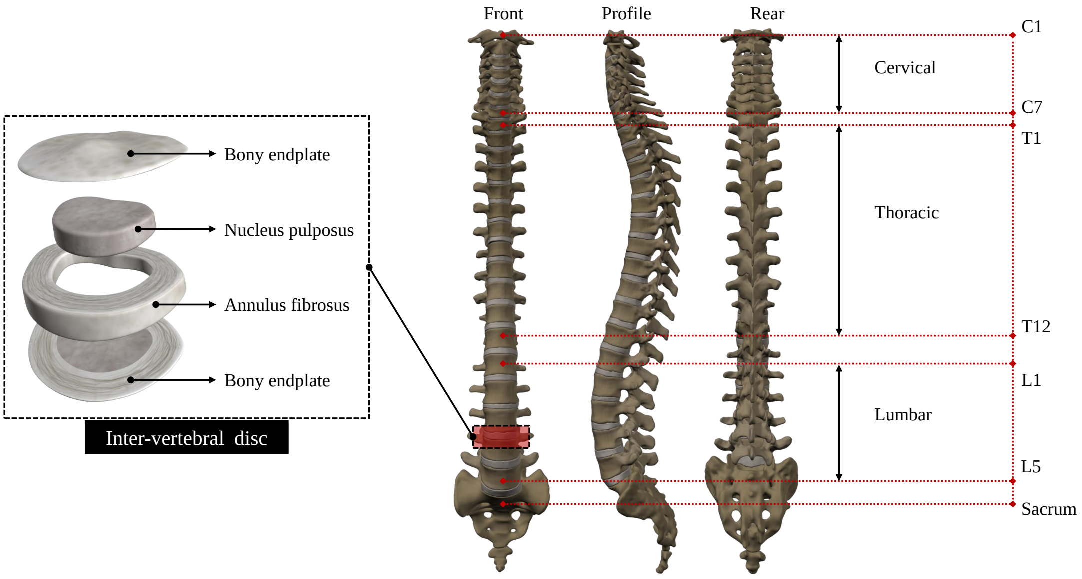


FIGURE 2.13 – *Spine model and disc composition.*

2.2 Modeling and simulation of the musculoskeletal system

The modeling and simulation of the human body has been a subject of much interest. This attention has produced a cornucopia of methods which are as diverse as the subject areas from which they arise : biomechanics, ergonomics, diagnosis, treatment planning, visualization, graphic, robotics, learning help, etc. Consequently, a wide variety of models have been proposed. Each of these works aiming for a different purposes and are consequently associated to different quality criteria.

The quality criteria represent the properties that are important to efficiently achieve the simulation goals. These criteria essentially include the accuracy, the computation speed (*performance*), the robustness, the visual realism and the ease to setup models considering various specific aspects.

We separate the literature models in three categories : those which address problems in medical and biomechanic area, those which address problems in animation and movie industry, and those whose the sake is to support the learning process.

The remainder of this section is as follow. Each subsection review the literature models of the main domains cited above. The quality criteria related to each of these domains are first described before giving further details of the musculoskeletal models proposed.

2.2.1 Musculoskeletal systems for biomecanics and medical area

The modeling and simulation of the musculoskeletal system for the medical and the biomechanic area aim to fulfill mainly the criteria of *accuracy*. Given that the goal of simulation is to reproduce real phenomena, the validation of these phenomena with the highest *accuracy* is essential since the objective is predict new states based on the simulated model. The purposes often involve health of an individual or important financial issues hence the interest in *accuracy* and validation of simulation against experimental or theoretical input data. Thus, these criteria are often chosen instead of computational efficiency or visual realism.

With this goal to reproduce the real behavior of musculoskeletal structure, the earliest mathematical model of skeletal muscle is suggested by Gasser and Hill in 1924 [HG24] to capture global muscle mechanical properties illustrated in Fig.2.12. This model is used to describe the macroscopic relationships between muscle actuation, force-length and velocity [Hil38] , along the muscle path. The force components are modeled from the measurement of isolated muscle fibers, which directly reflect the non-linear properties due to the sliding filaments, while the series elastic elements can be grouped with the tendon and removed from the model. This representation of muscle fascicles consist in three major components : *the series element (SE)* which represents the tendon elastic effects, the *passive element (PE)* which represents the passive elasticity of the muscle fibers resulting from the connective tissues within the muscle body, and *the contractile element (CE)* which accounts for the generation of active force that depend on the muscle length, the neural signal and the contraction velocity (cf (1) on Fig.2.14). This model is improved by Zajac et al [Zaj89] to a dimensionless aggregate model which can be scaled to approximate subject-specific musculo-tendon units and includes pennation effects (cf (2) on Fig.2.14). Schutte improves Zajac's model and adds a damping parameter into the force-velocity relationship [Sch93]. Thelen [The03b] adjusts the parameters to account for changes in muscle properties that occur with aging.

On the same way of representing muscles fibers using complex 1D spring, [Hux74] proposes an alternative to the Hill muscle model. This other representation shows a more accurate

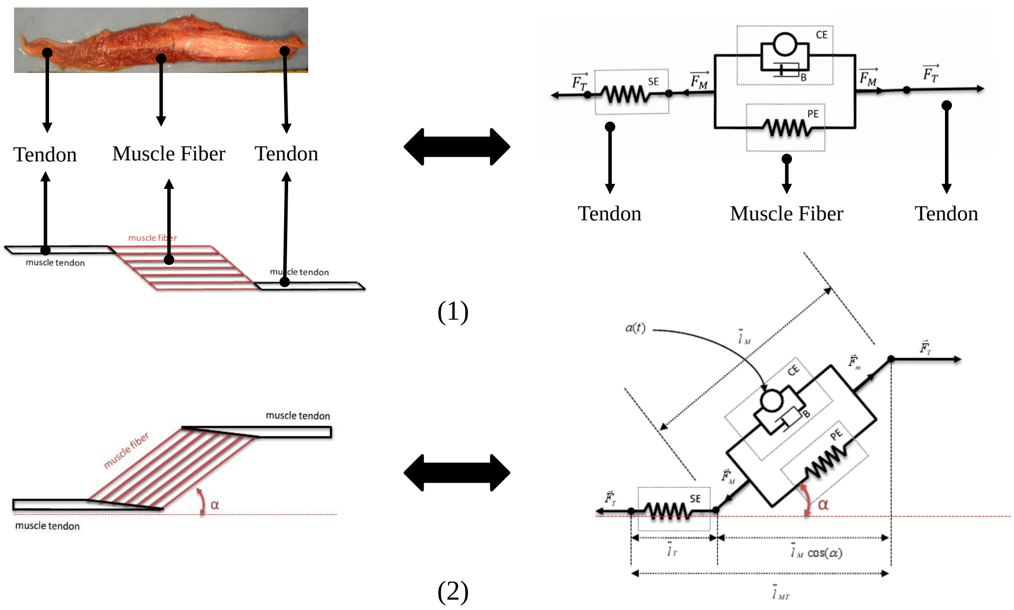


FIGURE 2.14 – *Mechanical representation of muscle : (A) Hill muscle model, (B) Zajac muscle model*

representation of the active behavior of muscle fibers, but remains less simple to use and more expensive in term of computation time than the Hill model [SRH*08]. These 1D muscles commonly named *Hill-type muscles* are widely used on study that involved the understanding of the musculoskeletal system mobility. Through [PD96], Piazza et al demonstrates that this representation of muscles can be used to understand the complex behavior of musculoskeletal structures during daily activities. They develop a model of the right lower limb with a simplified skeleton represented by a rigid body system, and actuators modeled as Hill type muscles (cf (1) in Fig.2.15). This model is introduced to examine the roles of muscles in determining swing phase knee flexion. They achieve this goal with a muscle-actuated forward-dynamic simulation of the swing phase of normal gait. In [TAD03], Thelen et al shows that Hill type muscles can be used to evaluate muscle control (cf (2) in Fig.2.15). They use static optimization with feedforward and feedback controls to drive the kinematic trajectory of a musculoskeletal model toward a set of desired kinematics. They prove the efficiency of their contribution by computing a set of muscle excitation that drive a 30-muscle, 3-degree-of-freedom model of pedaling to track measured pedaling kinematics and forces. While these model [PD96, TAD03] do not include all the fascicles of the lower limb and assume a simplified model for their studies, [AWLD10] introduce a more sophisticated model of the right lower limb with a representation of each fascicles represented as Hill muscle model and attached to a skeleton modeled as a rigid body system (cf (3) in Fig.2.15). Muscles trajectories are constrained by wrapping surfaces [BAG99] for a more accurate evaluation of muscle length. This model has been built based on data extracted from 21 cadavers and aims to the analysis of the human movement.

Although one-dimensional abstractions are sufficient in many applications, the complex muscle architecture, large muscle-to-bone attachment areas, as well as muscle contacts, can only be simulated using three-dimensional models. The most widely used technique is to subdivide organs into elements and to apply continuum mechanics, typically the finite element method (FEM). Chen and Zeltzer in [CZ92] model individual muscles as coarse linear finite elements and apply a Hill-type model to approximate their constitutive behavior. Ng-Thow-

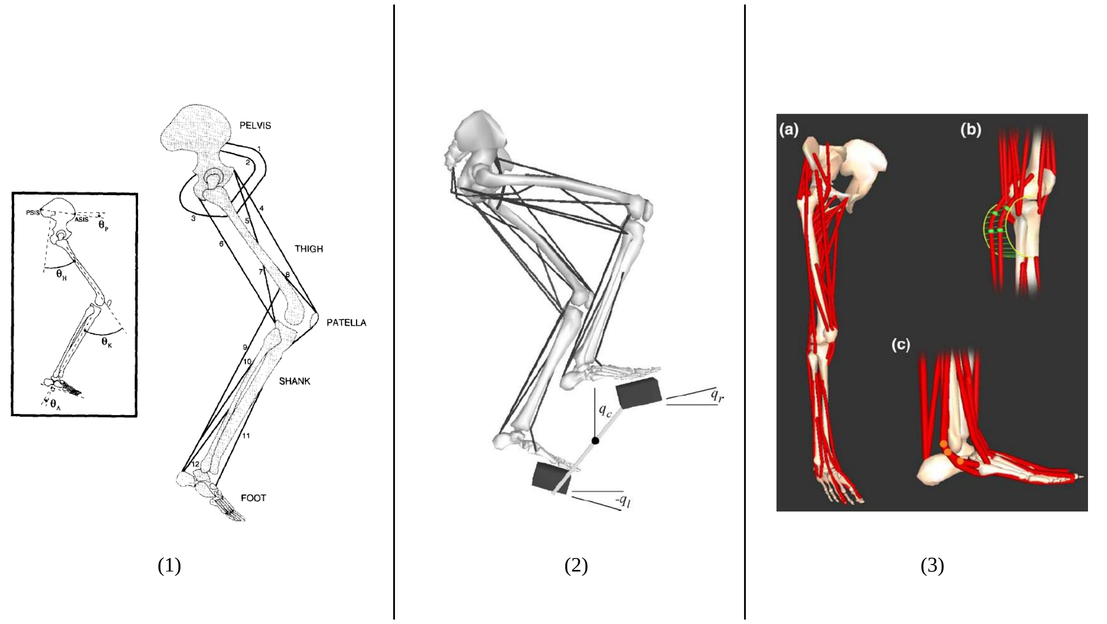


FIGURE 2.15 – *Use of Hill muscle models : (1) Right lower limb model develop by [PD96]. (2) Lower limb model for muscle control and analysis during pedaling by [TAD03]. (3) Detailed model of lower limb for movement analysis introduced by [AWLD10].*

Hing in [NTH01] chooses to model muscles as B-spline solids, with embedded Hill-type one dimensional elements. Blemker and Delp [BD05, BAGD07] develop and evaluate a new formulation for creating three-dimensional finite-element models that represent complex muscle geometry and the variation in moment arms across fibers within a muscle (cf (1) in Fig.2.16). This 3D muscle model has the advantage to represent complex muscle path motion without via points [DLH*90] or wrap surfaces [BAG99], but remains computationally expensive and impractical to use in real-time applications. More recently, [YB14] introduce a novel approach for simulating 3D muscles with complex architectures (cf (2) in Fig.2.16) with the aim to get the best model formulation in terms of computation cost and accuracy. Their approach consists in mixing different models for each component of the muscle. The passive behavior of volumetric tissue within the muscle is modeled with 3D finite element model (3D FEM). The active behavior and the fiber orientation are included in the model with oriented Hill contractile 1D element. All these active fibers are inserted on a membrane model accounting for aponeurosis tissue and simulated with 2D FEM. The separate models are mechanically binded using barycentric embeddings.

These volumetric muscles model are used in various musculoskeletal system to address numerous problems. The muscle model introduced by [NTH01] contributes to simulate skin deformation, and took into account the effect of fat layer, dermal layer, and musculature, in a finite element framework. Teran *et .al* [TEB*05] use a finite volume method (FVM) with quasi-incompressible, transversely isotropic, hyperelastic constitutive model to simulate muscular tissues contraction and deformation. B-spline solids are used to model fiber directions, and the muscle activation levels are derived from key-frame animations of a simplified system including rigid bones and Hill muscles (cf Fig.2.17). This model is built to reproduce the biomechanical behavior of the upper limb during a specified movement given in input. A framework for the construction of a 3D simulation model based on MRI data is proposed [ACS*09], focusing on the hip model subject and imposed bone skeletal motion.

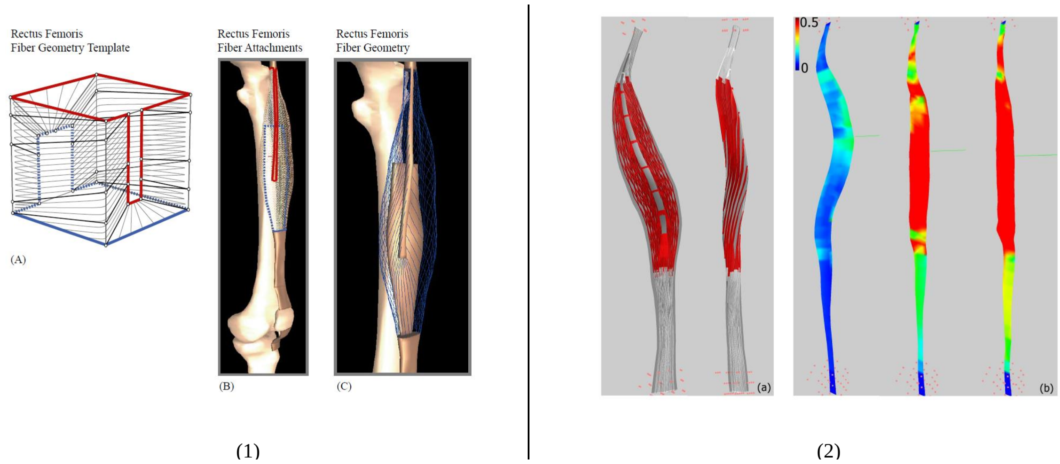


FIGURE 2.16 – *3d muscle models with complex architecture : (1) 3d muscle model based on MRI data and developed by [BAGD07]. (2) 3d muscle model that combine 1D contractile element, 2D FEM for the aponeurosis, 3D FEM for the passive effects of connective tissues within the muscle [YB14].*

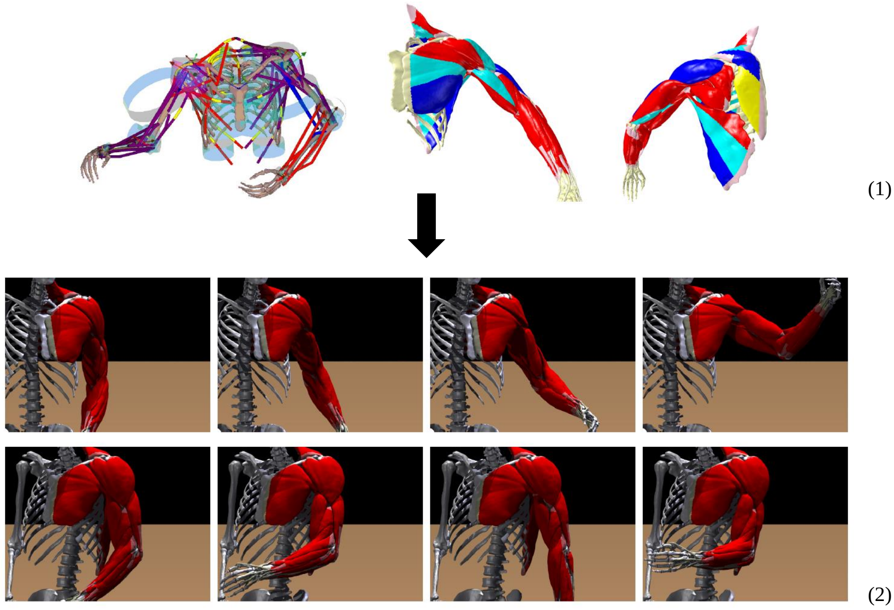


FIGURE 2.17 – *Model of left upper limb proposed by[TEB*05].*

The idea of using musculoskeletal systems to assist medical practice is not a recent concept, in 1998, [Kee98] introduce one of the first deformable model of facial tissue for craniofacial surgery simulation (cf (2) Fig.2.18). To achieve this task, they develop two different deformable tissue models into a surgical simulation system. The first one is an interactive tissue whose state is affected by the surgeon technique and that gives him the ability to realize nearly real-time simulations of the resulting soft-tissue changes in order to improve his planning process. The second tissue model is a more precise one which considers the exact physical properties of the facial soft tissue and used offline to verify the surgical procedure. Both simulations are validated using preoperative and postoperative MRI of skin. More recently, Stavness *et al.* [SLPF11] introduce an original 3D dynamic model for jaw-tongue-hyoid complex, consisting of an FEM model of the tongue, rigid jaw, and hyoid structures, Hill-type muscle actuators, constraints for bite contact and the temporomandibular joint. This model based on patient specific data has the novelty to combine different types of degree of freedom in a same model for a better efficiency. Physical models based on a finer discretization, allowing non uniform deformation along the fiber, have been recently proposed in [SKP08]. Beyond bones and muscles, [SKP08] demonstrate an impressive model including detailed bones, joints, skin, and tendons. The deformations of the skin due to the tendon actuators dramatically improve the resulting quality as shown in (2).(b) and (2).(c) in Fig.2.18. The windpipe is visible in an increasing number of feature animation characters, and the veins increase the realism of the skin. Schmidt *et al.* build a non linear FE model, with the parameters provided from a calibration process and validated with *in vitro* data [SHD*07]. This model is later used to understand several aspects of the behavior of lumbar spine units, such as the relation between ICR and facet joint force [SHCW08], the load combination that leads to a risk of disc prolapses, and the phenomenon of disc failure [SKH*07]. Through these types of work, they demonstrate how simulation can help the understanding of complex anatomical phenomenon. Shin et al. [SLK07] propose a finite element model of three lumbar spines segments L2-L5 to show that dynamic stabilization devices restore functionality of a spine with intact disc. The disc of the first segment is intact, while the second segment is a fused spine with a fixation device and the third segment is stabilized using a dynamic stabilization device. This study allows to understand the consequences of these two fixations device on the range of motion and on the adjacent disc of a segment during the long term use.

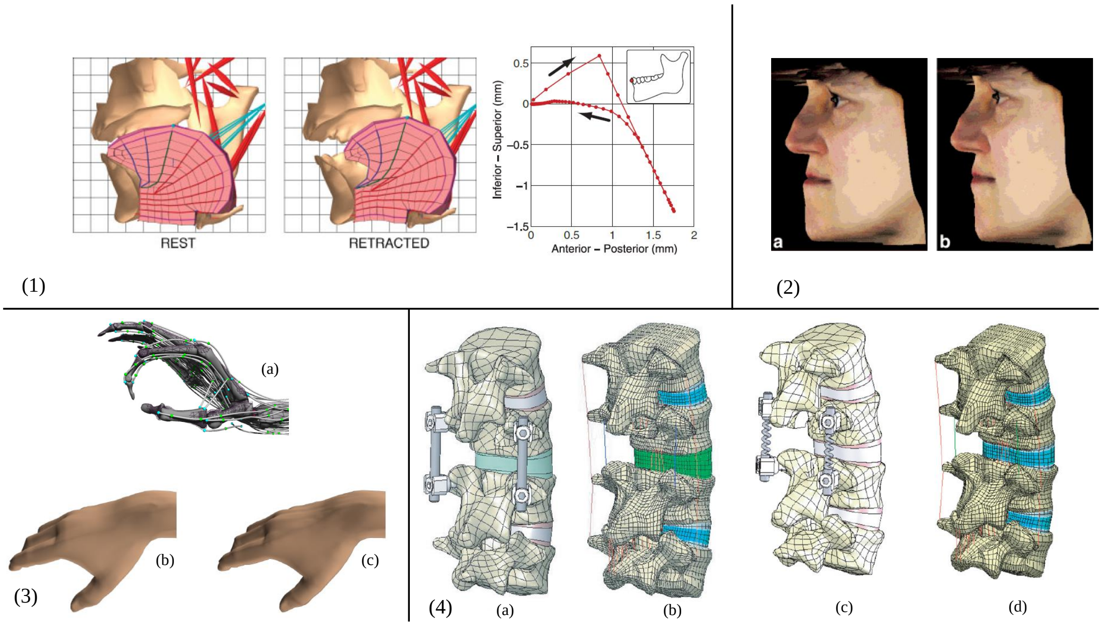


FIGURE 2.18 – *Examples of musculoskeletal systems for biomecanical purposes : (1) 3D dynamic model for jaw-tongue-hyoid from [SLPF11]. (2) Deformable model of facial tissue for craniofacial surgery from [Kee98]. (3) Hand model developed by [SKP08]. (4) Lumbar spine models for the [SLK07] study*

All these models representing musculoskeletal structures and systems have been modeled to reproduce the behavior of anatomical systems, with various types of components (*such as 1D elastic element, 2D and 3D FE model … etc.*) and for various purposes. Despite these differences, they all show some common limitations in addition to their inherent complexity in their construction. Accurate musculoskeletal simulations are achievable using detailed three dimensional representations and contribute to get high quality simulation, but at the cost of computational speed (*e.g. the method of [TEB*05] runs at 4 minutes per frame*), making these methods impractical in many real world applications. Moreover, heterogeneous models can not be easily combined or reused in different contexts, as they are not integrated into a common framework allowing for the modular design of simulation scenarios. Although one-dimensional model can fulfill various needs, they can not represent the whole reality of musculoskeletal system structures (*e.g the contacts between adjacent muscle fascicles*).

2.2.2 Musculoskeletal systems for animation

In computer animation, the modeling and simulation of musculoskeletal systems are increasingly used to obtain more realistic animation (e.g humanoid animation, reproduction of facial expression, morphing ...) and to create special effects (e.g tearing and cracking bodies). While in the medical and biomechanical area, the main goals of simulation are to reproduce and validate anatomical phenomenons against specified input data, computer animation focuses on modeling essentially the visual effects of these same phenomena. For instance, the muscle bulging or swelling during muscle contraction, which is the key of underlying factors for a better skin deformation. The main quality criteria in this domain are the *visual realism of the simulation*, the *computation efficiency* and the *easiness to set up simulation scenarios*. *Visual realism* remains the most important criterion because the purpose of an animation is first of all to make people believe and appreciate what they are watching. Equally important, *computational efficiency* and the *ease to set up simulation* contribute to accelerate the process of creating an animation, while allowing artists to exploit all their skills without considering the complexity inherent to the anatomical systems and the modeling tools. To perform simulations while respecting these criteria, a wide variety of tools and techniques have been established in computer graphic research. They rely on geometrically-based approaches and physically-based approaches.

In computer animation, skeletons are traditionally simplified into acyclic graphs, called animation skeletons or stick figures, composed of one-dimensional segments (limbs) connected by joints. Skeleton poses are then compactly expressed as a set of rotations along the most relevant joint degrees of freedom (i.e., reduced coordinates), which can be easily manipulated by an artist or animated based on real data such as motion capture. For visual applications, additional realism can be achieved using a boundary discretization of organs. Thus, these skeleton-based models are usually combined to other models to reproduce anatomical systems behavior. They have been used to control the motion of the human body or its interaction with objects, see e.g. [FvdPT01, ZMCF05] for full body, and [PZ05, KP06] for the hand. In [BS91], Bloomenthal et al use convolution surfaces to model the human hand and arm by approximating bones, muscles, tendons and veins close to the underlying skeleton. [CHP89, MT97] employs FFD (Free Form Deformations) to represent muscle deformation. The articulated skeleton transforms a surrounding FFD lattice, which in turn represents a muscles shape change. [SNTH03] uses surface bone models to generate more realistic skeleton animations based on possibly cyclic graphs and moving joint centers. Procedural methods can then be used to derive organ motion from the pose of an underlying animation skeleton (i.e. *Skeleton subspace deformation*, also knows as *vertex blending* or *skinning* [MTLT88, KvO07]). Baran et al [BP07] present a method for automatic rigging of character skins without internal anatomy, except for automatically inferred animation skeleton. In [KS12], they improve the skinning by introducing the concept *joint-based deformers* which allow to reproduce the elastic effects of physically based skin model (cf Fig.2.19.2), while using mathematical approach, with better computation time and an easier setup. Their key idea is to decompose the joint rotation into a twist (torsion) and a swing (bending), and 2D rotation interpolation is used for the twist, and the swing is blended linearly. This decomposition contributes to remove artifact of previous skinning technics. Geometrically-based techniques are employed because they are practical and efficient in terms of implementation of computation time [Lee10].

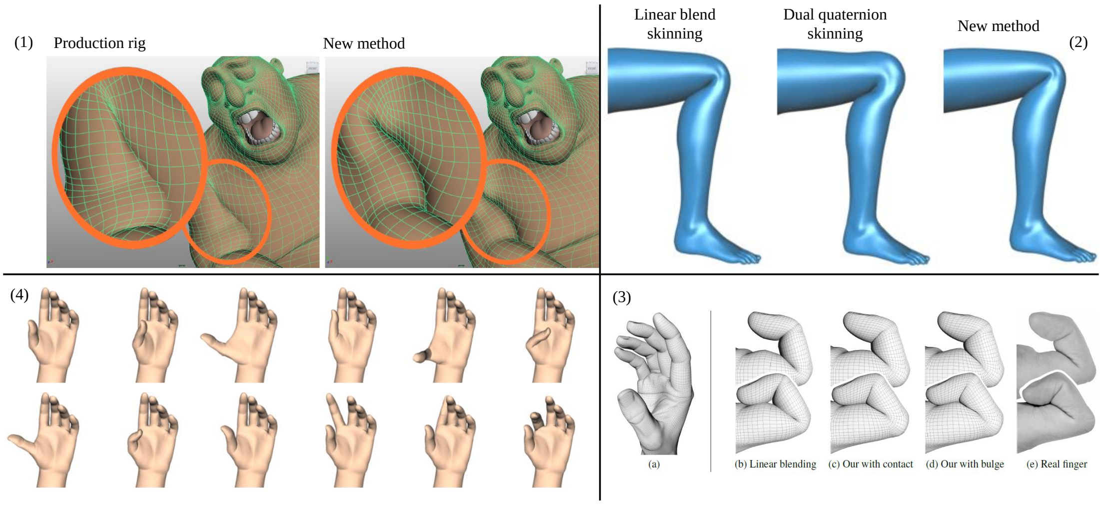


FIGURE 2.19 – *(1) Illustration of the effects of elastic skinning versus classical skinning (image extracted from [MZS11]). (2) Correction of artifacts of dual quaternion skinning and linear blend skinning using the new skinning approach introduced in [KS12]. (3) Illustration of the effects of implicit skinning versus classical skinning and how it allows to reproduce realistically effect of bulging on anatomical system in action (image extracted from [VBG13]). (4)Realistic skin deformations obtained on the thumb and the index finger with the "eigen skin" [KJP02].*

Musculoskeletal models have been proposed to animate muscle deformations [WVG97, AT01], to perform facial animation [Wat87, SNF05], to study or improve the control [LT06, WSP10, WHDK12], or to increase the quality of the flesh and skin deformations [LST09, SKP08]. Already in 1997, [SPCM97] introduces a framework to shape and to simulate human muscularity and its effect of skin with geometrical approaches. The skeleton consists of a hierarchical system of bones and joints where each bone is represented by simplified rigid surfaces, and moves relative to one another. Skeletal muscles composed by a contractile belly and two extremities : insertion and origin are modeled by an ellipsoid for the muscle belly. During animation, the muscle volume is conserved by an automatic adjustment of the muscle belly dimensions when its extremities move. The skin is produced by blending the individual primitive (muscles) represented by an implicit surface. The radius of influence of these implicit functions allows different thicknesses of fatty tissue under the skin. More recently, [VBG13] introduces the first purely geometric method handling skin contact effects and muscular bulges in real-time skin deformation framework as illustrated in (2) in Fig.2.19.3. They use implicit surfaces as underlying tissue, the advantages of blending functions are exploited to avoid several problems resulting from other skinning techniques, while reproducing better skin deformation in consequence to the flesh state. While geometrically-based models have proven to be sufficient for some graphical application demanding visually acceptable quality, their inherent simplicity and the need for the human intervention often makes them difficult to extend to complex scenes involving dynamics. Furthermore, they lack the physical and biomechanical accuracy often required for realistic modeling and simulation. To overcome these deficiencies, many works have turned to physically-based approaches in which physical simulation is employed to solve complex interactions involving anatomical structures. In [KHYS02], they propose a physical method based on shell elements to simulate thin muscles on an animated facial expression model. In [KJP02], they introduce the eigen skin which uses Principal Component Analysis to construct an error-optimal eigendisplacement basis for representing the potentially large set of pose corrections. The result is a stunning hand which is deformed in real-time, with nice rendering, while reproducing the bulging due to underling tissues (cf Fig.2.21.4). Lee et al [LA07] introduce a novel system for real-time skin deformation for articulated characters. The muscle shape is a function of an animated quadratic Bezier action curve and control rings derived from it. A single spring within the muscle shape drive its dynamic. The smooth-skinning metaphor is used to blend skeleton-driven deformation with muscle-driven deformation. A two-layer skinning approach is used to animate and add jiggling effects to the character skin, this skinning considers joints and underlying muscles states. McAdams et al [MZS11] introduces a novel interactive simulation of skeleton driven high resolution elasticity models of skin with a framework accounting self collisions and soft-constraints, the whole with new robust simulation algorithms. They demonstrate the efficiency of their skinning method on the animation sequence of a CG character playing piano, where the muscle bulging, the contact between arm and torso, and the skin elasticity are handled for a higher quality animation as illustrated in (1) in Fig.2.19.1, this sequence is used on the Disney movie *Tangled*. Along these lines, Faure et al. show that frame-based deformations with material-aware shape functions allow efficient simulation of anatomical system as it is demonstrated through a simulation of knee during its flexion [FGBP11]. Lee, Sifakis and Terzopoulos [LST09] introduce one of the most highly detailed biomechanical model of the human upper body composed of a dynamic articulated skeleton, numerous Hill-type muscle actuators which include the force-velocity relation, and realistic finite element simulation of soft tissues and skin deformation. They use target poses and co-activation as input to compute muscle activation using inverse dynamics. The activation is then used to simulate skeleton dynamics and soft tissue deformation. The skin and underlying soft tissues are also simulated using FEM. Although [LST09] appears in the biomechanical section (Sec.2.2.1) as [TEB*05, SKP08], we include it in this section to illustrate that these types of work with such quality in visual result are legitimated and sufficiently plausible to serve the movie and animation industry. Fan et al introduce an novel active volumetric muscle in [FLP14] illustrated in Fig.2.20.4. They use an Eulerian-on-Lagrangian discretization that handles volume preservation, large deformation, and close contact between ad- jacent tissues. Muscle activation model uses knowledge from the active shapes of muscles extracted from medical imaging data.

All these technical or types of approach are widely used in the creation of movie, animation. They are mainly integrated in the process of creation of special effets such as bone fractures or ventral tear as illustrated in (3) of the Fig.2.21 which is extracted from the movie *Immortals*, the creation of fictive humanoid character as Gollum in *The lord of the ring* or the Na'vi in the movie *Avatar* (cf (1) of the Fig.2.21), and the creation of photo-realistic animation such skin moving in function of underlying muscles and bones deformations as shown in the animation of *Hulk* arm illustrated in (2) of the Fig.2.21. If all this work address issues related to the movie industry, they are often hard to integrate in an animation pipeline or usual artist tools. Furthermore, they remain tedious and hard to be manipulated by an artist in addition to complexity of the required knowledge to create plausible anatomical systems.

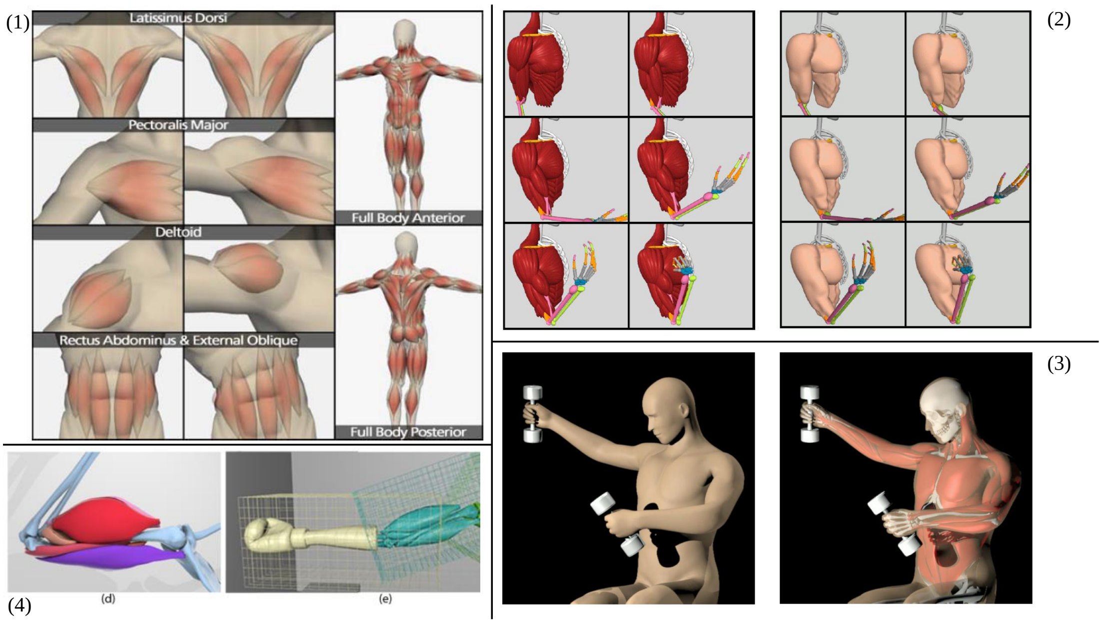


FIGURE 2.20 – *(1) Musculoskeletal system introduced by [LA07] to deform and to add secondary effects on skin during a specific movement sequence. (2) Upper limb muscularity and skin deformation deduced from the muscles state (images extracted from [SPCM97]). (3) Skin deformation based on underlying tissues deformations and animation key frames given as input (images extracted in[LST09]). (4) Active volumetric muscles in action [FLP14]*

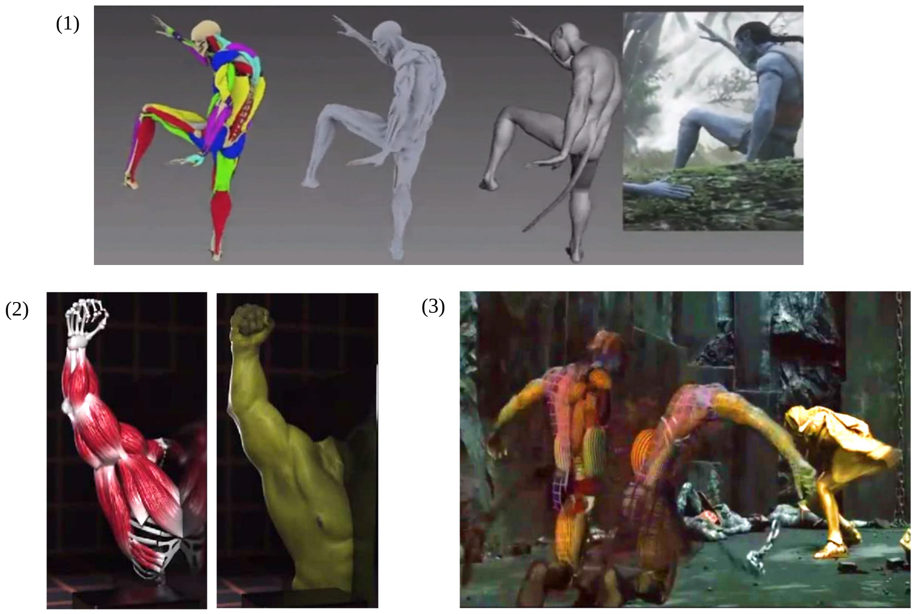


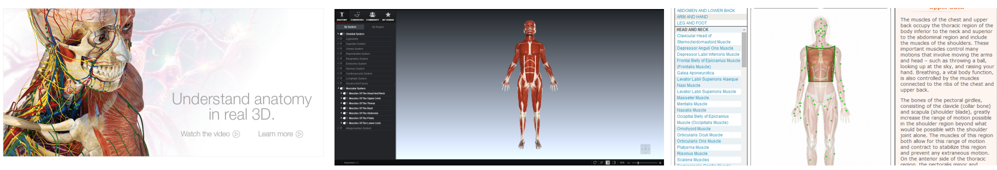
FIGURE 2.21 – *(1) Images extracted from the « making of » of « Avatar » to show how the musculoskeletal system and the other underlying tissues are made to produce such a high quality animations of the Na'vis during the whole movie. (2) Illustration of bulging and secondary effects produced on « Hulk » right arm during an animation sequence. (3) Image extracted from the « making of » of the movie « Immortals » to explain how the body tearing is realized using physically-based approaches.*

2.2.3 Musculoskeletal systems for learning

Learning anatomical structures, functional anatomy and musculoskeletal system kinematics is indispensable for medical studies, sport education as well as for education more generally. With the technological advances and the spread of devices such as smartphones, tablets, or even laptops which are more accessible today than twenty years earlier, a high energy has been used to find novel methods to improve and ease the learning. To fulfill these new needs and support learning process, a wide variety of innovative approaches to build and simulate musculoskeletal systems that meet quality criteria related to the education field have been proposed.

These quality criteria are : *the visual realism*, *the interactivity*, *the computation efficiency*, and more generally the *sensory feedback* that reflects reality (*visual, auditory, force feedback*). Simulation with the right *sensory feedbacks* contribute to reproduce the identical conditions of those of the medical practice. It offers obvious benefits to novices to learn invasive procedural skills, especially in a trend of decreasing clinical exposure. The *visual realism* enables an easier identification and comprehension of phenomenon or organ if their modeling and simulation reflect the reality. The *interactivity* as well as the *computational efficiency* further to act or to practice medical gestures in comfortable conditions to keep the student focus on the procedures. In the process of learning anatomy, two complementary aspects are found : *the theory aspects* and the *practice of medical procedures* (surgery, suture, repositioning of organs. . . etc).

Learning theory aspects involves observation coupled to the description of the anatomy and the physiology of organs and systems. However, this learning aspect remains tedious and difficult because of the complexity of knowledge that includes a large amount of structured in addition to static and dynamic notions. An important difficulty in the way of learning anatomy and physiology comes from the fact that it is still taught through static objects provided from books atlases, drawings of board, photos, or from dissection of dead bodies [GCB*07]. Understanding the musculoskeletal system and its mobility usually required 3D visuo-spatial abilities also known as *mental rotation*. More precisely, mental rotation (MR) is the ability to make a mental image of a two-dimensional (2D) or three-dimensional (3D) object turning (or static) in space. A lot of studies have been made on anatomy learning [DRC*06, PR13, KBSC14b, SM08], and the importance of 3d models for anatomy learning has gained momentum. This work highlight and prove through experiments how these new tools improve our perception of anatomical structure, their functions and how they can be used in addition to classical approaches of learning anatomy[HCD*14, HCR*09, GCB*07]. Among the propositions of models, simulations and animations of musculoskeletal system built to support learning, websites stand out : [bio, inn, zyg, hea, ani, vis] (cf Fig.2.22a, 2.22b, 2.22c). Some of them as BioDigital [bio] have developed a library of highly realistic animations and non-interactive models to illustrate procedures for educational purposes. Most of these websites rely on 3D models and descriptions of anatomical entities and systems shown on a web browsers. The existence of these new websites is possible thanks to the WebGL technology (2011). Animations and simulations shown use either geometrically-based method and physically-based methods.



(a) *visiblebody* **(b)** *biodigitalhuman* **(c)** *innerbody*

With the emergence of new technologies as the *Augmented Reality* (AR) systems (*AR helmet, kinect, google glasses*), new approaches of learning through interactions become increasingly used [FBE*13, PR13, KBSC14b]. These technologies ease the learning process and the construction of mental representations of these knowledge. Augmented reality (AR), which consists in superimposing data on visible objects, has been increasingly used in the medical domain [DHLD*02, JBC08, Spe12]. The fundamental idea is to combine or mix the view of a real human with additional virtual content of its anatomy. This virtual content can appeal to different senses such as sight, hearing, touch and smell. One of the best and unique illustration of these AR tools for learning anatomy is the *Magic Mirror project* [FBE*13, BKBN12]. The Magic Mirror allows to create an illusion that the user is standing in front of a mirror, which augments virtual anatomy information to the user's image (cf Fig.2.23a, 2.23b). The user controls it through hand gesture and speech command. It can superimpose an in situ visualization of bones from CT volume and 3D organs on the user's body. It also shows the selected slice image from CT, MRI or photographic volume corresponding to the relative position of the user's left-hand and body. To achieve that, together with orthopedic surgeons they have defined anatomical bone landmarks which users can touch easily on their body while standing in front of any sensor. These landmark positions allow the deformation and interpolation of the medical data correctly within the magic mirror and onto the human body, resulting in a more precise augmentation. In the existing version, they directly use the Kinect device to compute skeleton information, the scale factors and transformation matrix between VKH volume and user-specific torso.

(a) *Magic Mirror project* **(b)** *Magic Mirror project.*

Observational learning has historically played an important role in surgical training, constituting the first step in the time-honored « *see one, do one, teach one* » model [Kne05]. This way of practicing medical procedures involve some explanations of experts with practices on cadavers, animal or sometime real patients. Although medicine students use to learn in this manner, learning in workplace settings is sometimes too risky, difficult to organize, time-consuming and expensive. The complexity of the work environment may also be daunting for the trainee. Educational technology has proven its potential to offer a safe, suitable and cost-effective training setting in which whole, real-world training tasks can be practiced [AR12]. In such controlled environments, learners can make errors without adverse consequences, while instructors can focus on learners rather than patients. One advantage of simulators over simpler inanimate practice models is their ability to provide immediate, objective, automated performance feedback. Virtual reality (VR) training allows to transfer technical skills to the operating room environment [Sey02]. These supplemental training modalities are not meant to replace real operative experience or be used in isolation, but rather must be integrated into existing surgical training curricula. The efficacy of simulation as a training tool in the medical field has been validated in numerous studies [Gra05, Kne05, SVT*11, AR12]. These simulations have greatly improved training efficacy in participants. Among the research works which address this need of proposing tools for training surgical gestures, only few of them focus on the musculoskeletal system. Various works on simulating interaction between surgical instrument and internal organ as the viscera (liver [CDA99, FDD*12, HDP14, HC14], cf (1) in Fig.2.24), the eyes [DCC*13] (cf (2) in Fig.2.24), external systems as the skin [Kee98, Sif09] (cf (3) in Fig.2.24), thin organs as blood vessels [BMLS01] and other soft tissues [KWBBS98, All07, AR12] have been proposed. All these works propose simulation of these anatomical systems with real time interaction, while reproducing the mechanical behavior of the system components, the whole in a stable and robust simulator that allow students to practice in suitable conditions.

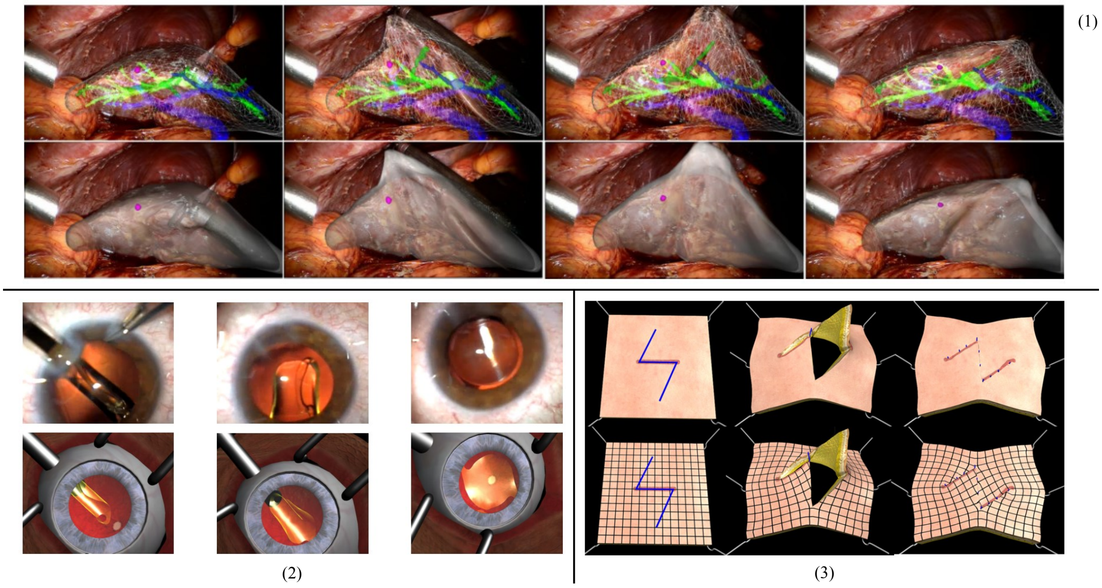


FIGURE 2.24 – *(1) Simulator of minimally invasive surgery on liver with augmented reality made by [HC14]. (2) Steps of the simulation of the intra-ocular lens implant injection and its deployment within the lens capsule (Top : reality. Bottom : [DCC*13]'s simulator). (3) Simulation of a z-plasty procedure for the elongation of a scar contracture (image extracted from[Sif09]).*

Concerning virtual system to practice gestures on musculoskeletal system, already in 1998, [Kee98] introduce an efficient system for cranio-facial surgery based on real patient data that allows to predict the effects of surgical procedures before operation. This work uses FE-method and mass spring system to model and simulate skin, underlying soft tissues and bones (*more details in Sec.2.2.1*). They validate their simulator with pre-operative and post-operative data. In [LSS12], they present an implementation of eyeball and extraocular muscles model for interactive surgery simulator. Their simulator includes the reaction of muscles in response to external force applied from surgical instruments for cataract surgery (cf (3) of the Fig.2.25). The application achieve the visual realism of the reaction of eyeball and extraocular muscle realistically. Tensile stress tests is carried out on human skin samples, and the resulting data is incorporated into a hyperelastic model for real time simulation of skin [LGK10]. This skin model is ultimately included in a simulator which anticipates coupling the software with a haptic feedback device to create a real-time plastic surgery simulation in an interactive virtual environment [LGK10].

One major issue with virtual training systems is the software issue. Several simulators that model plastic surgery procedures are currently available [AR12] but only few remain accessible for the general public. Among them, we find the software SOFA [FDD*12] which has been used for several works on liver, eye and skin [All07, JBB*10, HC14]. BioDigital [bio] develops a cleft lip simulator that can be run on a standard computers (cf (1) Fig.2.25). Their simulator uses data from CT scans to model unilateral or bilateral cleft lips, and allows the user to navigate the anatomy and explore each layer of tissue. The user is able to create and transpose tissue flaps, as the computer can accurately model tissue properties. This cleft lip simulator is based on the works produced by [OC05]. BioDigital system also include simulator for other specialized procedures as a simulator for latissimus [AR12]. This simulator models inter alia the full human anatomy and incorporates motion (a beating heart) and biomechanics (a golf swing) (cf (2) Fig.2.25). In [FDOM10, FDG*10], McCarthy et al has spearheaded the creation of the Interactive Craniofacial Surgical Atlas, a collection of simulators ranging from a frontal orbital advancement simulator to the Le Fort III advancement/distraction [NCvdW*08] simulator. The computer simulations are supplemented with features such as videos of live surgeries, audio voiceover animations, and 3D visualization to better illustrate the procedure and facilitate learning [FDOM10, FDG*10].

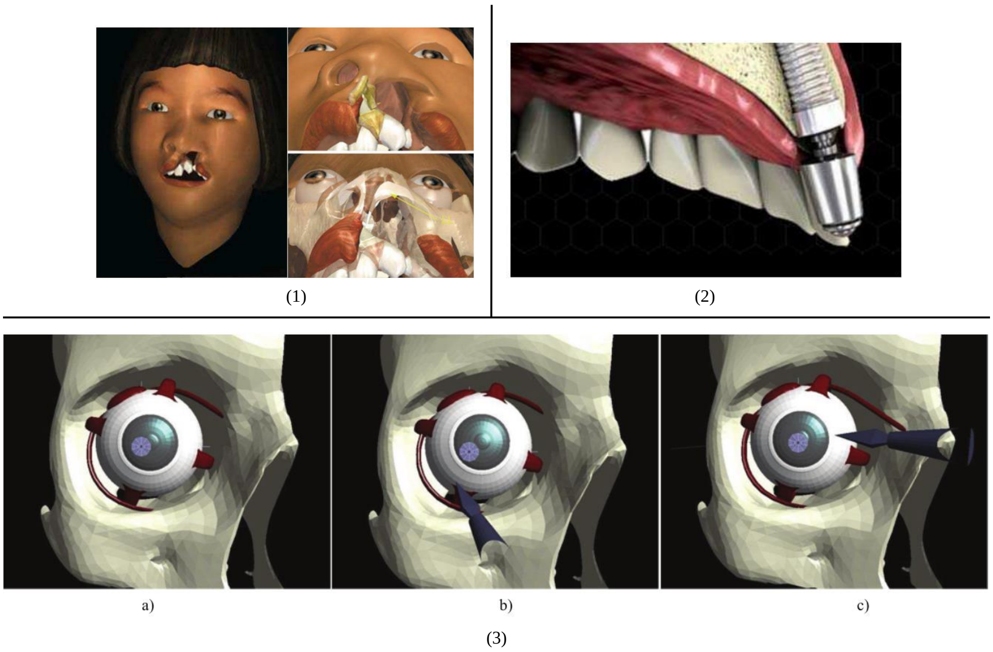


FIGURE 2.25 – *(1)Cleft Lip Simulator Image and 3D graphic animations to illustrate cleft lip and palate surgery (images extracted from [AR12], the model is made by [OC05]) . (2) BioDigital human 3D animated simulator [bio]. (3) Simulation of reaction from eyeball and extraocular muscle during a session on[LSS12] 's simulator.*

Numerous of these works demonstrate the positive effects of these news tools to support the learning process. Although the medical teaching remain traditional, the obvious benefits of the modeling and simulation of anatomical system make that they are more integrated into the learning process. If this can be seen positively, the complexity to build these system, and to create simulation scenarios with these tools remain affordable for only few experts.

2.3 ONTOLOGY-BASED MODELING OF MUSCULOSKELETAL SYSTEM

In this section, we remind and lay the foundation of what it is necessary to know on ontologies. We review these basic knowledge mainly because ontology is not commonly used in the computer graphics field in general. Since one of our contributions is based on the use of this tool, it is thus essential to well understand few points concerning it. These points or issues to address are :

- What is an ontology ?
- The advantages and benefits it brings.
- How creating a consistent ontology ?
- How to query one ?
- And which kind of works and approaches related to ontology have been explored in the computer graphics discipline (i.e biomechanic, animation, support learning . . . etc) ?

This quick survey has mainly its sources in the works of [Gru93, NMO01, RM03, JdB03, Noy04, CSH06, MBK13]. It does not intend to cover all aspects related to ontologies, but only those necessary to the comprehension of this report. For more details on ontology in general or on specific aspects such as *alignment, mapping, . . .*, we refer to the readers these literature works [NMO01, RM03, Smi03, SS10].

2.3.1 Generalities

The handling of huge amount of data represents a major issue in computer science today. Organizing data, efficiently and accurately access to specific knowledge, each of these aspects represent a challenge to tackle, with several consequences in various domains. To address these issues, numerous communities ranging from the AI community to the web semantic community rely on the use of ontology. In the context of this PhD, we use ontology to organize and query on anatomical knowledge with the purpose to construct anatomical systems.

The word ontology comes from the ancient Greek, it consists into two terms "onto" and logia" which respectively mean "*being ; that which is"* and *"science, study, theory"*. It refers to the philosophical study of the nature of being, or reality. This term was used to allude to the shared understanding of some domains of interest which may be used as a unifying framework to solve the above problems in the above described manner [UG96]. Traditionally, it deals with questions concerning what entities exist or can be said to exist, and how such entities can be grouped, related within an hierarchy, and subdivided according to similarities and differences.

More recently, [Gru93] introduces a definition which is still topical today while capturing all aspects covered by the term ontology. He defines *ontology as an explicit conceptualization of a domain* [Gru93]. If in computer graphic, ontology is little used, this is far from the case for disciplines such as Artificial-Intelligence (AI), Web Semantic or Library science. Whereas its use is more common in these fields, their literature contains many definitions of ontologies ; and many of these contradict one another. In the AI discipline, ontology is defined as a formal specification of the concepts of an interest domain, where their relationships, constraints and axioms are expressed, thus defining a common vocabulary for sharing knowledge.

Since this definition fits most of our needs, for the context of these research works, we prefer the definition of ontology proposed by [NMO01]. They define ontology as a formal explicit description of *concepts* in a *domain* of discourse (*this includes the relationships between concepts*), *properties* of each concept describing various features, *attributes* of the concepts, and *restrictions* on slots. *Concepts* are often called *classes*, *slots* are also known by the terms *roles* or *properties* and *restrictions* can be substituted by *facets, axioms or constraints* [NMO01].

More than the definitions, the best way to see the idea behind ontologies is through examples. Let's take a sentence that describes an anatomical organ : *the patella is a bone organ and it is a part of knee*. If we translate this descriptive sentence into an ontology, we get the hierarchical description illustrated in the Fig.2.26. In this example, the *classes* are *patella, bone organ* and *knee*, and the *relationships* between these classes are *is_a* and *part_of*.

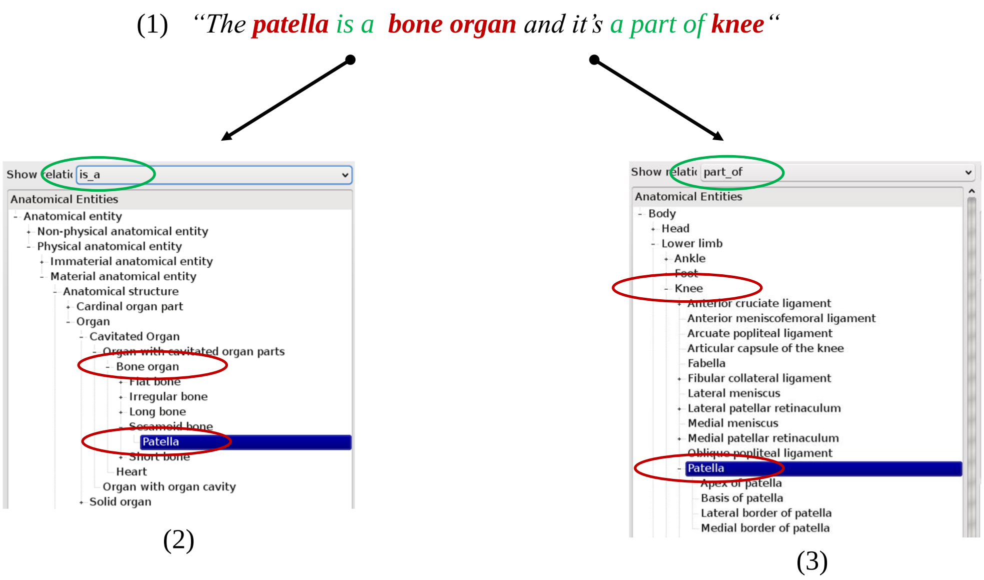


FIGURE 2.26 – *Translation of a descriptive sentence into an ontology*

This hierarchical description can be seen as a graph as shown in Fig.2.27. This illustrates how ontologies store definitions or descriptions.

2.3.2 Role and use of ontologies :

Given that ontologies are explicit and formal specifications of knowledge, they foremost help in disambiguating data and assist in finding correspondences between various information sources. This knowledge representation structure usually consists of a set of classes organized hierarchically describing a domain. The primary goal of ontologies is to identify the key elements of a domain that must be considered during its creation and evolution, so that it can facilitate the desired compilation and sharing of multi-modal and multi-dimensional data. Ontologies have gained popularity in the AI community as a means for establishing explicit formal vocabulary to share between applications. With the recent advent of the Semantic Web, the issues have increased, mainly because of the abundance, heterogeneity and independence of the various data sources. Thus, in this area today, data are annotated essentially using ontologies to address these issues. Ontologies hold in general a great importance to modern knowledge based systems. For instance, they constitute a powerful tool for supporting natural language processing, information filtering, information retrieval and data access. They also constitute an approach for knowledge representation that can be shared establishing a shared vocabulary for different applications hence their role as the backbone of the Semantic Web. Concepts in ontologies give meaning to data. Ontologies allow to capture knowledge in a way understandable to

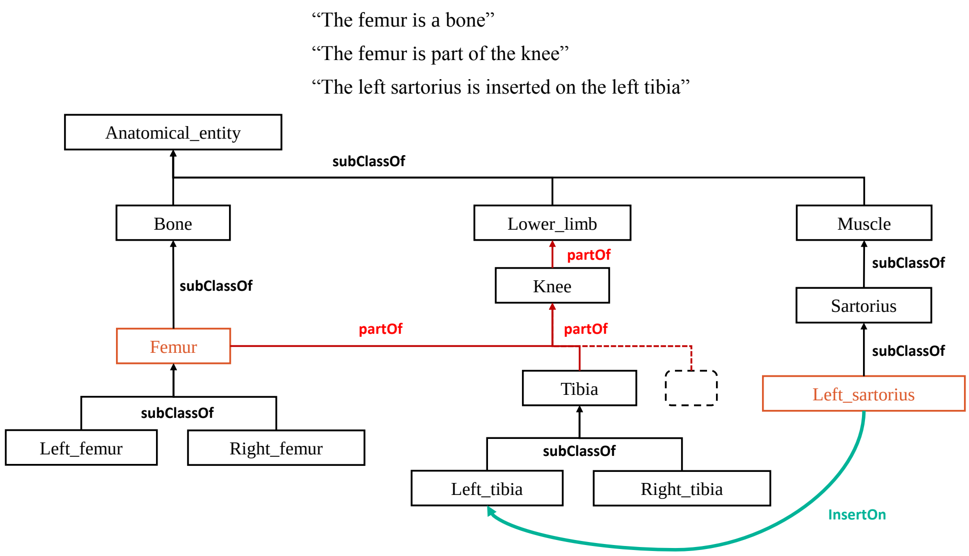


FIGURE 2.27 – *Translation of a descriptive sentence into a graph using ontology*

both humans and machines. They are machine readable in such a way that information systems are able to use them to represent and share the knowledge about an application domain. This enables a certain degree of interoperation between these data sources thanks to their shared terminology. Ontologies can be used for the annotation of multiple knowledge (or data) sources, not only limited to web pages, but also collections of xml documents, relational databases, etc. It is in this way that ontologies provide a common vocabulary between various applications. The relational definitions connect the various terms in a hierarchical fashion, and distinguish an ontology from a simple list of terms or a taxonomy. Ontology allows to constrain the meaning of a domain language, and creates a shared, formal semantics that describes how pieces of information interact with each other. Together, these characteristics help to structure logical inference and valid reasoning within a knowledge base. No less importantly, another purpose of an ontology is standardisation, at both the syntactic and semantic levels (i.e., the terms, concepts, and relationships). For collaborative purposes, it is important that experts agree upon, and properly implement a common language that incorporates mutually understood concepts and definitions. For such purposes, ontology is probably one of the unique tool that enable such possibilities. By establishing a standard framework to document and archive metadata, ontologies provide and propagate context when sharing data and/or processing and analysis tools.

The main reasons to develop an ontology according to [NMO01] are :

- To share common understanding of the structure of information among people or software agents
- To enable reuse of domain knowledge
- To make domain assumptions explicit
- To separate domain knowledge from the operational knowledge
- To analyze domain knowledge

The manual construction of ontologies is an expensive and time-consuming task because it requires professionals which usually are highly specialized. In addition to this difficulty, there is the difficulty in capturing the knowledge required by knowledge based systems, this obstacle is known as "knowledge acquisition bottleneck". An ontology is composed on one hand by concepts, taxonomic relationships (that define the concepts hierarchy) and non taxonomic relationships between concepts. On the other hand, we find concept instances and assertions about them. When developing an ontology, there are three important points that might be considered : *the clarity*, the *coherence* and the *extendibility*. *Clarity* is needed to effectively communicate the intended meaning of the included terms in part to all users, which is facilitated by making definitions as objective as possible [MBK13]. *Coherence* means that are to generate inferences that strives the ontology consistent with the definitions within the ontology [MBK13]. *Extendibility* criteria is indispensable because an ontology should be designed to anticipate the uses of the shared vocabulary. It should offer a conceptual foundation for a range of anticipated tasks, and the representation should be crafted so that one can extend and specialize the ontology monotonically [Gru93]. In practical terms, developing an ontology includes the following steps [NMO01] :

- Defining classes in the ontology.
- Arranging the classes in a taxonomic hierarchy.
- Defining slots and describing allowed values for these slots.
- Filling in the values for slots for instances.

The most popular languages for handling ontologies are *OWL (Web Ontology Language)* [MH04], the *RDF (Resource Description Framework)*, the *RDFS* (RDF Schema), the *HTML* and the *XML*. Theses languages are handled by software such as *Protégé-2000*, Chimaera, Ontolingua, KMgen, SemanticWorks, TopBraid Composer which are among the most popular ontology-editing environments.

2.3.3 Important terms :

Terms that are important to know and related to the ontologies and the subject of this work are :

- ***Concepts*** : They represent the entities of the domain being modeled. They are designated by one or more natural language terms and are normally referenced inside the ontology by a unique identifier. They are also known with the names *classes, sets, collections, concepts, types of objects, or kinds of things*.
- ***Entities*** : This term is used as a universal ontological term of art embracing objects, processes, functions, structures, times and places, and we distinguish among entities in general two special sub-totalities, called instances and classes, respectively [SR04]. In the context of this these, entities will most of time refer to classes.
- ***Relationships*** : They are primitives which asserts some kind of association between two or more entities. They define the concept hierarchy and express the ways in which classes and individuals can be related to one another.
- ***Function terms*** : They represent the complex structures formed from certain relations that can be used in place of an individual term in a statement.
- ***Restrictions*** : It formally states descriptions of what must be true in order for some assertion to be accepted as input. They specify additional constraints on the ontology and can be used in ontology consistency checking and for inferring new knowledge from the ontology through some inference mechanism.
- ***Instances*** : Objects or individuals which represent elements of a classes.
- ***Attributes*** : Aspects, properties, features, characteristics, or parameters that objects and classes can have.

– ***Rules*** : Statements in the form of a conditional sentence that describe inferences that can be drawn from an assertion in a particular form.
– ***Axioms*** : They are some assertions in a logical form that together comprise the overall meaning that ontology describes in its domain of application. The term "Axioms" also includes all the theories (*the term theory refers here to a tentative explanation of a portion of reality* [RM03]) derived from axiomatic statements.
– ***Events*** : The changing in the state of the attributes or the relations within an ontology.
– ***Knowledge base*** : It consists in an ontology together with a set of individual instances of classes.
– ***Taxonomy*** : Ontologies and taxonomies provide a structure to the concepts and language used to organize knowledge [MN09]. Although they are often confused, they refer to different concepts. Taxonomies are the classification scheme used to categorize a set of information items. They represent an agreed vocabulary of topics arranged around a particular theme, with a hierarchical or non-hierarchical structure. Considering taxonomies, ontologies are often equated with taxonomic hierarchies of classes, class definitions, and the subsumption relation, but ontologies need not be limited to these forms [Gru95].
– ***Ontology Aligning*** : It consist in bringing two or more ontologies into a mutual agreement. For instance, ontologies are kept separate, and one of the original ontologies is adapted such that the conceptualization and the vocabulary match in overlapping parts of the ontologies [JdB03].
– ***Ontology Mapping*** : Specifying how the concepts in the different ontologies are related in a logical sense. This means that the original ontologies do not changed, but additional axioms describing the relationship between the concepts are added [JdB03].

2.3.4 Ontologies for the construction and simulation of anatomical system :

If the use of ontologies is more common in domains mentioned above, its use is certainly not new in the computer graphic field and for the simulation of anatomical system, although this is not very frequent at present. The reference ontology for the human anatomy is FMA (The Foundational Model of Anatomy) [RM03]. This ontology is grouped into the Open Biological and Biomedical Ontologies foundry (0B0) [SAR*07]. FMA has been developed to fill the need for a generalizable anatomy ontology, which can be used and adapted by any computer-based application that requires anatomical information. It is a comprehensive description of the structural organization of the body, conformed to the definition of an ontology advanced by [Pis04]. The FMA main focus is the macroscopic and microscopic anatomy of the entire body, including neuro-anatomy. It is actually the largest anatomy ontology or terminology and one of the largest ontologies in the biomedical domain : it consists in more than 135,000 terms which point to 75,500 types, and which are interrelated by over 2.5 million iterations of 198 kinds of specific relations. Although it is the most complete ontology on human anatomy, it does not contain any information about function of anatomical system (physiology), for instance a formal description of the *hip flexion* (*i.e all the entity involved on this function and in which way*). These relations between anatomical structures and their functions appear to be a relevant knowledge if an ontology must be used for the purposes of musculoskeletal system simulation.

Alternatively to FMA, [DMT07] introduces a specialized ontology to understand the impact of pathologies of the musculo-skeletal system on the gait. This novel ontology is an extensible ontology in the biomechanics field, they named it OSMMI (Ontology of the musculo-skeletal system of the lower limbs). Its architecture includes 14 parts which are defined as the classes of the ontology : Nervous system, Ligament, Muscle, Tendon, Cartilage, Bone, Limb, Posture, Support of load, Diarthrosis Joint, Movement, Articular Contact, Contact of environment and Gait. The parameters of each part contains biomechanical information such as length, size, weight, mass center, ... etc. The ontology is validated with the WonderWeb OWL Validation Service. What is interesting with OSMMI is the fact that is built to make the bridge between anatomical system description, with their functional aspect while considering the biomechanics of these system. Its main limitation is that it is not compatible with the reference existing ontology which are widely established in various applications, all the terms within this knowledge base are then usable only with this ontology. This modeling choice restrain the domain of usability and extensibility of this work.

Another anatomical ontology we can find is the SNOMED CT [CR80]. In this ontology, anatomical concepts are linked to specific diseases or symptoms. This ontology is also part of the OBO foundry. As most of the ontologies register in the OBO foundry, it has been developed to address a specific need articulated by a community working with a particular model organism.

Ontologies are frequently used to provide formal description of phenomenons or structures, in [RBG*04], they propose to predict the anatomic consequences of injury (*penetrating injury*) on organ (*heart*) by using an ontology-based system. They create a geometric model of the organ that integrates biomechanical and anatomic knowledge handled by their ontology. They associate these geometric information with the formal descriptive knowledge, whose classes names are provided from FMA [RM03]. Hierarchy of abstract geometric objects is then created and represents organs and organ parts. These geometric objects contain information about organ identity, composition, adjacency, and tissue biomechanical properties. This integrated model support reasoning tools and thus allows complex queries. Their resulting system enables to predict the organs and organ parts that are injured.

Ontologies can also be used to assist the medical practice. In [CGMT07], *Charbonnier et al* design a medical-based ontology of the musculoskeletal system and system functions. This medical-based ontology is integrate in a system to help and guide orthopedists during clinical examination. They create this ontology to fulfill the lack of functional knowledge on existing ontology and to serve a clinical examination platform. The medical based ontology built extend FMA [RM03] and add some classes instances which contain specific attributes referring to functional (e.g., material properties, peak forces, etc.) or kinematical (e.g., joint center, joint axis, to define the etc.) parameters so as functional properties of the entity. By storing these parameters at a high level, it is thus possible to access them more quickly. All entities are linked to a 3D mesh and the system integrates conventional diagnostic support (MRI), visualization and simulation tools and the ontology interface between the constitutes application and the medical interpretation of data. Although this ontology extent FMA with mechanical and geometric data, the lack of ontology of organ function prevents its use for works on the musculoskeletal system function.

When one is interested in ontology and simulation of musculoskeletal (or anatomical) system together, we find My Corporis Fabrica (MyCF) [PBJ*09], this framework rely on an ontology which adds to FMA [RM03] new possibilities such as the support of topological, geometrical and functional aspects of individualized anatomy. Relations between entities and 3D models make explicit how anatomical structures are composed, and how they can be related to 3D complex objects describing patient-specific body parts, declared as instances of appropriate mesh 3D models used for simulation or 3D rendering. Thus, these features allow to query on anatomical entities in order to obtain formal description of structures in addition to 3D geometrical data and biomechanical parameters of each resulting concept. These additional data are coherently integrated into the ontology to keep its consistence while enhancing it. This add also enables the automatic construction of anatomical system for simulation purposes, by using the relationships between entities and systems. The efficiency of MyCF is demonstrated via an example of construction of a model of knee based on the information obtained automatically. The model is exported to a physical simulator that highlight the system benefits through a dynamic simulation of the effects of ligaments on the stability of the knee joint. Ontology can be used to guide automatically the segmentation process of anatomical entities and to label the resulting regions [HHP10]. To achieve such possibilities, [HHP10] uses MyCF [PBJ*09] and extends it with approximation of geometric shape (e.g plane, sphere, cylinder or cone) of the anatomical sub-parts (e.g Lateral face of the lateral condyle of the femur). The combination of localization information, adjacency relation, relative size and additional segmentation parameters allow to have better segmentation results as demonstrated with an example of femur segmentation on the paper. Our contributions introduced in the Chapter 4 also use the MyCF framework.

While many efforts have been made to address needs of structured anatomic knowledge, other approaches using ontologies have been taken in the computer graphic field. The works made in [GRVT*06] introduce a system that allows storing, indexing and retrieving facial animation by using an ontology that contain facial animation classes and relationship with emotion. The facial animation are registered in the ontology with MPEG-4 face animation objects which are associated to the domain concepts. On the same lines as [GRVT*06], [GGRT*07] present a new system to query on virtual human (VH) shape, animation and on human interaction with object. This system has the novelty that is is based on the use of a new ontology of virtual human that includes as main concepts : the human geometrical shape, morphological (or morphometric) information, animation of VH (autonomous or pre-set animation of VH) and interaction of VH with virtual objects. The system allows complex queries such as "*Does it have a long nose ?*", "*Is the model suitable for animation ?*" or "*Which are the standing (seating, walking) VH ?*".

What can be retained about the use of ontology to assist the modeling process for simulation purposes is that several works use anatomical knowledge bases to store and organize a wide variety of attribute (i.e 3D mesh model, (bio)mechanical parameter, geometric information, ... etc) in order to be able to get these parameters at a high level, and thus access more quickly on them for an automatic construction of model. This construction process take advantages of the relations and formal description within the ontology. This type of modeling system contributes to an easier access and manipulation of complex knowledge with simple queries. One lack on all these works of the literature is an ontology of function of the anatomical system, such an ontology will allow to get a formal representation of the systems functions which could be used to ease the construction of optimized model dedicated to simulate these functions. Such an ontology will required other layer of tools which will enable the construction of biomechanical systems for simulation or animation. Considering this lack that we introduce the second contribution of this thesis presented in the Chapter 4.

2.4 Registration and Re-targeting

In this section, we look at the registration methods which can be used to automatically model geometric representations of organ which could serve to setup patient specific simulation, by reusing existing models instead of remodeling them from scratch. Some registration methods are often call re-targeting or transfer in the computer graphics field. They contribute to improve the modeling process, since this stage of creating geometric models for simulation and visualization is a tedious and difficult task. This creation process is also time-consuming and needs to be performed independently for each subject, making such approaches impractical for graphics applications. In addition to the difficulties mentioned above, it requires an expert since the internal and external geometry of the organ must be well known, and adapted to the simulation methods if the aim is to obtain high quality results that reproduce as much as possible the organs behavior. Although registration methods do not remove from the modeling process this task of geometric modeling, at least one time, they enable to re-use the created models for the simulation of a wide variety of character with different morphologies. This aspect is what interested us on the registration methods.

In this section, we begin by a short reminder of what is the registration, the different stages during the registration process and why it is so important in the computer graphics domain. Then, we present the literature work that address this issues of creating geometric models for simulation or methods for adapting existing models from one character to another to quickly get a set geometric data for individuals with various morphology.

2.4.1 Generalities

According to [TCL*13], registration methods are the process of transforming multiple 3D datasets into the same coordinate system to align overlapping components of these sets. In the context of this work, it much more represents the process which given a source and a target inputs, find a motion that optimally positions points on the source surface into the scene in which the target data lies [HAWG08] (cf Fig.2.28).

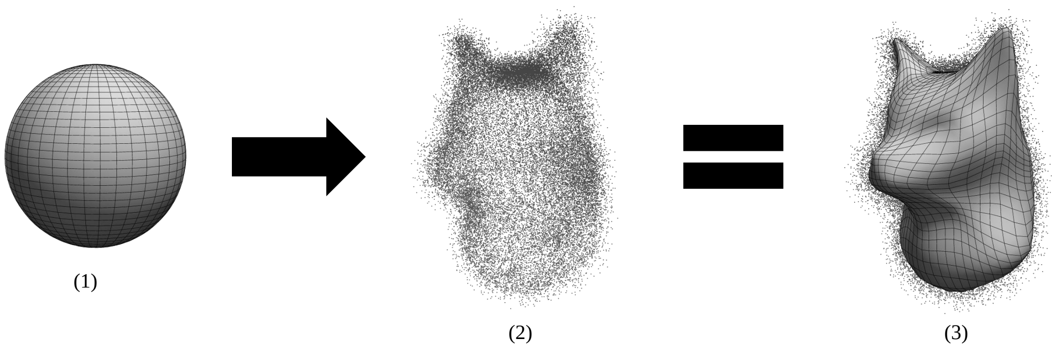


FIGURE 2.28 – *Principle of the registration : (1) source template, (2) target data, (3) source template registered on target data.*

Registration are often cast as an optimization problem [TCL*13], which aims at finding spatial correspondences between datasets, and find the optimal deformation field that aligns a template to a target dataset [GRP10]. It always involves two type of inputs : the *source* models and the *targets* models. The source inputs which is transformed are 3D mesh surfaces (triangle mesh, quad mesh) or volumes (tetrahedra mesh, voxel grid) in the context of this work. The target dataset includes measured points, 2D-images and 3D-images representing surfaces (and/or volume) of 3D objects or scenes. This type of registration with surfaces as input is called *"Surface registration"*, whilst registration with image as input is called *"Image registration"*. It is a central problem and an essential component of the 3D acquisition pipeline, fundamental to computer vision, computer graphics, reverse engineering and image processing communities. Registering templates to a set of deforming surfaces provides cross-parametrization, and facilitates texture and skeleton transfer, shape interpolation, and statistical shape analysis. Numerous applications also benefit from the continual research on correspondences and registration, including symmetry detection, articulated object matching, finding object correspondences, fractured object reassembly, sub-part identification, and skeleton and pose construction [TCL*13]. One major difficulty in the registration process are to find an adequate similarity measure that is as-convex-as-possible and a good parameterization of the deformation through the introduction of prior information [GRP10].

In surface registration, we distinguish two types of approaches : *the rigid registration* and *non-rigid registration*. Both may consider rigid and non-rigid shapes. The *rigid registration* assumes that two (or more) surfaces are related by a rigid transformation. In this type of registration, the input data are often subject to various difficulties. Among them, we essentially find *noise*, *outliers*, and limited amounts of *overlap* [ZF03, TCL*13]. In addition to the difficulties mentioned, variations in initial positions and orientations, as well as resolutions of data also affect algorithm performance, and must be taken into account when comparing rates of convergence, methods of correspondence determination, and approaches to optimization. The *non-rigid registration* as its name suggests, allows non-rigid deformation fields between inputs dataset. These transformations include affine transformation, morphing, articulation, etc. This type of registration process is more difficult than rigid registration, as it not only faces the above challenges but also needs to account for other deformations, so the solution space is much larger [TCL*13]. For these cases, establishing meaningful and natural correspondences is essential, however, it is a challenging problem in its own right. Non-rigid registration can be divided into *intrinsic* methods (i.e. *they consider the use of properties like surface distances and angles which are internal within a surface*) and *extrinsic* methods (*i.e. they consider use of external properties that can be applied onto the surface*). According to [GRP10], the registration methods are classified in a different way : the *variational methods* and the *pair-and-smooth approaches*. *Variational methods* globally evolve the model along its degrees of freedom in order to minimize the distance. *Pair-and-smooth approaches* locally minimize the distance through correspondence computations, and then regularize those displacements to satisfy deformation constraints.

Registration methods consist in three core nested components or stages [AFP00, ZF03, TCL*13] : *the model selection*, the search of *correspondences & constraints* and the *optimization*. In these terms, it is closely related to data fitting.

The model selection refers to the choice of transformation [AFP00], the selected model (i.e transformation) is guessed or provided by the user with a priori knowledge of the data. These models are related to the nature of the relationships between the source and the target models. They are based on an assumed transformation model, and are highly dependent on the applications. These cover both rigid and non-rigid transformations. The selected models are classified from the viewpoint of the assumed transformations. The main types of model are : *rigid transformations (translation,& rotation)*, *piecewise rigid deformations*, *general deformations* and *(nearly) isometric deformations*. Between the rigid and non-rigid models case, it exist the *rigid alignement with non rigid correction* which represent a transitional model.

Constraints are the properties that limit the search space for transformation. Together with correspondence, they allow to restrain the solution space. Finding constraints is the second stage of the registration process, and it consists on the one hand in computing a surface representations, and on the other hand in defining a matching criterion based on the computations on surface. It should include a similarity criterion which is sufficiently discriminating to associate homologous points unambiguously and efficiently, if the application dictates that algorithmic performance is an issue. In data fitting, the correspondences are incorporated within the data point, but in registration, correspondence information must be determined. Correspondences are defined for each vertex of the input models as the displacement that maximizes a certain similarity measure [GRP10]. In rigid registration, correspondences assist in further pruning the transformation search space, whilst in the non-rigid case, establishing correspondences is the essential step that drives alignment [TCL*13]. Among the constraints, we mainly distinguish *transformation-induced constraints, features, salience, search constraints* and *regularization*. We conserve the definition of these terms introduced by [TCL*13] :

- *Transformation-induced constraints* are the properties related to the transformation type that can be used to filter and sort the transformation space or correspondence search space. These include mainly distance (and angle) preserving properties, lower-bounding, affine ratio, principal axes, closest point criterion (CPC).
- *Features* are properties that can be quantified (e.g. principal curvatures), with the aim to describe a point. Multiple features can be concatenated to form a feature vector. When features of two points (or clusters of points) match, a meaningful correspondence may exist.
- *Salience* is the measure of local significance in a surface. For instance, salient points are those whose properties are unlike most of their neighbors. The salience denotes the contrast in properties of points or of cluster of points overlooked to their neighbors.
- *Search constraints* serve essentially for the efficiency of the optimization algorithms. They include the localization and the hierarchical approaches. Localization restricts the search at each iteration to the local neighborhood of the correspondence in the previous iteration, to avoid a global exhaustive search. Hierarchical approaches further combines the above with a coarse-to-fine hierarchical search technique.
- *Regularization* is the process consisting in adding penalty terms and a priori information in the optimization system, with the goal to improve search and avoid local minima during optimization.

The optimization stage is the last component of the registration process. This stage uses the search results for corresponding points or feature pairs (*based on the surface representation of the second stage*) and the computation of the optimal transformation as idealized in the first stage [AFP00]. The search for a match can be either a succession of comparisons of discrete candidates, as is frequently the case for feature pairs, or an iterative minimization of an objective function, As mentioned, registration methods are often considered as an optimization problems, among the optimization approaches, according to [TCL*13], we find two types : *the local deterministic optimization* and *the global deterministic optimization*. Local deterministic methods look for a solution that maximizes or minimizes an objective function locally. These techniques are efficient but also depend heavily on initialization and often converge to local minimum. Local optimization methods applied in the cases of non-rigid registration are mostly based on CPC formulate an energy functional with data and regularization terms. These type of methods can become stuck in local minima if the initial solution is far from the global solution. Global deterministic optimization tries to find a global solution and to avoid local minima. In the context of rigid registration, such an optimization type uses distance root mean squared error as a bound on CPC error. It is both tight and fast to compute. For non-rigid registration, it represents potential correspondences in a tree, and uses a self-deformation distortion measure for a set of correspondences to prune whole branches of a tree search. Global deterministic optimization contributes to find natural correspondences between two non-rigid 3D shapes of quite different kinds.

2.4.2 Registration methods for the simulation of anatomical systems

Registration is a fundamental problem in computer science, especially in the computer graphics fields [KZHCo11], where a lot of effort has been dedicated to solving problems at each stage of this process. This section reviews these works and their applications in daily life problems. It also highlights literature work related to each sub-part of the registration process more generally.

On the model selection and the evaluation of correspondences and constraints, one of the first registration method, the Iterative Closest Point algorithm commonly called *ICP* was based on the computation of the closest vertices between the source template and the target model [BM92]. In these ICP approaches, at each step of the registration, the correspondences are re-computed to find the best shape that approximate the target model. Unlike method of this type, feature correspondence requires propagation and smoothing steps to ensure a local consistency of the alignment [HAWG08]. The estimation of sparse correspondences, optimizing the extrinsic (e.g., closest points [BM92]) or intrinsic (e.g., [BBK08]) similarity have been explored. Intrinsic properties of shapes, such as geodesic distances, are interesting because they are quasi-invariant under object pose and current deformation [GRP10]. Shape embedding techniques have been developed to enhance such properties, for instance in spectral embedding methods [MHK*08], conformal mapping [LF09] or the medial axis transform [SSGD03]. Optimization of the pairing stage has been studied through voting techniques that minimize distortions of the template after registration [LF09, ZSCO*08]. In surface registration, researchers have considered the distance between rotation and translation invariant local shape descriptors, built from differential geometric quantities [HAWG08]. Intrinsic methods attempt to find common parameterizations between template and target surfaces. They are global but sensitive to topological noise [BBK09].

In image registration, pairing is performed by locally maximizing the image similarity between the template image and target image. Several correlation measures have been proposed for a range of imaging modalities [ZF03]. Although local pairing is not robust for large displacements, there are few degrees of freedom and a clear global minimum. Robust correspondences can be better achieved by computing similarity over the entire spatial domain. In image registration, features can be extracted from images based on the local intensity distribution [ZF03], such as the histogram moments [She07].

When correspondence and constraints are established, most surface and image registration techniques rely on a smoothly varying motion field over the spatial domain to avoid distorting the template excessively [ZF03]. In this sense, regularization techniques are used to enforce and to improve the result quality. At this regularization stage, a displacement model is often associated : ranging from rigid, affine, to as-rigid-as-possible deformation fields, possibly with extra constraints such as articulations [GRP10]. Rigid registration is among the simplest ways to recover the dominant motion. Among rigid registration methods, ICP [BM92] is one of the most used approach that finds, at each iteration, the best rigid transformation approximating the current target displacements. We also find the generalized gradient approach [CMP*07], is such approaches, target displacements are filtered to remove non-isometric components from the deformation field. Free Form Deformation (FFD) as deformation field to aligned the input models have been proposed as alternative approach [LCO*04], to minimize an energy functional that consider both similarities and smoothness measures. In computer graphics, various techniques have been developed to generate as-rigid-as-possible deformations in order to mimic elasticity. Along these lines, we find the work introduced by Sorkine et al. in [SA07], which iteratively minimizes an energy to define locally shape-preserving deformation. Similar approach have also been used in [ACP02]. Müller et al. [MHTG05] blend closest rigid transforms in a shape matching framework. While the assumption of smooth displacements is acceptable for a single object, it is inaccurate at boundaries when the relative displacement between objects is large, which is the case for rigid structure such as bones. To address these issues, researchers have tried to design piecewise quasi-isometric displacements fields. Arsigny et al. [ACPA06] introduced polyrigid and polyaffine transformations for image registration. Wang et al. [WHQ05] proposed a spline-based deformation technique that incorporates rigid components. For registering the skeletal system, articulated rigid motion has been considered in [KM04, STCK03, PDD*05]. Gilles et al in [GMMT06] use force fields coupled to collision and topological constraints to quasi-automatically register musculoskeletal system. Statistical analysis techniques are also used in registration techniques to exploit a set of existing or captured data as deformation modes [VBPP06]. These allow to compute models that captures the geometry and its variations relatively to specific attributes. In [OBJBH99], O'Brien et al. compute joint centers from markers motion (i.e., they register an acyclic chain of scalable rigid bodies). To smoothly deform surrounding soft-tissues from the articulated motion, numerous skinning techniques have been proposed in the graphics community. For skin registration, skeletal subspace deformation with an automatic tuning of influence weights has been presented in [CZ09, VBMP08, HAWG08]. To improve the robustness of the registration with respect to object poses (i.e., rigid transforms, isometry, etc.), different isometry invariant parameterizations have been proposed such as spectral embedding [MHK*08], conformal mapping [LF09], or functional maps [OBCS*12]. On the other hand, robustness to topological noise and to partial data can be achieved from extrinsic correspondences (i.e. established in Cartesian space) [LSP08, HAWG08].

Depending on the application, various use of registration have proved to be indispensable for different domains. Sumner et al [SP04] introduce the registration of deformations from one sequence of keyframe meshes of a source model towards a target model, in order to get the same motion sequence adapted to the target (cf Fig.2.29.1). Correspondence between the source triangles and those of the target are specified via markers positioned by the user. Affine deformation fields are used to modify the target mesh triangles according to the states of the source mesh triangles. In [BSBC12], rules such as the conservation of normal are used to keep the consistence of the retargeting objects, such as garments in this case. These rules are usually added as a constraint terms in the optimization system [CR03]. In [BP07, PP09], the rigging model is automatically computed and adapted to the boundary mesh that represent the character skin. Given a skeleton model for rigging, and a character skin as input, the skeleton is automatically adapted to the skin by registering the rigging model to the topology graph computed from the input skin. Such systems allow the re-uses of existing dataset model instead of recreating them for each new individuals. The registration algorithm introduced in [GRP10] allows the alignment of well segmented meshes onto volumetric data and poor quality surface meshes as shown in Fig.2.29.3.

While encouraging results have been demonstrated for transferring models from one character to another[SP04, PP09, BSBC12], little has been done in terms of volumetric geometry

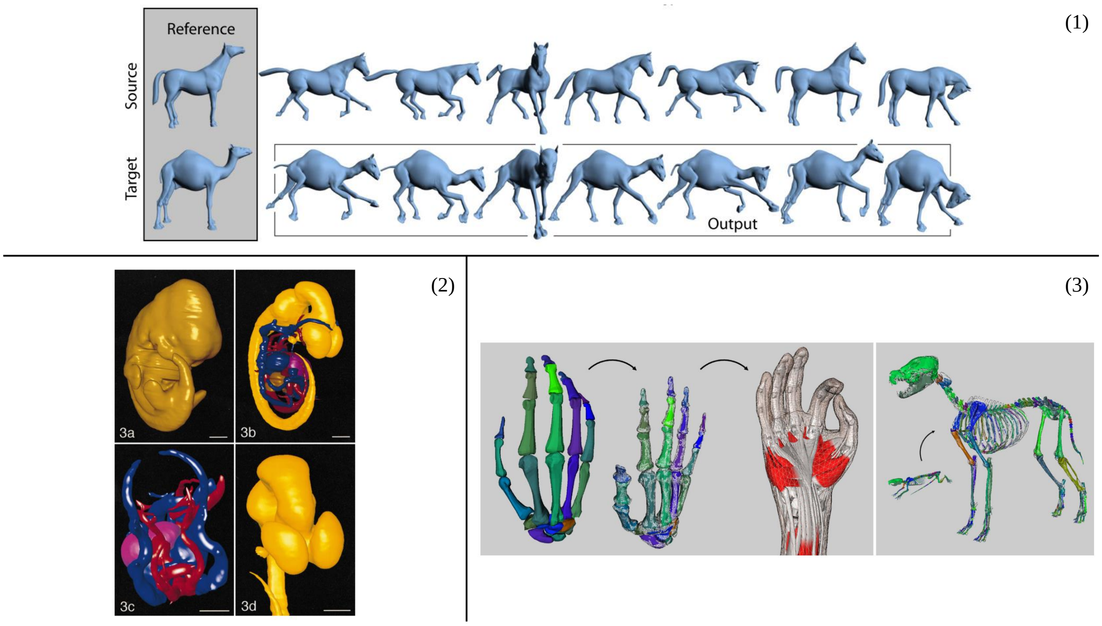


FIGURE 2.29 – *Examples of application of registration algorithms : (1) Registration of deformations introduced by [SP04], (2) Automatic registration of embriologic model from MRI data [WMSM98], (3) Registration of segmented meshes on reconstructed surfaces [GRP10].*

transfer across shapes. In [WMSM98], they demonstrate the importance of automatic registration method of embriologic model with 3D-MRI, the created models serves for teaching purposes (cf Fig.2.29.2). In [Fos05], they present an approach for automatic processing of three-dimensional (3D) CT images acquired during image-guided radiation therapy. With their system, they achieve automatic organ segmentation, and calculate the dosimetric effects of soft tissue motion. Folgoc et al in [FD14] introduce the *Sparse Bayesian framework*. This non-rigid registration method based on statistical tools from Sparse Bayes field, allows the automatic determination of the relevant deformation model among any preset, and the estimation of its parameters. They illustrate the efficiency of their approach on cardiac images from various sources : MRI, tagged MRI and 3D-US images. Registration methods are also used in [SVA*09] to investigate a markerless tracking system for real-time stereo-endoscopic visualization of preoperative computed tomographic imaging as an augmented display during robot-assisted laparoscopic partial nephrectomy. Image-based tracking technology tracked selected fixed points on the kidney surface to augment the image-to-model registration as illustrated in Fig.2.30.3. Augmented reality overlay of reconstructed 3D-computed tomography images onto real-time stereo video footage is thus possible using ICP and image-based surface tracking methods that do not use external navigation tracking systems or preplaced surface markers. Along these lines, we find the work presented in [HC14] and illustrated in Fig.2.24. In [PSD*13], they combine the motion estimated by image registration and a simulated model (cf Fig.2.31.2) to generate a synthetic and visually realistic time series of cardiac images. Input are an electromechanical model of the heart and real clinical 4D image sequences. Such a method is used to validate visual information on cardiac motion for diagnosis and therapy planning purposes.

Registration is used to reconstruct full 3D models from partial scans, creating statistical shape models, shape retrieval, and tracking [AFP00, ZF03, TCL*13, BRLB14]. Currently in

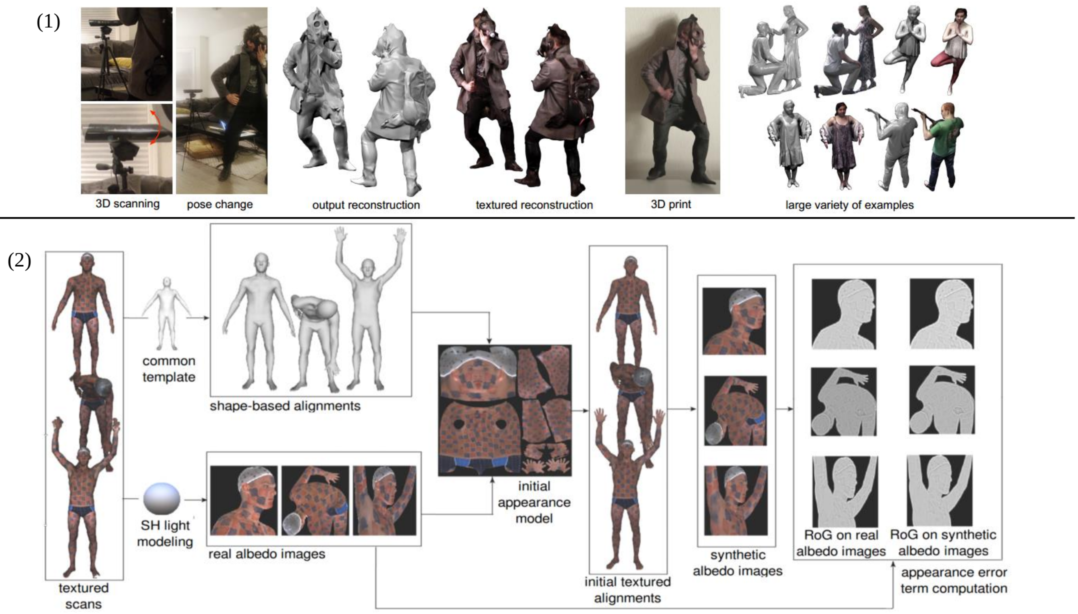


FIGURE 2.30 – *Examples of application of registration algorithms : (1) Registration of partial data from kinect sensor reconstructed to create 3D printed figurines [LVG*13], (2) Registration process used to create the FAUST database [BRLB14].*

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anthropology, they rely more and more on such approaches thanks to the recent advances in automatic processing. We find algorithms used for craniofacial reconstruction with the aim to recreate a likeness of the face of an individual at the time of death [CVD*10]. To perform accurate registration on full body scanned data, Bogo et al [BRLB14] paint the subjects with high-frequency textures and use an extensive validation process to ensure accurate ground truth. They register a corpus of scans of multiple subjects in multiple poses by aligning a triangulated template mesh to each scan. The deformations that fit the template mesh to a scan are regularized towards a deformable, statistical human body model. The alignment is performed in two steps : first, each scan are roughly registered and the parameters of the body model are extracted ; then, alignments are refined by introducing a novel appearance-based error term. This model database name FAUST is available for the general public. Animation from capture device are registered from real human to cartoon character [CBK06] to create realistic animations with such a method as shown in Fig.2.31.3. In [BP13, LVG*13, SFW*14], they propose some applications of registration methods which allow to reconstruct models from noisy and partial data (*extracted from kinect or other devices*), with the goal to fullfill needs of creating full body in 3D, printed figurines, and to assist the fast and easy creation of animation (cf 1 and 2 in Fig.2.30). These applications have the advantage to be within general public's reach. More generally, in motion capture, underlying poses of the animation skeleton are computed from fiducial marker correspondences using registration method.

Although a wide variety of sophisticated registration methods have been proposed to address problems on modeling of anatomical structures for simulation, almost none allows the quick creation of models for patient specific simulation on the fly. For such a purpose, we find in the literature some approaches as [SP04, PP09, BSBC12] which re-target objects, deformations and structures external to our body. For internal anatomy, and mainly for musculoskeletal

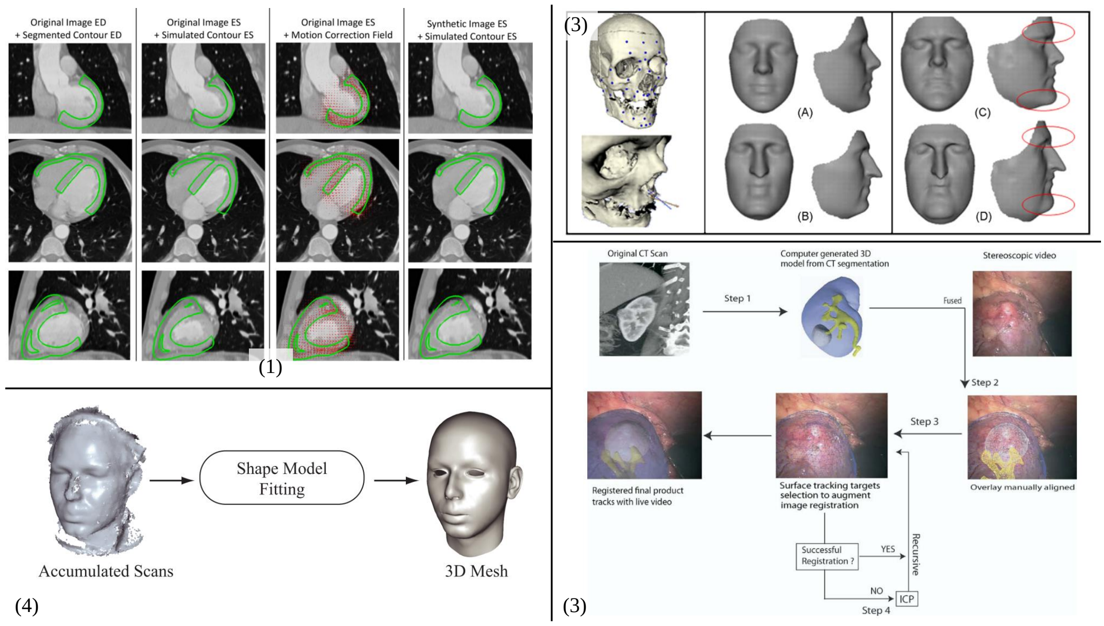

FIGURE 2.31 – *Examples of application of registration algorithms : (1) Pathological Synthetic 4D CT Created from a Clinical Sequence [PSD*13], (2) Registration of face template on skull to compute the initial face of the dead body [CVD*10], (3) Registration of 3D template mesh on scan data form capture devices [BP13], (3) Registration of reconstructed 3D geometric model from CT scan mapped to stereoscopic video data in order to create VR system [SVA*09]*

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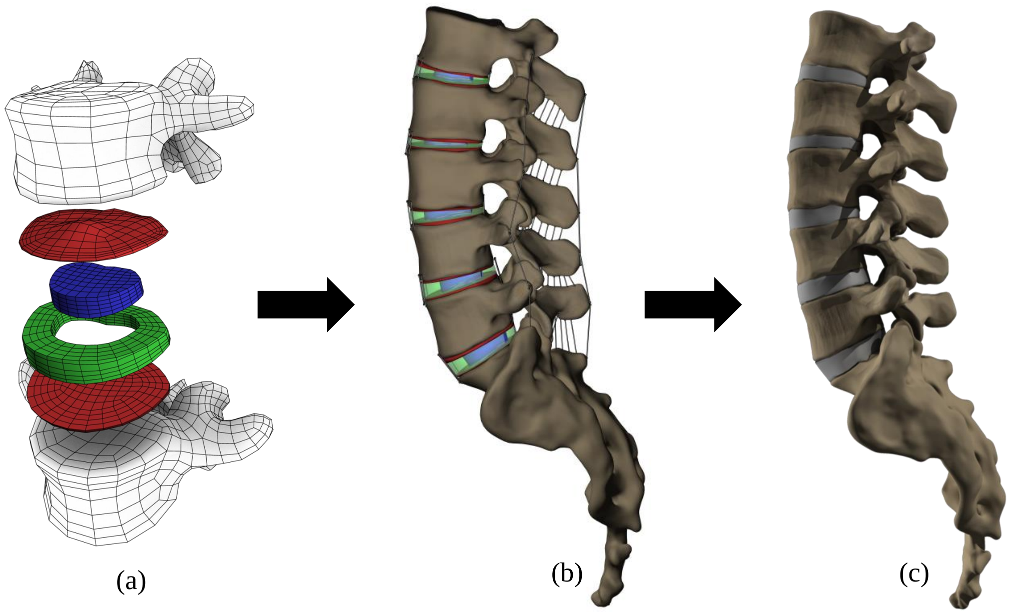
systems, we find works like those proposed by [GMMT06, GRP10], both of them are based on MRI data as input, which certainly give a huge amount of information, but these type of input particularly are not easy to obtain, are often subject to various problems (noise, acquisition quality, resolution issues, outliers, ... etc), and always require post-process stage. With boundary surface representing the external skin as input (*which represents the simplest and the most common way to represent a character in computer graphic*), until now, no method is available to register geometric models of internal anatomy (*for simulation*) from one source character to any target character, the whole while conserving the consistency and the properties that intrinsically define the model.

CHAPTER

3

HYBRID MODELING AND SIMULATION OF MUSCULOSKELETAL SYSTEM : APPLICATION TO LUMBAR SPINE



Teaser Figure : *Hybrid lumbar spine model : (a) FEM geometries of the disc component, (b) Complete lumbar spine model with rigid vertebrae, FEM discs, ligaments and zygapophyseal joints as non linear elastic elements, (c) Visual model after rendering.*

3.1 INTRODUCTION

The human spine is an important structure whose principal functions are to protect the spinal cord, to transfer load from the head and trunk to the pelvis and to enable the mobility of the upper body. Understanding the behavior of this complex and vital structure using computer simulations as well as in-vivo and in-vitro experiments is an important area of research in clinical applications, in treatment planning and surgical training. The lumbar spine is a sub-part of the spine, this complex system is composed of nearly rigid parts : the vertebrae, and soft to stiff tissues : the discs, the facet joints, the facet capsules, the ligaments and the muscles. All these components interact in synergy to allow us to smoothly perform a wide range of motions in our everyday life. In the field of health, simulations aim at reproducing real phenomena in order to understand, to predict and to prevent serious health issues. It is therefore essential to propose models that accurately fit the anatomical and physiological description of each component and interactions that occur within a system. Two approaches are commonly used for the simulation of anatomical system : approaches based on articulated rigid bodies system and approaches based on finite element method (FEM).

On the one hand, **full FEM approaches** accurately compute local forces and deformations, which is especially useful for complex soft objects such as lumbar discs. They require volumetric meshes composed of well-shaped elements of all organs, including bones, which can be difficult to build. Traditional FEM simulations are notoriously hard to set up, due to geometrical complexity and the difficulty of trading off precision, which requires fine meshes, with computational efficiency, which requires simple models. Furthermore, setting up interfaces between the volumetric meshes of the different objects, as well as tuning the laws and parameters of the different materials is tedious. Finally, the resulting equation systems may be large, depending on the resolution of the meshes, resulting in slow computation times. To spread the use of computer simulation across a large number of users, it is thus important to ease the modeling phase. Moreover, some applications such as medical and healthcare hardware design are based on trial and error approaches involving numerous simulations, therefore computation time is also an important issue.

On the other hand, in a wide range of applications vertebra can be safely seen as rigid, thus models composed of **articulated rigid bodies**, which are easier to set up and faster to simulate. Such types of system have been proposed to study motion [PAL04, YNT05, DASR07]. However, they suffer from a lack of accuracy in the reproduction of soft object's internal behavior which are common within our organism. This aspect becomes a limitation especially if the purpose is to simulate the discs and the joints between the rigid bodies which generally fail to accurately capture the relative motion between the vertebrae. Most of the spine studies involving simulation so far have used either the FEM or the rigid body approaches but never both together. This may be due to the lack of software able to efficiently and accurately combine the two models, and to the lack of validation of such hybrid approaches.

To get the best of both worlds, we introduce a novel 3D dynamic model of the lumbar spine that combines both FEM and multibody systems. This hybrid model is build with the aim to consider the advantages of both type of modeling : the easiness to setup model, the fast computation and the accurate internal behavior of soft structure during simulation. Hybrid simulations in the context of this work represent the simulations of model which combine different type of degree of freedom (DOF) within a unique system. Such an approach has been successfully applied to the simulation of the jaw-tongue-hyoid system [SLPF11], but never to the lumbar spine so far. More than proposing this novel spine model, we demonstrate that hybrid approaches allow to fit quality criteria required in the health area, while reproducing the behavior of such systems. We show that accurate simulations of the spine motion in the three anatomical planes can be obtained with easier modeling and faster computation times than using the traditional finite element method. A specific attention is paid to the construction of the model. Each vertebra is modeled using a rigid body. The position of the rigid body, the inertia matrix and the mass of the vertebra are computed based on the geometrical shape. Each rigid body controls a surface used for the contacts that occur between two vertebrae on the spinal process during the extension. The facet joints are modeled using 6D elastic joints that allow restricted relative translations and rotations. The ligaments are modeled using non-linear springs attached to the vertebrae. Each spine unit contains six ligaments : the anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL), the ligamentum flavum (FL), the transverse ligament (TL), the interspinous ligament (IL) and the supraspinous ligament (SL). The most complex object is the disc, modeled as a heterogeneous finite element model for two of its components : the nucleus and the annulus. The two bony plates are considered undeformable and are constrained in displacement and velocity to the rigid bodies of the vertebrae. These plates allow to connect the finite element model of the disc to the rigid bodies of the vertebrae and then allow to propagate load and force in the spine through the disc.

We validated the model by simulating the range of motion (ROM) and the position of instantaneous center of rotation (ICR) and comparing with the literature. To achieve these validations, three experiments was set up. In the first experiment, we apply different moments to the upper vertebra of the spine segment L4-L5 in the three anatomical main planes, and we show that all the range of motion (ROM) and the position of instantaneous center of rotation (ICR) are in agreement with the literature. In the second experiment, we extend the simulation to the whole lumbar spine L1-S1 to evaluate the ROM, and we show again that our results are in accordance with the literature. In the third experiment, we validate the disc model as well as its mesh refinement by comparing the range of motion, the ICR and the computation time of the simulation of one FSU using different disc models in different tests :

– In the first test, we use our new disc model composed of undeformable plates and an FE model of the nucleus and the annulus.
– In the second test, the four components of the disc are modeled using finite elements, at the same refinement (resolution) as in the first test.
– In the third test, the disc is a full FE model at higher resolution.
– In the last test, the disc is a full FE model at lower resolution.

The goal of this last validation is to show that the refinement of our mesh model is the best ratio between accuracy and computational efficiency.

The remainder of the paper is organized as follows. Section 3.2 briefly reviews previous work. The construction of the model, the construction of each of its components and the interactions between them are described in Section 3.3. The simulations and the validations are presented in Section 3.4. The model is finally discussed in Section 3.5.

3.2 Related work

Different spine models have been proposed depending on the aim of the study. Two types of models and associated software are often used : 3D multi-body models (OpenSim, SimBody, ADAMS), and finite elements (Catia, Ansys). Hybrid models have recently been proposed.

3.2.1 Articulated rigid bodies

Most of the 3D multi-body models assume that vertebrae are rigid bodies connected by joints that are often simplified as ball and socket joints with three degrees of freedom per joint and a fixed center of rotation [WAA*12]. Petit & al. [PAL04] present a method to identify patient specific mechanical properties of scoliotic spine using a flexible multi-body model of spine where each vertebra is represented as a rigid body and each intervetebral disc is defined by spherical joint and torsion springs. The purpose of this study was to allow the *in vivo* estimation of mechanical properties of scoliotic subject in order to discriminate flexible and rigid scoliotic curves. In the same way, Desroches & al. [DASR07] extend this approach to propose a rigid body model of the spine to find the optimal instrumentation configurations for a patient before surgery. These contributions highlight the great potential of the multi-body representation to assist the surgical process. Yoshimura & al. [YNT05] show that a multi-body dynamic model of the spine can be used to evaluate the vibration of the spine of a subject seated in a car. As in [PAL04, DASR07], they also consider vertebrae as rigid bodies, and each intervetebral disc as rotational springs and dampers. They model the cervical and thoracic spine using 3 DOF ((C1, C5, C7), (T1, T4, T8)), and the lumbar spine and the sacrum using 6 DOF (the segments L1-S1). Since the choice of articulated rigid bodies is a simplification in the representation of the spine, it is well adapted to the understanding of the musculoskeletal system behavior of the spine and his sub-parts as demonstrated in [CFLO12, HGG12, HGJL13]. This type of representation of the spine proved to be more effective than FE model in predicting changes in the shapes of spine resulting from different surgical instrumentation strategies or for the help in the design of device for automobiles[APS*03]. The rigid body systems are also easier to set up than the FEM systems while they are faster to compute.

3.2.2 Finite elements

Several recent studies of the lumbar spine segments have used the Finite Element (FE) method. FE models are often required for the understanding of the mechanical behavior of spine units in order to predict consequences of different diseases on discs or vertebrae. These contributions show that the FE method provides good results in agreement with experimental data from *in vitro* studies [SHD*07, SKH*07, SGR*12, SLK07, ADK*13, ZRB04, WAA*12] and are well adapted for all kinds of study that require high accuracy. One of the first FE model of the lumbar spine focused on the L2-L3 unit to study the mechanical causes of low-back pain. This requires the computation of stress and strain throughout the lumbo-sacral spine [SASA84]. To study the influence of ligament stiffness on intersegmental rotation and stress in the ligaments, Zander *& al* [ZRB04] proposed an FE-model of the L3-L4 unit with a heterogeneous disc, while the vertebrae were represented using hexahedral elements, facet joint and ligament were represented with one dimensional elastic element. Schmidt *& al.* built a non linear FE model, with the parameters provided from a calibration process and validated with *in vitro* data [SHD*07]. This model was later used to understand several aspects of the behavior of lumbar spine units, such as the relation between ICR and facet joint force [SHCW08], the load combination that lead to a risk of disc prolapses, and the phenomenon of disc failure [SKH*07]. Shin & al. [SLK07] proposed a finite element model of three lumbar spines segments L2-L5 to show that dynamic stabilization devices restored functionality of a spine with intact disc. The disc of the first segment is intact, while the second segment is a fused spine with a fixation device and the third segment is stabilized using a dynamic stabilization device. This study allowed to understand the consequences of these two fixations device on the range of motion and on the adjacent disc of a segment. B. Weisse*& al.* achieve a specimen-specific parametric study of the L4-L5 functional unit [WAA*12]. The results are used as input data for 3D multi-body musculoskeletal models in order to simulate vertebral motion and loading during daily activity. An interesting aspect of their work is the detailed description of the pipeline to develop an FE model to obtain stiffness parameters for estimating kinematic behavior of inter-vertebral joints. More recently, Alapan & al. [ADK*13] brought new insight on the relation between the quality of the motion of a spine unit and the ligament failure in the L4-L5 segment. They showed how the ligaments failure change the ICR positions during the motion on the three anatomical planes. While they are tedious to setup and slow to simulate, the FE model allows to accurately reproduce the internal behavior of all entities of the spine to understand their role and their impact in the physiology of this system.

3.2.3 Hybrid models

Hybrid models try to get the best of rigid and FE worlds using models composed of rigid and deformable parts. While this intuitive idea is difficult to extensively trace back in the literature, one the first general presentation of hybrid models was given by Sifakis *& al.* [SSIF07], using *hard bindings* or *soft bindings* to combine different models within the same object. This was later used to produce a detailed model of the upper body [LST09]. Alternatively, Faure & al. showed that frame-based deformations with material-aware shape functions allow efficient hybrid models [FGBP11].

Software is a major issue for hybrid models. While most of the reference software (Ansys, OpenSim, *etc.*) implement only one approach, FE or articulated rigid, recent open-source libraries allow to efficiently combine the two [LSF11, **?**]. Stavness *& al.* [SLPF11] used Artisynth to build a hybrid jaw-tongue-hyoid model, which is the most comprehensive, biomechanically validated hybrid model so far. They demonstrated that for some applications, hybrid models provide the best combination of accuracy and performance.

Concerning the lumbar spine segments, some attempts of hybrid models have been proposed in [SAP93, SA94] to study the difference between FEM and rigid representation of vertebra. They both suggest that vertebra represented by one rigid body tend to stiffen the response for coupled movements (*e.g flexion and internal rotation at the same time*). However, each vertebra can safely be represented by two rigid bodies : one for its anterior part and another one for its posterior part ; the two rigid bodies connected by deformable beam with high stiffness.

In this work, we use SOFA [FDD*12] to propose a new hybrid model of the human spine with validation on its response to different movements, and also on its quality of movement. We show how the attachment between rigid body and FEM are achieved to obtain the right behavior between the different components. Finally, we illustrate that comparable gains as [SLPF11] are obtained in term of computational efficiency with this hybrid model.

3.3 Methods

This section describes how we build the lumbar spine simulated model based on anatomical and physiological involvement of each component in the different movement. Our model of the lumbar spine is composed of the five lumbar vertebrae (L1-L5) that meet the sacral spine S1 on the base of L5 (see Fig.3.1). Each pair of vertebrae are separated by an intervetebral disc and are connected by a pair of facet joints, and by a set of ligaments (see also Fig.3.1). We used vertebrae surface meshes derived from medical images, of the bodyparts3d database [MFT*09]. We re-meshed these models to compute the FEM volumetric mesh of the discs and the texturing of each entity (see Fig.3.2,3.5).

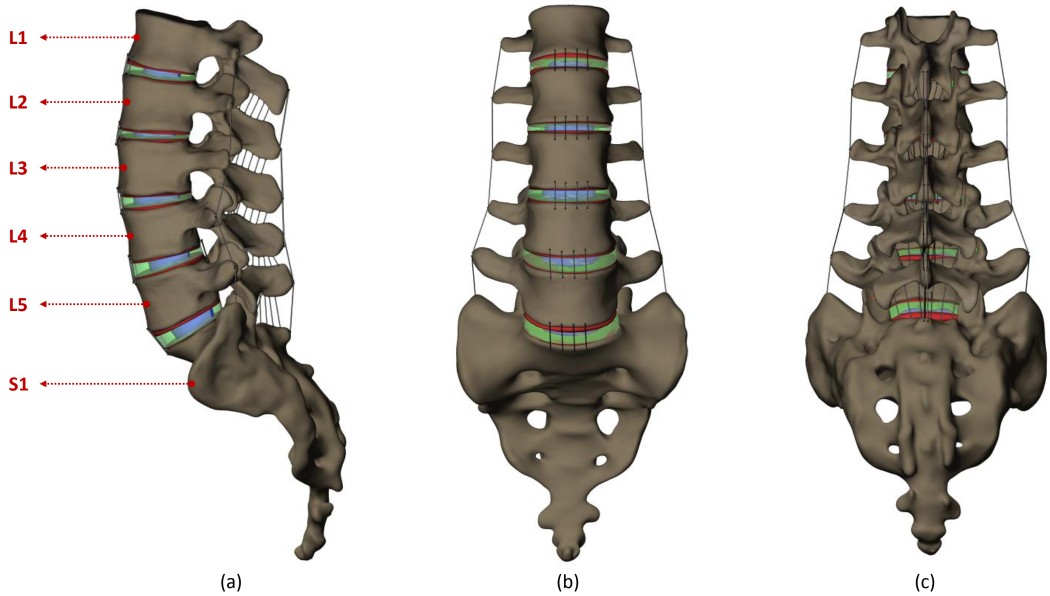


FIGURE 3.1 – *Full model of the lumbar spine on the sacrum : vertebrae (L1-S1), intervetebral discs and ligaments. (a) Profile view. (b) Front view. (c) Bottom view.*

3.3.1 Vertebrae

The vertebra are usually divided in 6 segments : the vertebral body, the arch, the pedicles, the lamina, the transverse process and the spinous processes. Each of them contain attachment regions of the spine ligaments and muscles. Vertebrae transfer the force produced by these entities within the all spine to initiate spine motion and to provide extrinsic stability. Since the main role of the spine is to support our upper body weight, the model must capture this process of the load transfer. The load is mainly propagated from one vertebra to another through the disc which is attached on the vertebral body via a cartilaginous end plate. These vertebral bodies are designed to bear mainly compressive loads and they are progressively larger caudally as the superimposed weight of the upper body increases [NF01], as shown in Table 3.1. The vertebrae of the lumbar region are thicker and wider than those in the other regions, so as to sustain larger loads [NF01]. During the extension, a contact through the ligaments occurs between the spinous processes of adjacent vertebrae to limit this specific movement. This bone is composed of two parts : the outer shell which is made of cortical bone, and the cancellous bone inside. The cortical bone is dense and very stiff. The cancellous bone consists of loosely knit structures that remind a honeycomb, and is softer. They are usually modeled using tetrahedron FEM for the cancellous bone and hexahedra for the cortical bone [SHD*07, SLK07, ZRB04, WAA*12]. Based on the anatomy and the physiology of a vertebra, we will retain that vertebra :

- transfer load from one vertebra to another via the inter-vertebral disc as a rigid structure ;
- transfer force of the ligaments and muscles within all the spine to enable his mobility ;
- limit the extension via the contacts that occur between two adjacent spinous processes.

Since vertebra deformations are negligible when studying motion, we model the vertebrae as rigid bodies as illustrated in Fig.3.4. The surface on the spinous processes creates contacts between vertebrae during the extension of the functional spine units (FSU). We consider these contacts frictionless. The two bony end plates considered undeformable (details in Section 3.3.2) are constrained in displacement and velocity to the vertebrae rigid bodies to allow the forces

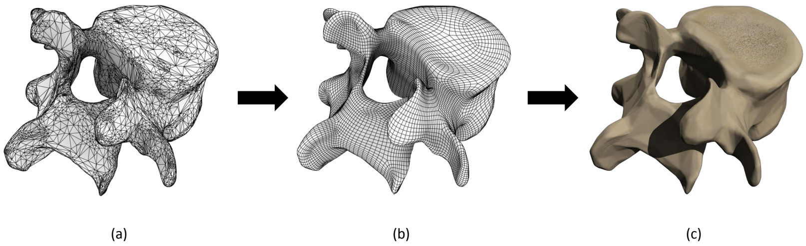


FIGURE 3.2 – *Reconstruction of the geometry : (a) Original model of the vertebra. (b) Remeshed version of the vertebra. (c) Textured vertebra*

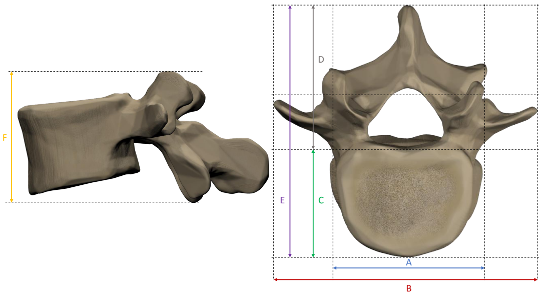


FIGURE 3.3 – *Vertebra dimensions. The dimensions A, B, C, D, E, F are summarized in 3.1*

and the loads to propagate from one vertebra to another through the vertebral body as described in the literature.

The advantage of a rigid body over finite elements is to highly simplify the nearly rigid simulation of this very stiff object. Rigid bodies under go only three translations and three independent rotations, compared with three unknowns per mesh node in finite elements. This simplifies the equations, and avoids numerical problems due to very high stiffness at the same time. The inertia matrix, the volume and the mass of each vertebra are computed based on the geometrical shape and the bone properties. This model captures the motion of the vertebrae, while enabling them to interact with FEM discs, as shown in Section 3.3.3. Our results highlight that they additionally limit the extension via contacts between adjacent spinous processes.

TABLE 3.1 – *Lumbar spine vertebrae dimensions*

Vertebrae	**Dimensions in cm**					
	A	**B**	**C**	**D**	**E**	**F**
L1	4,1	6,91	3,2	5,015	8,215	4,3851
L2	4,12	7,821	3,3	4,9241	8,2241	4,404
L3	4,383	7,9	3,433	4,734	8,167	4,138
L4	4,4	8,52	3,459	4,2611	7,7201	4,063
L5	4,8	10,004	3,729	3,8	7,529	4,178

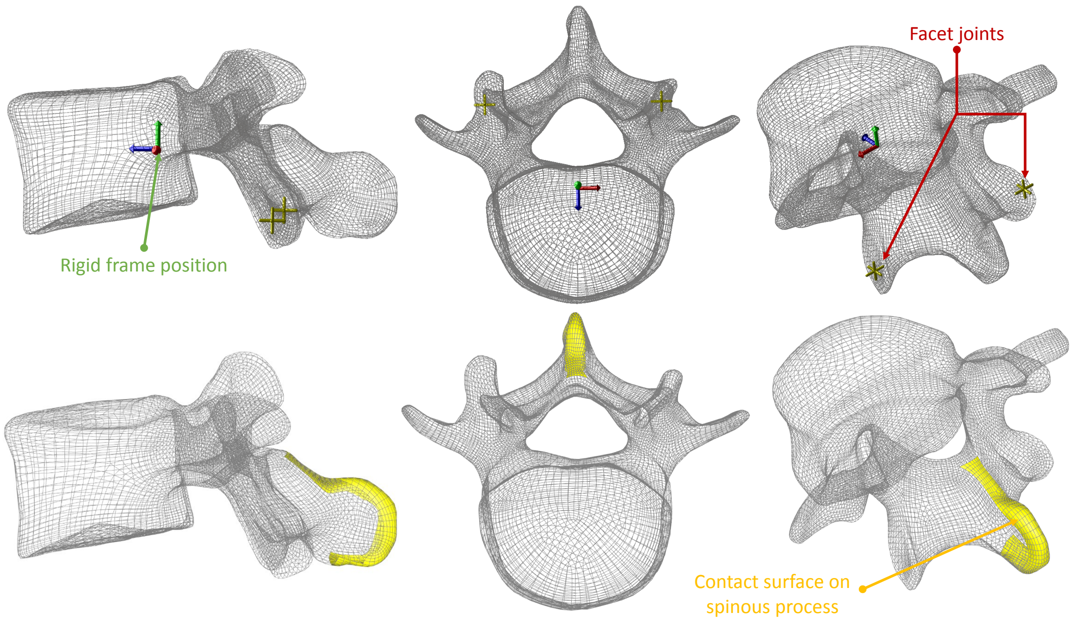


FIGURE 3.4 – *Mechanical model of a vertebra. On the top : the* ***rigid frame*** *that represent the vertebra, and in yellow on the top right* ***the two facet joint****. On the bottom : the blue surfaces represent* ***the attachment surface between the vertebra and the disc****. The yellow one is used to reproduce* ***the contact that occur between the spinal process*** *of two adjacent vertebrae during an extension movement.*

3.3.2 Intervetebral discs

Inter-vertebral discs are composed of four parts : the annulus fibrosus, the nucleus pulposus and the two bony endplates that link it to the vertebrae (see Fig.3.1). They are designed to bear and distribute loads, and to restrain excessive motion. They are well suited to this dual role thanks to its location between the two vertebrae, and also because of the unique composition of its inner and outer structures. The inner portion of the lumbar discs, the nucleus pulposus, is a gelatinous mass located in the posterior part (Fig.3.5). The nucleus is surrounded by the annulus fibrosus which is composed of fibrocartilage. The crisscross arrangement of the coarse collagen fiber bundles within the fibrocartilage allows the annulus fibrosus to withstand high bending and torsional loads. The endplate, composed hyaline cartilage, links the disc with the vertebral body.

The inclusion of all these anatomical and physiological aspects in disc models has been widely studied in previous work [WAA*12, SGR*12, ZRB04, FIN04, MPL09]. Our disc model keeps the subdivision of the disc in four components because of the large difference in the mechanical behavior and the role of each. For better computational efficiency, only the annulus and the nucleus are assumed to have linear elastic properties and were modelled using 4-nodes solid element (cf Fig.3.5) with the validated material parameters proposed of [SLK07]. We consider the bony end plates undeformable due to their high stiffness [WAA*12, SGR*12, ZRB04, FIN04, MPL09, SLK07] and the negligible deformations within them during the simulations. They are modeled using a set of points attached to the rigid vertebrae (cf gray elements in Fig.3.5). The material properties of all these components are summarized in Table 3.2.

FIGURE 3.5 – *The four component of the structure of inter-vertebral discs : the two endplates, the annulus and the nucleus. (a) : surface mesh from [MFT*09]. (b) : Our textured mesh. (c) : Our volumetric mesh.*

TABLE 3.2 – *Mechanical parameter of the deformable entities*

Material	Young's modulus (MPa)	Poisson ratio	Element Type	Reference
Vertebra			Rigid body	
Disc				
Nucleus	1	0,499	Tetrahedra (4-node solid)	[SLK07]
Annulus	8,4	0,45	Tetrahedra (4-node solid)	[SLK07]
End Plate			Undeformable structure	
Ligaments				
ALL	Force-displacement curve		2-node link	[CTB*85]
PLL	Force-displacement curve		2-node link	[CTB*85]
LF	Force-displacement curve		2-node link	[CTB*85]
TL	Force-displacement curve		2-node link	[SLK07]
IL	Force-displacement curve		2-node link	[CTB*85]
SL	Force-displacement curve		2-node link	[CTB*85]
Joints				
Facet joint	Force-displacement curve		6D Elastic Element	[SLK07]

3.3.3 Attaching FEM to rigid bodies

Since our vertebrae are modeled as rigid bodies and our disks as FEM to combine the advantages of both representations, we need to connect them while respecting anatomical constraints (namely perfect contact) and mechanical principles (namely two-way coupling). As noticed in [SLPF11], using Lagrange multipliers to attach FEM nodes to rigid objects would add computational complexity, while carefully leveraging these kinematic constraints actually allows to *simplify* the equations, by removing the attached nodes from the set of unknowns. In SOFA, we achieve this by introducing a diamond-shaped kinematic hierarchy, as illustrated in Fig.3.6. The top node represents the whole object, with a dynamics solver. The two children contain the independent degrees of freedom (DOF) which include rigid frames for the vertebra, and only the free nodes of the FEM. The motions are propagated top-down through the hierarchy, while the forces are accumulated bottom-up. The FEM nodes attached to the vertebra are entirely controlled (*mapped*, in the SOFA terminology) by the rigid motion of the vertebra. The multimapping takes input from the two particle sets and generates their union, at the bottom of the hierarchy. The FEM behavior laws such as inertia and stiffness are straightforwardly applied at the bottom level, making no difference between the particles. The inertia and elastic forces applied to these particles are mapped upward to their respective inputs. The particles mapped under the rigid body, in turn, accumulate their forces upward to the rigid bodies, where the rigid body inertia and forces (if any) are directly applied. This results in a two-way coupling with perfect attachment, while the elastic forces are automatically distributed to the independent DOFs.

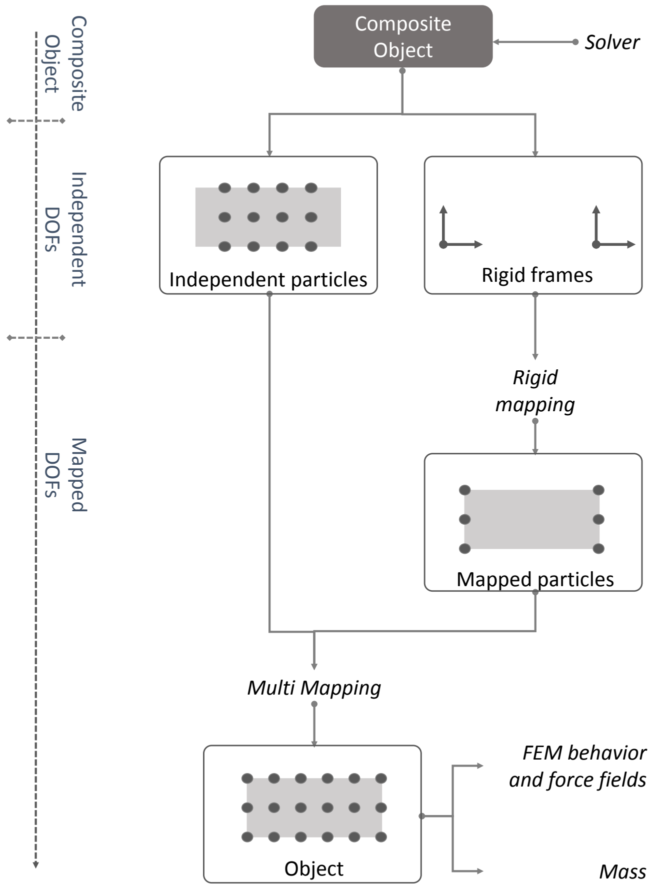


FIGURE 3.6 – *Mechanical connection between the vertebrae and the discs.*

3.3.4 Zygapophysial joint

The zygapophysial joints, also known as facet joints, are localized on the posterior part of the lumbar spine. They include the articular process of the two vertebrae that are coated with hyaline cartilage and surrounded by synovial fluid and the joint capsules. The capsules are composed of dense and parallel collagen fibers and irregularly oriented elastic fibers, which give them a mechanical behavior that reminds the behavior of the ligament hence the name capsular ligament. During the different motions, the capsules are stressed and they respond by twisting and stretching as an elastic joint to restrain the motion. Since the posterior portion of the FSU guides the vertebra movements, the type of relative motion possible at any level of the spine is determined by the orientation of facet joints with respect to the transverse and the frontal planes. This orientation varies along the spine main segment [MRD*04]. In the lumbar region, the facets are oriented at 90° angles to the transverse plane and at 45° angle to the frontal plane [PW80, NF01]. These orientations also vary from one person to another. This alignment allows flexion, extension, lateral bending but no axial rotation. The facet joints guide movement of each FSU and have load-bearing function. The load supported by the spine is mainly transfered across the whole spine through the discs and the zygapophysial joints [NF01].

Based on this, we choose elastic joints with six degree of freedom [De 00], three in translation and three in rotation, to model the elastic behavior of the capsules, with different stiffnesses for twist and stretch. To match to the anatomical position of the zygapophysial joint as accurately as possible, we set the elastic joints in the middle of the segment that pass through the centers of the two facets, as illustrated using yellow crosses in Figure 3.4. Their material properties are set based on the literature [SLK07] (see also Table 3.2).

3.3.5 Ligaments

The ligamentous apparatus of the spine mainly contributes to its intrinsic stability by allowing a balanced and restrained motion during the daily activities [ADK*13, NF01]. The ligaments connect pairs of bones, and are composed essentially of collagen and elastin with various proportions. The collagen limits the extensibility of the ligament during the spine motion. This specific composition provides the ligament with non-linear elasticity as shown in Fig.3.8. The stress-strain curve contains three main parts : the toe region, the linear region and the yield and failure region. Most physiological activity occurs in the toe and the linear regions.

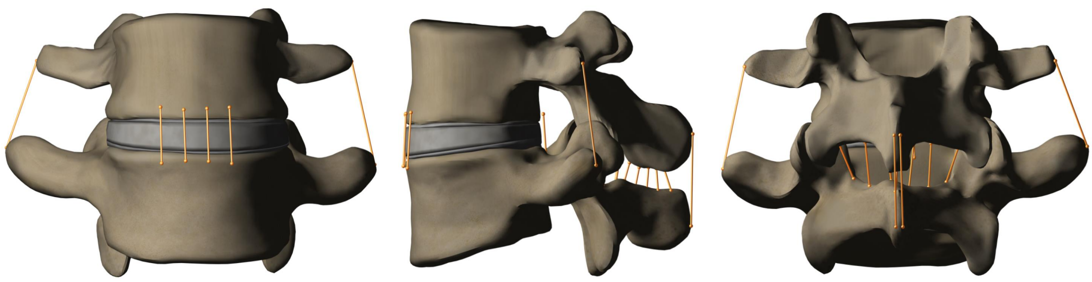


FIGURE 3.7 – *Ligaments in a functional spine unit of the model :The anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL), the ligamentum flavum (FL), the transverse ligament (TL), the interspinous ligament (IL) and the supraspinous ligament (SL).*

Our spine model includes six ligaments, as shown in Fig.3.7 : the anterior longitudinal ligament (ALL), the posterior longitudinal ligament (PLL), the ligamentum flavum (FL), the transverse ligament (TL), the interspinous ligament (IL) and the supraspinous ligament (SL). Each ligament is modeled using a set of one-dimensional tension-only spring elements (orange segments in Fig.3.7), and its elastic behavior is defined by a strain-stress function. Differentiable stress-strain laws are necessary to efficient implicit numerical solvers [PTVF07], while the experimental laws described in the literature [CTB*85, SLK07] are composed of discrete sample points. We thus approximate these using smooth curves optimal in the least-square sense, and we propose to use sigmoid curves, which fit the data reasonably well, as shown in Fig.3.8.

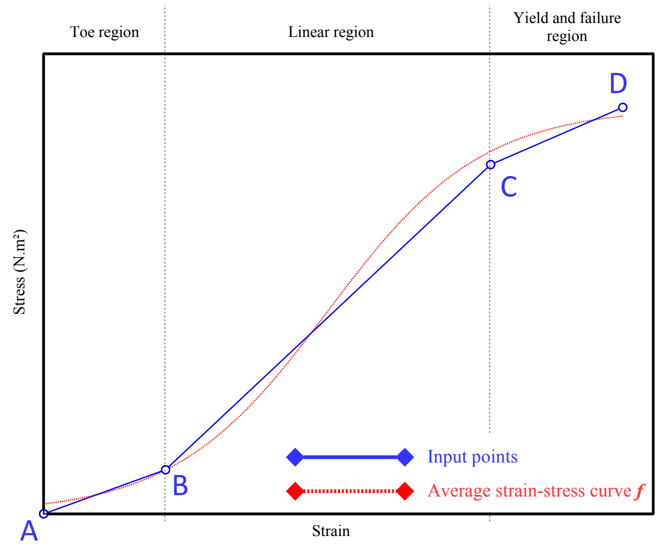


FIGURE 3.8 – *Approximation of the stress-strain curve of a ligament. Blue : the input data defined by 4 points (A,B,C,D). Red : the corresponding approximated sigmoid function.*

3.4 RESULTS AND VALIDATION

Three experiments were performed to validate the range of motion, the instantaneous center of rotation and the disc of the lumbar spine model. All of them were done using the SOFA library [FDD*12] on a laptop with a *Intel Bi-Core i7-3520M CPU @ 2.90GHz* and 8Go of RAM. The dynamic simulations were performed with small time steps (dt=0.0001s) and an implicit Euler integration scheme. The damping parameters of our Implicit Euler solver are respectively 0.001 for the Rayleigh stiffness and 0 for the Rayleigh mass.

3.4.1 Range of motion

The first validation of the model consists in simulating the range of motion (ROM) between L4 and L5 when a torque of 2.5, 5, 7.5 and 10 N.m is applied atop L4 (*cf the point of application shown in Fig.3.9(a)*) in the three anatomical directions (Fig.3.10). Torque is applied until the convergence to an equilibrium state. The angle between the two vertebrae (cf Section 3.4.1) is computed as shown in (Fig.3.9(c)). The range of motion is compared with those obtained by [WAA*12, TBA82, YPCO89, POYC94, SHD*07].

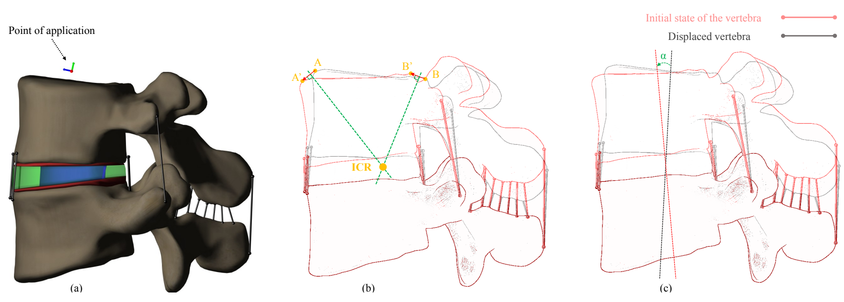


FIGURE 3.9 – *(a) : Point application of the pure moment. (a) : Determination of the ICR. (c) Computation of angle*

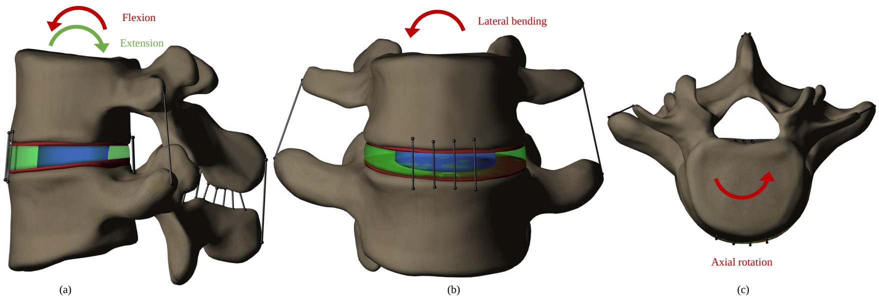


FIGURE 3.10 – *Functional spine unit and the motions in the main anatomical planes. (a) Flexion - extension. (b) Lateral bending. (c) Axial rotation.*

Our results are similar with the standard values of those found in the works to which the model is compared. For a torque of 7.5 Nm, the model performs a flexion of 5.73°, an extension of 5.7°, a lateral bending of 6.1° and an axial rotation of 2.4°. The results for the torques with different intensity are summarized in Fig.3.11, 3.11, 3.11, 3.11. It is interesting to notice from these graphs that the extension is limited to 5.8° despite the increase of the applied torque. This is due of the contacts that happen between the two spinous processes (Fig.3.11). Without these contact the angle between the two vertebrae in extension would exceed the 7° (Fig.3.11).

To validate the range of motion of the whole lumbar spine, we apply a torque of 7.5 Nm on a point atop L1 of the segment L1-S1 and compare the angles between each pair of adjacent vertebrae. Results are compared with those obtained by [YPCO89, WP90, RNP*07]. The angular amplitude between L1 and S1 is 49.95° for the flexion-extension. In this global movement, we get 20,34% of the ROM which is the flexion-extension of L1-L2, 16.17% is due to L2-L3, 18.15% is due to L3-L4, 21.65% is due to L4-L5 and 23.69% is due to L5-S1. The lateral bending shows 33.17° (24.31% L1-L2, 19.35% L2-L3, 18.86% L3-L4, 18.92% L4-L5 and 18.56% is L4-S1). The axial rotation shows 13.99° (22.64% L1-L2, 19.35% L2-L3, 18.86% L3-L4, 18.92% L4-L5 and 18.56% is L4-S1). The amplitudes of each pair of vertebrae on each global movement and the comparison with the literature data are detailed in Fig.3.12, 3.12, 3.12. As it was the case for L4-L5, the ranges of motion in flexion-extension of the hybrid spine model are in the standard values of the results of further works. The ROM in the coronal plane and in the transverse plane is slightly larger than the validation data but remains valid if the comparison is carried out for each FSU independently rather than in the global movement of the whole lumbar spine.

3.4.2 Instantaneous center of rotation (ICR)

To validate the quality of the motion resulting from our hybrid model, we compute the trajectory of the ICR at each time step of the same experiments performed to validate the ROM. These centrodes are computed using the method proposed by [PB88] (Fig.3.9(b)). In our experiment, between 50 to 200 steps of simulation are performed instead of the 10 steps in [SHCW08]. This large number is motivated by the need of accurately following the motion of the centroid. The ICR of L4-L5 are computed and compared with the results of [PB88, SHCW08]. Furthermore, the ICR of each FSU of the whole lumbar segment were computed (cf Fig.3.13) and compared with those obtained by [PB88, RBH*06, SHCW08, WHH*10, BCCR12].

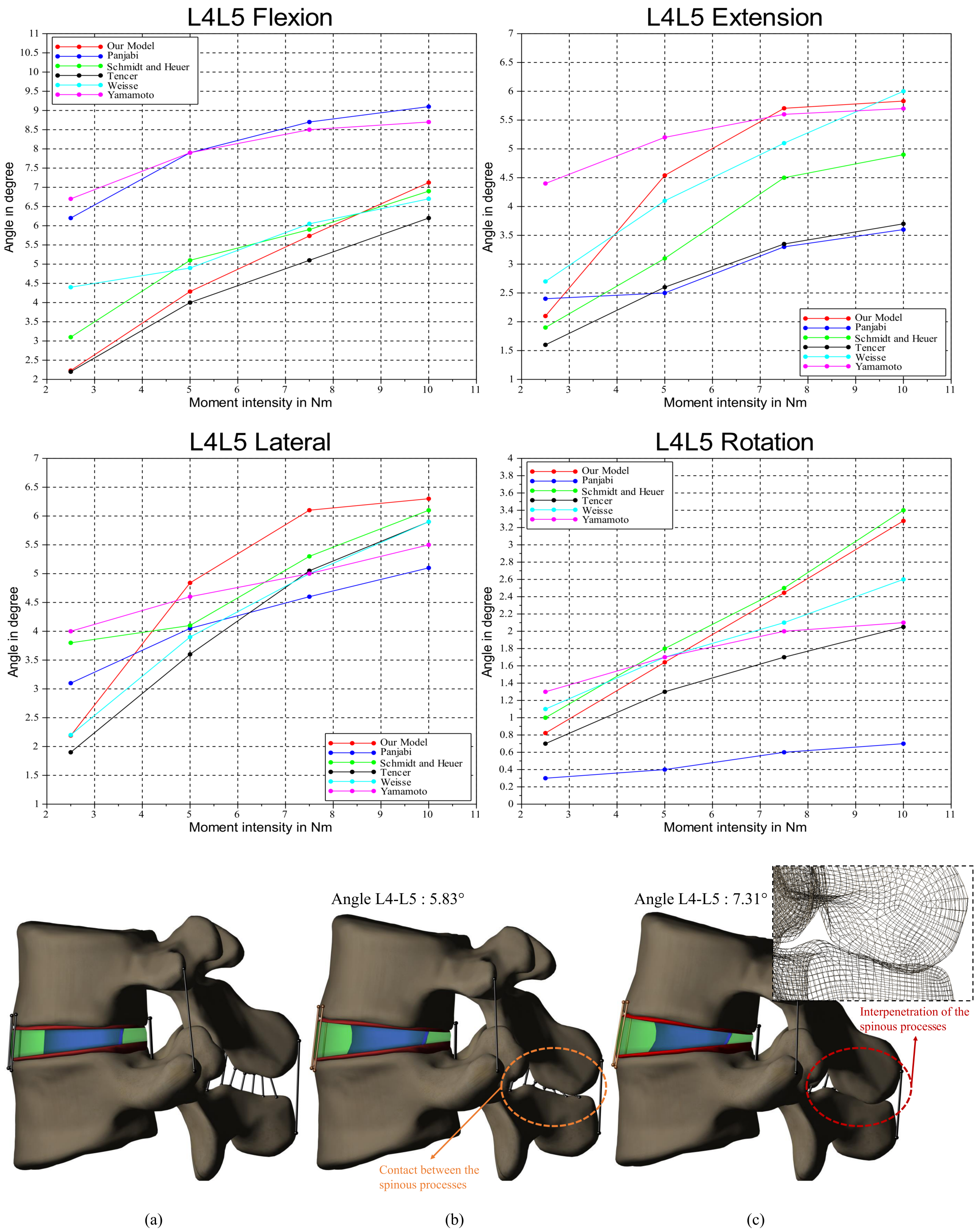


FIGURE 3.11 – ***Top :*** *Comparison of range of motion for L4-L5. Our model is represented by the red line.* ***Bottom :*** *Contact between spinous processes during the extension. (a) : L4-L5 initial state. (b) : Complete extension with the handling of the contact. (c) : Complete extension without the handling of the contacts.*

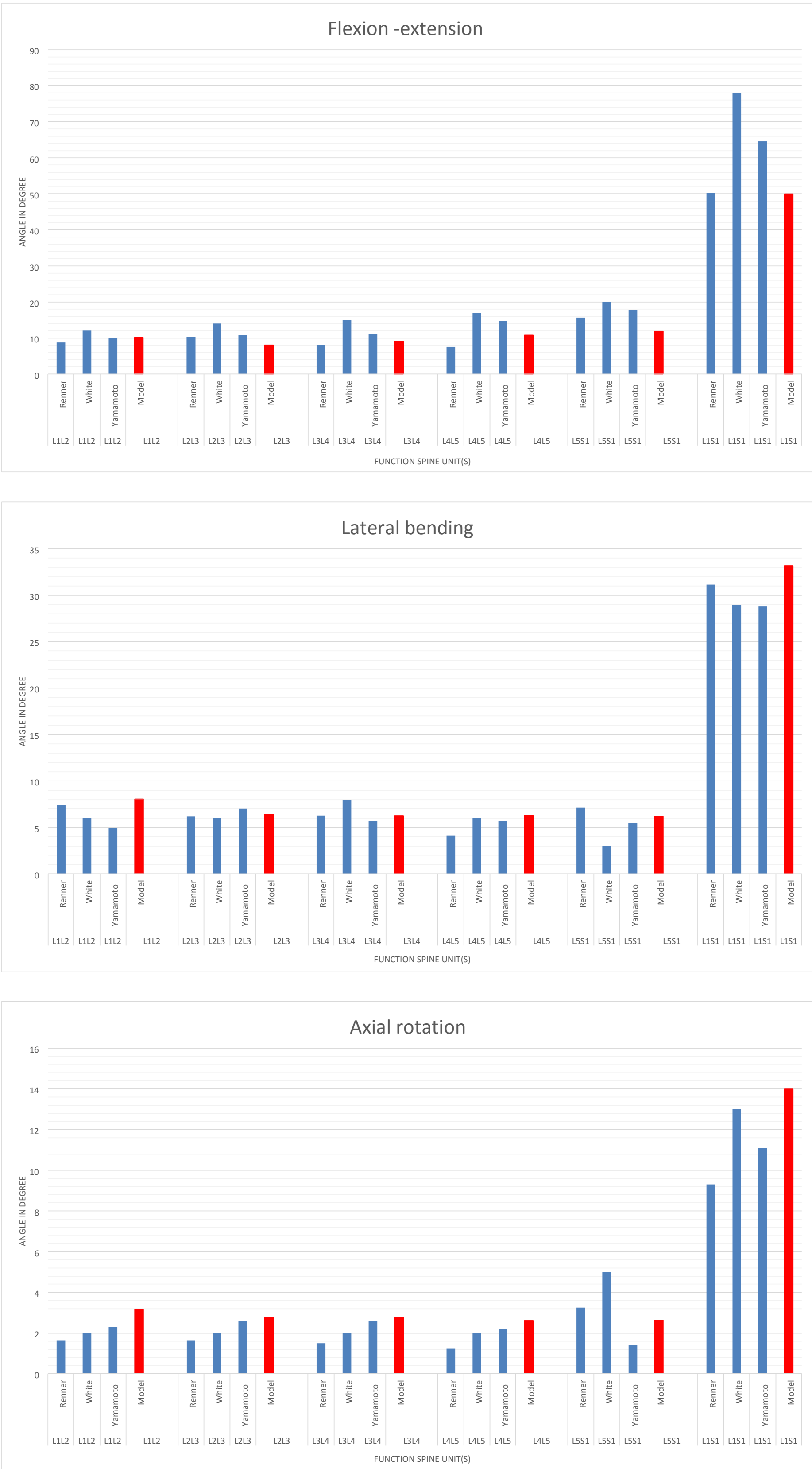


FIGURE 3.12 – *Comparison of range of motion for all FSU during a motion of the whole lumbar spine.*

In flexion, each FSU predicts ICR position in a region that starts from the center of the intervertebral disc on its posterior part, then migrates posteriorly across the disc to finally ends its pathway in the upper region of the vertebral body of the bottom vertebra (cf (A) in table 3.3). The points that describe the centrode motion are located in this region independently to the moment intensity and the FSU as it is shown in table 3.3 and Fig.3.13. These trajectories fit the results obtained during a flexion by [RBH*06, BCCR12] for L2-L3 and L5-S1, respectively. The localisation area of the ICR in flexion is also supported by the results shown by [PB88]. The ICR of the hybrid model are in the same region of those obtained by [SHCW08], but the pathway obtained remains quite different.

During the extension, the FSU predicts ICR positions in a same area as the ICR of the flexion, but their trajectory during the extension are different (cf (B) in Fig.3.3). The region of localisation of these ICR is supported by the results obtains by [PB88], but the position of the centrodes for L3-L4 and L4-L5 were not shown for the extension on its works. For L2-L3, the centrode motion obtained with the hydrid model have a same type of shape and are located in same area as those obtained by [BCCR12], the observation is the same for L5-S1 if the comparison is made with the results obtained by [RBH*06].

In lateral bending, the ICR was located almost in the upper part of the disc, the shape of the trajectory reminds a bell shape that starts on the bottom part of the vertebral body, continue across the disc until its center and then migrates toward the side of the bending (cf (C) in Fig.3.3). The intensity changes the centrode trajectory but the global shape of the centrode movement remain the same. The bell shape of the ICR and the area of the ICR position during a lateral bending are validated by the results obtained by [SHCW08].

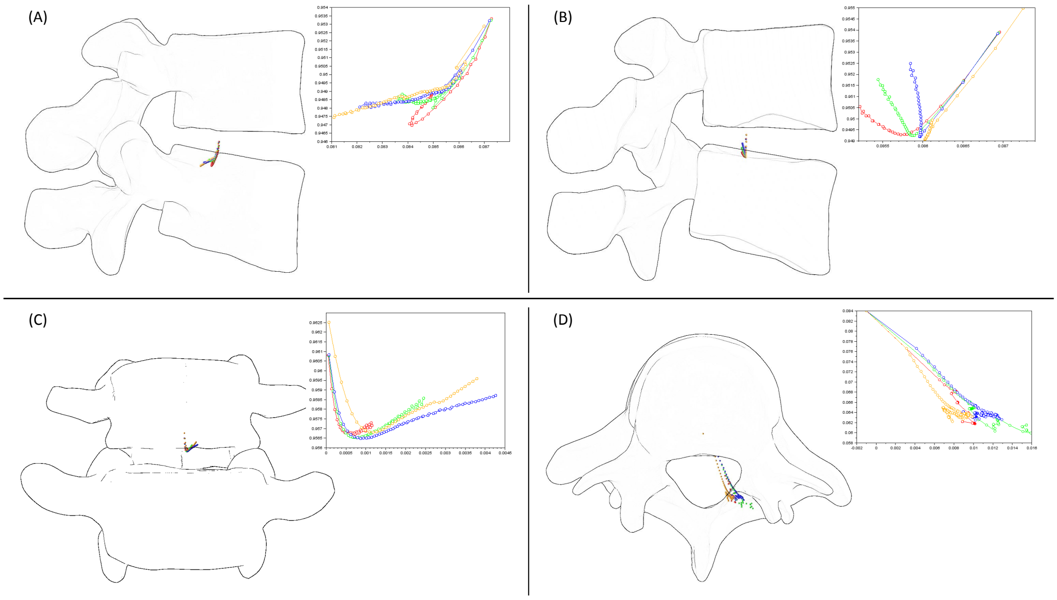
In axial rotation, to validate our ICR position, we add a compression preload of 500 N before to apply the moment as [SHCW08] since it is has been proved that the shape of centrodes during axial rotation are greatly dependent to the pre-load (compression, pre-flexion/extension) [WHH*10]. This compression is applied on the surface atop the vertebra L4. The trajectory of the axis of rotation follows a pathway that starts in the posterior part of the vertebral body. With the increase of the rotation angle, the centrodes migrate posteriorly to cross the disc and the spinal canal, to ends its pathway near the facet joint as illustrated in (D) in table 3.3. This result is validated by those obtained by [SHCW08] and [WHH*10].

3.4.3 Disc mesh refinement

This last stage aims at validating the FE-mesh refinement (i.e resolution) of our hydrid disc (Fig.3.14(c)). To achieve this, we create three other discs and compare the ROM and the trajectory of the ICR of the L4-L5 FSU with each of these disc models for flexion,extension and lateral bending. The first disc has the same resolution in the nucleus and the annulus as the hybrid model, but instead of undeformable plates, we model the two bony plates using an FEM model as is usually done. This model, shown in Fig.3.14(b), contains 6357 solid elements. Since our goal was to simplify the disc for a more interactive simulation, the obvious simplification is to coarsen the FEM model to improve the performance. This model with lesser elements (1088 solid elements) is the second one in Fig.3.14(d). We also create a model with higher resolution (29339 solid elements). The mechanical parameters of all these models are the same as those proposed by [SLK07].

The results are summarized in tables 3.4 and 3.5. We can see that considering the plates as rigid reduces the mobility by only 1.5%, with an increase of computation time by a factor of 5 (Table 3.4). The ICR trajectories of these two models remains really close as it is illustrated in the table 3.5. Therefore, rigid plates seem a reasonable and useful simplification. Reducing

Table 3.3 – *ICR of the L4-L5 FSU subject to different moment.* ***Top left*** *: the ICR during the flexion.* ***Top right*** *: the ICR during the extension.* ***Bottom left*** *: the ICR during the lateral bending (Left). On the* ***Bottom right*** *: the ICR during the axial rotation. The red curve correspond to a result with a moment of 2.5 Nm, the blue correspond to 5 Nm, the green correspond to 7.5 Nm and the orange correspond to 10 Nm.*



the number of elements leads to a loss of more than 20% of mobility, and significantly affects the trajectory of the ICR. Increasing the resolution results in a 2.4% increase of the mobility, which remains consistent with the literature results. However, the computation time is 38 times higher. Our medium resolution model seems a good trade-off between precision and speed.

Table 3.4 – *Comparison of the range of motion of L4-L5 spine unit with different type of disc.*

	High resolution	***Medium resolution***	***Hybrid disc***	***Low resolution***
Range of Motion (in °)				
Flexion	6,05	5,84	5,73	4,59
Extension	5,83	5,77	5,70	4,94
Lateral Bending	6,46	6,25	6,10	4,65
Axial Rotation	2,47	2,43	2,44	2,38
Other values				
Number of Elements	29339	6357	4025	1088
Frame per second (fps)	0,2	1,4	7,6	30

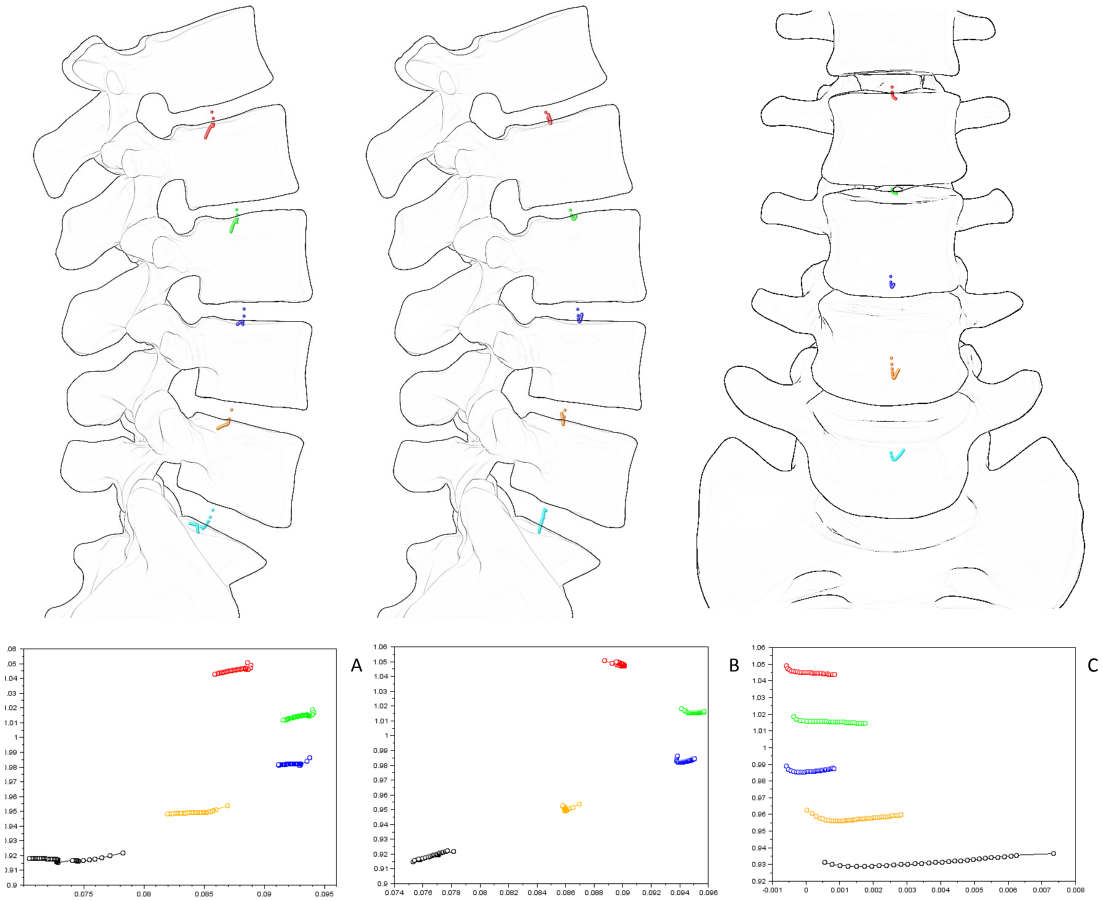


Figure 3.13 – *ICR of all the lumbar spine. On the top :* ***flexion****,* ***extension*** *and* ***lateral bending on the left side****. On the bottom :* ***A : flexion****,* ***B : extension****,* ***C : lateral bending (Left)*** *and* ***D : axial rotation***

3.5 Discussion

The main objective of this work was to build a novel hybrid model of lumbar spine which enables a better computational efficiency while respecting anatomical and physiological description of this complex organ.

As [SLPF11], our results of simulations reported here demonstrate the effectiveness of the coupling between rigid bodies and finite element model. The model shows that the assumption of simplifying the bones (vertebrae) as undeformable rigid bodies does not lead to any loss of accuracy in the quality of the movement and in the range of motion of the lumbar segment in the three anatomical planes. The validations we provide show that our model has a physiological movement with a motion amplitude that fit anatomical reality. All these results are in agreement with the literature.

One of the main advantages of the hybrid model is first of all its ease of construction. The only entities that require a meshing stage are the disk, and it is well known that this step is usually time consuming and a complex task. Create different configuration for medical experiment become easier and quicker. The position of the rigid body, and all the parameters that describes each vertebra are automatically computed based on the geometrical shape of the vertebrae. The other main advantage of the model is the computation time of the simulation. The

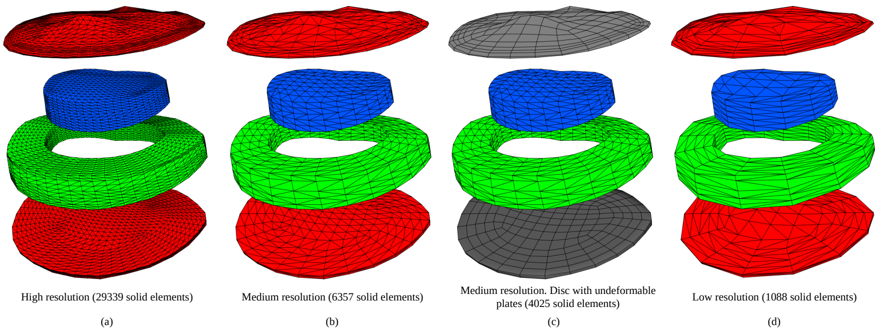
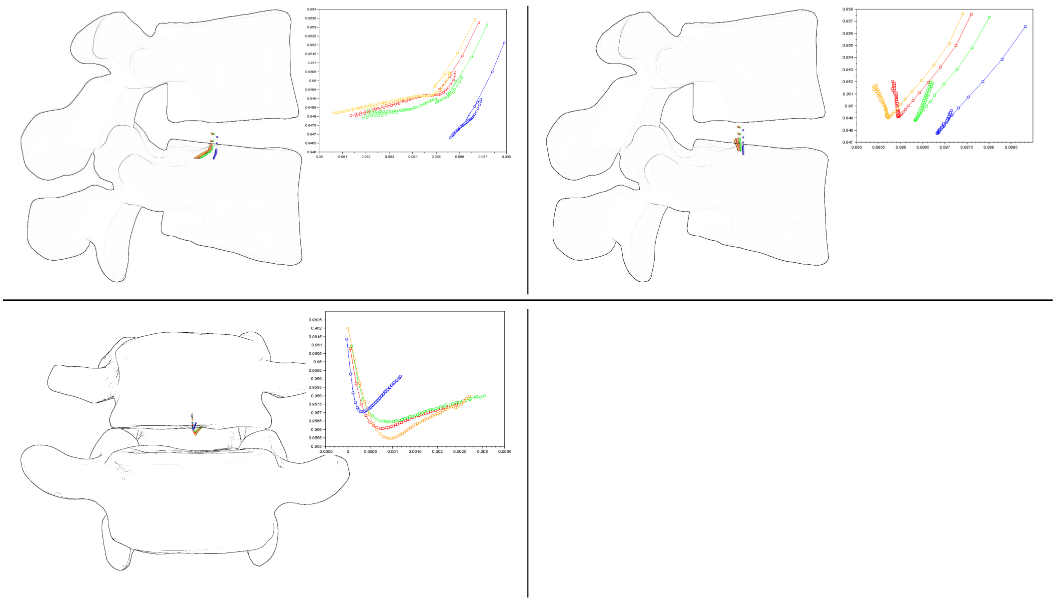


FIGURE 3.14 – *The four discs created for the validation of the hybrid disc*

TABLE 3.5 – *ICR of the L4-L5 FSU subject to different moment with different disc models. On the **Top left** : the ICR during the flexion. On the **Top right** : the ICR during the extension. On the **Bottom left** : the ICR during the lateral bending (Left). The red curve correspond to the result of a disc with the medium resolution, the blue correspond to the disc with the low resolution, the orange correspond to the disc with the high resolution and the green correspond to the hybrid disc.*



simulation of one FSU from the beginning of the movement until its stabilization spends less than 6 seconds. For the simulation of the whole spine, the simulation takes less than 7 minutes from the beginning until the stabilization. The acceleration of the time computations is mostly due to the lower number of degree of freedom that the model contains. The FSU L4-L5 made by [ADK*13] contains 270 324 solid elements for both disc and the two vertebrae, for the same FSU our model only required 4025 solid elements for an equivalent result that is proved by the validation.

Despite the difference in the load and the force transfer we can observe between an FEM and a rigid body more generally [SA94], our hybrid segment of spine shows through the ICR trajectories the quality of the movement of the model which is a consequence of the transfer function within the spine segments. These results are supported by the in-vitro and simulation experimentations of further works. It is interesting to notice that theses accurate movements are largely due to the behaviour of the zygapophysial joint that confirmed that these entities can be represented as elastic joints with six degree of freedom per joint : three translations and three rotations. One other physiological aspect in our model is the extension movement limit of the model due to the contacts occurring between the spinous processes which is a phenomenon that usually happens in reality, but it is not described in simulated models.

However, our hybrid model shows some limitations. It is not the most suitable representation of the spine for some studies like the impact of the bone porosity in the spine transfer function or any study that involves some changes in the internal behaviour of the vertebra. But even for this type of study, the concept of hybrid model may be beneficial. This kind of work is usually done on a pair of vertebra, only the mobile vertebra can be represent with an FEM since the other one is fixed by constraint and then can be model with a rigid body. Except these cases, our hybrid lumbar spine best suited for all other studies with its easier and quicker modeling process due to the lower number of input parameter and DOF. The simulations are for the same reasons faster without any lack of accuracy as we demonstrated in Section 3.4.

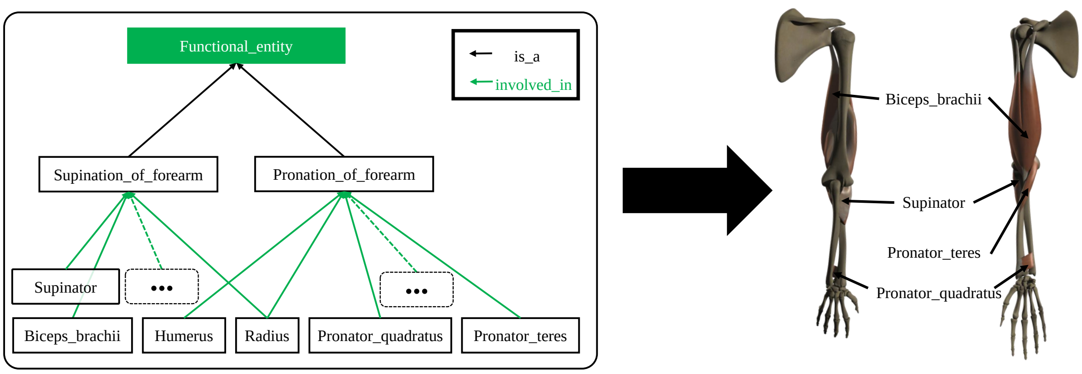
3.6 Conclusion

We have presented a novel hybrid model of the lumbar spine which combines both rigid bodies and FEM in the same model for the sake of computational efficiency. We described how the model have been constructed based on the anatomical and physiological behavior of each component of this segment of spine. The model has been validated in agreement with the literature. This stage of construction and validation of the model was the first step before its use for medical and biomecanical purposes. In future work, we plan to perform more validations as those of facet joint force, and intradiscal pressure during the main spine movements. All the previous and future validations will be compared against those presented in the recent work introduced by Dreischarf & al [DZSA*14]. In this work, eight FE-model of lumbar spine are compared against each other for a better evaluation of each, and for a better understanding of the lumbar spine. We also plan to experiment mesh-less, frame-based deformations [FGBP11], to remove the last step of meshing that remains in our modeling process and further accelerate the computations.

CHAPTER

4

ONTOLOGY-BASED MODELING OF MUSCULOSKELETAL SYSTEM



Teaser Figure : *Formal description of forearm functions in MyCF ontology, with geometries associated to each anatomical entity which contributes to the forearm functions.*

4.1 INTRODUCTION

The simulation of the human musculoskeletal system is ubiquitous in Biomechanics, Computer Aided Medicine and Computer Graphics. However, most of the models used for these purposes are relatively simple and generic due to the difficulties inherent to their construction. One of the biggest difficulties in the setup of simulations is the selection of the relevant anatomical entities (ae) (*i.e various bones, muscles, ligaments and organs*) which may have an important role, depending on the phenomenon or function to simulate. This is further complicated by the fact that a plethora of physical parameters are required for correct numerical simulation. Moreover, it is often necessary to incorporate additional subject-specific data.

The work presented here is initially motivated by the need to speed up and to ease the design process of specific human bodies for simulation purposes. This design approach can thus be used to simulate anatomical systems in movement, or the interactions of medical devices with body parts. For instance, studying how a hip or knee prosthesis interacts with individuals with various morphologies during flexion/extension of the knee, or during abduction/adduction of the hip. The first step for this approach is to create a biomechanical model specific to the subject. However, this is a very time-consuming and error-prone task, generally difficult to achieve for a mechanical engineer working on devices.

In this work, we introduce of a novel pipeline for tackling this challenge. This pipeline allows a user to select a set of desired functions to simulate, and based on a functional description, a list of approriate anatomical entities is constructed. This construction is automated using formal knowledge structured in ontology. Intuition of using ontology for creating simulation and manipulating (bio)mechanical parameter is not a novel concept in computer graphics. Previous works such as [HHP10, PBJ*09, GGRT*07, CGMT07] demonstrate the great potential of ontologies as organized knowledge bases to manage and to create complex models. For the purposes of this work, we introduce a new ontology of anatomical function. This novel ontology is constructed as a part of My Corporis Fabrica (MyCF) ontology [PBJ*09]. It extends FMA [RM03] to the formal description of organ functions, while remaining consistent with this reference ontology. The terminology of the taxonomy is validated againt the ICF [Org07] (*International Classification of Functioning, Disability and Health endorsed by the World Health Organization*). This ontology of our body physiology allows to generate the list of anatomical entities involved in a given function in response to a simple query. The relations within it make explicit how anatomical structures are composed, how they contribute to functions, and also how they can be related to data describing patient-specific body parts. Data such as 3D complex objects, or (bio)mechanical parameters are declared as instances of classes, and can be used for simulation or visualization. Using the resulting formal description, a physical model based on a reference geometry and mechanical parameters is created. Thanks to the pipeline, this mechanical model can then be personalized using subject-specific parameters, and registered to subject-specific geometric data such as volumetric images, to create a subject-specific model of the given anatomical function. The simulation of arbitrary biomechanical phenomena being beyond the scope of this work, we focus on producing computer simulations of standard tasks such as anatomy joint functions to illustrate the pipeline possibilities. However, the functional nature of our modeling pipeline permits extension to a much wider range of scenarios.

Our pipeline, illustrated in Fig.4.1, is targeted to non-specialists as well as expert users. We show how to generate simulations using standard open source simulation software, and to obtain visually pleasing results. The remainder of this chapter is organized as follows. The anatomical knowledge base and the modeling pipeline are presented in Section 4.3. The simulation tools used to perform and to validate the modeling are presented in Section 4.4, and results are finally shown and discussed in Section 4.5.

4.2 OVERVIEW

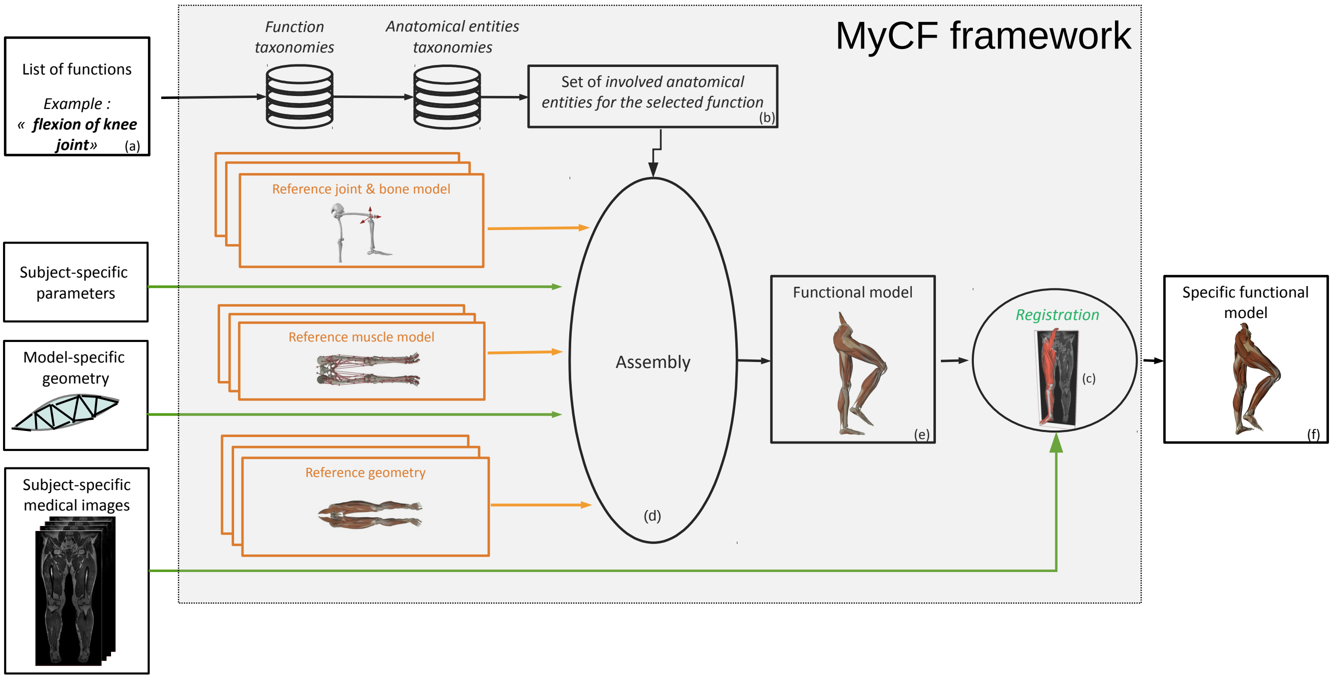


FIGURE 4.1 – *An overview of our modeling framework. On the left, the user input is a list of functions to simulate, optionally complemented with specific data. On the right, the output is a mechanical model ready for simulation. The modeling pipeline uses symbolic knowledge to select anatomical entities to assemble. The final model can be composed of a mix of reference and specific parameters and geometries.*

Our modeling framework is summarized in Fig.4.1. The main input is a set of anatomical functions that the user wishes to simulate (*e.g. hip abduction, elbow flexion, wrist extension, walking . . . etc*). Our system then produces a list of anatomical entities involved on these functions. Just like the anatomical functions, anatomical entities can also be the input of the framework. Each entity within the system is linked to an instance of the reference model with the mechanical parameters and geometric data required for the simulation. Our reference model is the zygote body [zyg]. A set of mechanical parameter for this reference model required for simulation have been computed and included in the system. A generic model is then created. This model can be complemented with subject-specific information (*e.g register geometries, specific mechanical parameters, attributes . . . etc*) in order to obtain a subject-specific simulation. The output contains the mechanical parameters and the geometric representation of the relevant anatomical entities involved in the chosen functions. This section presents in more detail the different modules of the pipeline. Examples of construction of entities and functions will be used to illustrate the role of each module of the framework for an easier understanding of the pipeline. Furthermore, a specific attention is paid to the construction of the ontology of function set up to complete MyCF and also to address this problematic of building models of anatomy in movement for simulation purposes.

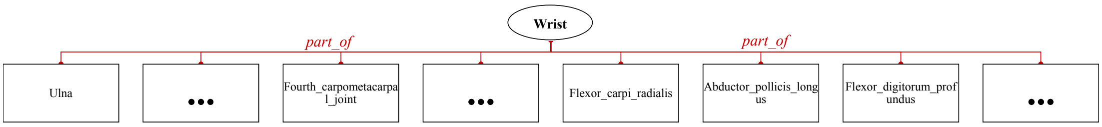
4.3 Modeling Framework based on ontologies

4.3.1 Symbolic description of anatomical functions and entities

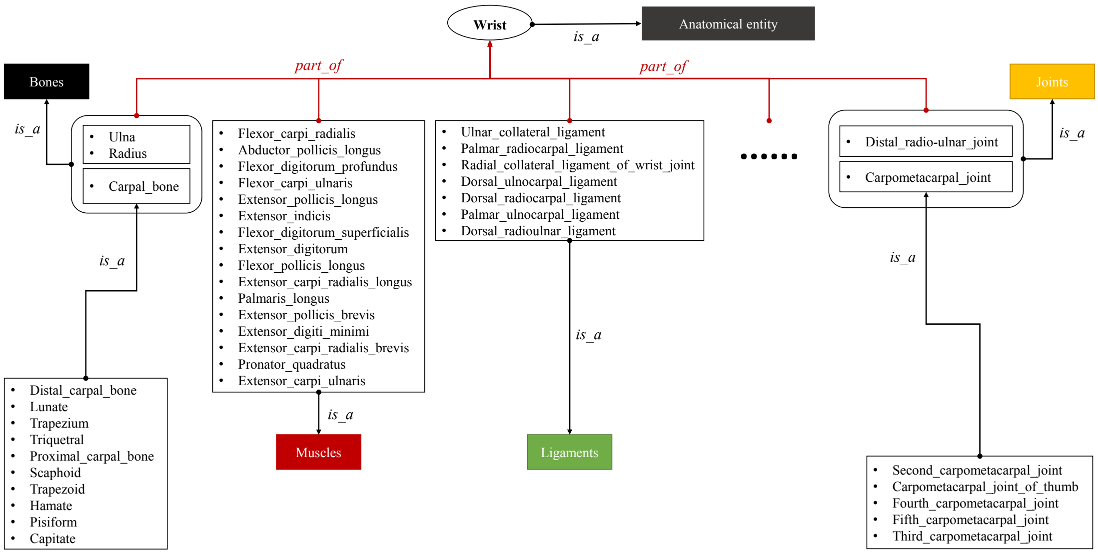
Biomechanical simulations require an accurate anatomical description. We base our framework on symbolic knowledge of the human body, organized in a widely accepted ontology of human anatomy called My Corporis Fabrica [PBJ*09]. This ontology describes our body as a set of anatomical entities(**classes**) connected by **relations**. MyCF is based on the Foundational Model of Anatomy (FMA) [RM03], which is the reference ontology for the human anatomy [RM03]. FMA consists in more than 135,000 terms which point to 75,500 types, and which are interrelated by over 2.5 million iterations of 198 kinds of specific relations. From all these elements of FMA, MyCF considered the entire anatomy taxonomy, and only two relations. It kept the hierarchical relations which allow to navigate through this complex set of entities describing consistently the anatomy, and the relations which allow to construct complex model of organ. In addition to what is extracted from FMA, it add one new relation : *inserted_on*. The relations in MyCF are :

- ***is_a :*** This relation describes each entity as a child of another one. It refers to the inheritance of property.
  Example *: The **femur** is a **bone organ** ↔ **femur** is_a **bone organ**.*
  femur *is an anatomical entity (class),* ***bone_organ*** *is an anatomical entity (class), and* ***is_a*** *is the relation between those.*
  This relation is the most common one in MyCF, it is used to classify concepts. Every entity within the ontology is a child of another entity by the *is_a* relation. In the context of the construction of anatomical system, this relation is used to determine which is the type of a concept, for instance it enables to find if an entity is a bone, ligament, muscle, joint, . . . etc. Thereby the right mechanical models or properties can be associated as instance of the concept. This specific aspect allows to ensure some consistency in the process of creation of simulation model, by restricting the representation choices for each entity. For example, it would be impossible to associate a Hill muscle model to the concept *right femur* because it is a bone, and not a muscle. It also eases the search of components by type in the ontology and in the simulation scenes, for instance *which are the bone in the wrist ?* For such a query, the system will look within all the entities which compose the *wrist*, and will return only those which have a bone as parent on their hierarchy. During the construction of a model, the *is_a* relation allows to classified component by type for a better readability of the created scene (*scene refers to the model created to represent an organ for simulation*).
- ***part_of :*** This relation represents each entity as a part of another one, and consequently denotes the partonomic inclusion [SR04].
  Example *: The **patella** is a part of **knee** ↔ **patella** part_of **knee**.*
  The *part_of* is used to make explicit the subparts of anatomical entities, which represents an important anatomical knowledge for the construction of system. It is a hierarchical relation. To assist the modeling of anatomical systems, this relation enables the extraction of all the anatomical components within a system. In this sense, it has been widely used to assist the segmentation process in [HHP10]. The relation *part_of* is complementary to the relation *is_a*. By using the relation *is_a*, the system can classify all the concepts from

an output of a query based on the relation *part_of* as illustrate in the following example.
Example *: Consider the following query : what are the sub-part of the wrist ? The system output is the graph illustrated in Fig.4.2a which contains a list of 123 components (i.e bones, bones sub-parts, tendons, muscles, ligaments, joint . . . etc). This long list of items is tedious to analyze, and could be longer for other entities. If we sort the output concepts by using the is_a, we obtain the result illustrated in Fig.4.2b, which is obviously preferable if we desire to associate mechanical parameters according to the type of each concept. This classification also eases the output readability for educational purposes for instance.*



(a) *MyCF output to the query "which are sub-part of the wrist ?"*



(b) *MyCF output sorted with the relation is_a. This output is the answer to the query "which are sub-part of the wrist ?"*

FIGURE 4.2 – *MyCF output to a query using the relation* $part\_of$

This relation can also be used to complete a system, for instance, if we consider all the sub-parts p_i of a system S, by looking each p_i is a sub-part of which other systems S_i, the framework can first of all determine all the adjacent systems S_i to S, and if necessary can extend S to these additional systems.
Example *:* ***femur*** *part_of* **knee** $\rightarrow$ ***femur*** *part_of* x *?* $\rightarrow x = \{hip\}$.
By adding the hip and knee into a unique model, we can create a system that allows to better understand the role of the femur, or the impact of the knee function on hip, and . . . etc.
Both *is_a* and *part_of* are ubiquitous in bio-informatics ontologies and terminologies. With these two relations, it is already possible to build various complex anatomical models for the visualization or simulation purposes. Connections between entities de-

duced through inference using logical tools are considered within the framework (e.g $A\ is\_a\ B, B\ is\_a\ C \rightarrow A\ is\_a\ C \ldots$ etc). In the way of using them, their potential and some uses of these two relations are described in more detail in [SR04].

– ***inserted_on :*** This relation is mainly used for the ligaments, muscles, tendons and other soft tissues. It is used to specify attach points of anatomical entities, and describes the fact that one entity is inserted on another one. It is also used to locate parts of our anatomy.
Example *: The **vastus medialis** is inserted on the **femur** ↔ **vastus medialis** inserted_on **femur**.*
This relation is essential in anatomy and for biomechanical simulation purposes. It is specific to MyCF, and has been introduced because of its great value in the process of learning musculoskeletal system. To assist the modeling, it speeds up the determination of the entities on which each muscle and ligament is attached during the simulation set up, in order to know how to manage attachment constraints.
Example *: In which entities the right_sartorius is inserted on ? ↔ right_sartorius insert_on x ? → x = {right_tibia, right_hip_bone}.*
The *inserted_on* also helps to complete system by finding in which other anatomical systems or entity (*i.e bones or bony part*) a musculoskeletal component is connected, to better understand this role, observe the following example.
Example *: Lets build a model of the right_knee by using the relations part_of and is_a. We obtain all the muscles and ligaments as illustrated in Fig.4.3.*
If we simulate this system, numerous muscles within the system will be attached only on their side that are included on the right_knee (i.e right_femur, right_patella, right_tibia, right_fibula), their other side will be free and thereby can skew the simulation if no additional constraint are applied.
*Let's look at which bones each component is inserted on ? ↔ x part_of right_knee, x is_a muscle, x inserted_on y ? → y ={right_femur, right_patella, right_tibia, right_fibula, right_hip_bone}. In addition to the knee bones, we can thus add the **right_hip_bone** to the existing system (cf Fig.4.3) .*

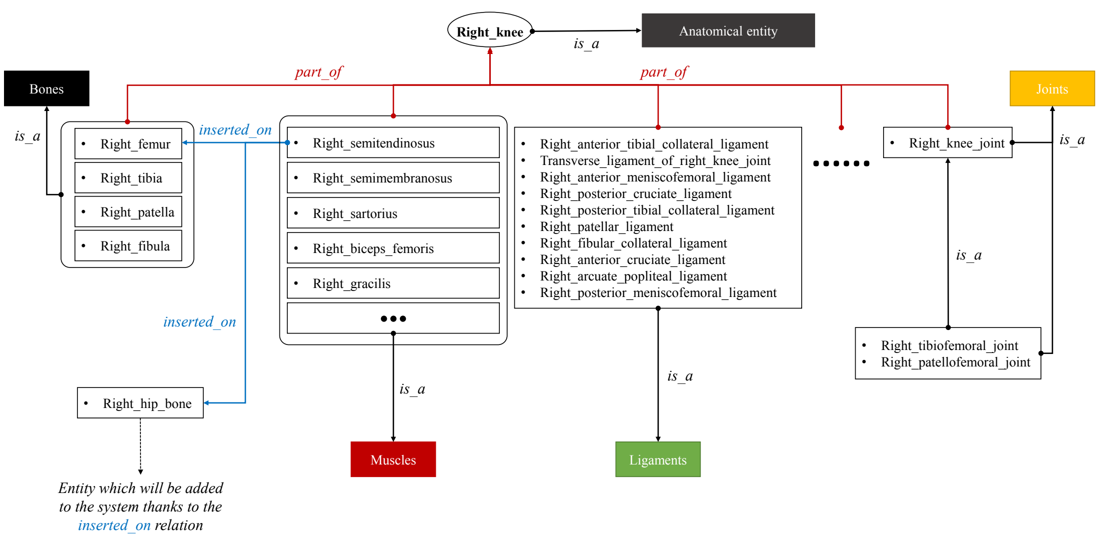


FIGURE 4.3 – *Completion of a right_knee system with the use of the relation inserted_on. The relation is only illustrated for the right_semitendinous muscle.*

In addition to the structural anatomy considered in FMA, MyCF extends it to graphical and to mechanical representation of anatomy [PBJ*09]. It also provides tools to visualize and to build complex models using the potential of ontologies as illustrated in the few examples above. The general purpose of an ontology is to provide a means of classifying individuals and data associated to them, in this sense, patient-specific data are defined as instances of canonical anatomical entities stored in the ontology. Anatomical entities are thus linked to data, and can be linked to **attributes** in the same way.

In order to complete the ontology and to ease the simulation of anatomical functions, we introduce a new ontology of human body functions consistent with FMA. This novel ontology consists of a taxonomy of human functions, along with the new relation ***involved_in*** used to represent that an ontology class is involved in a given function.

***Example** : **rectus femoris** is involved in **knee flexion**.*

*In this example : **rectus femoris** is the anatomical entity (class), **knee flexion** is the biomechanical action or function (class), and **involved in** is the relation between those.*

For the semantic validation of the terminology of organ functions, we base our classes description on the ICF standards [Org07], which defines the terminology of functions, and describes the various functions of the human body that should be taken into account in current medical practice. With this additional content, MyCF currently has two roots nodes : the *Anatomical_entity* and the *Functional_entity* whose classes are inter-connected by the relation ***involved_in***. In this taxonomy of anatomical functions, each organ function F is described as a set of sub-functions $\{f_i, f_i\ involved\_in\ F\}$ and a set of anatomical entities $\{e_i, e_i\ involved\_in\ F\}$ which participate to F (*cf the example illustrated in* Fig.4.4). Only the classes which actively participate are considered in the description. For instance, in the *flexion_of_knee_joint*, only the motor muscles appear into the output list because they are mainly responsible to the movement (agonist muscles), those which contribute to the stability, or the antagonist muscles do not figure in the function description. For the completion of the model, the other relations cited above can be used as shown in Fig.4.4. This ontology of functions, currently consisting of 4200 functions, was created using the *Protégé 4.2* software and was stored in *OWL* format.

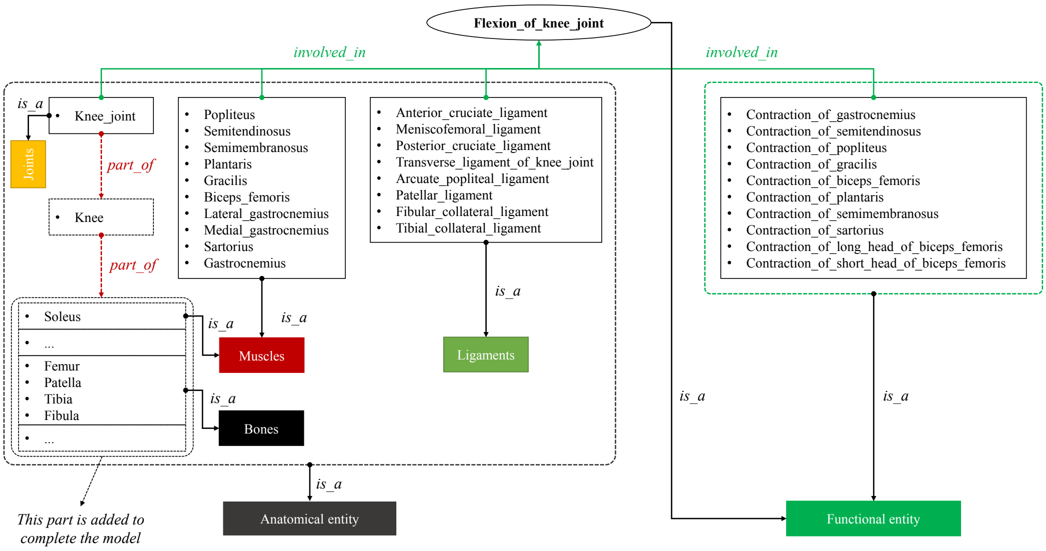


FIGURE 4.4 – *Functional description of the Flexion_of_knee_joint.*

The querying system to manipulate the whole framework is based on a web server system illustrated in Fig.4.5, with a MVC (Model-View-Controller) architecture for the application. Concerning the database structure, information can be found in [PBJ*09]

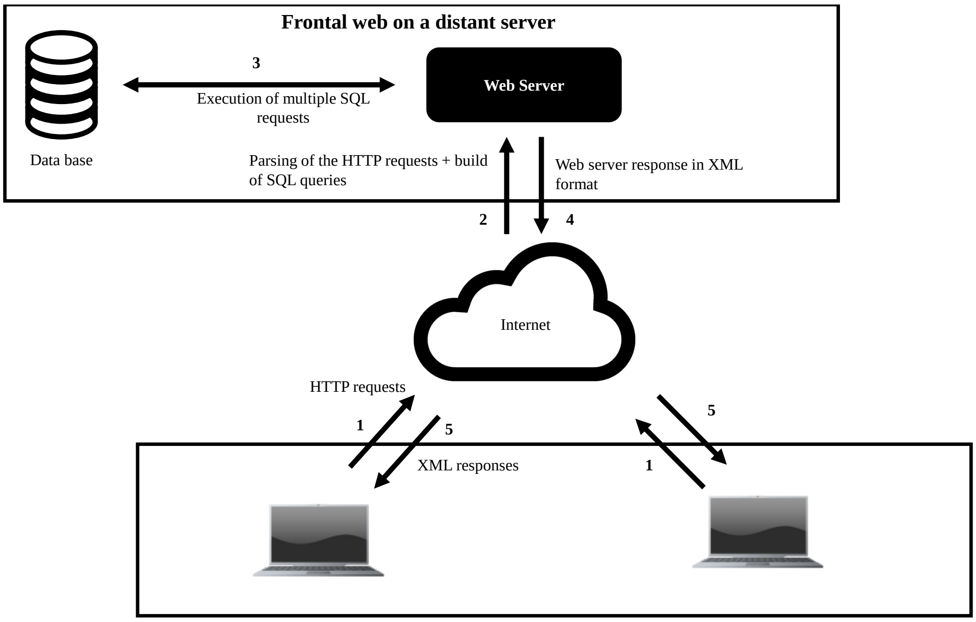


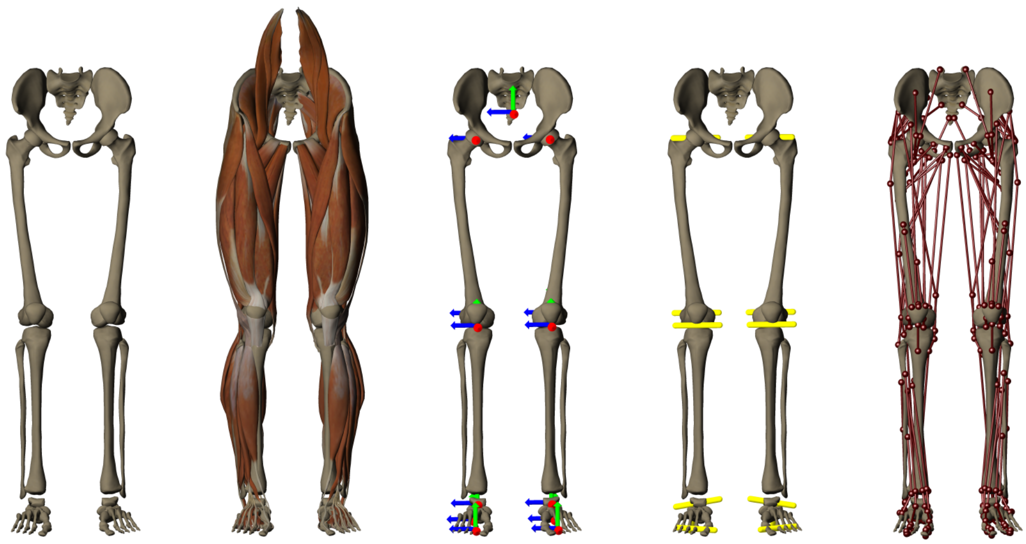
FIGURE 4.5 – *Web server functioning*

In this first stage of the pipeline, all these 4 relations described above are used to generate as output a formal description of the chosen inputs. This formal description is a classified list of the anatomical entities and the sub-functions relevant for the simulation.

4.3.2 Reference model

One key feature of our approach is the coupling of a universal symbolic description of anatomy with numerical data and parameters. For versatility, different levels of coupling are proposed. First, a collection of reference data is associated with entities in the ontology and in a second step, this reference model is enriched with custom data. The first version of our reference model includes :

- **Geometry** : some template bone and muscle surface meshes (cf Fig.4.6a, 4.6b). For now, we have chosen to use the commercially available zygote model [zyg] as our template for its completeness. It represents an average normal subject.
- **Joint and bone model** : we have defined joint centers and rotation axes with respect to surface models based on data from [mod], and registered into our reference model (cf Fig.4.6c, 4.6d). The joint limits are also based on [mod]. To model bone inertia, an average bone density is specified.
- **Active muscle model** : one-dimensional muscle paths equipped with functional parameters (cf Fig.4.6e). Currently, these models are adapted from Opensim [The03b], a widely used framework for musculoskeletal simulation. Because we focus on the functions of musculoskeletal systems which describe movements, we thus use this muscle representation since it is efficient for studying movements, while being fast to simulate, easier to implement and to set up. All the muscles parameters come from [AWLD10]. The data has been registered to the zygote model.



(a) *Zygote skeleton* **(b)** *Zygote muscles.* **(c)** *Rigid body DOF* **(d)** *Joint models* **(e)** *Hill muscle models registered on the zygote body*

FIGURE 4.6 – *Reference models*

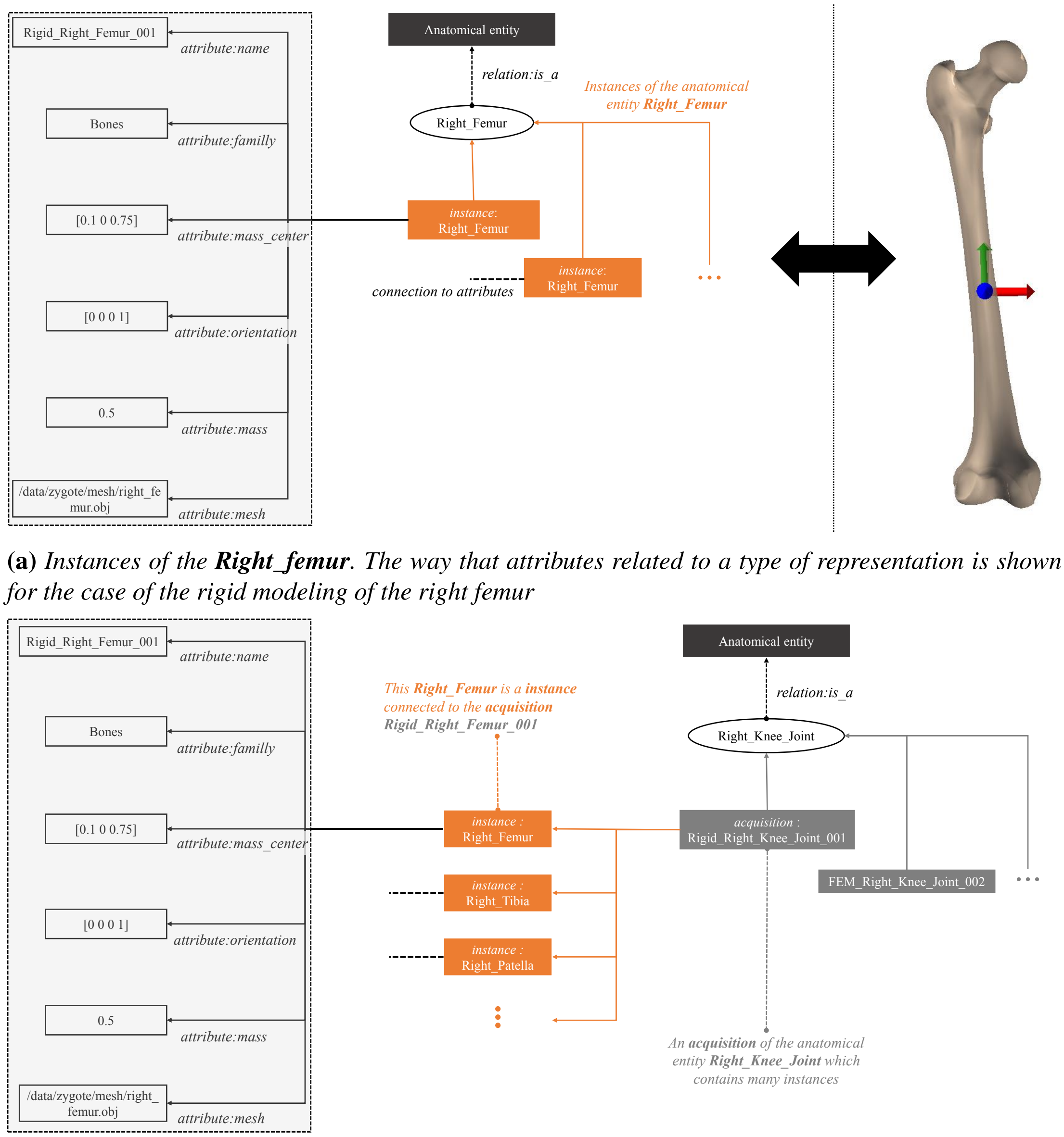
Only the lower limb is considered in the examples of simulated models, since only mechanical parameters for this sub-part of anatomy was available. This data collection is easily extendable, and we intend to include more detailed functional information in the future, such as local fiber orientations and rheological properties. Moreover, to make our reference model both generic and parameterizable, we will soon collect subject-specific data, to be averaged and integrated into a statistical atlas.

4.3.3 Assembly

Our framework allows the selection of the geometry, functional parameters, and even the nature of the representation (*e.g., Hill-derived muscles, surface or volumetric meshes with different discretizations, etc.*) as shown in the left of Figure 4.1, among a choice of built-in or external models. Once the selection of anatomical entities is achieved, an assembly stage is necessary to connect them to the anatomical entities and build a fully functional model that can be automatically registered to medical data or simulated.

In this assembly stage, each mechanical model describing an anatomical entity (class) is defined as an instance of the class, linked to a set of attributes or properties as illustrated in Fig.4.7a. Each attribute describes an aspect of the instance (*e.g the name, the type, the familly, etc*) or a parameter of a mechanical model (*e.g the mass, the mass center, the mesh representing the visual model, the maximum isometric force etc*). Numerous attributes remain specific to some types of model to ensure some consistency during the assembly stage. Templates of mechanical models containing a set of specific attributes are stored in the system to ease the addition and the handling of data, additional templates and attributes can be added to the framework. The attributes within the system are all the parameters of the models described in Sec.4.3.2.

To achieve the assembly stage, we introduce in MyCF the concept of *acquisitions*. An *acquisition* represents an anatomical entity or an anatomical function as a set of instances and



(a) *Instances of the **Right_femur**. The way that attributes related to a type of representation is shown for the case of the rigid modeling of the right femur*

(b) *Acquisition which describes a rigid body system representing the right knee joint*

FIGURE 4.7 – *Examples of instance and acquisition.*

a set of attributes (cf Fig.4.7b). Each acquisition can be seen as a simulation scene, which will contains the functional data related to an anatomical system. We introduce this concept to handle data according to the description provided from the ontology. Acquisitions are exported and parsed to be used in simulation softwares. An acquisition can be composed of instances extracted from other acquisitions.

Concerning the mechanical model, the connections are composed of pairs of matching points (for attachments) or reference frames (for simple joint models). These features can be located with respect to others in hierarchies, as allowed by the *mapping* components in the SOFA framework [FDD*12] discussed in Sec.4.4. In our prototype, these include the location of joint frames and insertion points with respect to bones, and non-rigid attachment of muscle geometry to the bones. In the future, more general interpolation functions will be implemented.

For instance, the way muscle fibers will be embedded into a volumetric mesh, or the distribution of actuation intensities in a group of co-actuated muscles.

At the end of this assembly stage, a functional model of the selected anatomical functions or anatomical entities is obtained as output. This functional model includes all the instances describing the mechanical models of the relevant entities and functions. This output can be registered to patient-specific data, or can be directly simulated.

4.3.4 Registration to subject specific data

We rely on existing registration techniques to fit our generic models to subject-specific data. Registration is a central problem in the computer graphics and image processing communities [ZF03]. It proceeds by iterating two basic steps : first, the distance between the current position of the reference shape and the target is estimated ; secondly, this distance is minimized by transforming the reference model. The main difficulties are to find an adequate distance measure that is as-convex-as-possible and a good parameterization of the transformation through the introduction of prior information. This introduced constraints in the deformation mechanism describe the allowed geometric excursion from the reference model. In the context of the musculoskeletal system, articulated bones can undergo large discontinuous displacements, soft tissues are highly deformable and due to shape variability in the population, geometric differences between the reference model and the subject can be relatively large. However, organ topologies, their relative positions and muscle attachment sites are stable in healthy subjects. Moreover, models must remain collision-free. We leverage this knowledge in our reconstruction framework.

The specific registration technique we use here is described in [GRP10]. This technique is well suited to the musculoskeletal system because it decomposes the unknown displacement field into :

(a) A global articulated motion of the bones.

(b) Global bone deformations from a statistical shape model generated from multiple subjects.

(c) A locally smooth as-rigid-as-possible deformation field.

The additional constraints are the range of motion of joints, the maximum amplitude along each statistical mode, and the stiffness of the as-rigid-as-possible deformation technique (see [GRP10] for details). Given some reference manual segmentations, a prior appearance model, containing image intensities in the neighborhood of organ surfaces, is learned. Maximizing the similarity between this appearance model and the current observed appearance in the subject images, the reference model automatically converges to the target contours, by evolving in the constrained deformation space. In our study, we use T1-weighted MR images of the complete lower limbs, acquired with an isotropic spatial resolution of $1.2mm$. Slice-by-slice manual delineation was done once on one dataset (about 15 hours of work) to reconstruct a reference model for automatic registration (about 2 minutes of computation). This approach is hierarchical : the skeleton is first registered, then muscles. Based on comparisons with manual segmentation, the final accuracy of our model is about $1mm$. As shown in [GRP10], when target geometric models are available through manual segmentation or other sources, a similar registration procedure can be used to perform surface-to-surface registration. Structures that are not visible or outside the field of view do not undergo image-based constraints ; they are automatically extrapolated based on the regularizing internal deformation constraints previously described. Our method has been implemented in the SOFA framework, allowing an easy and modular configuration of the registration process.

4.4 SIMULATION FRAMEWORK

We rely on a number of simulation techniques tied together using the SOFA framework. Our simulation pipeline is summarized in Figure 4.8. It combines a skeletal dynamics simulation module with Hill-type muscle actuators, and procedural methods for soft tissue deformation. Such approaches for modeling musculoskeletal system are common in the biomechanical area [Zaj89, DLH*90, Sch93, The03b, SLPF11]. The independent degrees of freedom of our simulation model are the positions and orientations of the bones, while the input parameters are the function actuation levels which can be interactively changed by the user.

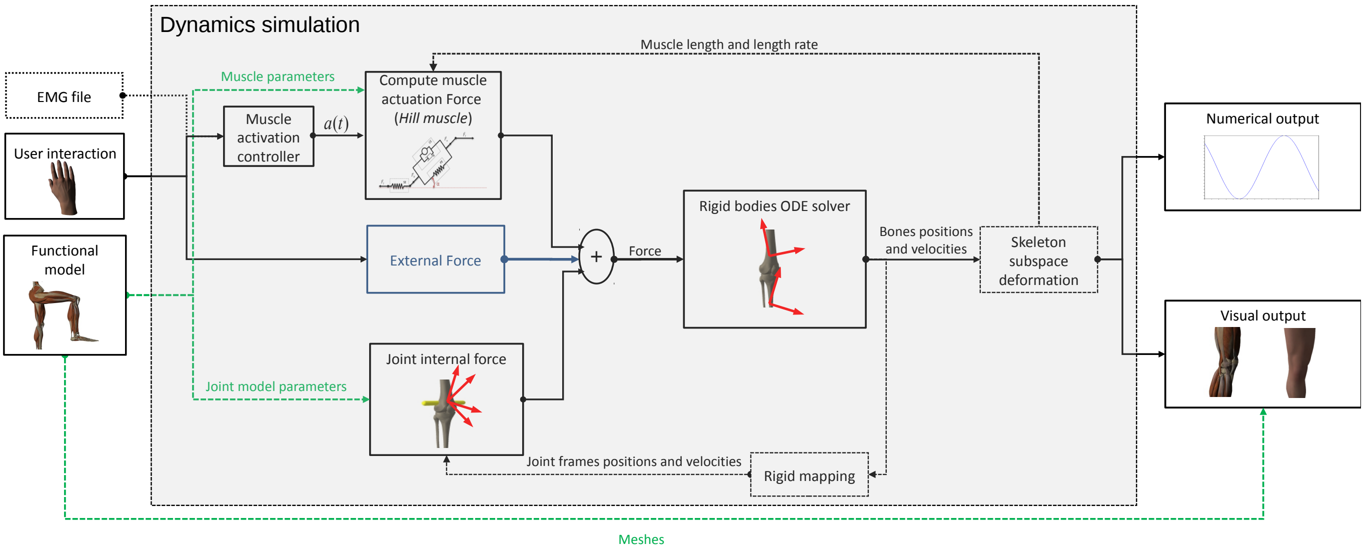


FIGURE 4.8 – *Overview of our simulation pipeline*

4.4.1 Skeletal model :

Each bone is represented as a rigid body equipped with a center of mass and an inertia matrix. To model joints, frames are rigidly attached to bones using a SOFA *RigidMapping* component. In most biomechanical frameworks [DLH*90], bone positions are expressed using reduced coordinates (the classical serial approach of robotics). In contrast with these simplified models, we use world coordinates and stiff penalty forces to enforce soft articular constraints. This allows a simpler handling of kinematic loops (such as the femur/patella/tibia complex), and it is more realistic, since slight translations are present in real biological joints. Note, however, that a traditional robot-like link hierarchy could be applied to the same functional model. Reference joint angle and shift limits, as well as joint stiffnesses are provided by the reference model, although they are easily customizable by the user. Stable bone dynamics with large time steps is achieved using an implicit Euler integration scheme.

4.4.2 Passive soft tissue model :

Based on bone positions, we compute the deformation of soft tissue (e.g. skin, muscles) using a classical *Skeleton subspace deformation* approach [KvO07]. The transformation of a given soft tissue vertex is computed as a combination of bone transformations. In Figure 4.9, point P is associated with each bone in the reference configuration (left). During the simulation, bones are displaced as illustrated in the right of Figure 4.9. If P was moving rigidly with respect to each bone, different target positions would be obtained. In linear blend skinning, the final position of P is obtained by computing the weighted sum of these target positions. The influence of each bone is modulated by its weight function. Weights are computed as a function of the closest distance d_i to each bone i. Here we use the function $w_i(P) = tan(-d_i^2+\pi/2+\epsilon)$, and subsequently normalize the weights to sum up to 1, but other functions are possible. Other techniques such as dual quaternion blending [KvO07] have been proposed and are applicable in our context. The vertex blending allows to preserve the approximate distances between the muscles, and therefore the fat is implicitly taken into account. The external forces acting on the soft tissue are mapped to the bones using the principle of virtual work (transpose of the Jacobian of the kinematic relationship).

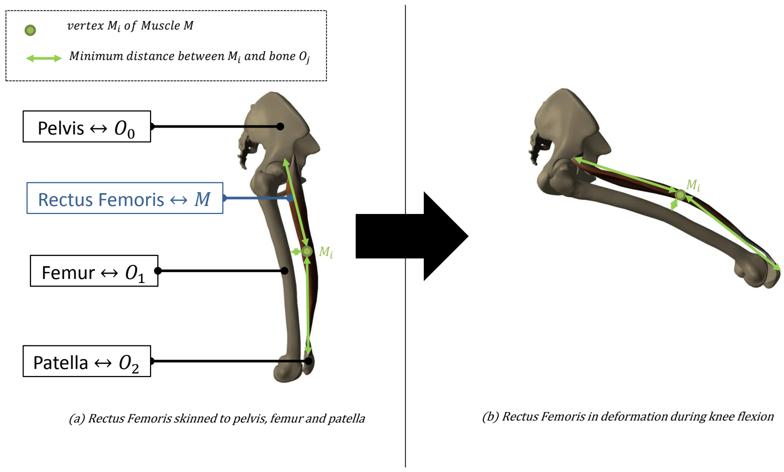


FIGURE 4.9 – *Skinning of a point P of the rectus femoris muscle by the patella, pelvis and femur bones. On the left, the original configuration used to compute the closest distance and the weights. On the right, the deformed configuration where P is computed as a weighted sum of rigidly transformed positions.*

4.4.3 Muscle actuation model :

In our prototypes, we use the one dimensional Hill-type model proposed by D.G Thelen [The03b], which is a variant of the popular Zajac model [Zaj89] accounting for age (cf Fig.4.10.a, 4.10.b). Tendon and passive muscle elasticity is modeled using exponential stress-strain laws followed by linear functions as respectively shown in Fig.4.10.e and in the red curve in Fig.4.10.d. Active muscle forces are modeled using a stress-strain Gaussian function (cf blue curve in Fig.4.10.d), based on the actuation level. Dependence on contraction velocity is also taken into account (cf Fig.4.10.c). The macroscopic model is composed of the serial assembly of the tendon elastic component and the muscle component, which is composed of the passive and active units in parallel. In summary, the macroscopic muscle model can be considered as a generalized spring that generates forces on bones at insertion points, based on its internal length, velocity and actuation level. The intrinsic muscle parameters, controlling the shape of strain-stress curves, have been adapted from the OpenSim framework [SLDT07]. Specifically, they are scaled to fit the subject-specific morphology at the registration stage. In our method muscle paths are automatically deformed using skinning methods as for their external shape. Via points [DLH*90, BAG99] are considered in the model to ensure that the right muscles length is computed during large movement such as the flexion of knee joint. For more details on the implementation of the muscle model, we refer the reader to the annexe A.1, and the following documents [The03b, mus].

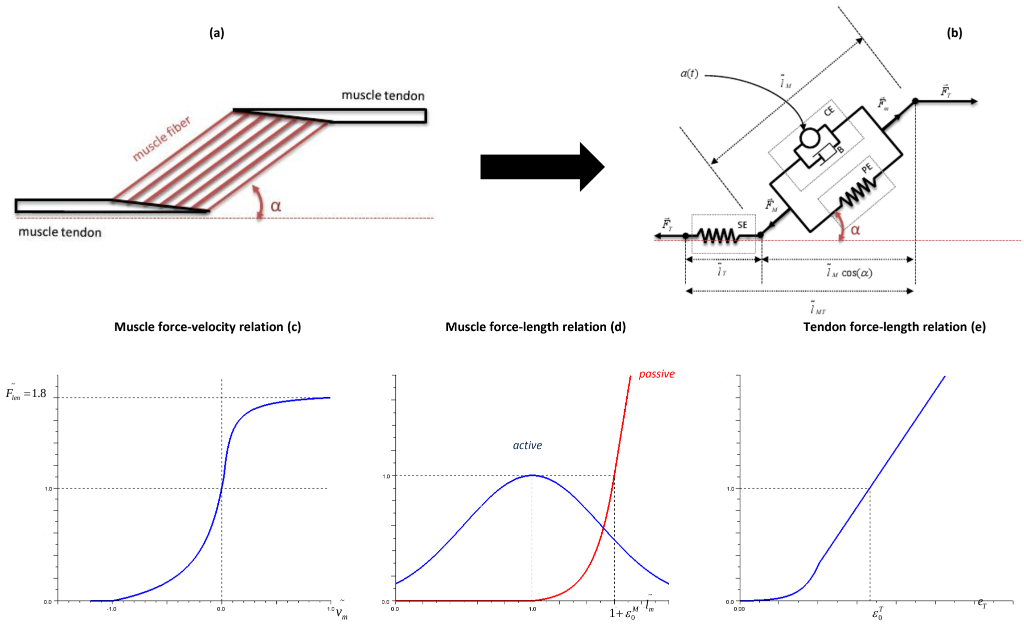


FIGURE 4.10 – *(a) Muscle composed by the tendon and the oriented muscle fibers, (b) Representation of muscle as a complex spring in series, (c) Relation force-velocity, (d) Force-length relation of the contractile component (CE) and the passive component (PE), (e) Force-length relation for the tendon (SE).*

4.5 RESULTS

We present some examples of results obtained using the described framework. As explained, we start by selecting the functions we want to model and to simulate. We choose *the knee flexion and extension, the hip adduction and abduction, the hip flexion and extension, and the hip internal/external rotation*.

In response to a query, the system based on the ontology returns a description of the functions in the form of a set of anatomical entities and sub-functions participating to the requested functions (cf Fig.4.11.a, 4.4). In the case of *knee flexion*, we obtain all muscles contractions directly involved as well as the joints, bones, muscles and ligaments of the body part.

This description is used in the assembly stage to generate an acquisition (graph shown in Fig.4.11.b) by looking through the ontology elements, and by adding to the acquisition the attributes, the mechanical and geometrical data to which each element is linked. This graph (tree) organizes entities according to their original type (or family) in the ontology : muscles, ligament, bone, joint, function (e.g contraction), etc.

Based on this, the output is a lower limb composed by 8 rigid bones linked by 7 joints, 44 muscles actuators, 108 muscles and 23 bone geometries as shown on Fig.4.11.c.

Given this data as input of the simulation pipeline presented in Section 4.8, we obtain a model allowing the realistic simulation of the targeted functions for the reference model,

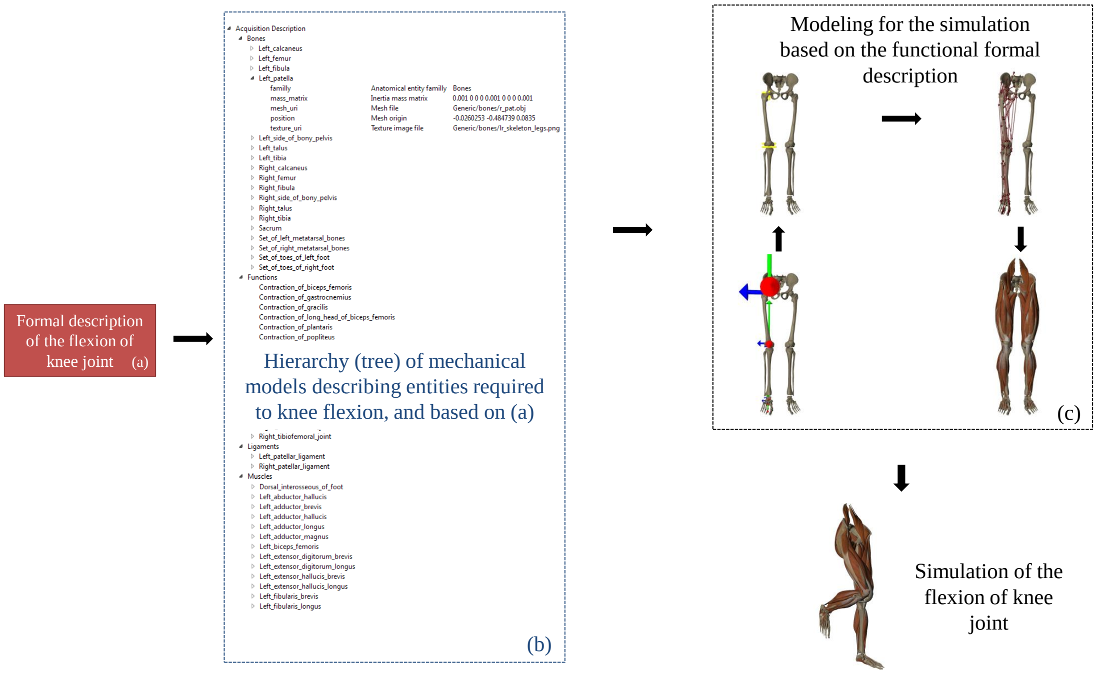


FIGURE 4.11 – *The ontology use for the modeling and the simulation of the reference model : (a) Formal description of the function resulting from the queries on ontology. (b) Generation of the tree containing the instances of functions and anatomical entities involved in the input function. (c) Models based on (b) used for the simulation.*

as shown in Fig.4.12. The muscles can be controlled using sliders to change the levels of activation, or using an EMG file containing the time varying activation level for each muscle.

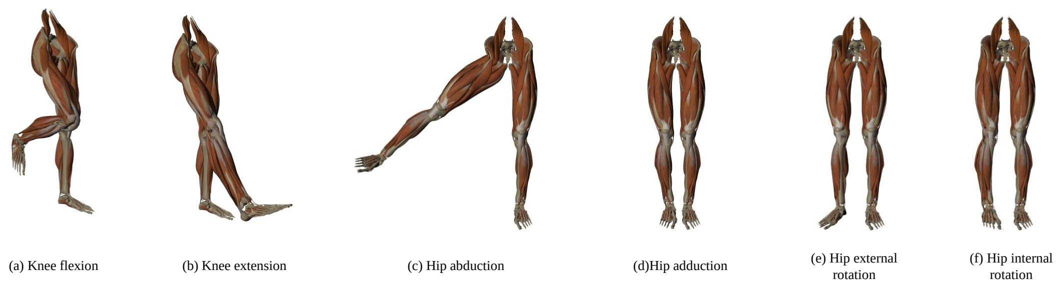


FIGURE 4.12 – *Simulation of the* ***reference*** *model. (a), (b), (c), (d), (e), (f) : hip and knee functions.*

In order to simulate the same functions for a specific subject, we register the generic model (cf Fig.4.14.a) to the subject-specific MRI data as shown in fig4.13.b. The output of this registration is the set of data ready to be simulated as shown in Fig.4.14.c.

This registered model is used in the same simulation framework. The results are shown in Fig.4.14, and a live capture of the modeling an simulation pipeline is available in the accompanying video (`https://www.youtube.com/watch?v=xxMxOJaBjl4`).

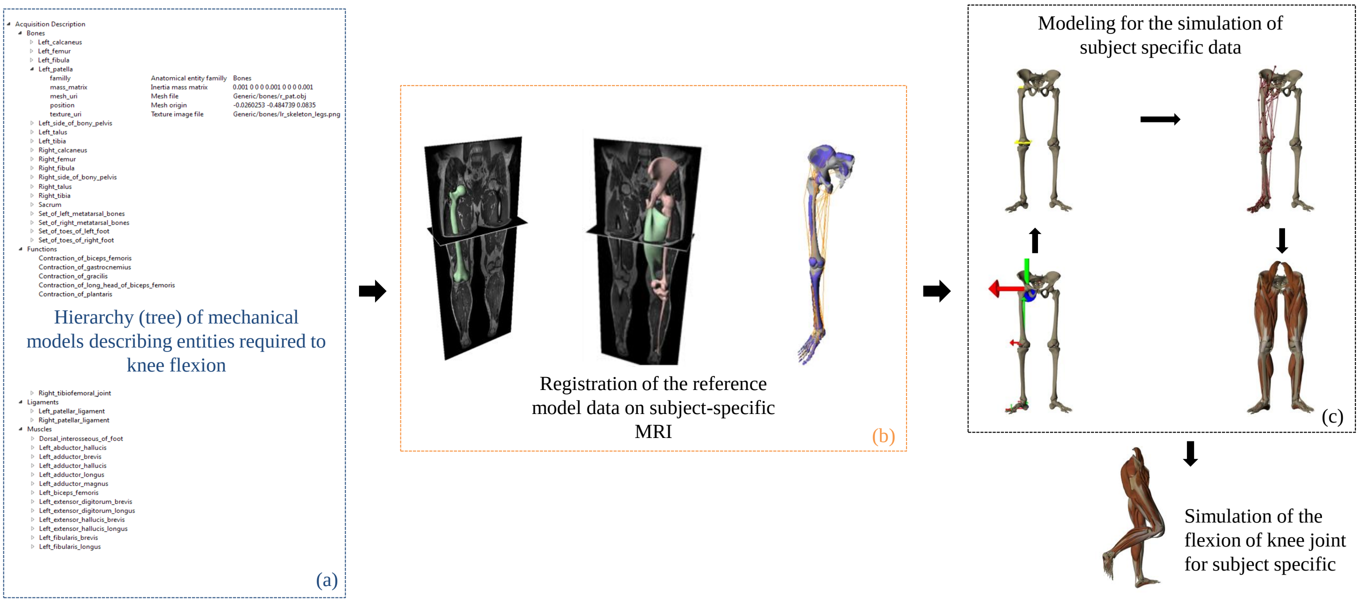


FIGURE 4.13 – *The ontology use for the registration of data and the simulation of the subject-specific model : (a) Tree containing the instances of function and anatomical entities involved in the input function. (b) Registration of data described in (a). (c) Subject-specific model for the simulation.*

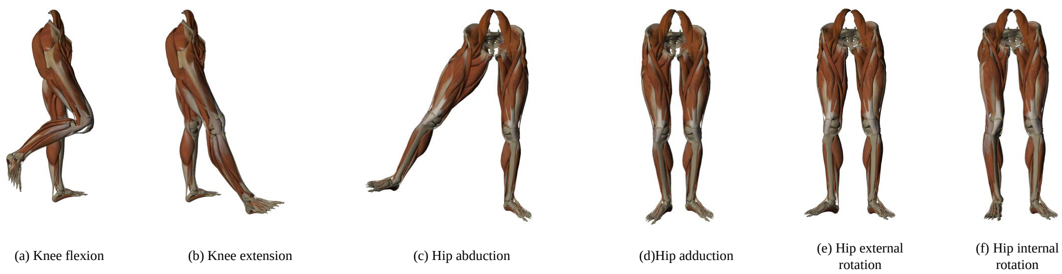


FIGURE 4.14 – *Simulation of the* ***subject-specific*** *model. (a), (b), (c), (d), (e), (f) : hip and knee functions.*

The simulations are performed with the SOFA Framework [FDD*12], and run interactively at 8 frames per second on a computer with a *core i7 720qm* at 1.60 GHz and an *NVidia quadro 1800M* graphics board with 1 Gbyte of memory. While not achieving the same level of physical accuracy as detailed Finite Element or Finite Volume simulations, the precision of this simulation is comparable with OpenSim [SLDT07], with subject-specific geometry and improved visual realism.

4.6 Discussion

We have presented a versatile modeling and simulation framework based on ontologies of human anatomy and reference models. By combining novel functional information with efficient registration techniques, our method can produce interactive, subject-specific musculoskeletal simulations. We will use this framework to generate subject-specific biomechanical models interacting with various devices as discussed in the introduction.

This novel approach of anatomical modeling raises new questions. The optimal selection of involved anatomical entities, given a user-specified musculoskeletal function, is an open question. For instance, the muscles directly involved in knee flexion (i.e., 1-ring neighbors in the ontology graph) are : *Biceps femoris long head, Biceps femoris short head, Gastrocnemius lateral head,Gastrocnemius medial head, Gracilis, Sartorius, Semimembranosus, Semitendinosus*, as illustrated in Figure 4.15. These entities are not sufficient to create a fully functional model to simulate this function, because some features such as the presence of stabilizing and antagonistic muscles are not automatically present in the output. Some additional knowledge and input is required, such as the concatenation of functions (e.g. muscle flexion *and* extension) in this example.

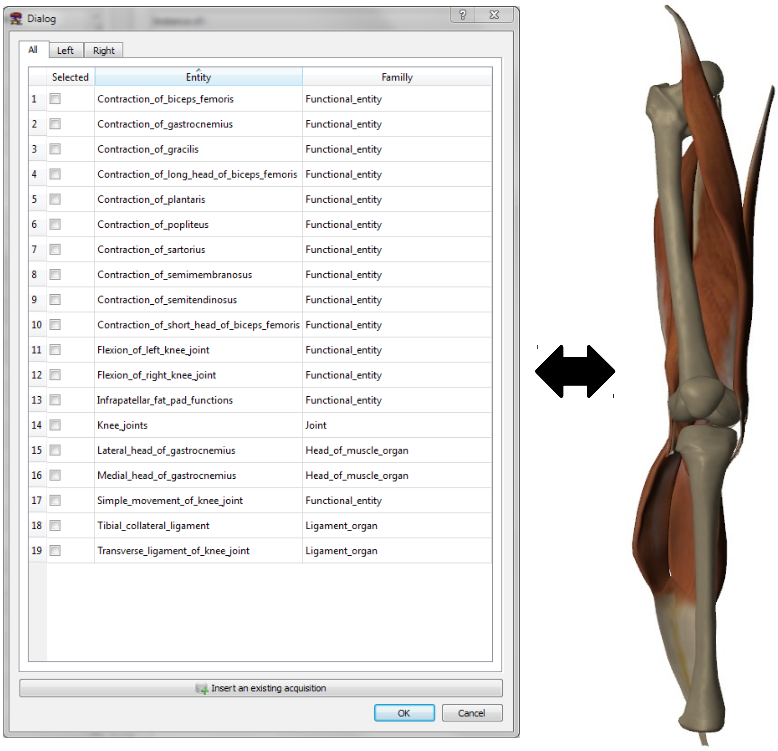


FIGURE 4.15 – *Partial model obtained with a one-level propagation in the ontology graph from the term "knee flexion"*

The automatic completion of models is ongoing work.

More accurate, three-dimensional mechanical models such as Finite Element meshes are necessary where more accuracy is needed. The associated challenge will be to finely tune each model type and resolution (thus its computational cost) depending on the function of interest, and on the purpose of the simulation.

In this work, we handled mechanical models and geometries as instances of the ontology classes, each mechanical model being represented by a set of attributes describing intrinsically the model. This choice of representation of data within our system restrains the possibilities on the queries we can make on the instances. To extend these possibilities, we plan to create an ontology of mechanical models. This new ontology will enable a better handling of all aspects of the modeling and simulation process. New features such as a reasoning tools for complex queries, the handling of constraint on mechanical models and geometries, the possibilities to check the compatibility between representations, and numerous others possibility will then be easier to manage. The main purpose of this novel ontology will be to consider entirely all aspects during the modeling of anatomical systems for simulation : *contact, interactions, constraints, boundary condition, behavior law, ... etc*. Coupled to our anatomical ontology, it will contribute to a more efficient construction of models.

We believe that by allowing the easy creation of anatomical models combining reference and specific data, our framework opens avenues for new applications of biomechanical simulation, and that it is a significant step toward an anatomical knowledge and modeling system to help users from different domains with various levels of expertise to share, customize and re-use models.

The original version of the work presented in this chapter has been published in [DGFP13]. Recently, an improved version of the pipeline has been presented in [BUD*14]. This novel version address numerous issues of the initial work. Most of the new features of this recent version of MyCF are based on the works introduced in [PUF*14]. In summary, these features are :

- A re-writing of the anatomical and functional ontology to make it more understandable, and easier to use.
- News relations for the anatomical taxonomy :
  - ***subClassOf*** : This relation replaces the relation ***is_a***.
  - ***rightSubClassOf*** : It is a specialized version of the relation ***subClassOf***, which is used to access to entities and functions of the right side of the body.
  - ***leftSubClassOf*** : It is a specialized version of the relation ***subClassOf***, which is used to access to entities and functions of the left side of the body.
- The relation ***involved_in*** role changes, in addition to this relation, we find new relations for the functional taxonomy :
  - ***involved_in*** : It now plays a role analogous to that of the ***part_of*** relation between anatomical entities, but for the functions.
  - ***participatesTo*** : This relation specifies that a given anatomical function contributes to the realization of a given (set of) function(s).
  - ***contributesTo*** : This relation specifies that a given anatomical entity contributes to the realization of a given (set of) function(s).
  - ***hasFunction*** : This relation relates an anatomical entity with the function(s) that it realizes (eg, knee *hasFunction* flexion_of_knee_joint).
- A set of reasoning tools for complex queries, to constrain models, and to infer information with rules.
- Use of the RDFS format for a more intuitive queries, instead of the classical SQL tools used in the previous version.
- New web tools to manipulate the anatomical knowledge, the ontology and the data as illustrated in Fig.4.16 and in the video shown in the following link : http://mycfbrowser.inrialpes.fr/mycf/
- Construction of models using frame based simulations [FGBP11] as also shown in the video in http://mycfbrowser.inrialpes.fr/mycf/

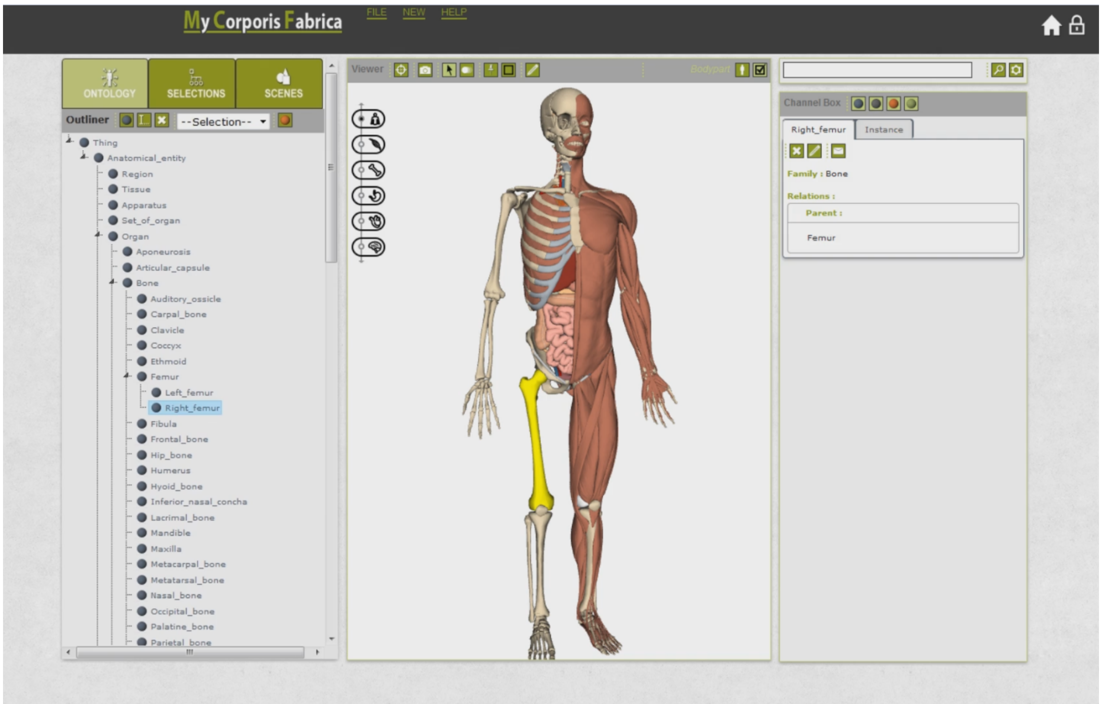


FIGURE 4.16 – *New version of MyCF with web tools.*

CHAPTER

5

A NOVEL APPROACH TO REGISTER INTERNAL ANATOMY : THE ANATOMY TRANSFER

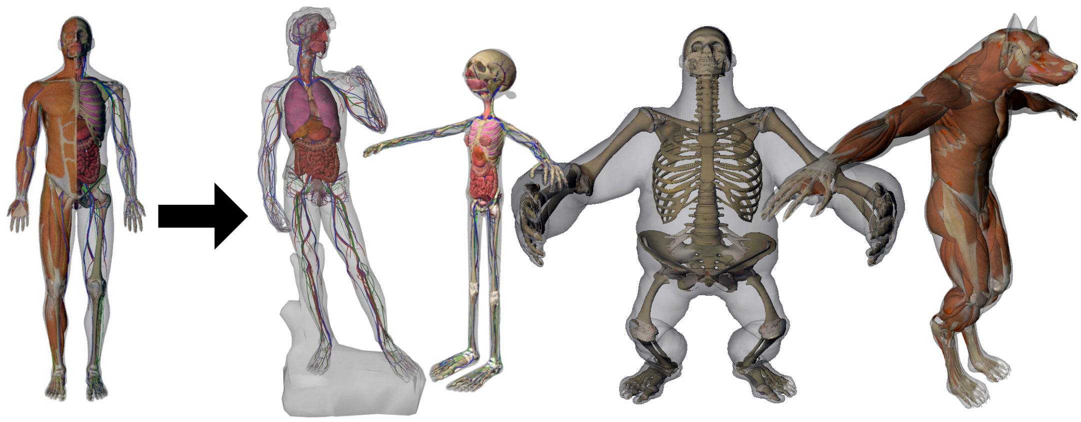


FIGURE 5.1

Teaser Figure : *A reference anatomy (left) is automatically transferred to arbitrary humanoid characters. This is achieved by combining interpolated skin correspondences with anatomical rules.*

5.1 INTRODUCTION

A high level of anatomical precision is necessary in many Computer Graphics applications, from visualizing the internal anatomy for education purposes, to anatomical simulation for feature films, ergonomics, medical, or biomechanical applications (e.g. optimizing muscle energy). Highly realistic animations showing muscles or tendons deforming the skin typically require precise anatomical models. Moreover, the control of the fat distribution is important for achieving the associated secondary dynamics effects. While a lot of research addresses the challenge of fast and accurate simulation, we focus on the upstream part of the pipeline, modeling anatomy.

The current tools available for artists to model anatomical deformations [MM13] as well as early academic work [WVG97, SPCM97] extensively rely on user input, essentially amounting to setting up the musculature from scratch. Recent years witnessed huge improvements in anatomically-based simulation, especially in terms of computational efficiency [PMS12]. However, the cost of setting up a 3D anatomical model for a given character remains. This task is very time consuming and tedious, as it requires modeling of the bones, organs, muscles, and connective and fat tissues. With real humans, it is possible to take advantage of 3D imaging, such as MRI [BAGD07]. However, this route is difficult or even impossible for fictional characters, ranging from Popeye to Avatar's Na'vi.

A naive idea to solve the problem would be to transfer the anatomy from a reference character to the target in a purely geometric way. It is obvious this route has a number of shortcomings : humanoids are made of bones, viscera, muscles, and fat tissues. Specific anatomical rules need to be preserved in order to generate a plausible anatomical structure : bones should remain straight and symmetric, and the distribution of fat, which may vary from one individual to another, should be taken into account while transferring muscles and viscera. CG characters can also contain non-anatomical or stylized components, such as hair, a shell, or even clothes. A specific problem is to prevent the internal anatomical structure to fill these areas, as we want our method to work even in these challenging cases.

We propose a semi-automatic method for creating the internal anatomy of any target character by transferring the internal anatomy of a highly-detailed anatomical model with minimal fat layers (Zygote body). Our method starts by registering the skins (outer boundaries) of the two models to define correspondence. An initial deformation between the two volumes is established using Laplacian deformation. The Laplacian is however uninformed about the anatomy and can, e.g., bend or otherwise unnaturally deform the bones. Therefore, we impose a number of anatomical constraints, such as requiring the bones to remain quasi-rigid. We also provide a tool for carving out the fat layers as well as the non-anatomical parts of the volume of the target model, before transferring the muscles and viscera.
Our specific contributions are :

- a novel registration method to transfer a source anatomy to characters with very different shapes while exploiting anatomical knowledge to get a plausible result ;
- the use of a texture, specifying non-uniform distribution of fat under the skin of a character, and a robust method to erode the internal volume accordingly ;
- a user-friendly tool for editing the fat distribution texture, if needed, on a per bone basis.

We exploit prior knowledge about human anatomy, e.g., we require that bone shapes and sizes remain as close as possible to human, by restricting the deformation modes and enforcing symmetry during registration.

In the context of this PhD work, the anatomy transfer greatly simplifies the simulation setups of each new character. This is mainly due to the simplicity of its input model, which is the most common representation existing for geometric data : *a surface mesh*, while registration methods such as the one used in the previous chapter (cf Sec.4.3.4) are based on MRI data (*i.e not easy to obtain, and often subject to various issues resulting from the capture conditions, and all the post-process required*). Surface meshes give the benefit to be simple to create, and in this sens that such an input accelerates the whole process of creation of patient-specific simulation, since all the specific mechanical models can be obtain by transferring them within the patient skin.

The journey towards realistic Computer Graphics humans starts with modeling. To our knowledge, this work is the first attempt to address the challenging goal of semi-automatic anatomy authoring. While many limitations and open questions remain, we hope that our method opens the door to inexpensive anatomy authoring tools and helps to promote and democratize applications leveraging anatomically-based simulation and visualization.

The remainder of this chapter is organized as follows. An overview of the anatomy transfer is described in the Sec.5.2. Each stage of our registration method is detailed from the Sec.5.3.1 to the Sec.5.3.6. The results are presented in Sec.5.4. This chapter ends with discussion and the presentation of future works in the Sec.5.5.

5.2 Overview

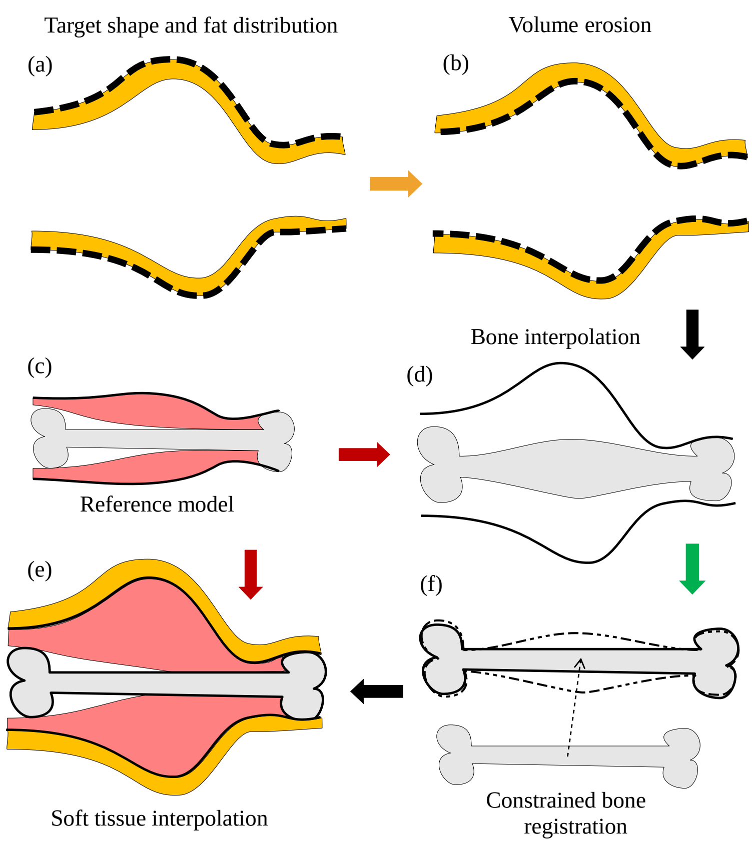


Figure 5.2 – *Anatomy transfer pipeline.*

The anatomy of a living body depends on numerous physiological constraints. The huge variability of anatomy is constrained by critical anatomical rules. We propose semi-automatic modeling of humanoid anatomy that uses some of these rules to constrain the resulting volumetric deformation, aiming to achieve as-anatomical-as-possible results. These rules are :

– **Rule 1, 2 (R1, R2) :** For the skeleton, our pipeline relies on the rule that bones must remain straight at the end of the anatomy transfer (R1), and symmetric across the sagittal plane (R2). The rule (R1) ensures that bones (*e.g, the femur, the humerus, the cubitus, etc*) are not bent after the skeleton registration. The rule (R2) is mainly applied to sets of bones such as the ribs, the spine, etc to ensure their symmetry. This second rule can not be taken into account for the limbs, since it is possible to find on one side of the body limbs longer (or shorter) than those of the other side. This is the case for real humans, and even more for cartoony characters.
– **Rule 3, 4 (R3, R4) :** The third and the fourth rule relate the fact that there is no relation between the quantity of fat tissue and the size of the bones [MD99]. For example, a fat character has the same skeleton as a lean one (R3), but the muscularity is proportional to keep up the body (R4) [MG08].
– **Rule 5 (R5) :** The fat layer is localized mainly between the skin and the muscles (R5) [MG08]. They can be interpreted as a stock of energy and therefore, the amount of fat tissue can be very variable.
– **Rule 3, 4 (R6, R7) :** During anatomy transfer, anatomical structures cannot disappear (R6), and the muscular and ligament insertion points are preserved (R7).

Our anatomy transfer pipeline implementing these rules is illustrated using a didactic anatomy piece in Fig. 5.2. The method can be seen as a partial registration process, where skin surfaces are first registered to define input correspondences, and based on that, the interior is estimated using interpolation and anatomical rules. More precisely, as input, the user provides the skin of a target character and may add a user-defined distribution of sub-skin fat (possibly including other non-anatomical structures) modeled using a thickness function in texture space (Fig. 5.2.a). As an alternative to user defined fat map, this information can be also extracted from real MRI data. Our method requires that the source (reference character) and target skin share the same (u, v) texture space. The first step, not shown in the figure, is thus to compute the registration of the source and the target skin (Sec.5.3.1).

Our source model (Fig. 5.2.c) is composed of bones, skin, muscles and viscera, and it includes almost no fat. We therefore erode the volume of the target (Fig. 5.2.b) according to the thickness of the fat layer, to warp our "lean" source anatomy to the sub-fat part of the target volume (Sec. 5.3.3), following rule (R5). The user can create the thickness data for stylized and cartoony characters using our new semi-automatic tool (Sec. 5.3.2).

The displacement of the skin from the source to the eroded target is then interpolated within the volume to transfer the internal anatomy (Sec. 5.3.4). This, along with a reasonable choice of fat thickness, enables us to follow rules (R3) and (R4). However, naively interpolating the skin deformation generally results in visible artifacts in the internal anatomy, especially in the skeleton, which may exhibit bent or inflated bones. We thus use this interpolation (Fig. 5.2.d) as an attractor for a constrained registration (Fig. 5.2.f), where the constraints express anatomical properties, such as the symmetry of the skeleton about the sagittal plane (Sec. 5.3.5). This allows us to incorporate rules (R1) and (R2). This constrained registration provides us with a plausible skeleton which fits the shape of the target character while following the anatomical rules.

Finally, we compute a new interpolating deformation field, using the internal skeleton as well as the eroded shape as boundary conditions (Fig. 5.2.e). This allows us to interpolate the remaining anatomical entities in between. This preserves all the anatomical structures, their relative locations, and satisfies rules (R6) and (R7).

5.3 Methods

5.3.1 Computation of correspondences : skin registration

The first step of our pipeline is to establish surface correspondences between the source and target skins as illustrated in Fig.5.3. Because skins of different subjects are not isometric, we focus on extrinsic correspondences for registration. For simplicity, we compute closest point correspondences such as in the popular Iterative Closest Point algorithm [BM92]. Based on correspondences established at each iteration, a smooth as-rigid-as-possible deformation field for the source skin is updated. As in [GRP10], we use the shape matching deformation method [MHTG05] which is both efficient (being based only on geometry) and controllable. Skin stiffness is progressively decreased during the registration to reduce sensitivity to local minima. Manual initialization is performed in the case of large differences between the pose of the source and target characters.

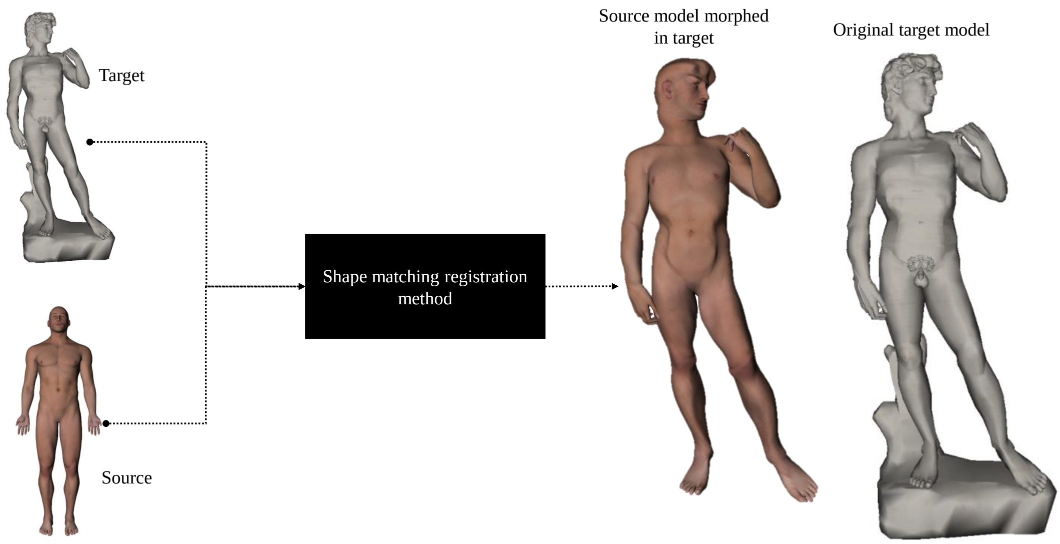


Figure 5.3 – *The source skin is registered into the target skin. The transformed mesh model is used instead of the initial target mesh during the transfer. Thus, with this new input, all the correspondence between the source and the target are known.*

5.3.2 Edition of the fat distribution texture

In order to generate fat distribution textures for arbitrary characters, we created a "fat editor" that provides both physically plausible initialization and full artistic control. Based on the observation that fat distribution is close to uniform around each bone, we adopted the idea of bounded biharmonic weights [JBPS11] to create smooth fat distribution maps. We set the bones as boundary constraints and minimize biharmonic (Laplacian) energy subject to these constraints. This way, we smoothly spread influence from the bones to the skin and obtain a

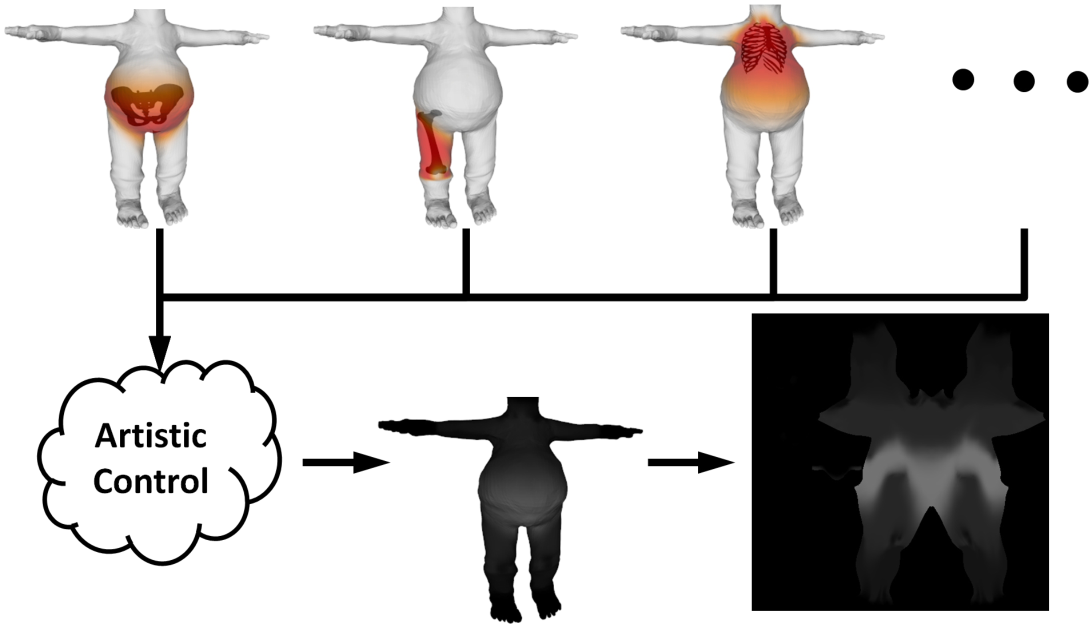


FIGURE 5.4 – *Fat distribution texture generation for the character. Top (initialization) : we use bounded biharmonic weights to compute skin weights corresponding to each bone. Bottom (fat editing) : artists can set fat parameters and generate fat distribution map. Brighter regions correspond to thicker fat layers.*

fat map by tuning only a few parameters. Choosing a reasonable thickness leaves space for a realistic amount of muscle tissue (rule R4).

Fig. 5.4 shows the pipeline of our fat editor. The editor first loads the target character model and its bones calculated using our bone registration. Similarly to [KS12], we compute bounded biharmonic weights using a regular voxel grid, obtained using the Binvox program [Bin13]. After pre-computing the weights, users can very quickly tune the amount of fat distribution around each bone. For example, we can assign 0.4 to the pelvis and sacrum bones to model the fat around the character belly, but give 0 to the skull because there is no fat beneath his scalp. The editor computes linear combination of pre-computed bone weights with the control parameters set by the users to generate the final fat distribution map.

The fat editor does not have to use only the anatomical bones. If artists want to control the fat around a certain region with more details, they can add fictional bones inside that region and tune the new parameters introduced by the fictional bones. We did not do this in our examples because we were satisfied with the results using only anatomical bones.

5.3.3 Volume erosion

The internal volume of the target character is composed of the skeleton and the soft tissues modeled in the source anatomy, along with a significant volume of fat tissue, which is usually not explicitly represented in anatomical models, including ours [1], and therefore difficult to model. We thus consider only a sub-skin layer of the fat tissue, which separates the skin from the rest of the anatomy. This layer, which may have a significant thickness depending on the target

1. www.zygote.com

character, reduces the available volume for the skeleton and muscles. The layer of fat below the skin is not uniform around the body. It is well-known that men and women exhibit different distributions, and this distribution may also vary between individuals [GL18]. With realistic human models, we make the simplifying assumption that each gender can be associated with one scalable distribution.

The simplest way to model the distribution of fat is to add a channel to the texture of the skin to represent the local thickness of the fat layers. We compute this thickness using the MRI image of a real person. We tag the voxels corresponding to the skin and to the fat layer using segmentation technique, and export the distance between the two layer. The output is a texture image containing the fat thickness at each point of the mesh representing the skin (cf Fig.5.5.a). This image texture representing the non-uniform distribution of fat can be edited as discussed in Section 5.3.2, and then used to perform volume erosion (Fig. 5.2.b).

To remove the fat layer, relying on the local normal to associate each voxel of the skin to a thickness would not be reliable due to skin curvature and imperfections in the input data. Therefore, based on the thickness texture (cf Fig.5.5.a), our approach to remove this layer of fat consists in reducing non-uniformly the skin volume. We rely on discrete data, exploiting voxel neighborhoods. Thus, we first convert the skin surface into a volumetric image (cf *Fig.5.5.c where one slice of the image is shown*). We compute a forest of shortest paths from the skin voxels to the fat voxels, where each skin voxel is the root of a tree. Then for each skin voxel we set the local fat thickness to the maximum distance to the leaves of its tree. Based on the texture coordinates of the skin voxels and the associated thickness, we interpolate the value at each pixel of the thickness texture. For each vertex of the target mesh (cf Fig.5.5.a), we compute the local depth using the texture coordinates and we move the vertex by this distance following the forest of shortest paths. An example result is shown in Fig.5.5.d and in Fig.5.8.

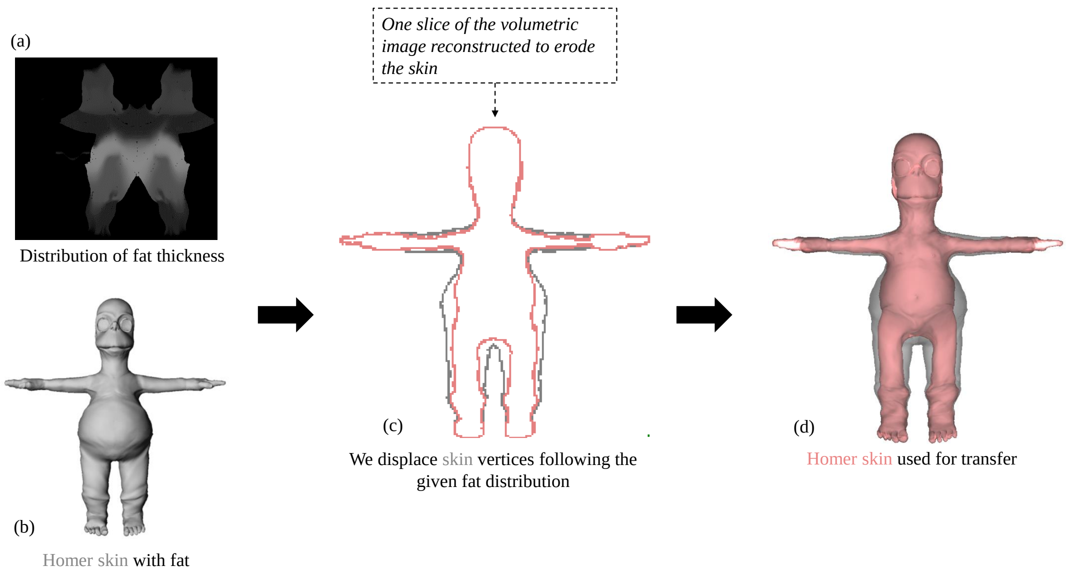


FIGURE 5.5 – *Removal of the fat thickness : (a) Fat thickness distribution represented as a texture image. (b) Target mesh skin. (c) Computation of the eroded volume (represented by the pink layer) based on the fat distribution. (d) Projection of the skin mesh vertex on the eroded volume (pink mesh)*

5.3.4 Interpolation

In our framework, volumetric interpolation is required at two stages of the method :

1. To initialize bones inside the eroded skin.
2. To transfer soft organs, using both the eroded skin and the bones as boundary conditions.

This section describes the interpolation method we use in both cases.

Given boundary conditions on the displacement field (i.e, *the deformation field between the source mesh and the target mesh resulting from the skin registration*), we solve for the displacements in the interior by minimizing the harmonic energy, also known as Laplace interpolation [PTVF02]. The principle is to compute as linear as possible interpolation by requiring zero value of the Laplacian of the displacement field at each unconstrained voxel. The boundary displacements $\bar{f}$ are incorporated as hard constraints :

$$\nabla^2 f(x) = 0 \text{ , } x \text{ inside} \quad (5.1)$$
$$f(x) = \bar{f} \text{ , } x \text{ on the boundary} \quad (5.2)$$

The discretization on our grid results in a large sparse system of linear equations, which we solve using the Conjugate Gradient solver from the Eigen library [GJ*10] . More sophisticated methods such as a multigrid solver with an efficient handling of irregular boundaries [ZSTB10] could be used to further accelerate the computation.

5.3.5 Bone registration : The affine optimization

Bones directly deformed using the method presented in Section 5.3.4 may become non-realistically stretched or bent, as illustrated in Fig. 5.2.d . The difference of shape between real or plausible characters and the reference anatomy is due to a different size as well as a different amount of soft tissue around them. Changes of character size mainly scales up or down the bones, while the changes of soft tissue do not modify the bones. We thus restrict each bone transformation to an affine transformation, using the initial interpolated bone as an attractor to a plausible location inside the body as shown in Fig.5.6. Moreover, the symmetry of the trunk is enforced by deforming it using transformations centered in the sagittal plane. We did not impose symmetry constraints for pairs of corresponding bones to allow the input of non-symmetric target characters, such as the David model in our examples. The constrained minimization is performed by attaching all the voxels of the reference bone to a common affine frame and attracting them to their interpolated position using linear springs. We have not noticed any visible artifacts due to the possible shearing modes introduced by the affine transformations. We use an implicit solver to ensure stability [BW98].

Organ intersections do not occur when the interpolation is foldover-free, which is the case in all our examples : during the semi-rigid bone registration, the offsets between the interpolated bones and the registered bones mostly occur in the off-axis directions, so we have not encountered any intersection.

If necessary, this issue could be addressed using standard collision handling routines.

Fig. 5.7 illustrates the benefits of bone registration compared to simple interpolation. Notice the bent bones in the legs, the oddly inflated bones in the arms of the interpolated skeleton, as well as the broken symmetry of the rib cage are fixed by the affine registration. Moreover, the shape of the skull is influenced by the hair during the interpolation. This deformation is also filtered out by the affine transformation.

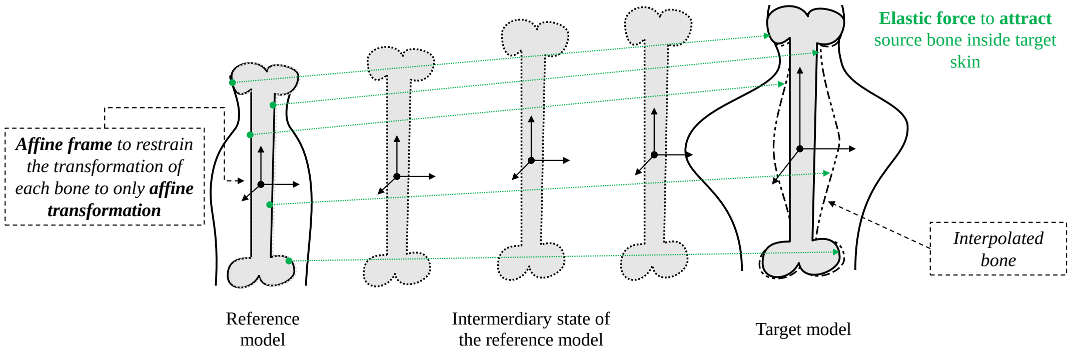


FIGURE 5.6 – *Reference bone B_{source} registered into the bone B_{target} by using the affine optimization. B_{target} results from the Laplacian interpolation of B_{source} into the target skin.*

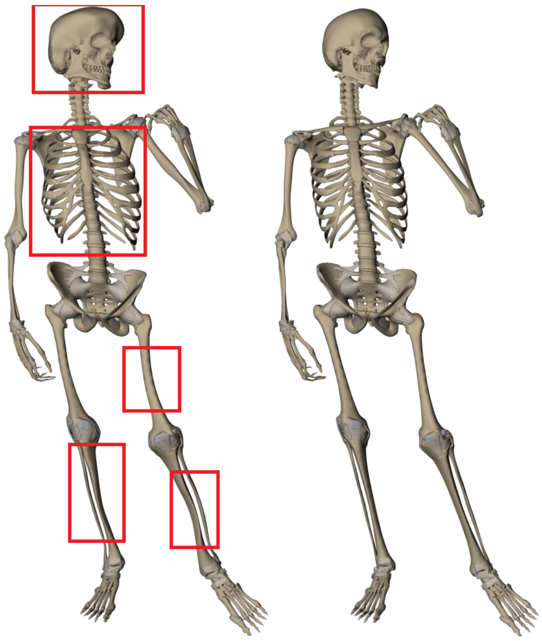


FIGURE 5.7 – *The benefits of bone registration. Left : after interpolation only. Right : after affine registration.*

5.3.6 Soft tissue registration

To transfer the soft tissues, we use the Laplacian interpolation with different boundary conditions than those in the bone registration. To keep the relative distances between the organs and the skeleton, the relative positions (*e.g, the lungs inside the ribs cage, brain within the skull, etc*), and to conserve the muscular and ligament insertions on bone as they are initially on the source character, we use the skeleton of both source and target characters, in addition to the eroded skin from the target plus the source skin to define the boundary conditions of the interpolation process. This allows to ensure that the rules R6 and R7 are respected. Because of this stage, the anatomy transfer can be seen as an hierarchical approach where it is indispensable to first register the skeleton before to obtain soft tissues consistent with the target morphology, and valid against anatomical descriptions.

5.4 RESULTS AND DISCUSSION

We have successfully applied our framework to both realistic and cartoon characters, as can be seen in Fig. 5.1. Cartoon characters were not intended as a primary motivation for anatomically-based modeling, but they are a challenging *stress test* for the system, showing how far from the input model we can go.

A nice feature of our method is that what we actually compute a deformation field, which can be used to transfer arbitrarily complex internal geometry. Once this computation is achieved, we are able to transfer a complete anatomy including bones, muscles, ligaments, viscera, blood vessels, nerves etc. very quickly. Our fat editor allows an artist to tailor a distribution for a specific target character, as shown in Fig. 5.8.

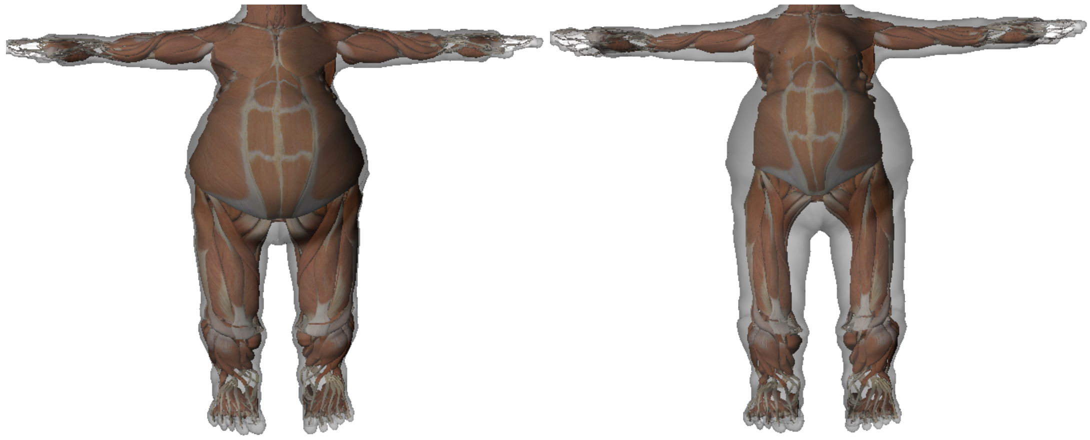


FIGURE 5.8 – *Transfer to a fat character. Left : without erosion. Right : a preliminary erosion accounts for the fat and results in a more plausible muscular system.*

Other examples of anatomy transfer are shown in Fig. 5.9.

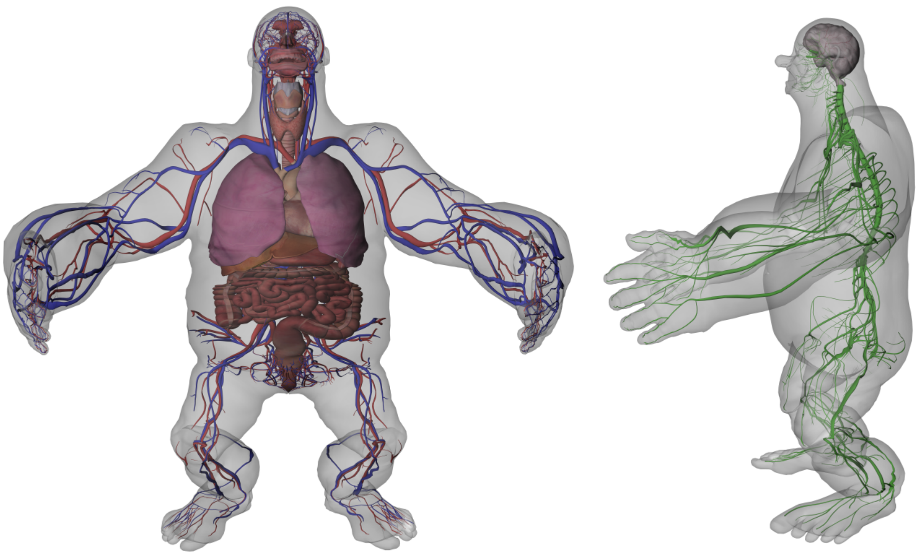


FIGURE 5.9 – *Brutus blood vessels and nerves.*

The reconstruction of Popeye in Fig. 5.10 exhibits a surprising chin, which could be mitigated using fat. Note, however, that his forearm bones are realistic despite the odd external shape.

FIGURE 5.10 – *Popeye and Olive.*

Fig. 5.10 also shows the reconstruction of the anatomy of Olive, a very thin character. Notice how close her muscles are to her skin while her skeleton remains thin, but well adapted to her morphology. To see how far we can push the concept of anatomy transfer, we transferred our reference model into a werewolf (half human and half animal). Fig. 5.11 demonstrates how the human anatomy fits accurately within the body of this monster despite the different morphology. To go even farther in this approach, we transfer the human anatomy within a real dog (a doberman) as illustrated in Fig.5.12. It is interesting to notice the similarity between the skeletons of the real doberman (*shown in the center of the Fig.5.12*) and the human skeleton transferred within the corporal envelop of this dog (*shown in the top of the Fig.5.12*). The bottom of the Fig. 5.12 validates our method by comparing the results we get with the musculoskeletal system of a real wolf (Canis lupus), shown on the left.

Fig. 5.13 shows the reconstruction of a real male based on his MRI image. The muscles are suprisingly well captured in the lower legs, the bottom cheeks and the trunk. Some muscles are not accurately reconstructed, due to different relative sizes in the real person and our reference model and to errors in skin registration. The latter are also responsible for inaccuracies in the fat layer.

The goal of this reconstruction attempt is not to compete with established segmentation methods, but to suggest that anatomy transfer may provide a useful initial estimate. Moreover, a lot of thin anatomical structures which cannot be seen in the volumetric image are present in our model. In future work, complementing our framework with sparser but more accurate segmentation methods may provide useful constraints to insert in our interpolation, to accurately infer the positions of the features invisible in the MRI.

Our methods provides significant improvements over a shape matching method like [GRP10], which is based on different premises. They assume noisy MRI input and therefore employ approximate volumetric shape matching, while our method assumes exact correspondence between the input and the target surfaces, i.e., the deformation field has to interpolate rather than

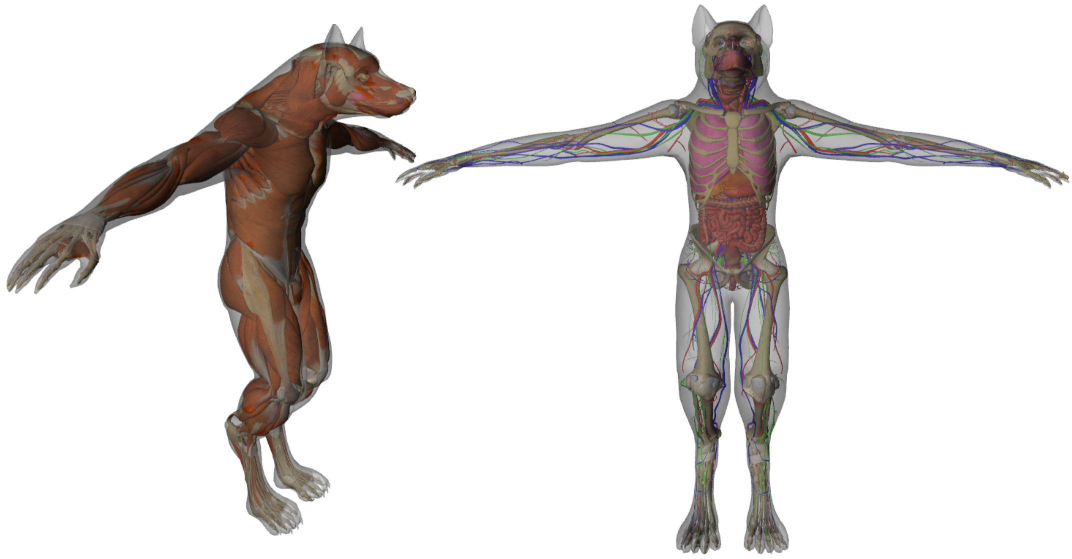


FIGURE 5.11 – *Werewolf muscularity, skeleton, and internal organs.*

approximate the boundary. To make [GRP10] as interpolant as possible, we need to make the shape matching stiffness and cluster size small enough, thereby slowing down the convergence and requiring a sufficiently dense mesh. A comparison is shown in Fig. 5.14. In the result of [GRP10] the internal tissue intersects the skin (lower arm, chest) and the matching is less accurate, as can be seen near the biceps, the shoulder, and the neck. Moreover, symmetry is visibly violated in the lower abdominal muscles and between the arms. Finally, the computation time was 30 minutes for [GRP10] due to the small size of the clusters, while our Laplacian solution converged in only 3 minutes.

In Fig. 5.15, 5.16, 5.17, we present examples of useful anatomy transfers. In Fig. 5.15 we show a transfer of an articulated system, animation and skinning [KvO07]. The joint orientations match the character posture, and the resulting motion is similar for all characters, as can be seen in the accompanying video. All the animations of cartoon characters in the accompanying video have their rigging models computed using the anatomy transfer.

Fig. 5.16 shows a transfer of muscle lines of action [The03b] for physical simulations. Using the same muscle activations, we are able to create similar movements, such as knee flexion or hip rotation, for both the reference model and the target (see the accompanying video). These action lines attached to the bones at both ends could be transferred directly. However, more realistic muscle paths include via points along muscle center lines or around warp surfaces on bone geometry, and this requires full volumetric transfer, because these points cannot be entirely defined with respect to skin and bone surfaces.

In Fig. 5.17 we present a transfer of deformation, mimicking bicep bulging in David's arm.

We use a standard laptop computer with an Intel CoreI7 processor at 3 GHz and 8GB of RAM. For each character, the total computation time ranges from a couple of seconds to less than five minutes with our current implementation. Fat erosion takes about one minute in a $64 \times 171 \times 31$ volumetric image, and the first Laplacian interpolation takes 15s in the same grid. The bone registration takes about 3 minutes. Most of the computation time is spent in the final Laplacian interpolation, which requires a finer resolution to get a smooth result. In a $309 \times 839 \times 142$ grid, it takes less than 5 minutes. Once the displacement field is computed,

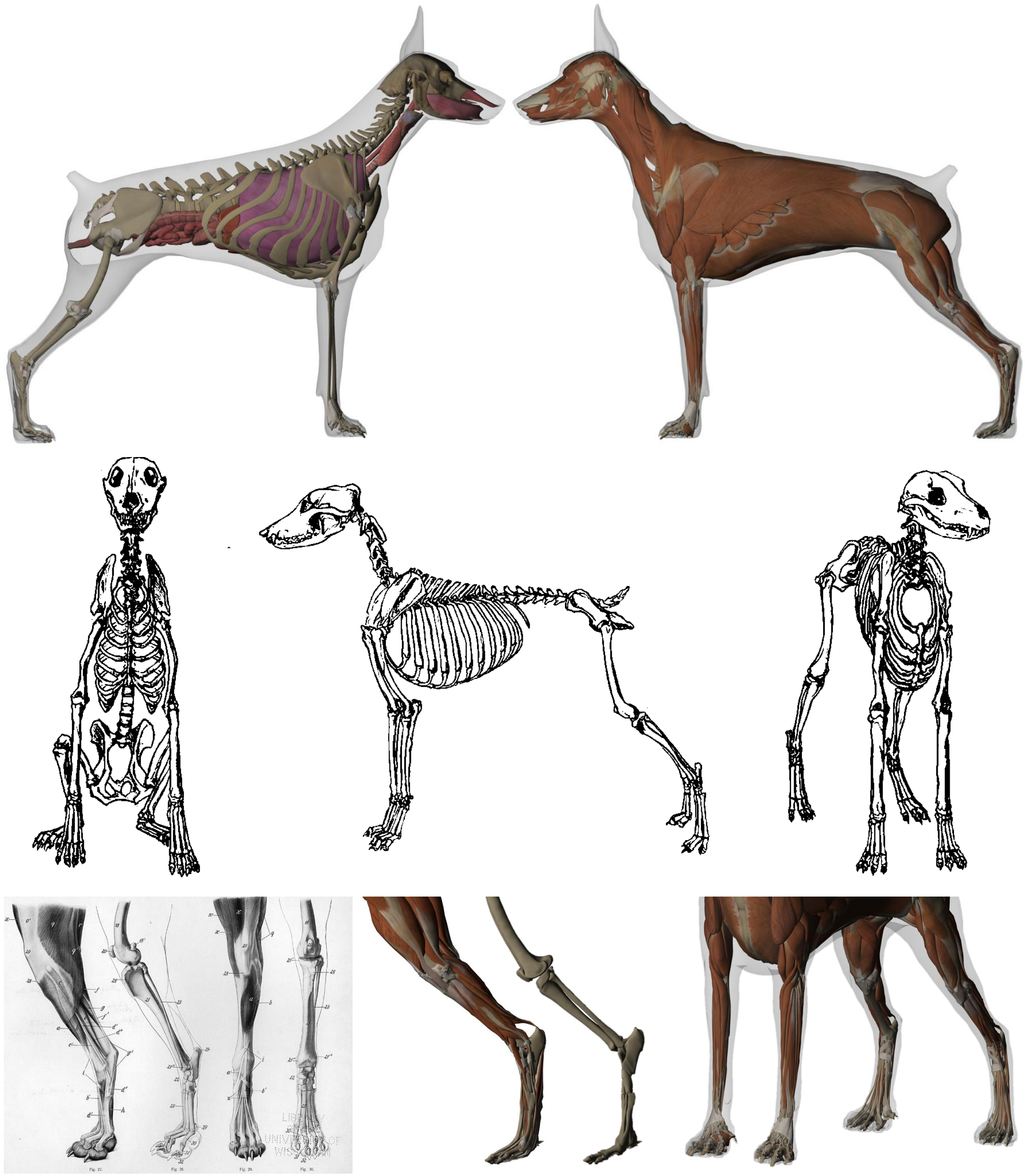


FIGURE 5.12 – *Top : Dog muscularity, skeleton, and internal organs. In between : doberman skeleton (images from [dog]). Bottom : Comparison between dog lower limb and real wolf lower limb*

transferring the 500MB of geometry of our model takes less than a minute. Our method has a number of limitations.

Firstly, automatically inferring non-standard distributions of fat from the morphology of the character would be an interesting extension. Standard human morphograms (i.e. classes of shapes : big belly, big chest, or completely skinny) are available in the literature, but so far we found no precise information on the corresponding fat distribution. Moreover, we do not model the fat tissue distributed anywhere else than directly below the skin.

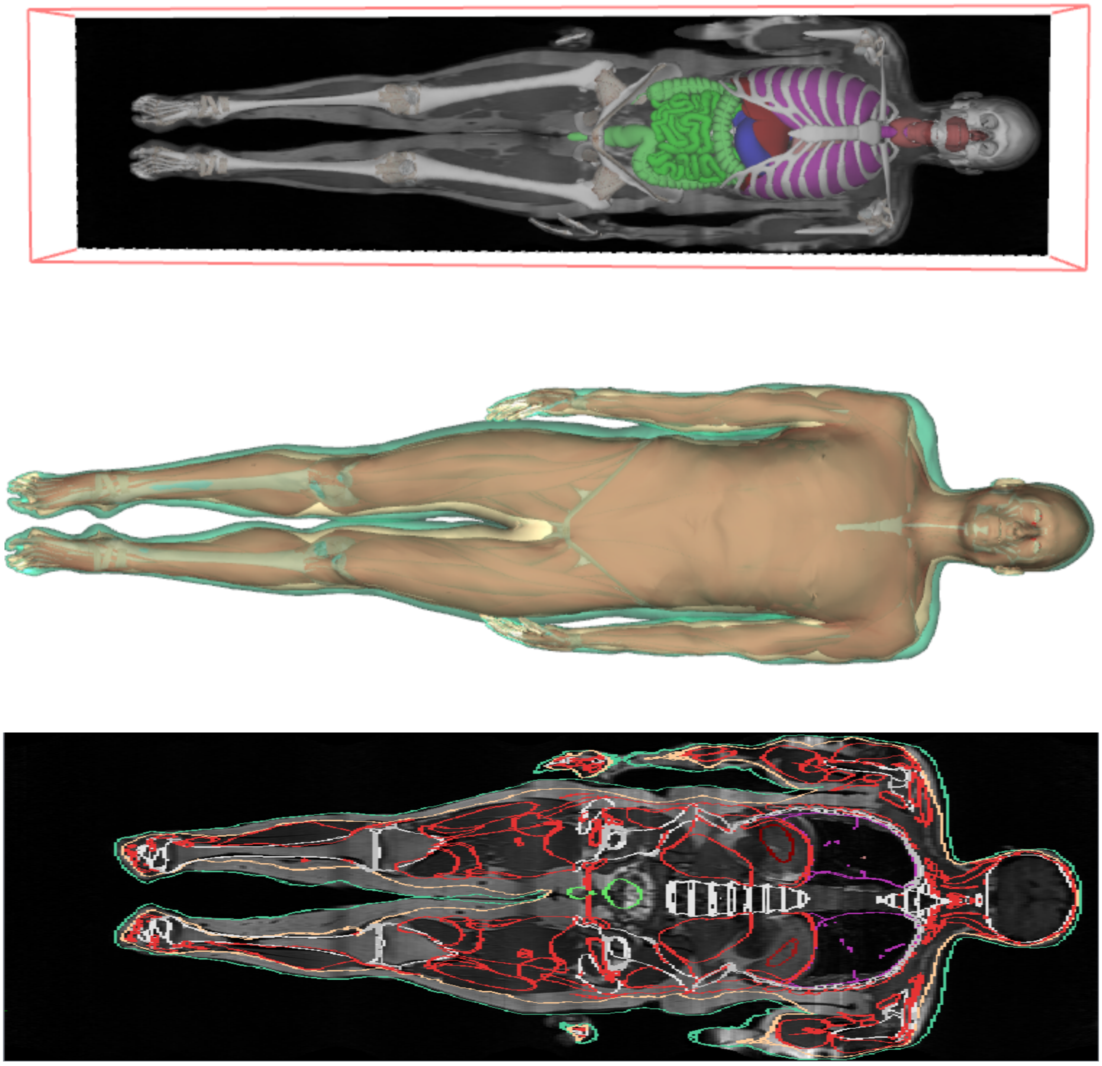


FIGURE 5.13 – *Transfer to an MRI image of a man laying on his back. Top : reconstruction of internal organs and skeleton within one slide of MRI Data. Center : reconstruction of muscular system. Bottom : comparison with the data. The green lines highlight our reconstructed surface, the beige lines correspond to the eroded volume, while the red line is muscle reconstruction, white and gray lines are bones and connectives tissues, the purple represents the lungs and the bright green represents the small intestine.*

Other practical limitations are related to the registration. The skin correspondence is inferred on a proximity basis. This sometimes creates wrong results when the source and target characters are in different poses.

Our volumetric interpolation method does not guarantee foldover-free displacement field : although we did not observe overlapping between internal structures in any of our examples, it could occur in theory.

The skin registration fails when the target character has a different morphology from the reference anatomy. For the example shown in Fig. 5.8, we had to create a five-fingered variant of the target character.

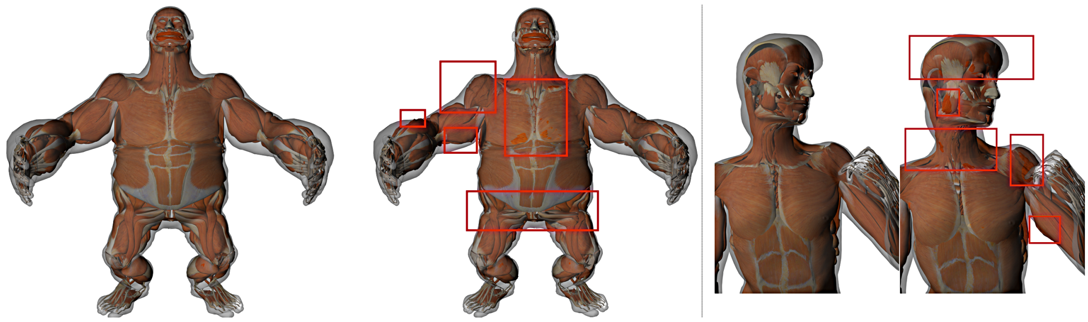


FIGURE 5.14 – *Left : our method. Right : the method of [Gilles et al. 2010] based on shape matching. Notice the artifacts of the latter method, e.g., the upper arm muscles intersecting the skin and asymmetry of the abdominal muscles.*

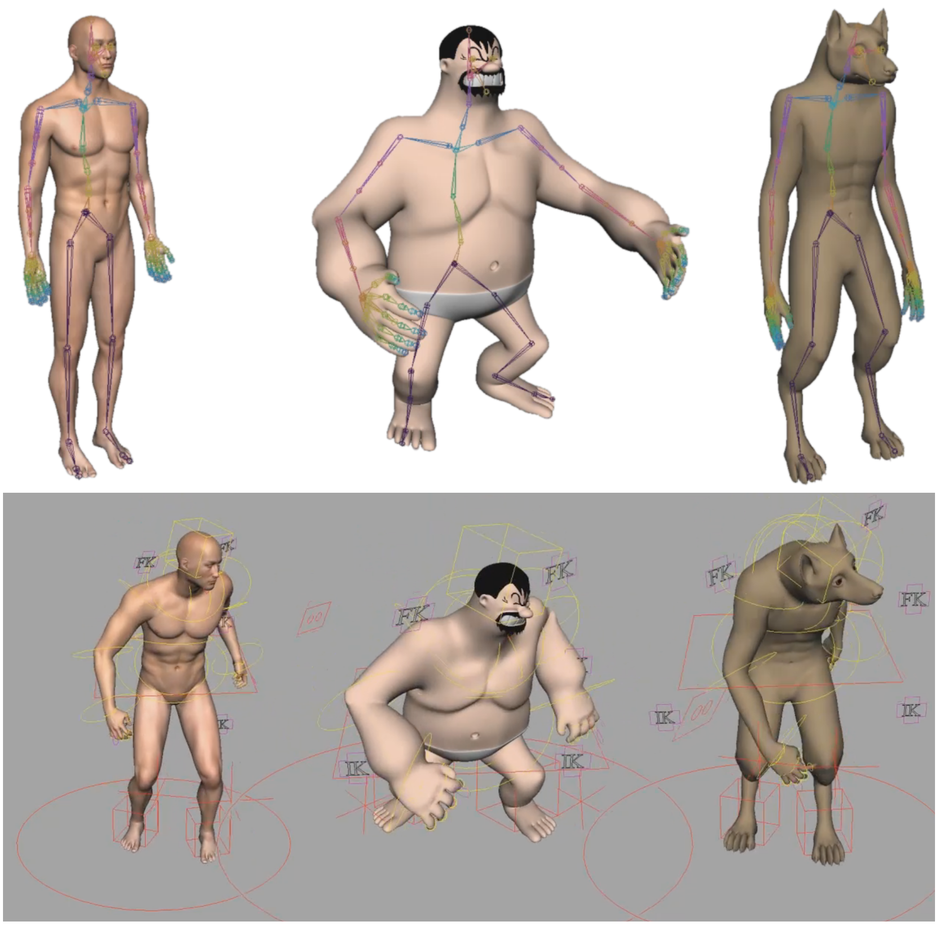


FIGURE 5.15 – *Top : Transfer of an articulated system. Bottom : Transfer of animation.*

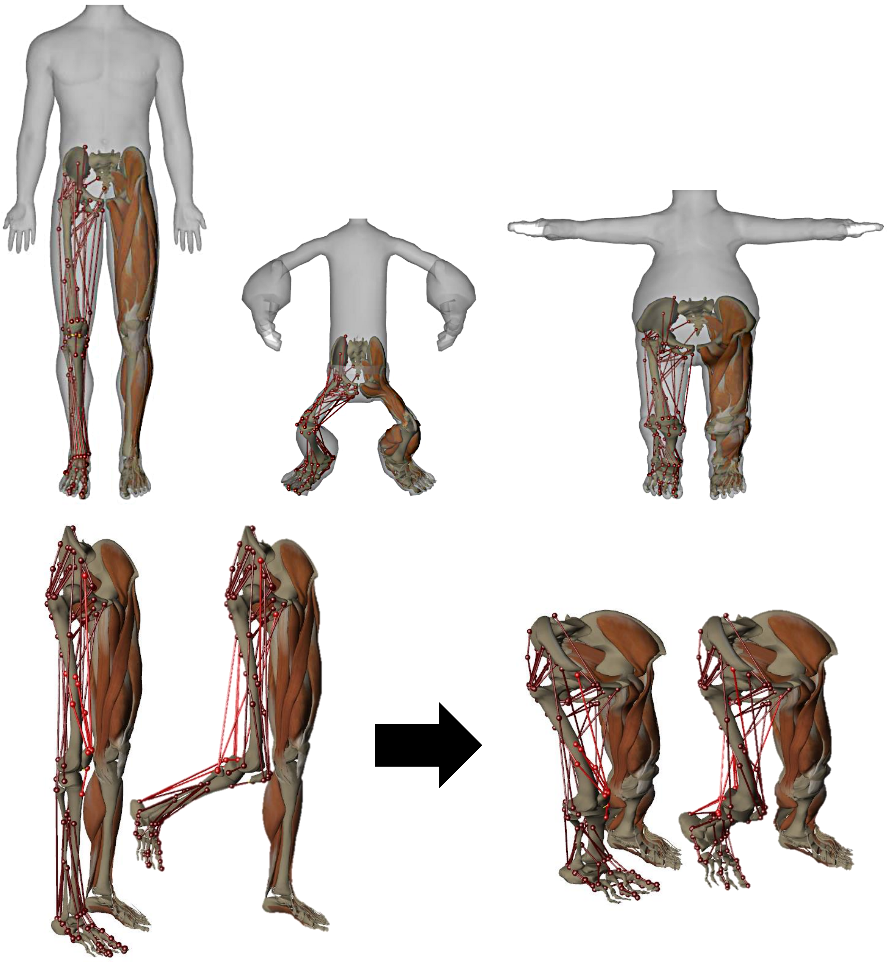


FIGURE 5.16 – *Top : Transfer of muscle lines of action. Bottom : knee movement using muscle control on both the source (zygote) and the target.*

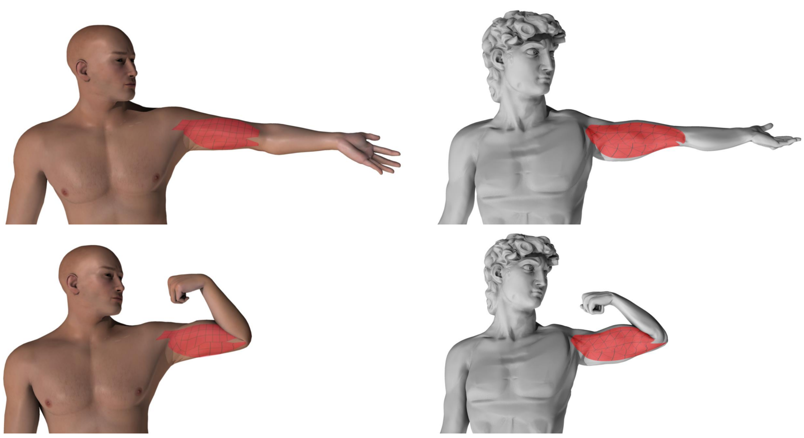


FIGURE 5.17 – *Top : Transfer of muscle and skin animation by using transferred muscles.*

5.5 CONCLUSION

To address the high costs associated with anatomy authoring, we have presented the first method for quickly creating a plausible anatomy for any target character. For realistic humanoid models, we transfer both the internal anatomical structures from a reference model, as well as the fat thickness information extracted and retargeted from MRI data. Our method is thus purely automatic. For cartoony characters, we offer a user friendly editing tool enabling the user to tune the fat tissues of the target character. Transferring the internal bones, viscera and muscles is then automatic. We show that the anatomy transfer allows to register mechanical models for subject-specific simulation as we initially target according to the objectives of this research work. We illustrate the robustness of the method by showing how far we can deviate to real human anatomy with the werewolf and dog examples. While it is obvious that these species do not have the same anatomy structures, the results and comparison to real anatomy shows the quality of the transfer results.

We have shown that direct Laplace interpolation, perhaps sufficient to generate simple effects such as muscle bulging, leads to objectionable artifacts when used to transfer the full anatomy. Our specific pipeline ensures that basic anatomical rules are preserved, and also shows the benefit to be simple to re-implement.

In future work, we would like to take advantage of more anatomical knowledge to constrain the interpolations. Furthermore, we would like to improve the computation of correspondences between both source and target, despite the fact that this step is upstream the anatomy transfer. We plan to replace our interpolation solver with a highly parallel GPU interpolation coupled to an efficient multi-grid solver to better handle the interpolation stages. We also plan to deal with transfer of partial anatomy, or in individuals in poor condition (*e.g individual with broken limbs, disable, etc)*. We believe that our method could also help the processing of body scans by computing a first guess to the segmentation process, and complementing the final result with thin structures, invisible in the volumetric image, as shown by our validation example (Fig. 5.13).

5.6 SUPPLEMENTARY MATERIAL :

Video `https://www.youtube.com/watch?v=ddp996DIZOk`

CHAPTER

6

CONCLUSION

In this thesis, we presented three new contributions to tackle the challenge of the fast and efficient construction of musculoskeletal systems for simulation purposes. We first proposed a novel hybrid model of the lumbar spine, for the interactive simulation of this complex organ ; we secondly developed an innovative system based on a new ontology of the human physiology to create models for the simulation and the visualization of organs functions ; lastly we introduced the anatomy transfer, a registration method to reconstruct internal anatomy of any character based on its skin mesh surface.

In the first contribution, we have introduced a novel hybrid model of the lumbar spine based on the combination of different type of degree of freedom, with the aim to simulate interactively this vital organ. We validated the model against literature works by showing that it reproduces the same amplitudes of movements with the right kinematic. This kinematic behavior has been validated through the computation of instantaneous center of rotation (ICR). We show that considering bone as totally rigid does not lead to any loss of accuracy, and any loss in term of quality in the kinematic of each vertebra during the movements on the three anatomical planes. The ICR trajectories support this claim. The results obtained with the model support that zygapohiseal joints although modeled as non-linear elastic elements, can be represented as a 6D elastic joints with non-linear elastic properties, in order to mimic the twist and stretch of the capsules. This representation contributes to the reproduction of the right movements of the whole lumbar spine and its functional units taken independently. This hybrid spine has been designed with the ultimate goal to test disc prostheses. With the simplicity to attach objects to rigid DOF, it offers the benefit to simplify the testing phase of these medical devices, since the disc geometry can be design without any consideration on the geometry of any interfaces or vertebra. The performances obtained with the model during the simulations also go along the lines of these aims. The simplicity of this hybrid model also enables an easier tuning of initial condition often necessary while testing device. More than proposing a new model, this work supports that hybrid approaches can fulfill quality criteria specific to the biomechanical and medical domain. Since we focus on the presentation and the validation of the model, the next stages will focus on using it for medical and biomechanical purposes.

To address issues related to the complexity of creating anatomical model, we developed a new system based on an ontology to model systems aiming to simulate and visualize organ functions. Through this work, we showed that prior knowledge can be used to handle functional dataset according to anatomical knowledge, for efficient and fast construction of complex mechanical systems with simple queries. To achieve these goals, we introduce the first ontology of the human organs functions, validated against ICF terminology, and consistent with the standard existing ontology of anatomy FMA. This system enables to build system which re-transcribes formal descriptions of anatomical structures and functions as they used to be formalized by anatomists and medical practitioners. These descriptions highlight the relevant anatomical structures and sub-functions of each requested functions or organ. This novel system offers the benefit to greatly simplify the modeling process, since the user only cares of which component or function he wishes to simulate or study, and the system handles all the required stages, and returns a functional model, with the possibility to visualize all its structures. More than assisting the modeling for simulation purposes, our modeling tool MyCF also represent a great support to the learning and teaching of anatomy. We add some registration tools to our system to manage the creation of patient specific simulations. These tools are based on MRI data as input. We illustrate the success of our approach via the simulation of the lower limb functions of specific patient by using his MRI data as input.

Although registration methods based on MRI allows to obtain patient-specific simulations, we realized that this approach does not represent the fastest way to model specific dataset, because of various issues mentioned in the previous chapter. Therefore, we developed the anatomy transfer to remedy these drawbacks. With this novel registration approach, we transfer all the internal anatomy from one reference character into any other character simply represented by the surface mesh of its corporal envelop. To overcome the lack of information as those we can have with MRI or scan data, we based our method on the used of several anatomical rules to keep some consistence during the registration process. The method works semi-automatically, and the only manual stage consists in defining the fat distribution for fictional characters, while this can be computed from MRI data for real individual. We bring with the method a new tools allowing to assist the creation of these fat distributions. We present some results to demonstrate the potential of the approach, and we compared it to existing methods to show the benefits of such technics. The anatomy transfer is simple to implement, fast to execute and robust. We show through some examples how far we can deviate to real human anatomy, and how close we can be to real human data only by relying on rules to respect during the process. The anatomy transfer significantly contributes to the fast and efficient creation of patient-specific simulations.

In terms of applications, the two last contributions are actually used and combined to perform what they was originally designed to perform. The recent publication of *Bauer & al* [BUD*14] represents a good illustration of this claim. In this work, we can see the new version of MyCF with its new web interface, and its tools to manage dataset from several characters for whom the anatomy has been reconstructed using the anatomy transfer. For these characters, it is possible to construct complex models thanks to the new querying and reasoning systems. As it was already the case in the work presented in this thesis, users can visualize the anatomy at educational purposes, or export them for simulation purposes. These two works are part of the recent project : *The Living Book of Anatomy* (https://persyval-lab.org/fr/sites/lba) which aims to propose tools to assist the learning process. They will be also used in the future startup *Anatoscope* (http://www.anatoscope.com/) to create patient-specific models.

Concerning the limitations of our approach, we focus on developing tools to fast and efficiently construct subject-specific simulations, but we only handle the subject-specific geometries and few mechanical information such as *the mass, the joint angles etc*. Concerning the functional parameters of mechanical model such as *stiffness, muscle isometric force, pennation angle, young modulus, etc*, numerous issues remain, and are aware that deducing these parameters from geometry is probably not the right solution. Although some geometry problems have been address, finding these mechanical parameters for each new character models represent an open problem for the researchers community.

In the future, we would like to address the issues mentioned for each work presented in this thesis. Concerning the lumbar spine model, we plan to perform more validation such as the validation of the pressures within the discs finite element models, and the the validation of the forces applied by the zygapophyseal joints. As mentioned above, we plan to use the model to test artificial discs on various subjects with different morphologies, thereby, the model will be registered to the morphology of these characters.

Concerning our ontology based modeling, we are hardly working on a new ontology of mechanical model we call My3D. This novel ontology will allow to manage all aspects present when one is interested in the modeling process for simulation. Thus, this new ontology will allow to query on mechanical models, to check the consistency of representations and to reason on knowledge such as *interactions, force fields, behaviors laws, geometry representation, contact, etc*. The ultimate goal of this new ontology will be to be coupled to MyCF in order to improve the modeling process of systems for simulation. Regarding the drawbacks mentioned of our current approach, we truly believe that developing this new ontology is the right way to go.

About the anatomy transfer, we plan to improve the stage upstream the transfer : the computation of correspondence, to make it more simple and automatic. We will improve the interpolation solver to make it more faster. We also plan to add more anatomical rules to better constraint the whole process. Lastly, we will work on partial transfer and transfer on disable people.

The work presented here has been initiated to address issues related to how we can manage existing model to create adequate subject-specific systems aiming to be simulated, we are aware that a very small part of the problem has been treated in this PhD, while bringing new questions, but we hope that the contributions we have made will help those who work and who will work on these issues.

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APPENDIX

A

ANNEXE

A.1 THELEN MUSCLE MODEL

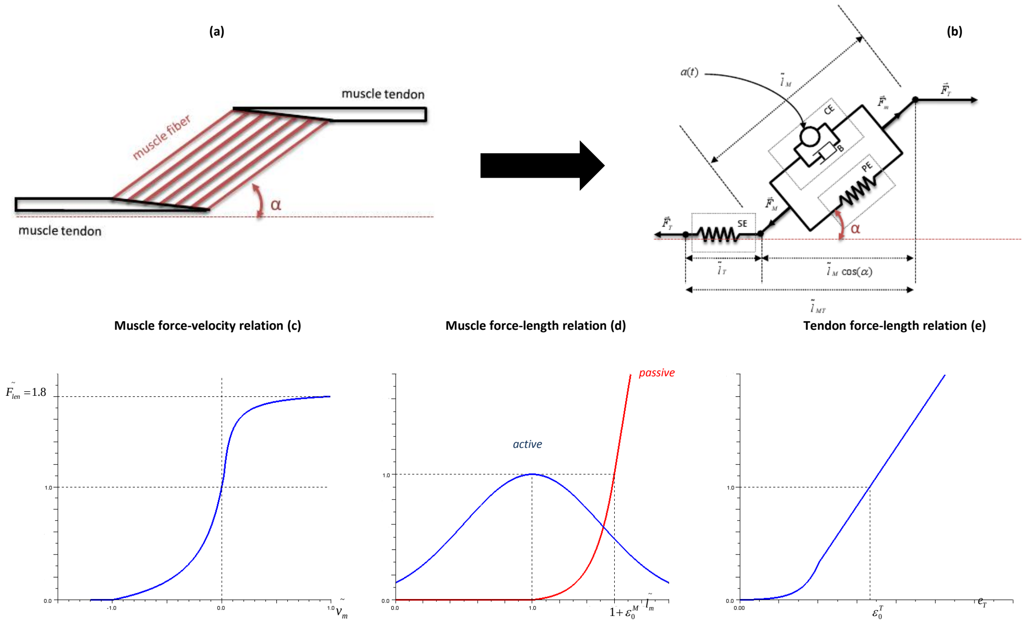

FIGURE A.1 – *(a) Muscle composed by the tendon and the oriented muscle fibers, (b) Representation of muscle as a complex spring in series, (c) Relation force-velocity, (d) Force-length relation of the contractile component (CE) and the passive component (PE), (e) Force-length relation for the tendon (SE).*

Muscle is an extraordinary machine able to transform accumulated biochemical energy into mechanical energy. We use them for their actuator role. Researchers used to model it as a complex spring with non-linear behavior laws (cf Fig.4.10). For our simulation, we use the Hill-type muscle model (cf Fig.4.10.b) proposed by D.G Thelen in [The03b]. He improves the model created by Delp et al [DLH*90] in order to account change due to age. The elastics effect of tendon and the intrinsic elasticity within the sarcomere are represented using an exponential function in a given region, and a linear function afterwards. (cf Fig.4.10.e). The normalized tendon-strain force or **serie element** force (SE) (cf Fig.4.10.b) is :

$$\overline{F}^T = \begin{cases} \frac{\overline{F}^T_{toe}}{e^{k_{toe}}-1}(e^{k_{toe}\epsilon^T/\epsilon^T_{toe}} - 1); & \epsilon^T \leq \epsilon^T_{toe} \\ k_{lin}(\epsilon^T - \epsilon^T_{toe}) + \overline{F}^T_{toe}; & \epsilon^T > \epsilon^T_{toe} \end{cases} \tag{A.1}$$

$\overline{F}^T$ is the tendon force normalized to the maximum isometric force ${F_0}^M$. ϵ^T is the strain. The value for the stiffness k_{toe} is $k_{toe} = 3.0$. The transition from the exponential to the linear behavior occurs at $\overline{F}^T_{toe} = 0.33$. To maintain continuity of slopes at the transition region, transition strain is $\epsilon^T_{toe} = 0.609\epsilon^T_0$ (ϵ^T_0 *is the tendon slack length*) and the stiffness k_{lin} is $k_{lin} = 1.712/\epsilon^T_0$.

To represent the **parallel element** (PE) (cf Fig.4.10.b) or the passive elasticity of the muscle resulting from the penetration of connective tissues into the muscle body, an exponential function is used :

$$\overline{F}^{PE} = \frac{e^{k^{PE}(\overline{L}^M - 1)/\epsilon^M_0} - 1}{e^{k^{PE}} - 1} \tag{A.2}$$

$\overline{F}^{PE}$ is the normalised passive force (cf Fig.4.10.d), $k^{PE} = 5.0$ represent the exponential shape factor, ϵ_0^M is the strain at the maximum isometric force $F_0{}^M$ and $\overline{L}^M$ is the normalised fiber length.

The contractile element (CE) (cf Fig.4.10.b) accounts for the generation of the active force, and it is dependent to the time-varying neural signal $a(t)$, the fiber length L^M and the fiber contraction velocity V^M. The properties of this element can be represented using three functions.
The active force-length relationship which is define as a Gaussian function (cf Fig. 4.10.d) :

$$\overline{F}_L = e^{-(\overline{L}^M - 1)^2/\gamma} \tag{A.3}$$

The contraction velocity which equals :

$$V^M = (0.25 + 0.75a)V^M_{max}\frac{\overline{F}_M - a\overline{F}_L}{b} \tag{A.4}$$

V^M_{max} is the maximum contraction velocity and b is force velocity damping computed differently depending on whether the fiber is shortening ($\overline{F}_M \leq aF_L$) or lengthening ($\overline{F}_M > aF_L$). And the active force-velocity $\overline{F}_V$ (cf Fig.4.10.c) :

$$\overline{F}_V = 1.8 - \frac{1.8}{1 + e^{(0.04 + \overline{V}^M)/0.18}} \qquad where \quad \overline{V}^M = V^M/V^M_{max} \tag{A.5}$$

$\overline{F}_M$ is the normalized active force or **contractile element** force express as :

$$\overline{F}_M = \overline{F}^{CE} = a\overline{F}_L\overline{F}_V \tag{A.6}$$

At each time step of the simulation, we compute the actuation force $\overline{F}_T$ applied on bones, and update the muscle state using the following equations :

$$\overline{F}^T = (\overline{F}^{CE} + \overline{F}^{PE})cos(\alpha) \tag{A.7}$$

$$\overline{L}^{MT} = \overline{L}^T + \overline{L}^M cos(\alpha) \tag{A.8}$$

$$\alpha = \arcsin(L_0^M \sin(\alpha_0)/L^M) \qquad where \quad \alpha_0 = \alpha(0). \tag{A.9}$$

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