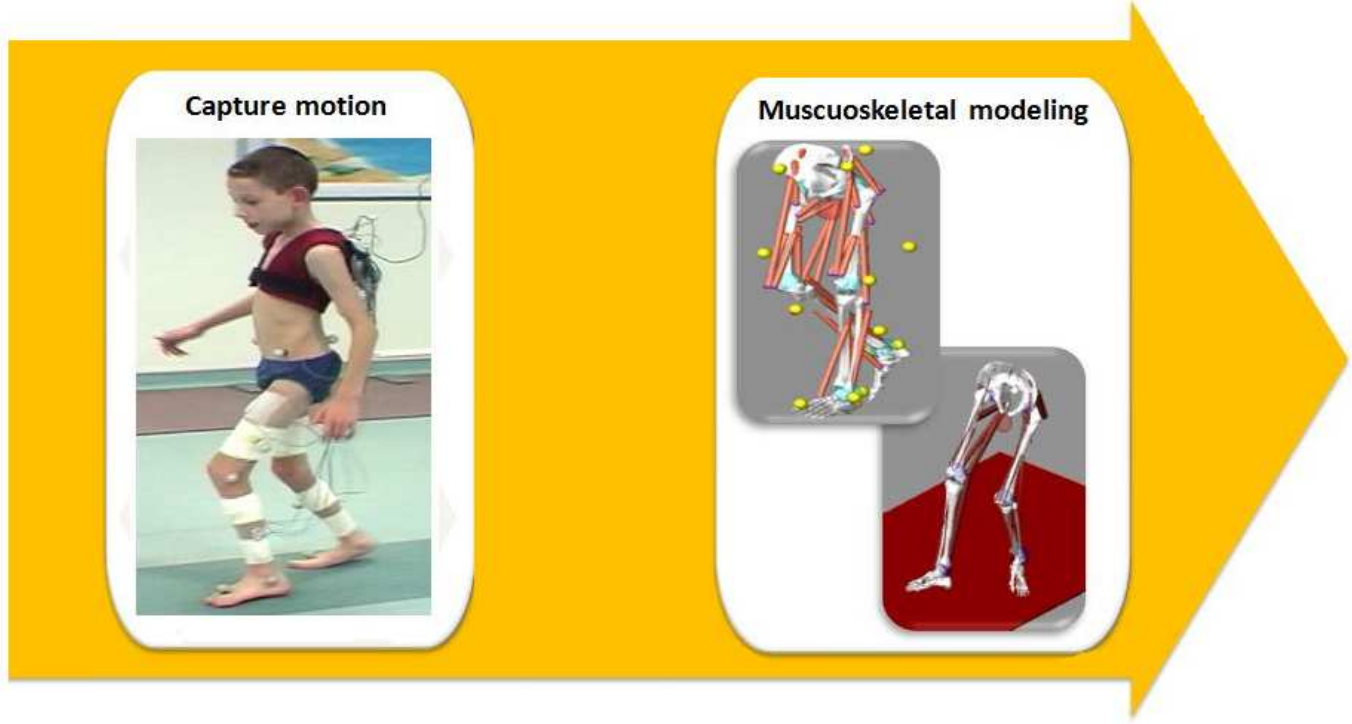
Université de Technologie de Compiègne
Biomechanics and Bioengineering field

PhD thesis

submitted in fulfillment of the requirements
for the degree of Doctor of science
in Biomechanics and bioengineering field

by Taysir REZGUI

Musculoskeletal Modeling of Cerebral Palsy Children



<u>Main supervisor</u>: Professor Frederic Marin

Submitted: January, 17th, 2012

Laboratoire de biomécanique et bio-ingénierie UMR 6600
Centre de recherche de Royallieu – Université de technologie de compiègne
Rue de Dct Schweizer BP 20599 – 60205 Compiègne cedex France

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Jury members:

Professor Marie Christine Ho-ba-tho, UTC (Examiner)
Professor Ana Presedo, hopital Robert Debré (Examiner)
Doctor Fabrice Megrot, UCAMM – Bois Larris (Examiner)
Doctor Valérie Kromer, Université de Nancy (Reviewer)
Professor Multon Franck, Université de Rennes (Reviewer)
Professor Frederic Marin, UTC (Main supervisor)

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To my Family

With all my great love and respect

Acknowledgements

The research work of this thesis was carried out in the University of Technologies in Compiegne (UTC) in the Biomechanics and bioengineering laboratory (BMBI – UMR 6600) during the years 2008-20011.

At the end of this work, I would like to address my deepest gratitude to my supervisor Professor Frederic Marin, for his precious guidance, encouragement and availability during this work to bring it to its completion. His enthusiasm, his continuous support, experience and friendship, have a great influence in my first steps in researches and it will be greatly helpful in my future carrier. I am deeply thankful to my supervisor because he believes in me and care about my future in every minute during my thesis.

Special thanks are offered to Dr. Fabrice Megrot for his valuable advices and constructive assistance. I'm really grateful for his constant and precious help and also for his kindness and warmly welcome during my visits to Bois Larris.

Thank you to Laetitia Fradet, for her availability, precious advices and support during my work and to Nicolas Vignais and Khalil Ben Mansour for this assistant and help.

I would like to thank the members of the jury, Pr. Marie Christine Ho-ba-tho (UTC), Dr. Valérie Kromer (ESSTIN- Nancy), Pr. Franck Multon (UFR STAPS - Rennes2), Dr. Ana Presedo (Robert Debré) and Dr. Fabrice Megrot (UCAMM), to accept taking part of my jury of defense and for their constructive comments and advices.

I would like to thank Picardy Region for its financial support (grant N° 0812006893) that gives me the opportunity to achieve this research work.

Many thanks go to my colleagues for their participation and patience during my long experimentations. My gratitude goes also to my colleagues who accepted to accompany their children for a gait capture movements in the laboratory: to Pr. Frederic Marin and his daughter Zoe and his son Emrique, to Dr. Sofiane boudaoud and his son Zyed, to Pr. Marie-Christine Ho-ba-tho and his son Charles, to Mme Sylvie Lugez and his son Mathis.

I would like also to thank all my friends in the University of Technologies in Compiègne. Special thanks go to Leatitia Debernard for her generosity and kindness, to my dearest friends Gwladys, Jean-Sebastien, Aly, Mahmoud, Fouaz, Kais and Hassen for their support and encouragement and also for all the good times daily shared together. I would like to thank my laboratory' colleagues: Walid, Helen, Laurie, Nicolas, Tuan.

I want to thank all the partners who, directly or indirectly, contributed to the better achievement of this thesis: Marie-christine Ho-ba-to, Catherine Lacourt, Catherine Marque, Sofiane Boudaoud, Didier Gamet and Philippe Pouletaut.

I cannot achieve this part without thanking my supervisor for planning monthly CINDYAH meetings to share our works and learn new challenges in biomechanical researches and the lovely and friendly dinners with our little group's families: Laetitia., Frederic, Cornelia, Zoe , Emrique, Nicolas, Audrey, Elliot and the little Azélie.

Finally, and most importantly, I am very grateful to my family that never stopped overwhelming me with love, support and encouragement. My warmest thanks go to my parents for being a constant source of encouragement, for their patience and precious caring; to my brother for his caring and comprehension, and to my lovely sister for his continuous love and infinite support.

Thesis abstract

Musculoskeletal modeling of cerebral palsy children

The analysis of pathological gait using musculoskeletal modeling is a promising approach to qualify and quantify the pathology as well as to monitor the potential recovery after therapy. When dealing with cerebral palsy, its specific neurological disorders and consequently bones deformities, specific-subject musculoskeletal models has been developed. The imaging techniques are still unaffordable in clinical practices. So, using the LifeMod software, we aimed to develop musculoskeletal model in a retrospective study to evaluate the accuracy of surgical treatments on cerebral palsy. Two principles studies are performed. First, relying on the accuracy of a rescaled generic adult skeleton, the musculoskeletal modeling limitation have been determined when applying normal gait and pathological crouch and jump postures, imitated by healthy adults and children. Second, calibration technique had been developed to refine the model's parameters based on data collected from the subject. Results from musculoskeletal modeling are compared to gait analysis data. As results, even if the model outputs gave correct results with healthy adults, the standard rescaled musculoskeletal modeling showed limits on predicted kinematics and muscle forces for healthy and CP children. The refinement of subject-specific joint parameters and driving the model with the experimental GRF data have a huge influence in model outputs and improve quantitatively the predicted muscle activations and forces. This work pointed out that the parameters of a rescaled generic musculoskeletal model can be refined and personalized to improve model's outcomes. It may represent a new perspective in clinical applications.

Key words: Cerebral Palsy, Musculoskeletal modeling, Calibration, Gait analysis

Abstract de thèse

Modélisation musculo-squelettique des enfants paralysés cérébraux

La modélisation musculosquelettique est aujourd'hui utilisée dans de nombreux domaines tels que l'analyse de la marche pathologique et la simulation des traitements thérapeutiques et chirurgicaux. Dans le cadre de la paralysie cérébrale (PC), la prise en considération des spécificités des patients, des troubles neurologiques et des déformations osseuses est nécessaire. Etant donné que les techniques d'imagerie médicale sont encore marginales en routine clinique, le recours aux modèles génériques reste donc indispensable. Notre étude rétrospective vise le développement d'un modèle musculosquelettique (MMS) générique adapté aux enfants PC. Une première étude détermine les limites d'un tel modèle pour la marche normale, les marches pathologiques des enfants paralysés cérébraux, et les postures pathologiques imitées par une population saine. Une seconde étude propose une technique de calibration pour raffiner les paramètres du MMS à partir des données recueillies de l'analyse quantifiée de la marche (AQM). Ainsi, on a pu déduire que, même si les résultats estimés sont représentatifs pour les adultes sains, le MMS standard présente des limites concernant la cinématique et les forces musculaires prédites pour les enfants sains et les enfants PC. D'autre part, la procédure de calibration influe de façon positive sur les données prédites comme les activations musculaires et les forces musculaires. Ce travail montre que le MMS générique peut être calibré à partir des données de l'AQM afin d'améliorer les résultats du modèle. Cette technique pourrait représenter une nouvelle perspective dans les applications cliniques de la modélisation musculosquelettique.

Mots clés: Modélisation musculosquelettique, paralysie cérébrale, calibration, analyse de la marche

Sommaire

List of abbreviation

UCAMM	Unité Clinique d'Analyse de la Marche et de Mouvement – Centre de Médecine Physique et de Réadaptation de Bois Larris
CinDyAh	Cinématique et Dynamique des Articulations Humaines
CP	Cerebral Palsy
MSM	Musculoskeletal model
GRF	Ground Reaction Force
EMG	Electromyography
MRI	Magnetic resonance Imaging
CMC	Canonical Measure of Correlation
PCSA	Physiological cross-section Area
PID controllers	Proportional- integrator- derivative controllers
PD controllers	Proportional- integrator- derivative controllers
ROM	Range of Motion

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General Introduction

Cerebral palsy is defined as a clinical syndrome characterized by disorders of movement and posture caused by a non-progressive lesion of a developing brain. The primary characteristics of the CP are the altered motor control and abnormal muscle tone usually due to spasticity (primary effects). The presence of such aberrant muscle coordination induces shortening of specific muscles and alters the joint range of motion. The functional abilities of the child with spastic CP are profoundly affected and often deteriorated during childhood growth by the development of many lower limb bone deformities (secondary effects) and compensation mechanisms instated spontaneously helping the child to have his proper gait autonomy.

Pathological gaits observed in CP children are results of interferes over time between these effects. It is important, for clinical evaluation, to define primary effects which are permanent and to discriminate between bone abnormalities which can be corrected and the compensatory mechanics which disappear as soon as they are no longer required.

Facing the CP problems, clinicians have to determine the best prognosis and to select the appropriate treatment leading to increase the child's quality of life. The clinical gait analysis exam, combined to the clinical history of the patient, is decisive for planning surgical and rehabilitation treatments for these disorders. The description of gait by kinematics, kinetics and muscle activation using surface electromyography (SEMG) quantitatively documents the gait disorders and helps clinicians in understanding the abnormal pattern and assists them into the clinical decision making. Although this approach has led to a more objective assessment of locomotion biomechanics, its ability to quantify muscle function is limited. Muscle activations, recorded by surface electromyography (SEMG) systems, determine only whether a muscle is active or not. There is any established correlation between the level of a measured SEMG signal and the amount of force that the muscle might be producing during a dynamic movement such as walking. Also, the muscular system is very redundant and SEMG is only used to measure the principal muscle groups in lower limbs, it could not inform and quantify the action of individual muscle contribution during gait, which may help clinician understanding the pathology. Musculoskeletal modeling appears as a complementary tool, in order to estimate isolated muscle forces that are difficult to obtain by direct measurement in vivo or from a gait analysis experiment. Musculoskeletal models are also used to predict post-treatment clinical outcomes.

Previous researcher studies have determined that results from musculoskeletal models are very sensitive to inter-individual variations in its biomechanical parameters (maximal muscle forces, joint stiffness, PCSA's muscle,…). Nevertheless, most studies still rely on rescaled generic musculoskeletal models, generated from anthropometric database of literature. Recent studies focused on specific- subject modeling (e.g. musculoskeletal geometry, bone deformities, muscle insertion) demonstrated that these elements are to be taken into consideration when studying pathologies. However, musculoskeletal models have common parameter hypothesis of visco-elastic joints and muscles. Biomechanical results might be highly sensitive to the parameter hypothesis and would possibly provide offsets or wrong results. Biomechanical parameters of the model were set using data gathered from literature and especially deduced from averaged data of cadaveric measurements in a healthy adult population. It cannot represent a normal gait of healthy children neither Cerebral Palsy ones. Few researches demonstrated that biomechanical parameters have highly consequences on the obtained results. Therefore, the accuracy of these parameters is highly important. These parameters must have physical meanings and clinical interpretations, which helps correcting the set of parameters values and improving prediction of the kinematics and kinetics data. It would help the clinical understanding and transfer in patient-specific treatments.

Starting from these observations, with the collaboration of F. Megrot, responsible of the gait analysis platform of the Red Cross institute in Bois Larris, and with the financial funds from the Picardie region, the thesis project aims at developing musculoskeletal models for cerebral palsy children as a tool for a retrospective study to evaluate the accuracy of surgical treatment done previously in this clinical center. This last condition imposed the use only if the information gathered from the clinical gait analysis exam and the clinical examination.
Therefore, the objective of my PhD Thesis will be to answer to the following questions:

- Can standard generic musculoskeletal modeling provide satisfactory results when studying cerebral palsy pathological gaits?
- Can musculoskeletal modeling parameters be refined and calibrated only by the use of data gathered through a clinical gait exam?

In this project, two studies are developed to answer these issues. The first one consists of using the standard rescaled generic model to define limits of such modeling for healthy adults and children and also cerebral palsy children with spastic diplegia. The second one requires

the calibration of musculoskeletal model's parameter values to data collected from patient. The parameters taken into consideration are viscoelastic parameters of the joints, parameter of the contact with the ground, other intrinsic parameters of the model such as, the parameters defining the motion agents, and parameters of the controllers conditioning the forward dynamic procedure.

For healthy population, musculoskeletal models are developed, firstly, for a normal gait to determine the impact of rescaling on child skeleton model and secondly, we studied the influence of altered muscle activation on model's results. This condition was performed when healthy subject imitated representative CP pathological gait, crouch and jump gaits.

The PhD Thesis report will be organized into chapters as follows:

- The chapter 1 develops a general literature background of human gait, the cerebral palsy and its specificities and finally a literature review of musculoskeletal models developed for CP cases.

- The chapter 2 deals with the materials and methods used to develop musculoskeletal models, going from gathering data of gait analysis exams to the numerical simulation with LifeMod software and data analysis tools.

- Results from Musculoskeletal simulations are presented in the chapter 3 and subsequently discussed in the chapter 4. Conclusions drawn from our studies and future work recommendations are finally given.

Chapter 1.
Literature Review

Normal walking is a complex movement which consists of highly complex interactions between the mechanical structure of the body, the muscles, the nervous system and the other physiological systems. Neurological pathologies, such as Cerebral Palsy, affect the way of walking and biomechanical analysis of CP gaits reveals its complexity. The principle goal for cerebral palsy children is increasing the mobility and decreasing the pain. Objective clinical function measurements, based on physical examination and the clinical gait analysis examination, are important to assess suitable treatments. In addition, musculoskeletal modeling is progressively drawn on as tool giving complementary quantified muscle information.

In this chapter, we will focus on the literature background concerning essential parts of the musculoskeletal modeling. A description of the anatomy of lower limbs (§1.1) and the normal gait are presented (§1.2) followed by the pathology of Cerebral palsy and its specificities (§1.3). Finally, musculoskeletal models developed to better understand Cerebral Palsy pathological gaits, are detailed (§1.4).

1.1. Anatomical review of lower limbs

A wide variety of movements carried out by the human musculoskeletal system are performed and controlled through interaction between skeletal system, joints, muscles and the central nervous system. The §1.1.1 presents the anatomical axis, the §1.1.2 details the musculoskeletal anatomy and finally, the §1.1.3 presents the central nervous system and motor control strategies.

1.1.1. Anatomical plans/ axis

The anatomical position is the universal starting position for describing human body part positions and movements. A three dimensional coordinate system consisting of three anatomical planes, sagittal, frontal and transverse planes (Figure 1.1), is used to identify an anatomical relationship of structures relative to one another and to itself in space**:**

- The sagittal plane is the only plane of symmetry in the human body. This vertical plane, passing through the midline of the body from front to back, divides the body into left and right parts.
- The frontal plane, also called the coronal plane, is a vertical plane perpendicular to the sagittal plane which divides the body into anterior and posterior sides.

- The transverse plane, also called the axial plane, is a horizontal plane, parallel to the ground, which divides the body into superior and inferior parts

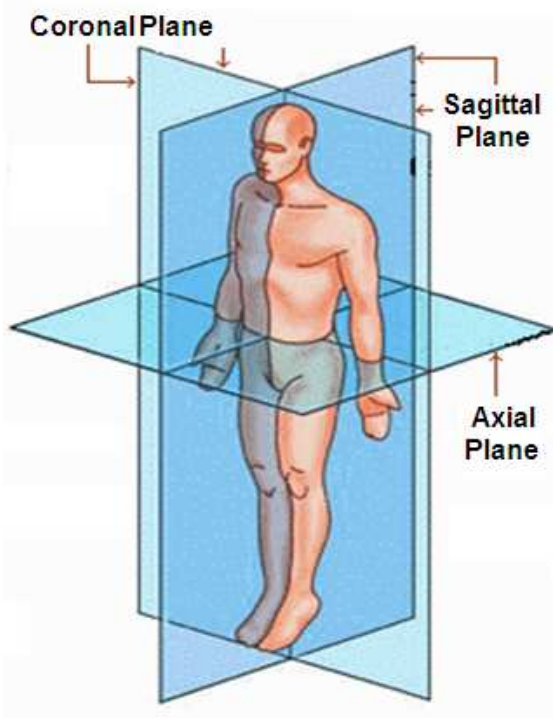


Figure 1- 1: The Three Primary Anatomical Planes of the Human Body in the Standing Anatomical Position. (© www.spineuniverse.com)

1.1.2. Musculoskeletal Anatomy

The lower limb skeleton anatomy is composed of four distinct parts: a pelvic girdle, the femur, the tibia and the foot (Figure 1.2), linked together through several joints: the hip, knee and ankle joints. A movement is usually a collaboration of a set of muscles, coordinated and controlled by the central nervous system.

The main role of the lower extremities is the support of weight, adaptation to gravity, and locomotion. The foot provides an additional stable support in the upright posture.

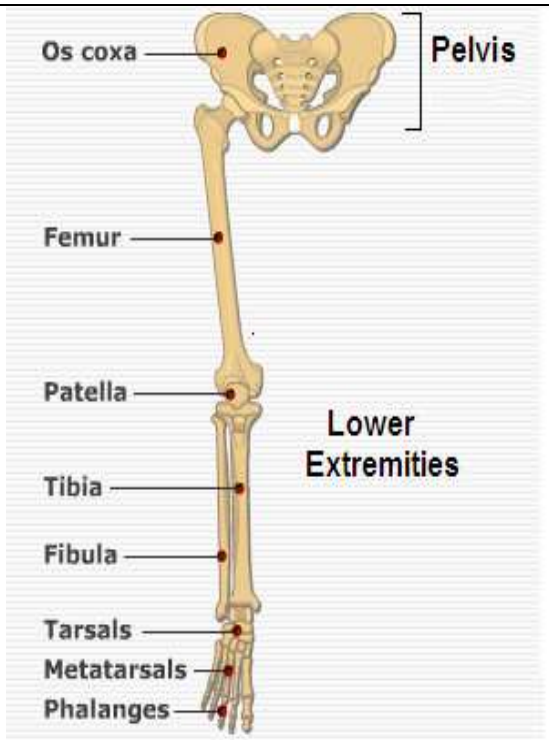


Figure 1- 2 : Frontal View of the Lower Extremities of the Human Skeleton
(© www.getbodysmart.com)

1.1.2.1. Bones

a) Pelvis

The pelvis forms a bony ring that connects the torso and lower extremities. It is composed by the two hip bones attached to the Sacrum and to the Coccyx, the last two elements of the vertebral column. The pelvic girdles of female are more flexible, broader and shallower than those of male as an adaptation for pregnancy (childbearing). A detailed description of the pelvis anatomy is shown in the figure below (Figure 1.3).

The pelvis gains its strength and stability through the surrounding ligaments and muscles and its primary function is to protect the abdominal organs and to support the upper body when sitting or standing.

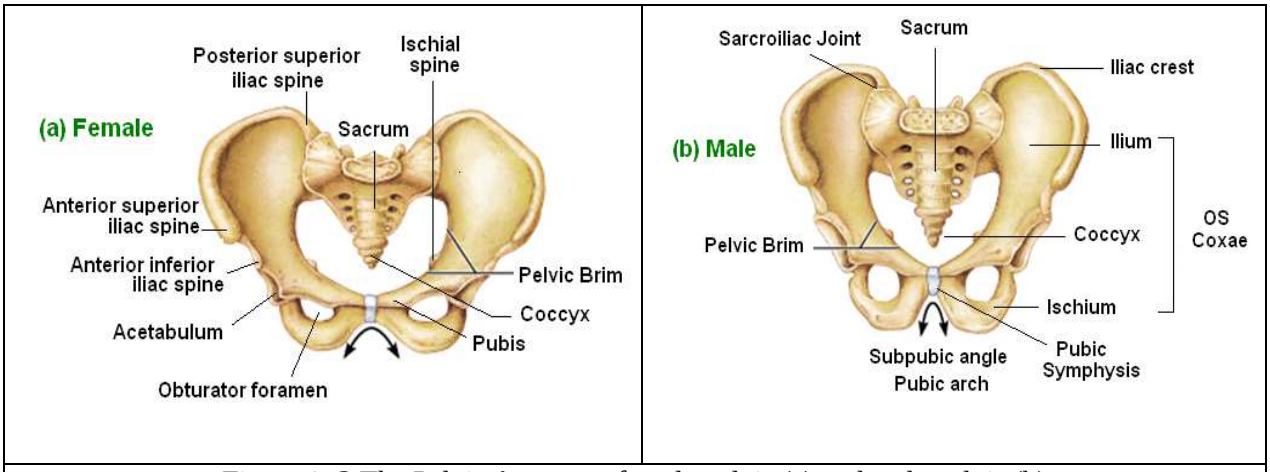


Figure 1- 3.The Pelvis Anatomy: female pelvis (a) and male pelvis (b)
(© Marieb E. 2010)

b) Upper leg

- Femur

The femur, also called the thigh bone, is the longest and the heaviest bone of the human skeleton located between the hip bone and the knee. It makes up part of the hip joint on the acetabulum of the innominate bone and of the knee joint on the tibia (Figure 1.4). The femur is composed of four parts: the head, a shaft, greater trochanter and lesser trochanter, which give attachment to muscles. The head of the femur joins the pelvis and the other end articulates with the tibia of the leg at the knee joint.

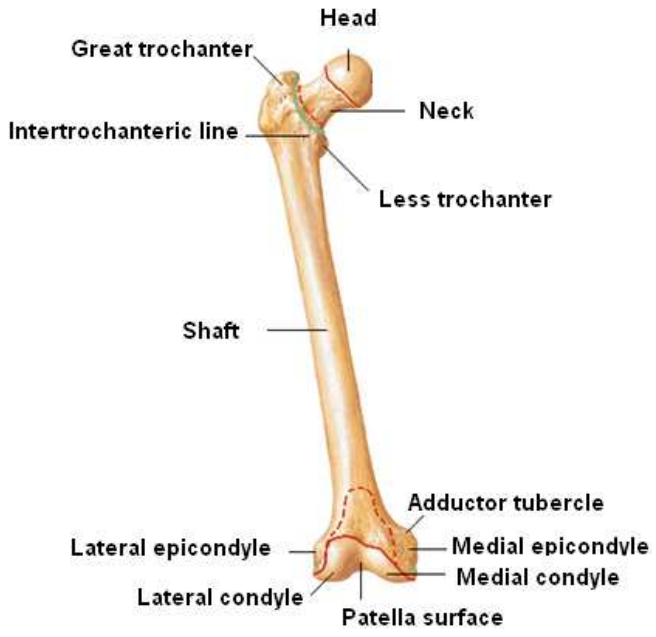


Figure 1- 4. The upper leg bones: Anterior View of the Femur (© www2.ma.psu.edu)

- Patella

The patella, also known as the knee cap, is a triangular shaped bone found between the femur and fibula (Figure 1.5).

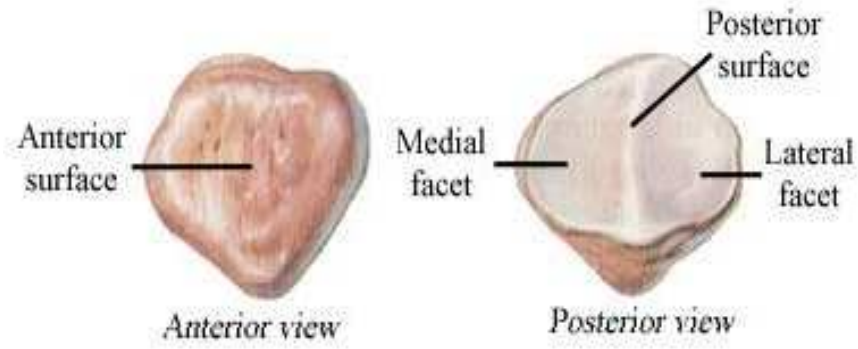


Figure 1- 5. The Upper Leg Bone: Patella (© www.kine-formations.com)

It is a sesamoid bone developed in the tendon of the quadriceps extensor muscle. It is a relatively thick bone consisting of a rough anterior surface and a smooth posterior side articulating with the patellar surface of the lower extremity of the femur. Its primary function is to protect of the knee joint.

c) Lower leg

Together with the fibula, the tibia forms the lower leg. They are commonly treated as a single skeletal structure, connecting the patella and the ankle (Figure 1.6). The fibula is parallel with the tibia on its outer side and does not form a part of the knee-joint.
The tibia and fibula are further connected both head extremities by ligaments and joined throughout their lengths by an interosseous membrane between the bones.

The upper extremity of the tibia consists of medial and lateral condyles, connected to the femoral condyles to form the knee-joint and it represents the attachment surface of the ligamentum patella. The inferior surface of the tibia makes part of the ankle joint. It is grooved by tendon attachments and connected to the talus through the lateral surface of the medial malleolus. The tibia and the fibula provide support for both the calf muscles and the Achilles tendon.

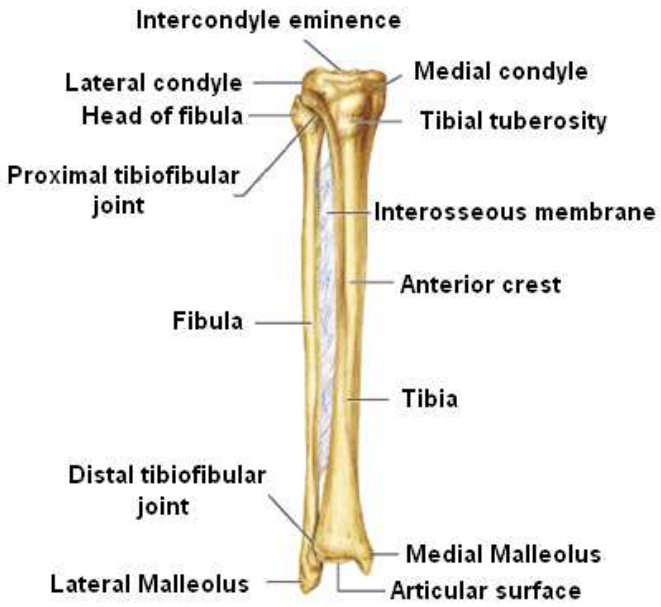
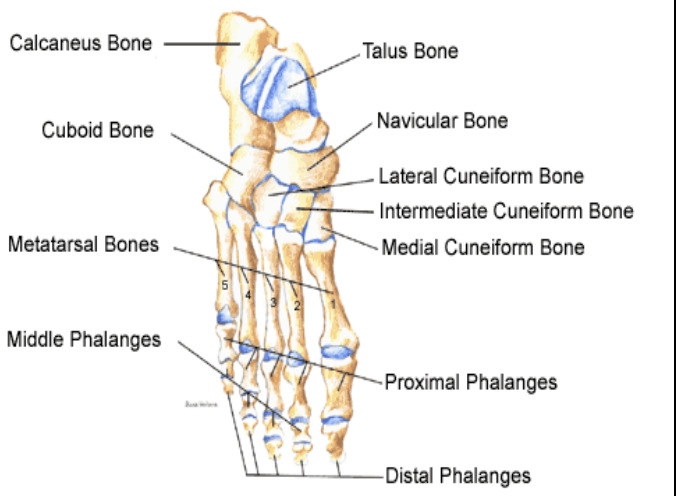


Figure 1- 6. The Lower Leg Bones: Anterior View of the Tibia and the Fibula (© www2.ma.psu.edu)

d) Foot

The human foot is an important functional part of the anatomy. Its fundamental functions are supporting the body's weight and propelling the body forward when walking and running and it is constantly exposed to high level of mechanical stresses.

The bone structure of the foot is divided into three parts: the forefoot, the midfoot, and the hind-foot bringing more flexibility (Figure 1.7).



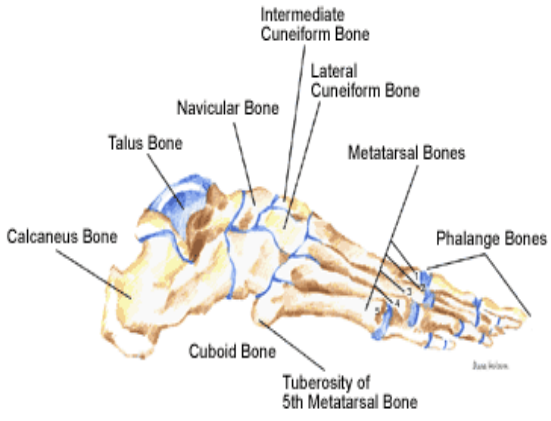


Figure 1- 7. The Foot Structure (© www.healthcommunities.com)

1.1.2.2. Joints

There are three principle joints in the lower extremities of the human skeleton, which are the hip joint, the knee joint and the ankle joint. These joints are assumed to be synovial joints.

a) The hip joint

The hip joint forms the connection between the lower limb and pelvis. It is a multi-axial ball-and-socket synovial joint, where the ball is the femoral head and the socket is the Acetabulum (Figure 1.8).

The hip joint is a very strong and stable articulation. It is surrounded by powerful muscles and a dense fibrous capsule, which is strengthened and reinforced by five ligaments. The principal external ligaments are, the Iliofemoral ligament preventing from over-extension movement, the Pubofemoral ligament preventing from over-abduction movements and the Ischiofemoral ligament preventing the hyper-extension of hip joint. Internally, there are two ligaments namely: the ligamentus teres, and the traverse acetabular ligament, which help limiting hip adduction and hip displacement.

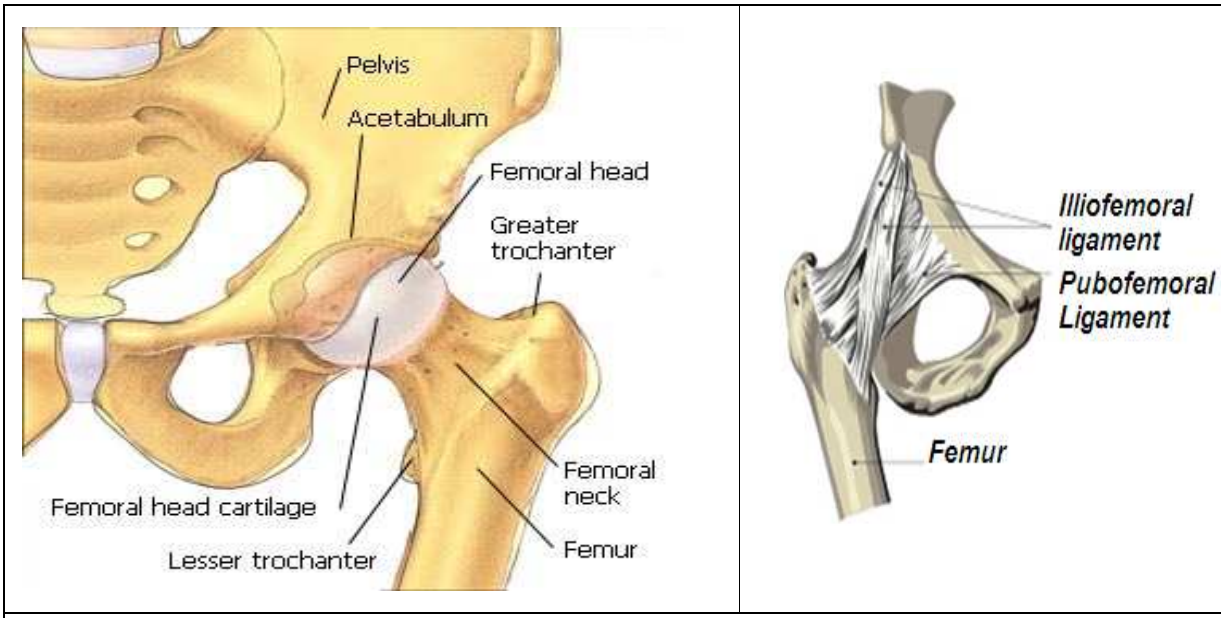


Figure 1- 8. The Hip Joint (© www.hipsurgery.co.il)

b) The knee joint

The knee joint is a condylar articulation between the condyles of the femur, those of the tibia, and the patella (Figure 1.9). As first approximation, it could be represented as a hinge joint for extension and flexion accompanied with some gliding and rolling with rotation on vertical axis.

The integrity of the knee joint is secured and stabilized by the sets of ligaments connecting the upper and lower leg bones. The Cruciate Ligaments (Anterior Cruciate Ligament, Medial Collateral Ligament, Posterior Cruciate Ligament and Lateral Collateral Ligament) are responsible for a significant degree of the stabilization at the front of the joint and the Anterior ligament is resisting forward displacement of the tibia on the femur.

The stability is ensured due to surrounding muscles and tendons. The most important knee stabilizers are the quadriceps femoris, the knee cartilage and the Medial and Lateral meniscuses. These last anatomical structures provide shock absorption as well as assistance in the reduction of the friction that could otherwise occur when bones come into contact.

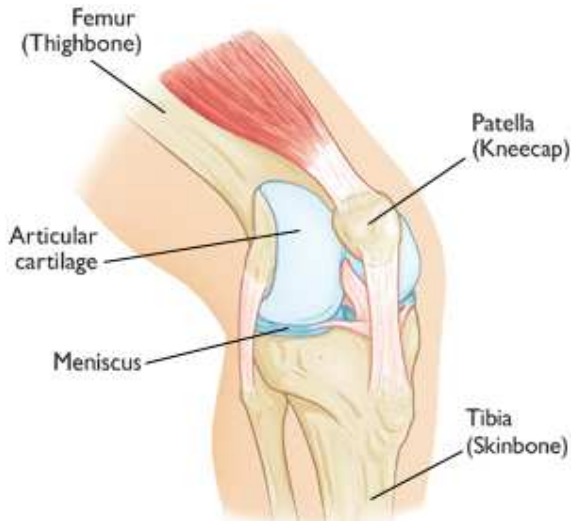


Figure 1- 9. The Knee Joint (© www. orthoinfo.aaos.org)

c) The ankle joint

The ankle joint is a hinge joint connecting the tibia, the fibula, and the ankle bones, which are secured and reinforced by a protective structure, composed of three separate sets of strong

ligaments (Figure 1.10). The structure of the joint and the organization of ligaments permit the ankle to be rotated, flexed, and extended in all directions.

The ankle joint allows, by its sophisticated structure, dorsiflexion and plantar flexion around an axis that passes approximately through the malleoli, extension and rotation in all directions.

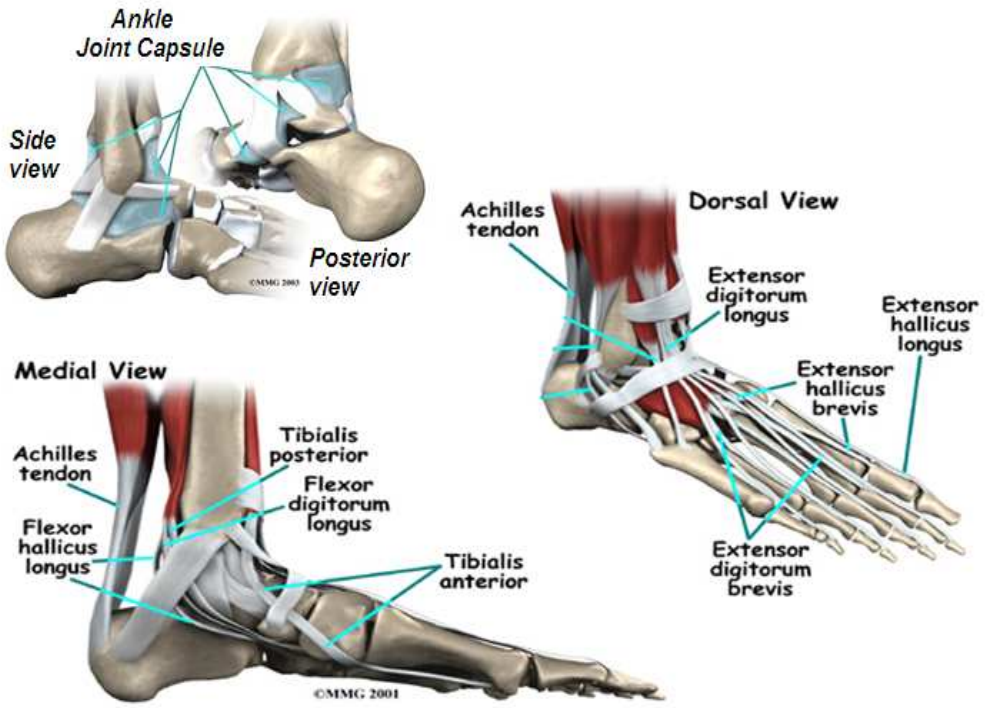


Figure 1- 10. The Ankle Joint
(© www.sportspodiatry.co.uk - www.parkwayphysiotherapy.ca)

d) Secondary joints

- **Tibiofibular joint**

The tibiofibular joint**,** connecting lower leg bones, is composed by two joints: proximal and distal; and interosseous membrane. In proximal view, the joint is a plane type of synovial joint between fibular head and lateral tibial condyle, strengthened by anterior and posterior ligaments of fibular head. It ensures gliding movements during dorsiflexion and plantar flexion. In distal part, the joint is a fibrous joint, essential for the stability of ankle joint. It keeps lateral malleolus against lateral surface of talus and it is strengthened by tibiofibular ligaments and inferior transverse ligament.

- **Foot joints**

Inversion and Eversion of the foot take place at the talocalcaneal articulations and at the mid-tarsal joints between the calcaneum and the cuboid and between the talus and the navicular. The talocalcaneal joint is the more important and the other tarsal joints are not of clinical importance, they allow slight gliding movements only, and individually. The metacarpophalangeal and interphalangeal joints are basic plan joints allowing flexion – extension and they are tightly joined by ligaments that allow only slight movements.

Several tendons and ligaments surround the foot securing it, like the large Achilles tendon, the posterior/ anterior tibial tendons, small tendons bending the toes down, the lateral malleolus tendons helping turn the foot outward and many small ligaments holding the bones of the foot together.

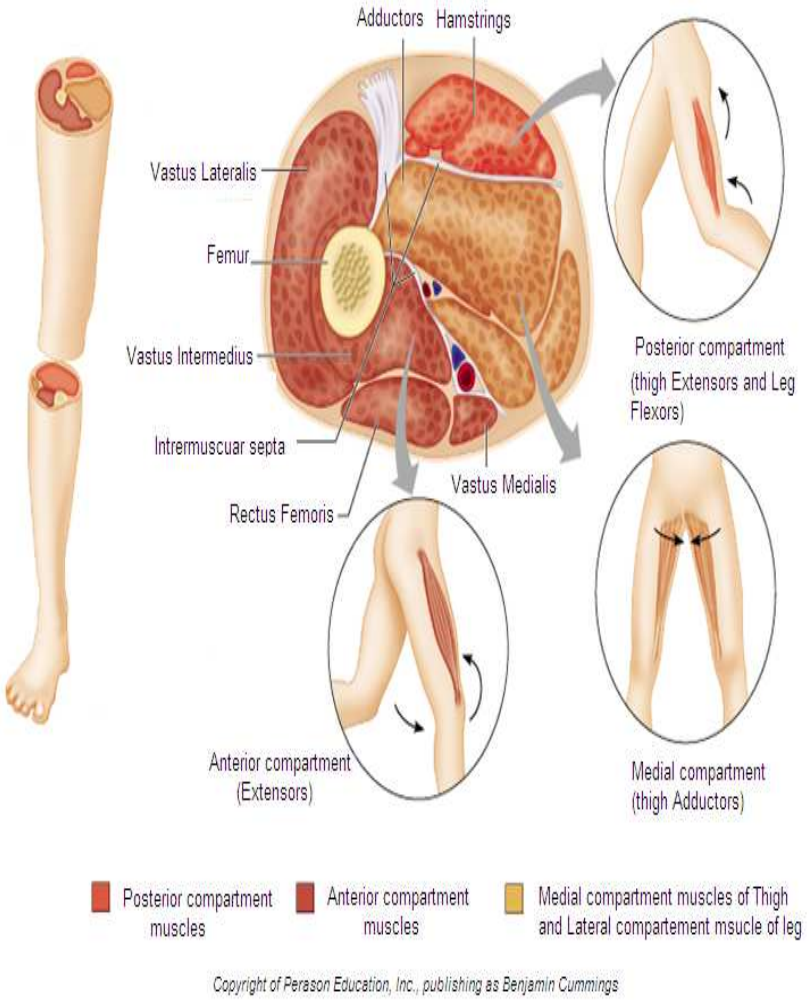
1.1.2.3. Muscles

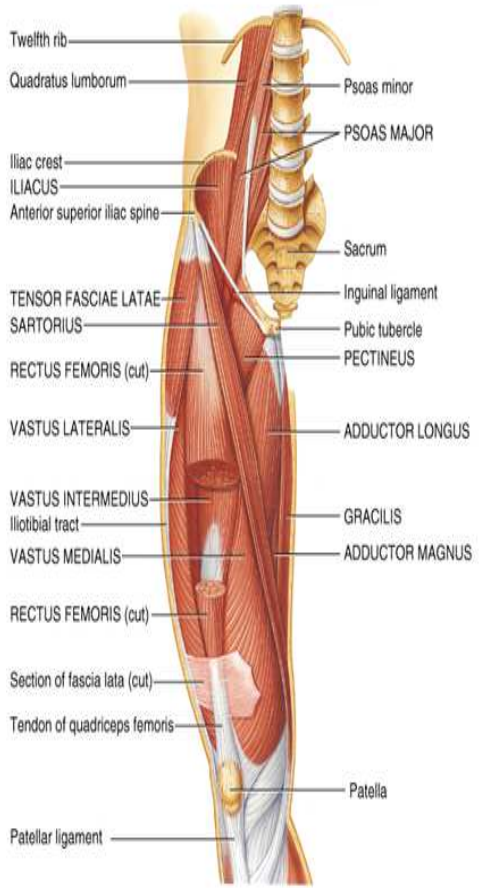
a) The upper leg muscles

The thigh comports the chief muscle acting on both the hip and the knee (Figure 1.10-11). On the anterior side of the thigh, the principal muscles are the iliopsoas, quadriceps femoris, and Sartorius; they mostly represent the flexors of the hip and the extensors of the knee. On its posterior side, the main muscles are the hamstrings (biceps femoris, semitendinous, and semimembranosus), the major extensors of the thigh and flexors of the leg, especially during walking. On its medial side, the major muscles are mostly the adductors of the thigh (pectineus, adductor longus, brevis, magnus, and gracilis).

Quadriceps Femoris forms the prominent muscle mass, located on the anterior side of the thigh. It comprises the rectus femoris and three vasti (lateralis, medialis, and intermedius). They are the principle flexor of the hip and the main extensors of the knee.

Gluteus maximus is the main extensor hip muscles. The gluteus medius, and gluteus minimus are the main muscle group of abduction and medially rotation of the thigh and also supporting the pelvis in walking and running. These muscles originate at different locations on the hip bone and insert on the femur.





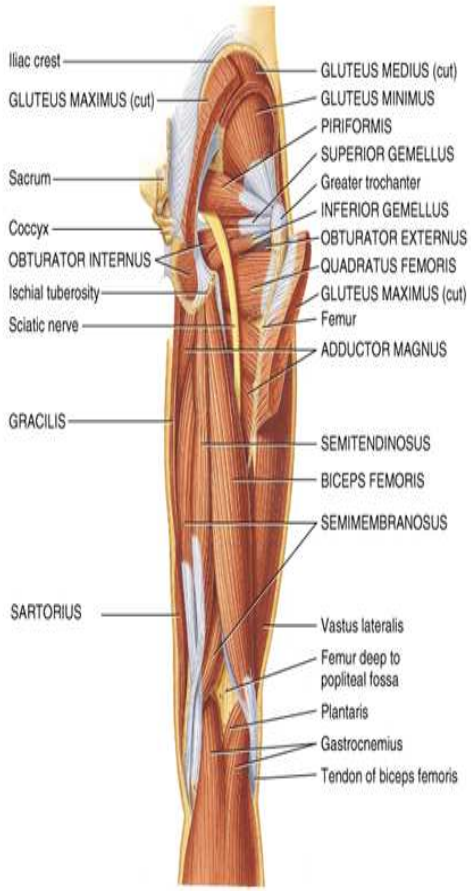


Figure 1- 11. Upper leg: Thigh muscles and their functional actions (© Marieb E. 2010)

Hamstrings are the primary muscles located at the posterior of the thighs and play an important role to the overall muscular balance of the knee joint. They are formed by the semimembranosus, the semitendinosus, and the biceps femoris. Together with the gluteus maximus, they represent the extensors of the hip which are responsible for contracting and extending the lower leg. Hamstrings, assisted by gracilis, gastrocnemius and Sartorius, represent the main flexors of the knee.

The hip adductors are located on the medial compartment of thigh and formed by several monoarticular muscles: the adductor magnus, longus and brevis assisted by gracilis and pectineus muscles.

b) The Lower leg muscles

The calf muscles and the Achilles tendons are especially responsible of Ankle dorsi-flexion and plantar-flexion and also foot inversion and eversion (Figure 10-12).

The muscles of the anterior leg are the tibialis anterior, extensor digitorum longus, peroneus tertius, and extensor hallucis longus. These muscles are dorsiflexors of the ankle joint and extensors of the toes.

The muscles of the lateral side of the leg are called the peroneus muscles and hold the peroneus longus and brevis muscles. These muscles pull the foot outward and assist in foot plantarflexion.

Muscles of the posterior side of the leg are principle plantar-flexors of the foot and have an important role in both posture and locomotion. The superficial muscles hold the large muscles, that are most commonly known as the calf muscles, the gastrocnemius and soleus, together called also triceps surae and attached to the Achilles tendon. The deep muscles are the flexor digitorum longus, flexor hallucis longus, and Tibialis Posterior, responsible for toes flexion. All these muscle assist the calf muscles in foot plantar-flexion movements.

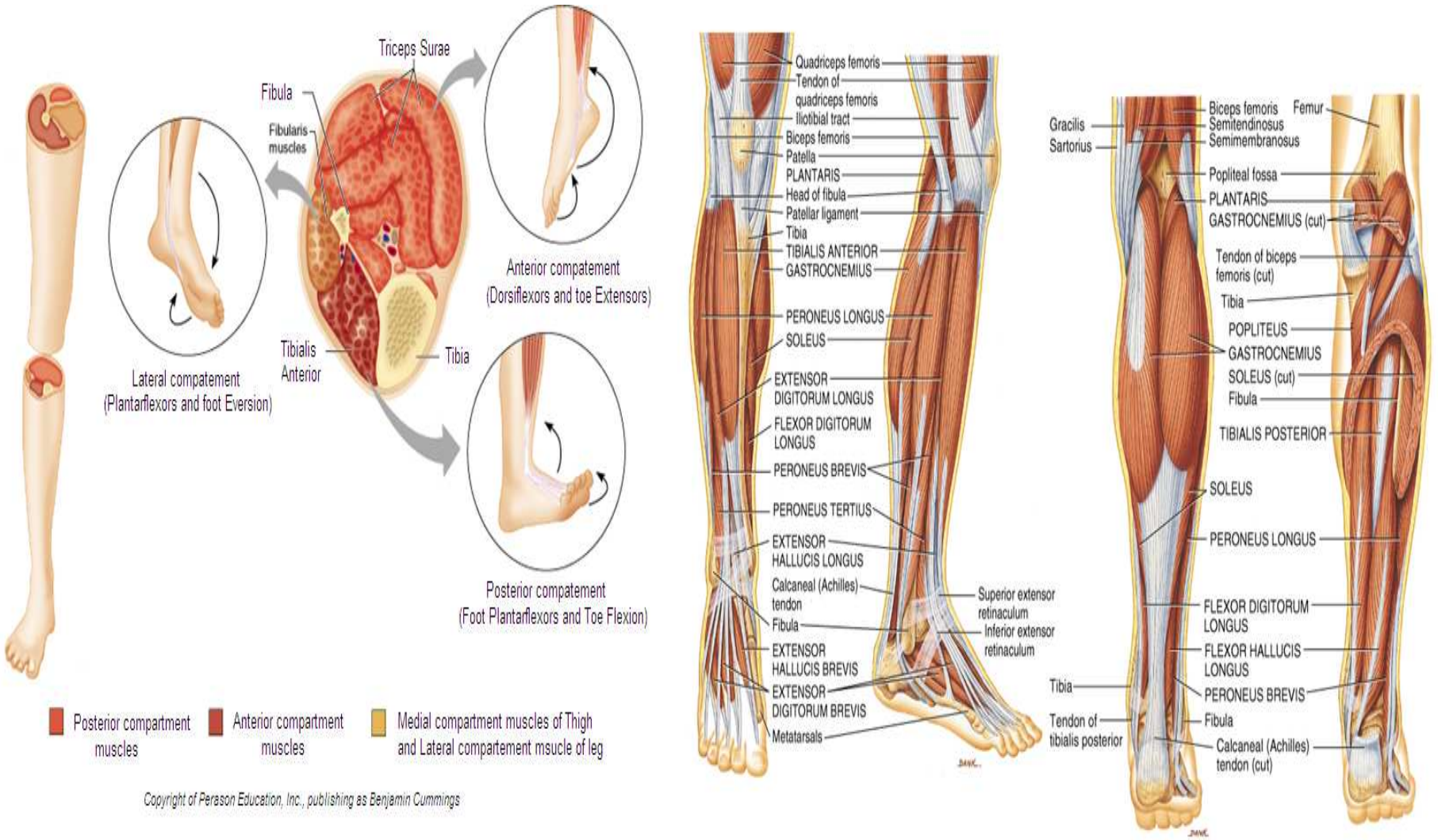


Figure 1- 12. Lower leg: Shank muscles and their functional actions (© Marieb E. 2010)

c) The foot muscles

Most of the motion of the foot is supported by the lower leg's muscles connected to the foot through strong tendons. The foot Inversion is carried out by tibialis anterior and posterior and assisted by the long extensor and flexor tendons of the hallux. The foot eversion is the function of peroneus longus and brevis (Figure 1.10). There is a single dorsal foot muscle, the extensor digitorum brevis, which extends the toes.

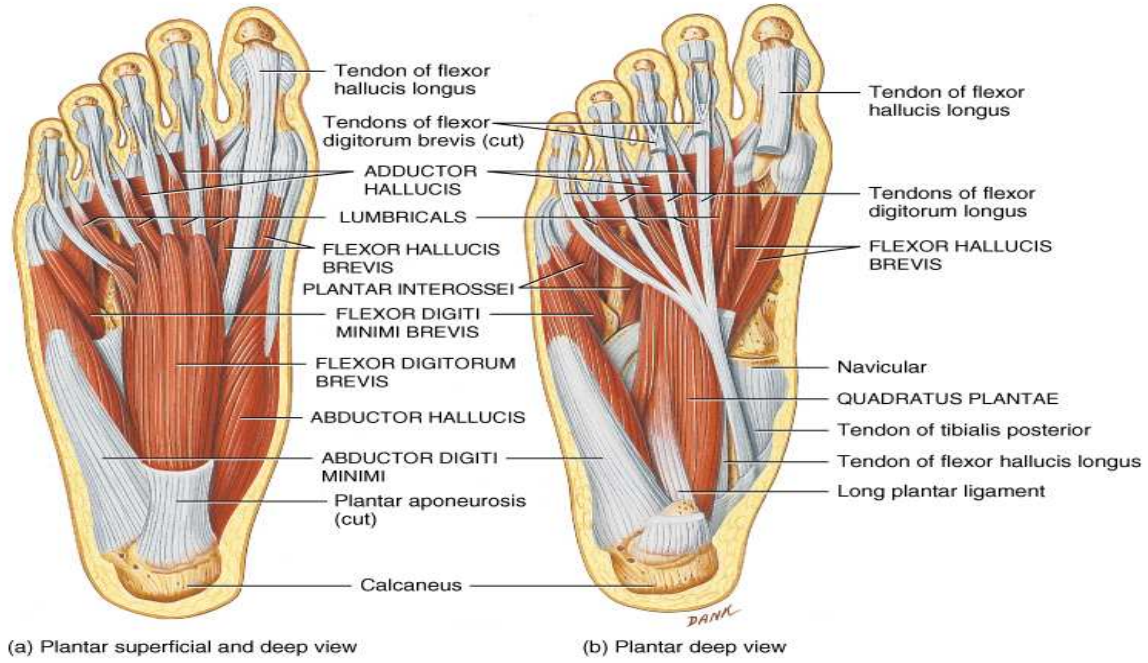


Figure 1- 13. Intrinsic Plantar Muscles of the Foot (© Marieb E. 2010)

There are numerous small plantar muscles in the foot, arranged in four principle layers on the sole of the foot. They are responsible for moving the toes (Figure 1.13). These muscles are collectively important in posture and locomotion, and they provide strong support for the arches of the foot during movement.

1.1.3. Central Nervous System and Motor Control

The voluntary body motions are achieved through coordinated skeletal muscle activities acting on a multi-articulated skeleton in a controlled manner to accomplish the predetermined task requirements. The muscle contractions are simulated and controlled by the nervous

system, efferent nerves and sensory neurons connected with skeletal muscles and skin (Figure 1.14).

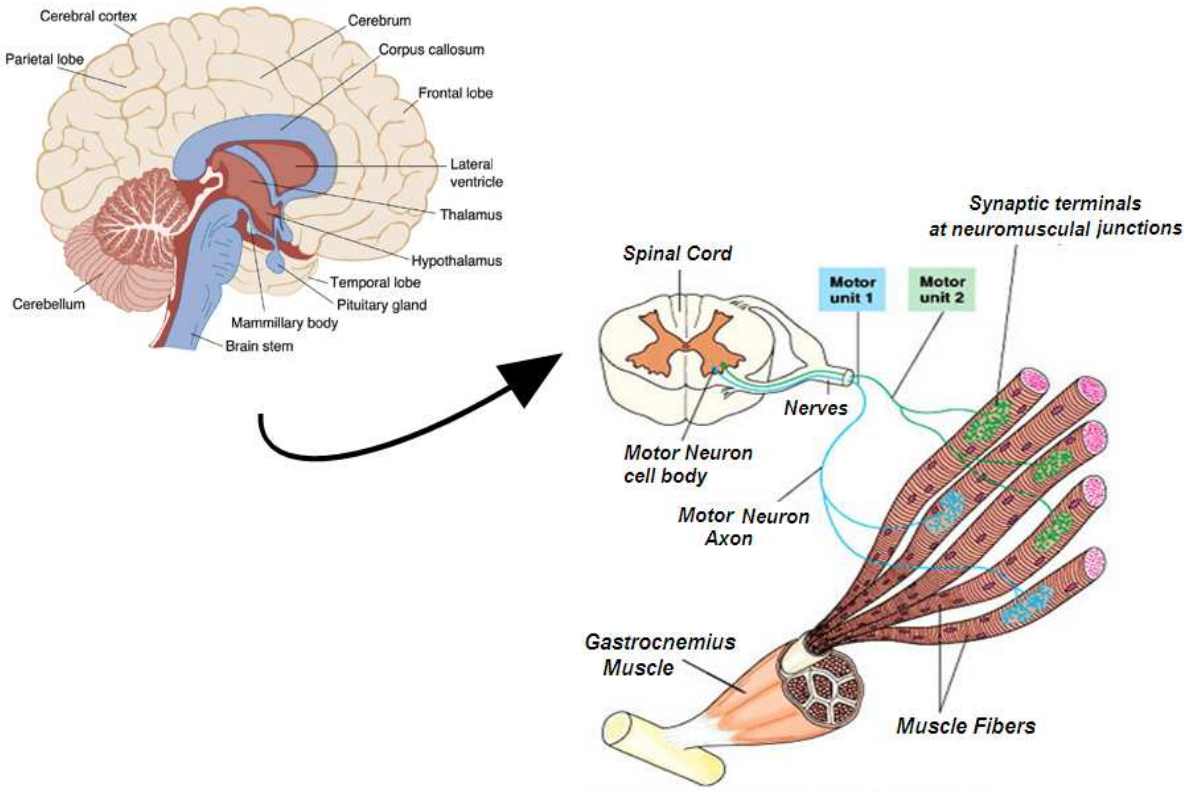


Figure 1- 14. The Nervous System Controlling Human Movement (© binhasyim.wordpress.com)

The nervous system consists of two components, the central nervous system (the brain and the spinal cord) and the peripheral nervous system which is responsible for controlling and coordinating all the functions of the body. The motor cortex, the primary responsible for starting movements, receives and processes information and impulses from peripherical nerve cells and sends back instructions and signals to muscles. Three types of nerve cells or neurons, sensory neurons, motor neurons, and inter-neurons, are important in regulating the signals between the muscles and tendons and the brain and spinal cord.

When muscles are stimulated upon receiving a signal, they contract. This signal may be voluntary stimulus that the muscle receives from the brain in response to a person's desire, a reflex, or an involuntary stimulus. Muscles work usually in harmonious collaboration responding to central nervous system's recommendations to achieve the desired movement.

When neurological responses or joint movements are altered, the entire structure is compromised which influence the growth and development of the skeleton (Figure 1-15).

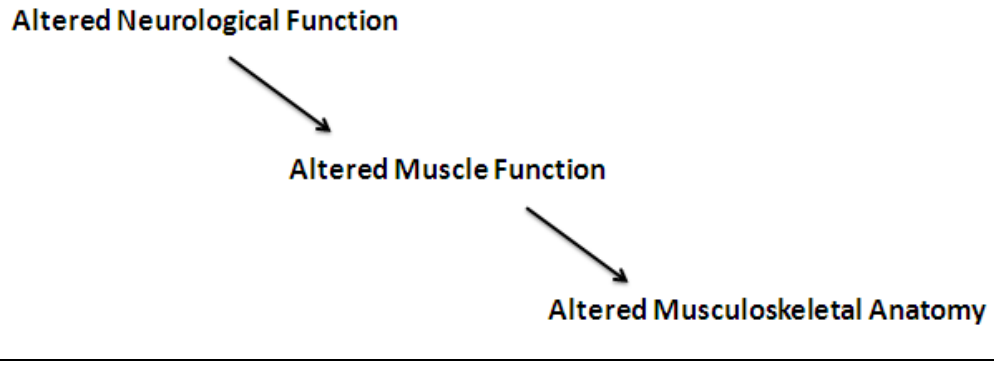


Figure 1- 15. The Basic Sequence of Altered Brain Function (© Gage 2010)

1.2. Normal gait

Since it is primordial to have a look into the characteristic of the normal gait and the history of walking maturity in order to understand pathological gait in young children, this section presents different gait characteristics and gait maturity process.

1.2.1. Gait characteristics

Bipedal gait is the specificity of the human and it is the fundamental system of human locomotion. It is a complex activity requiring a good motor control to ensure smooth lower limb motion and stability.

Walking is a repetitious pattern of lower limb movement resulting from the periodic leg movement moving each foot from one position of support to the next. It is a symmetric, cyclic and three-dimensional activity, but, most of the movements occur in the sagittal plane. Because of its cyclic nature, the description of walking is provided by the repetitive basic unit defined as the gait cycle or stride, which represents the period of time between any two identical events in the walking cycle. The initial contact with the ground, or heel strike, is usually considered as the starting and ending event.

Normal gait has five attributes or prerequisites, which are: stability in stance phase, sufficient foot clearance during swing, appropriate swing phase prepositioning of the foot, an adequate step length and energy conservation in order to maintain balance during smooth and painless body motion. According to Anderson et al. (2001), the normal gait with a comfortable gait velocity is assumed to be the most efficient in terms of energy consumption. Gait prerequisites have to be acquired during childhood maturity but they are frequently lost in pathological gait.

1.2.2. Gait Cycle

The human gait cycle (GC) has been divided in two primary parts: **stance phase,** the time when the foot is in contact with the ground, constituting about 60 percent of the gait cycle and the swing phase, which denotes the time when the foot is in the air, constituting the remaining about 40 percent of the total cycle (Figure 1.16).

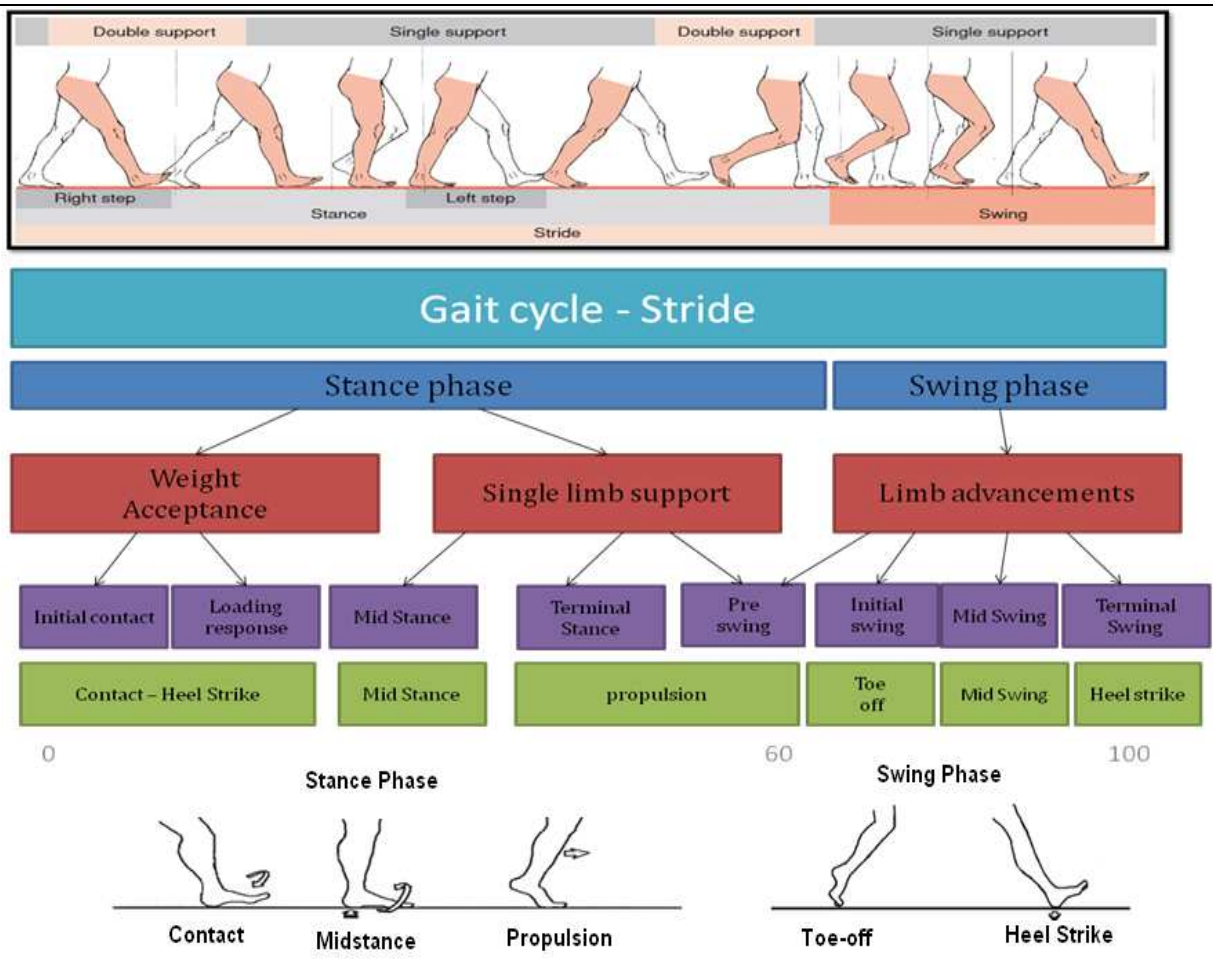


Figure 1- 16. The Gait Cycle (© Bérard C. 2008)

The stance phase is subdivided into three intervals according to the sequence of ground contact. The first period of double support (0%-10% GC), occurs immediately after the initial contact when the heel touches the floor (0%-2% GC) and continues until the toe-off of the second foot, representing the loading response (2%-10% GC). It represents the period when the shock of the impact is absorbed by quadriceps contraction and the body is stabilized for a single stance support. The single stance lasts about 40% of GC. The mid stance (10%-30% GC) represents the body progression beyond the supporting foot and ensure the limb and trunk stability. The terminal stance (30%-50% GC) begins with the heel rise and ends with the initial contact of the second foot (contralateral limb). The stance phase ends with a second double support period, called also the pre-swing period (50%-60% GC) which represents a loading phase of the swing limb and ensures the body weight transfer from the stationary foot to the other. The muscles that are active during the stance phase include the dorsiflexors and plantar flexors, the quadriceps femoris, the hamstrings, the hip abductors and the gluteus maximus (Figure 1.17).

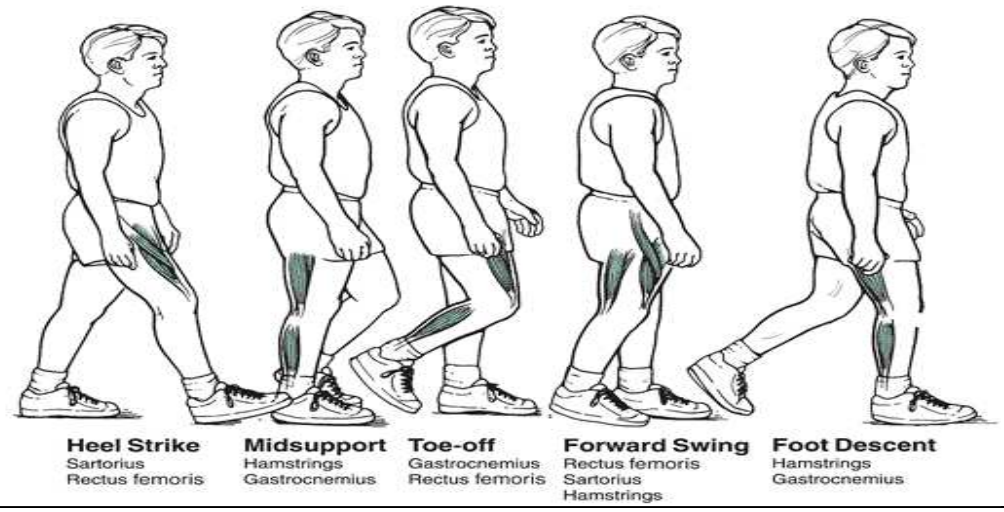


Figure 1-17. Illustration of Muscle Activities During a Gait Cycle (© Hamill et al. 2009)

The toe off defines the beginning of the swing phase, generally divided into three sub-phases. The initial swing (60%-73%) represents the period of limb advancement and foot clearance. The mid swing occurs from 73% to 87% of GC, and ends when the swing limb is forward and the tibia is vertical. The final period is the terminal swing (87%-100% GC) representing the deceleration of the foot movement preparing to the next heel strike. It is controlled by the hamstring and dorsiflexion muscles.

The progression over the supporting foot is divided into three functional rockers: the heel rocker, the ankle rocker and the forefoot rocker occurring respectively during the loading support, the mid support and the terminal stance periods.

1.2.3. Spatio-Temporal Gait Measurements

Walking activity can be also characterized with spatio-temporal parameters visualized using foot prints. The temporal parameters are: stride time (time between initial contact of one limb with the ground and the next initial contact of the same limb), the step time (time between initial contact of one limb with the ground and the initial contact of the contralateral limb), the cadence (number of stride or steps per minute) and the gait velocity. Spatial parameters are step length and stride length, which represent respectively the distances covered during their respective times.

1.2.4. Gait maturity

Independent and mature gait is the major motor development task during the first two years of child's life. Walking behavior's development passes through several postural changes during which the child gains the motor control necessary first to assume and to maintain an upright posture, and finally to walk independently (Figure 1.18).

Figure 1- 18. Walking behavior's development (© www.growthgraph.com)

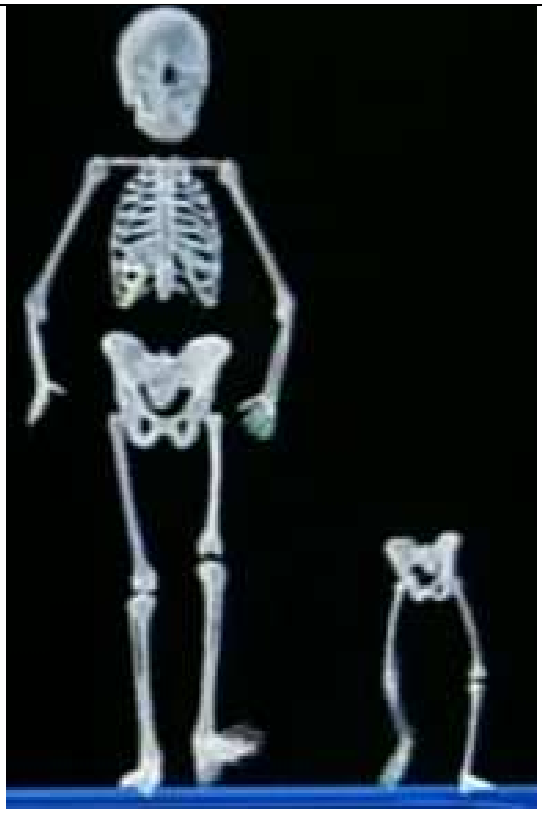


Figure 1-19. Motion Capture for child walking vs Adult walking
(© Thierry Berrod 2010 – from Documentary film “du bébé au baiser”)

Walking usually starts at about one year old; initial efforts at walking are usually characterized as stiff legged and jerky. In earliest gait, the child walks with relatively stiff knees, a wide base of support with feet relatively far apart and pointed outward and outstretched arms for balance (Figure 1.19). As walking matures after two years of learning, at least three year and a half of age, the child develops balance and equilibrium to reach a stable adult gait patterns. The base of support gradually narrows and the feet are placed within the lateral dimensions of the trunk and an adult heel toe gait takes place. Arm movements gradually become synchronous with the walking stride [*Sutherland 1980, Sutherland 1988, Malina 2007*].

The independent walking does not indicate the achievement of the mature walking pattern. The mature process brings stabilized gait at about four years old. By about five years of age, the adult walking pattern is established for the majority of children. However, the stride dynamics are variable among children and vary with walking velocity. In initiated walking, all spatio-temporal parameters increase and movements show greater reproducibility as the walking pattern becomes more like an adult pattern. Sutherland (1980), Holf (1996) and

Vaughan (2003) pointed out that these stride dynamics, presented as dimensionless gait parameters, are invariant after 80 months of ages, which show evidence of both central nervous system maturation and growth (Sutherland, 1997). Neuromuscular maturity is gradually established and the mature walking is progressively attained. The adult-like dynamic joint angles and kinetic patterns for the hip and knee were attained by approximately 5-7 years of age (Figure 1.20), whereas adult-like ankle patterns were not achieved until nine years of age or older [Sutherland 1997;, Cupp 1999, Ganley 2005, Victoria 2007, Viel 2000].

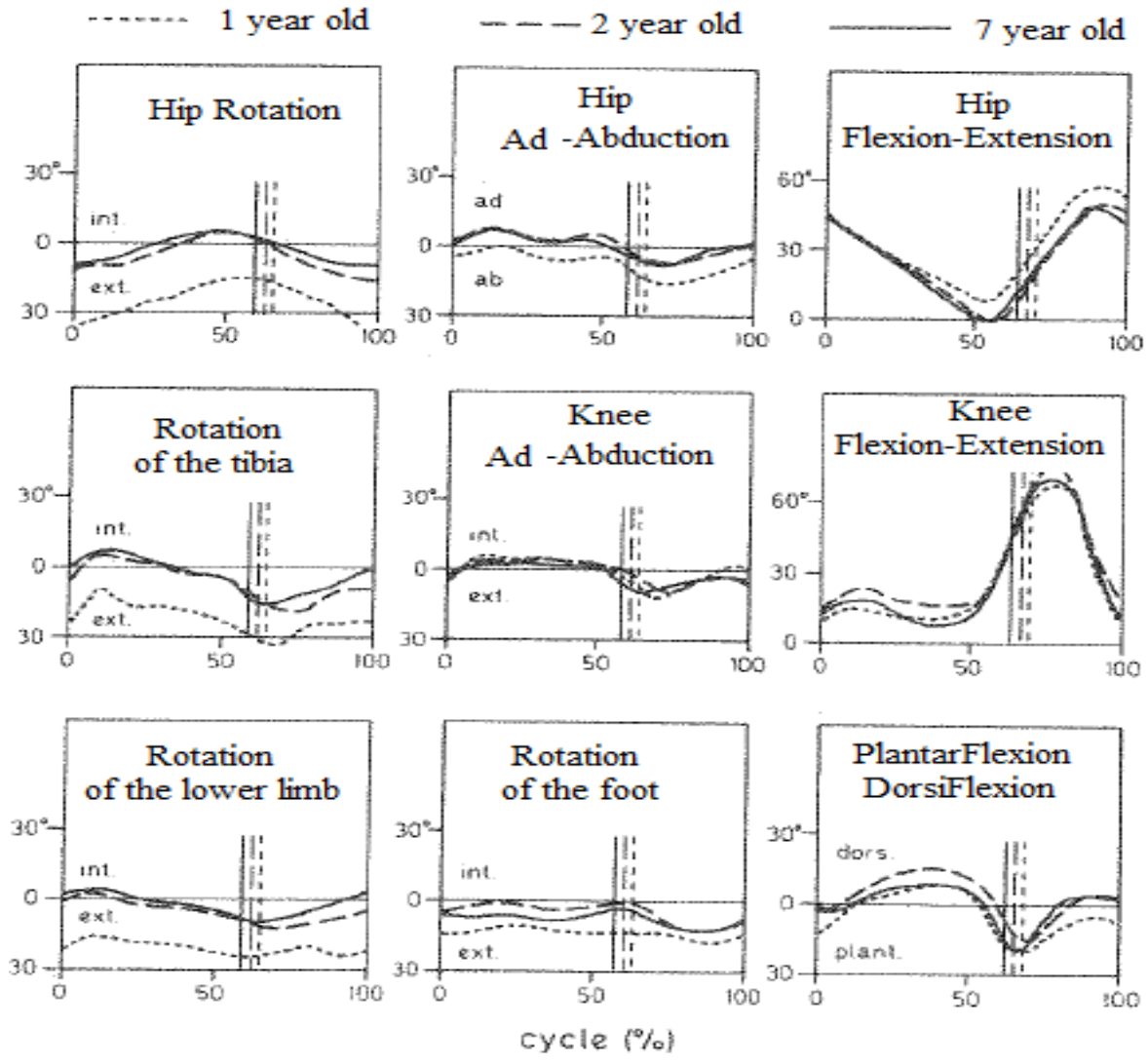


Figure 1-20. Joint kinematics for healthy children of one, two and seven years old (© Viel E. 2000)

During childhood, the central nervous system and musculoskeletal development simultaneously progress. Therefore, it is important to understand the natural history of walking's maturity in order to detect and then interpret pathological gait in young children. In children with neurological impairments, the maturity process is altered and progressively delayed because of the development of musculoskeletal malformations [*Johnson 1997, Katharine 2002, Forssberg 1992, Bell 2002*].

1.3. Cerebral Palsy

1.3.1. Definition

Cerebral palsy, a range of non-progressive syndromes of posture and motor impairment, is a common cause of severe physical disability in childhood. Nowadays, it is estimated that about 764,000 children and adults manifest one or more of the symptoms of cerebral palsy in the United States, about 650.000 persons in Europe and 125.000 persons in France [*Seuret 2007*]. Currently, about 8,000 babies and infants are diagnosed with the cerebral palsy each year. The worldwide prevalence and incidence of the disorder are not clearly known. It is about 0.6 - 4 per 1000 live birth yearly, with variability rates between girls and boys [*Koman 2004, Himmelmann 2006; Seuret 2007, Bache 2003, Cans 2002, Winter 2002, Mongan 2002, Merberg 2004, Jessen 1999, Dolk 2006, Hagberg 2001, Colver 2000]*.

Defining the cerebral palsy was challenging over years. Since 1843, several definitions of cerebral palsy (CP) have been proposed in literature [Cans 2000, Blair 2005, Stacey 2005, Bax 2005] and a universal definition is established by 2005. Subsequently, the Cerebral Palsy (CP) is defined as "a group of permanent disorders of the development of movement and posture, causing activity limitation, that are attributed to non-progressive disturbances that occurred in the developing fetal or infant brain. The motor disorders of CP are often accompanied by disturbances of sensation, perception, cognition, communication and behavior, by epilepsy and by secondary musculoskeletal problems" [*Bax 2005*].

The static alteration of brain function can include loss of selective motor control, abnormal muscle tone, imbalance power between agonists and antagonists and impaired balance and coordination mechanisms which increase over time. When altered tone, power and control are

imposed on the growing child's muscles and bones, the clinical expression of this pathology is subjected to change as child matures and grows, leading progressively to musculoskeletal or orthopeadic problems such as muscle/tendon contractures, reduced muscle elasticity, reduced joint range of motion (ROM) and disturbed bone and joint development [*Koman 2005, Stacey 2005, Soo 2006, Garne 2007, Penneçot 2009, Gage 2010]*.

1.3.2. CP Clinical forms and Classification

Subjects with CP show a wide variety of symptoms that may differ both in type and severity, depending on the magnitude and location of the brain damage. The severity ranges of CP may involve the whole body and lead to a complete inability to control the movement and to walk. There are many classifications of the cerebral palsy syndromes, taking into consideration the quality of the movement disorder and the topographical distribution of the affected area [*Murphy 2003*].

According to the topographic distribution of limb involvement, classification of CP leads to three principal groups: Hemiplegia characterized by the involvement of one side of the body and usually the arm is more affected than the leg, Diplegia in which both lower limbs are severely affected and Quadriplegia which describes the case when all four limbs and the trunk are involved.

For Subjects with CP, the quality of muscle tone and involuntary movement are evaluated function of the location of the brain injury. According to the type of movement disturbance, CP subjects may be classified as spastic, athetoid, ataxic or a mixted CP (Figure 1.21).

The most common type of cerebral palsy is **spasticity** which represents nearly 80 percent of all cerebral palsy cases. Children with spastic cerebral palsy have stiff and jerky movements caused by the stiffness of the muscles and their permanent muscle contraction, which limit movement. They often have a hard time moving from one position to another. The "spasticity" is defined as a clinical condition in which certain muscles are continuously contracted, causing stiffness or tightness of the muscles. It may be associated with spinal cord injury [*Lance 1980, Crenna 1998*].

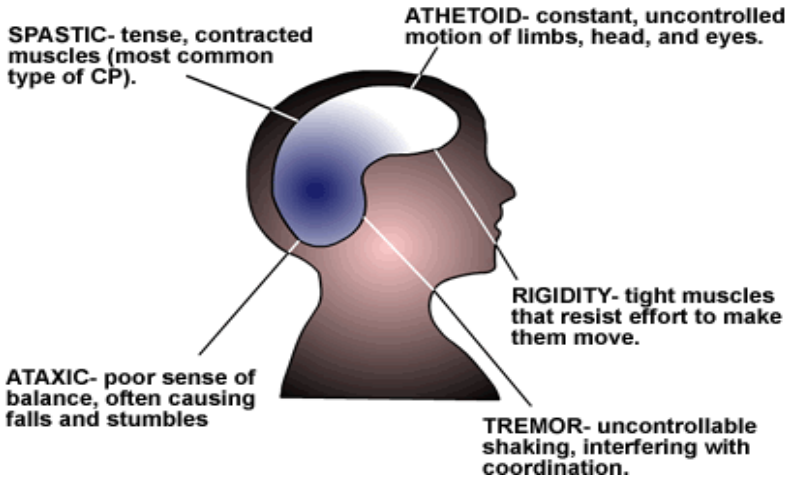


Figure 1- 21: The Location of the Brain Damage and Relative Cerebral Palsy Types:
a) Damage of cerebral cortex is generally the cause of Spastic Cerebral Palsy, b) Damage in the basal ganglia in the midbrain (cerebellum) causes the Athetoid Cerebral Palsy, and c) Damage of the cerebellum causes the Ataxia Cerebral Palsy. (© livingwithcerebralpalsy.com)

1.3.3. Cerebral Palsy and Gait

Cerebral palsy is difficult to diagnose during early infancy. As the infant matures, poor development, weakness, spasticity, or lack of coordination becomes noticeable. Early signs of cerebral palsy usually appear before 18 months of age, specifically abnormal development of motor skills including rolling over, sitting up, crawling, talking, and walking which obviously occurs in normal children at 12 months. However, most children with CP can be confidently diagnosed by 18 months [*Murphy 2003*].

By the age of eight, most patients with CP reach the plateau of motor control development and then a mature and independent walking [*Evans 1995*].

Disorders caused by the brain injuries in the case of cerebral palsy are not static and tend to progress with growth. The primary effects of this neurologic impairment are the altered motor control and abnormal muscle tone usually due to spasticity. Because of muscular spasticity and inappropriate muscle activations, cerebral palsy children almost have joint contractures and bone deformities (e.g. equinovarus and Equinovalgus foot, excessive femoral anteversion and spinal deformities). As consequence, the functional abilities of the child with spastic CP are profoundly affected and often deteriorated during childhood growth by the development

of many lower limb bone deformities, called secondary effects, and compensation mechanisms, called also tertiary effects, instates spontaneously helping the child to have his proper gait autonomy. According to Sutherland et al. (1993), the most common gait abnormalities in diplegic CP patients are:

- Jump knee gait, characterized by an increased knee flexion in early stance phase, through initial double support,
- Crouch gait, characterized by an excessive knee flexion throughout the stance phase and frequently accompanied by an increased hip flexion and internal rotation,
- Recurvatum knee gait, described as an increased knee extension in mid-stance and late-stance phase,
- Stiff knee gait, characterized by a decreased range of motion of knee and delayed peak knee flexion in swing phase, hindering foot clearance.

In addition to these gaits representatives of CP, Rodda et al. (2004) and Wren et al (2005) showed that other types of gaits may be observed: intoeing gait (excessive internal foot progression), Equinus gait (insufficient ankle plantarflexion during stance phase, with or without hindfoot and/or forefoot varus or valgus), Excessive hip flexion (a flexed instead of an extended hip in terminal stance) and Excessive internal hip rotation (excessive internal hip rotation with excessive external foot progression).

1.3.4. Clinical evaluation and Management

The treatment of cerebral palsy deals with the management of the impairments and disabilities resulting from brain injury. At present, there is no cure for CP. However, various treatment possibilities are available aiming to establish a normal motor development and function, to prevent of contractures and deformities and improve child's capabilities.

1.3.4.1. Cerebral palsy management

As CP is usually associated with a wide spectrum of developmental disorders, a multidisciplinary approach is most helpful in the assessment and management of such children [*Bose 1975, Sussman 1992, Renshaw 1995, Sharan 2005*]. It may include:

- physical, occupational and speech therapies help children acquiring developmental skills and specific functional tasks and teaching them to lead towards independence in real life settings.
- pharmacotherapy (dantrolene sodium, and baclofen and Botulinum toxin) to / control spasticity or painful spasms.
- orthopedic devices (walkers, bracing or handling a wheelchair) to maintain stable joint positions, stretching muscles and balance.
- orthopedic surgeries (bone osteotomies, muscle lengthening, rectus femoris transfer, etc.) to treat and to prevent serious skeletal deformity and address muscle, ligament, tendon, and surrounding soft tissue contracture.
- neurosurgical procedures (neurectomy and rhizotomy) to control muscle tone and reduce spasticity.

Since the goal of surgical intervention is the correction of a deformity of the musculoskeletal system, the biomechanical factors responsible for the functional gait limits have to be correctly identified. The clinicians have to differentiate between primary deviations caused by the CP and the natural compensatory mechanisms established because of the primary gait deviations. Nowadays, the 3D gait analysis, combined to data from clinical examination, represents a common management procedure for CP.

1.3.4.2. Clinical evaluation

Clinical assessment is used either as a control or as the basis for classification of CP types and included joint range of motion (ROM), muscle strength and selectivity, rotational alignment of extremities, spasticity, and other clinical parameters.

Evaluation of a child with cerebral palsy (CP) needs many numbers of considerations to better understand the orthopedic and neurological impairments confronting the patient. Several information have to be interpreted together, the medical history, a detailed physical examination and functional assessment, clinical gait analysis exam, and consideration of patient's goals [*Bérard 2008, Gage 2010*].

The clinical evaluation of gait of cerebral palsy children occurs within the context of detailed history and a complementary evaluation of walking skills performed through physical examination and motion analysis exam.

a) Medical history

The medical history is an important part of the clinical decision. It includes information regarding birth, developmental milestones informing about maturity, medical problems, functional skills, orthopedic status (deformities and compensations), surgical interventions, physical therapy treatments and medication and may reveal complaints of pain, weakness and instability and give awareness about underlying neurological and musculoskeletal troubles.

b) Physical examination

The physical examination aims at determining the degree of impairment of selective motor control and tightness or strength of isolated muscle groups, to evaluate joint's range of motion during slow static contracture (Figure 1.22), to evaluate muscle tone as a response to a passive stretch which gives information about muscle spasticity or rigidity (Figure 1.23), to estimate bone deformities (the degree of genu valgum or varum, femoral anteversion, bone torsions and foot deformities) and finally evaluate proprioception skills.

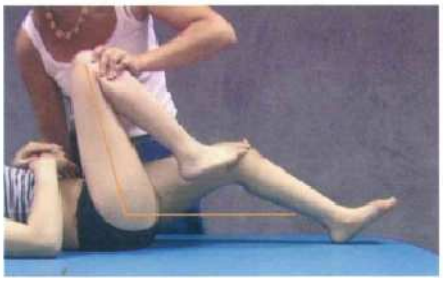
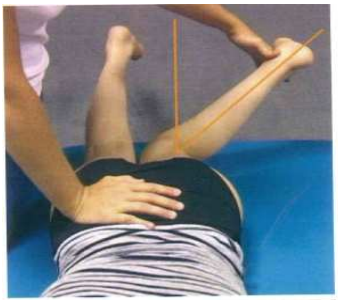
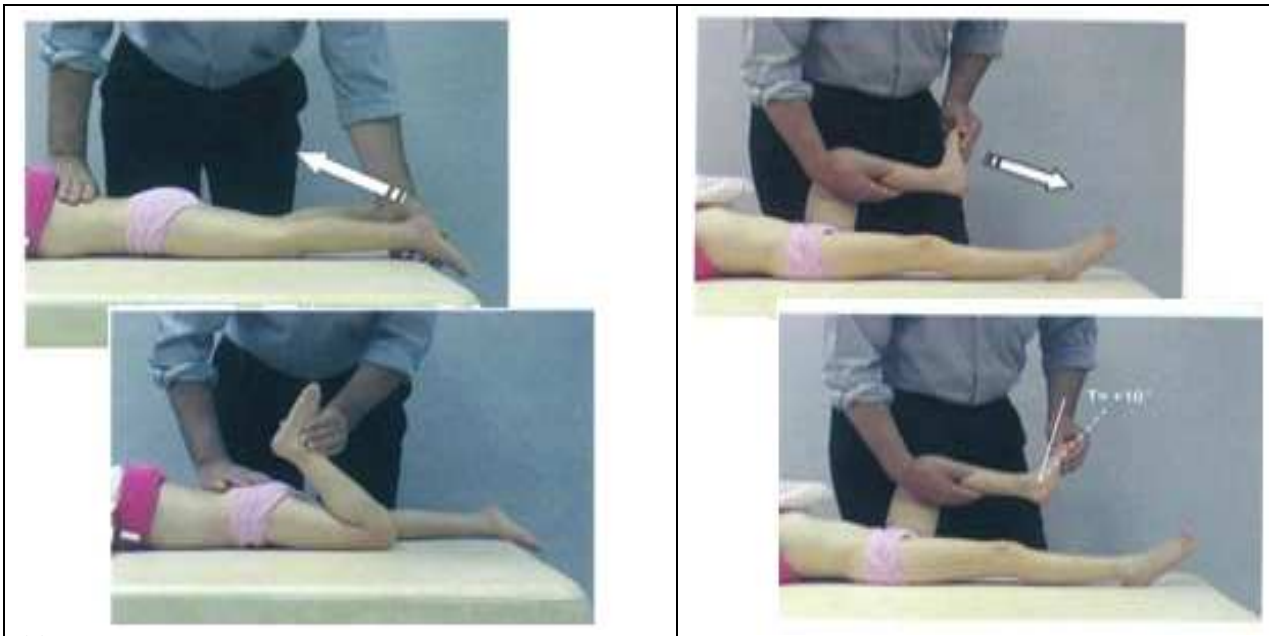
(a)	(b)	(c)
		

Figure 1- 22. Illustration of Physical Exam for Cerebral Palsy Children to Measure the Joint ROM: Maximum hip flexion measurement (a); Internal rotation of the hip measurement (b); Maximum knee flexion measurement (c) (© Bérard 2008)



(a) (b)

Figure 1- 23. Illustration of Physical Exam for Cerebral Palsy Children to Evaluate Muscle Spasticity or Rigidity: Evaluation of the rectus femoris spasticity(a); Evaluation of dorsiflexors spasticity (b) (© Bérard 2008)

c) Gait analysis exam

The gait analysis is a measurement tool providing information about individual's walking skills of patients with specific gait-related problems and its deviation from the normal patterns. It is principally used in treating subjects with neuromuscular disabilities (Figure 1.24). This procedure takes part of the gait abnormalities understanding and becomes an indispensable tool for treatment decision-making and therapy in CP children. With static and dynamic studies, this exam gives the clinician the opportunity to better understand the pathology (which joints and muscles are involved) and then separate the primary causes of a gait abnormality from compensatory gait mechanisms.

A complete understanding of joint kinematics is important in the diagnosis of joint disorders, treatment decisions and when studying locomotion. The fundamental information obtained from gait analysis usually include spatiotemporal parameters, such as velocity, cadence, stride and step lengths, the kinematic of the limbs and joint motion, the ground reaction forces, the calculations of the moments and the power production occurring in major of lower limb joints (kinetics) and the dynamic electromyography defining the on-off signals sets of individual muscle or groups of muscles. These data, gathered from gait analysis, can predict which joint

is mostly affected and which type of muscle or tendon would be most managed. Post-operative gait analysis exam could be used to evaluate the success of orthopedic surgery and suggest changes in a person's rehabilitation program.

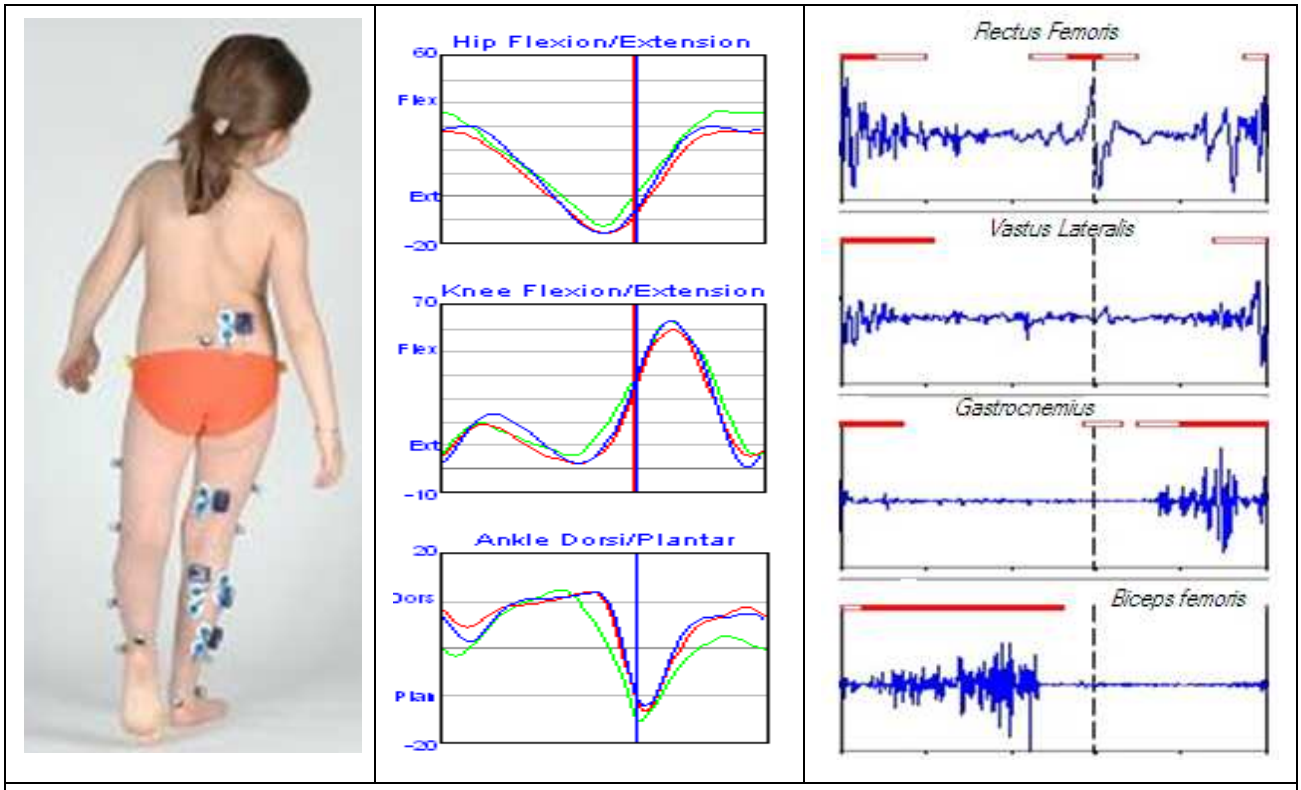


Figure 1- 24. Gait analysis exam and results

Indeed, the Gait Analysis Exam is used to understand and to evaluate the validity of cerebral palsy classifications of gait deviations observed in children with cerebral palsy [*Sutherland 1969, Gage 1994, DeLuca 1997, Davids 2006, Steinwender 2001, Kay 2000, Rodda 2004, Desloovere 2006, Chang 2006, Narayanan 2007, Dobson 2007*], has a potential role in the pre-surgical assessment, can affect treatment decisions [*Gage 1983, Gage 1996, Cook 2003, Wren 2009, Kay 2000, Greene 2000*] and serves as a following up reporting tool. Several studies have assessed how clinical gait analysis improves patient outcomes by evaluating the impact on post surgical orthopedic decisions and treatment planning [*Lee 1992, DeLuca 1997, Kay 2000, Cook 2003, Wren 2005, Molenaers 2006, Chang 2006, Lofterod 2008, Filho 2008*]

Because of cerebral palsy, walking performance is often compromised, leading to pathological, jerky and instable movements that need compensatory strategies. Motor impairment can be clinically assessed, but underlying muscle function cannot be directly measured. In addition to gait analysis, computer simulations have demonstrated that

biomechanical parameters derived from a musculoskeletal modeling (muscle forces, moment arm length and muscle-tendon length, individual muscles' function in the gait performance) have a unique potential to predict treatment outcome [*Delp 1995, Schmidt 1999, Neptune 2000, Jonckers 2003, Arnold 2004, Arnold 2005a-b, Higginson 2006, Reinbolt 2008, Jonkers 2010*].

1.4. Musculoskeletal modeling

Because of the complexity of the human body's structures, biomechanical analysis of gait relies on simplified mathematical representation of the lower limb musculoskeletal system to understand the dynamics involved during movement, the interaction between different biological structures (bone, muscle, joint, etc.) and to estimate biomechanical quantities, such as muscle forces, that are difficult to obtain by direct measurement in vivo.

In musculoskeletal modeling, the morphology and the functional characteristics of bones, joints and muscles are numerically represented with a set of anatomical parameters, defined through cadaveric studies or measured directly using sophisticated imaging techniques.

According to studies' needs, many musculoskeletal models have been developed for the lower limbs to improve knowledge about the normal gait and to study Cerebral Palsy pathological gaits.

1.4.1. Musculoskeletal modeling procedure

1.4.1.1. Musculoskeletal Model Description

In general, musculoskeletal modeling techniques describe the geometry of the musculoskeletal system as multilinked rigid segments rotating around mechanical joints and connected with set of muscles. Muscles are usually defined as straight line spanning between an origin point and an end point attached to segments.

Several lower limb musculoskeletal models have been presented in literature. Many of them relying on rescaled generic models, in which standard geometry data are given from cadaveric studies for healthy male patient and then rescaled using regression equation based on subject's

anthropometry [*Delp 1990, Arnold 1997, Schmidt 1999, Hicks 2008, Desailly 2008, Correa 2011*]. When studying movement abnormalities, generic musculoskeletal models may have several limits on predicted results. More realistic body's morphology is very decisive and challenging. Therefore, recent studies have developed more sophisticated techniques, based on clinical imaging, to determine subject-specific geometry [*Schyes 2008, Dao 2009, Oberhofer 2010*].

A musculoskeletal model is defined by several sets of elements: skeleton, joints and muscles, as follows.

a) Skeleton model

A lower limb is modeled as several linked rigid body segments: pelvis, femur, patella, tibia and foot segments [*Delp 1990, Arnold 1997, Anderson 2001, Pandy 2001*]. The tibia and fibula are generally defined as single rigid segments. Most of the time, the foot are considered as single segment or divided in sub-segments for talus, calcaneus and toes. Generally, the upper body (Head, Arms and Torso) is represented as a punctual mass at the hip or as an additional rigid segment [*Neptune 2007*]. Each segment is defined by its inertial parameters, located at the center of inertia, required to calculate joint kinetics and generally determined from cadaveric studies [*Winter 1990*], and its attached local frame references (Figure 1.25) [*Wu 1995*].

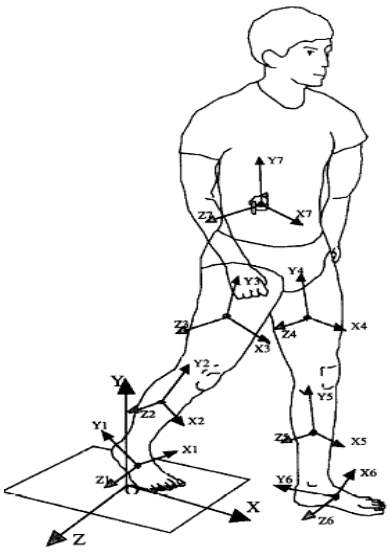


Figure 1- 25. ISB Recommendation of Local Reference Frame and joint Coordinates System (© Wu et al. 1995)

i. Rescaled generic geometric model

The common approach described in the literature is the generic musculoskeletal modeling. Geometry data have been extracted from various cadaveric studies of healthy male adults or medical images, assumed to be representative of unimpaired adults. Segments anthropometry, joint geometry, and muscles attachment sites are, then, determined using scaling techniques to corresponding anthropometric parameters of subjects [*Delp 1990, Arnold 1997, Schmidt 1999, Hicks 2008, Desailly 2008, Correa 2011*].

Lower limb musculoskeletal models have been indispensable to improve understanding of the characteristics of both normal and pathological gait. To take into consideration specificities of gait abnormalities, specific subject impairments have been introduced in musculoskeletal models such as femoral anteversion and tibial torsion deformities for Cerebral Palsy children [*Arnold 2001, Hicks 2007, Scheys 2008, Correa 2011*].

ii. Personalized Geometric Model

As an alternative to generic musculoskeletal modeling, computer tomography (CT) and non-invasive magnetic resonance imaging (MRI) techniques can be used to integrate subject-specific details in the musculoskeletal models (Figure 1.26).

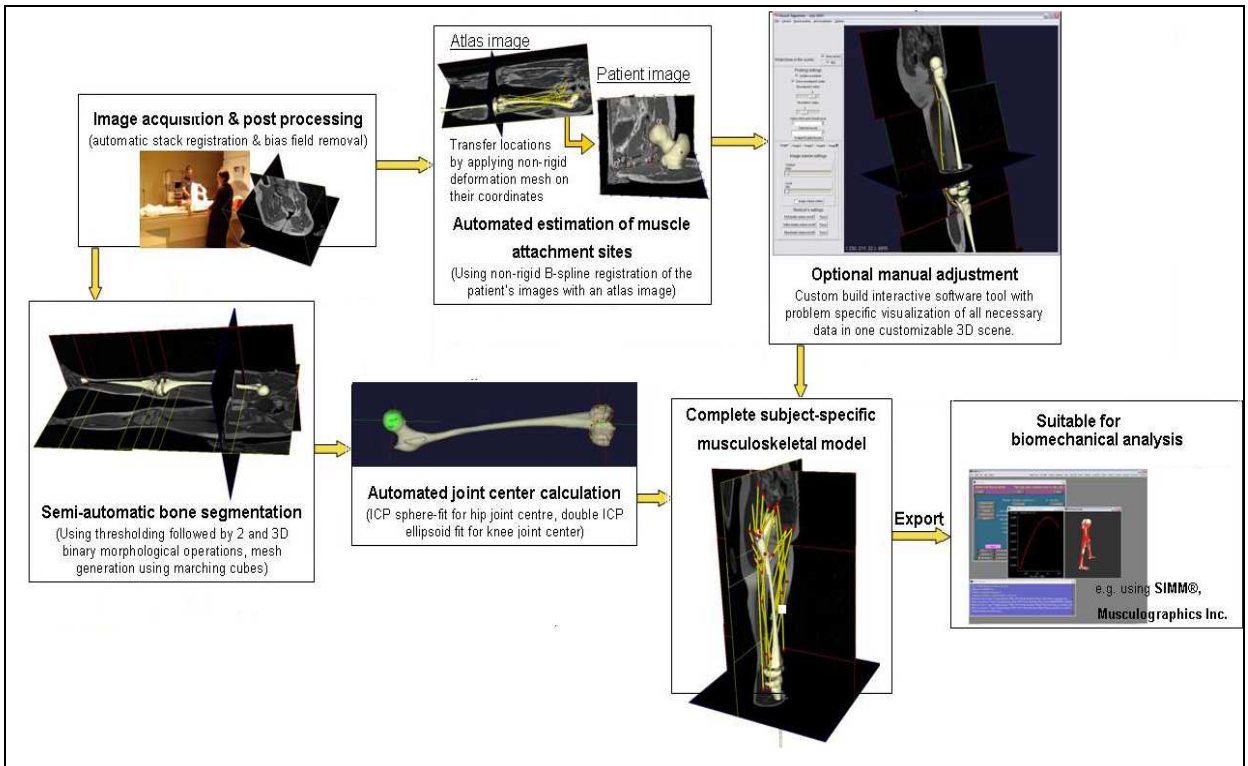


Figure 1- 26. An Overview on Personalization Procedure for MRI-based Subject Specific Musculoskeletal Modeling. (© Scheys et al. 2010)

Using these techniques, in vivo anatomical structures are detected and geometrical model parameters can be accurately determined to describe bone and soft tissues shapes, muscle line of actions and insertion sites, joint locations and bone deformities, allowing more accurate biomechanical analysis. [*Arnold 2000, Arnold 2005, Blemker 2007, Hicks 2008, Scheys 2008, Oberhofer 2010, Dao 2009, Klets 2010, Jonkers 2010, Correa 2011*].

Despite the efficiency of such techniques, MRI-based musculoskeletal modeling remains prohibitively time expensive to be implemented in clinical environments because MRI scan is too costly and manually image processing needs [*Blemker 2007, Dao 2009, Scheys 2010*].

iii. Deformable sketeton

Few studies tried to combine the musculoskeletal modeling and the finite element techniques in order to introduce the deformability aspect of bones. Al. Nazer et al., 2008, integrated a flexible tibia in a generic musculoskeletal model of a healthy adult to estimate the dynamic bone strains during walking (Figure 1.27). Shefelbine and Carriero, 2010, determined muscle and joint forces applied into finite element model of deformable femur to determine stresses and strains in the growing femur and then predict changes in bone growth with different pathological gait patterns for cerebral palsy children (Figure 1.28).

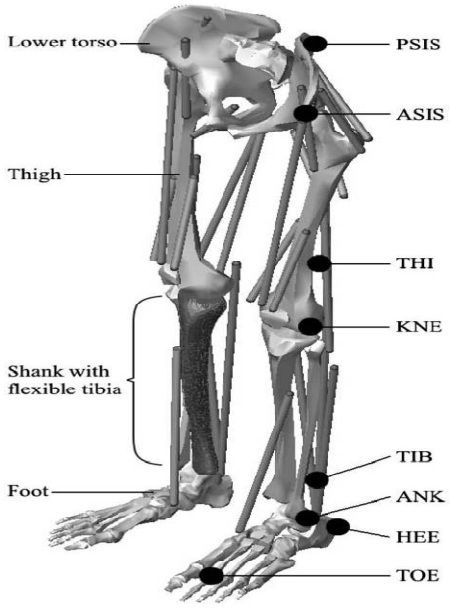


Figure 1- 27. Graphic representation of the lower body musculoskeletal model with a flexible tibia (© Nazer 2008)

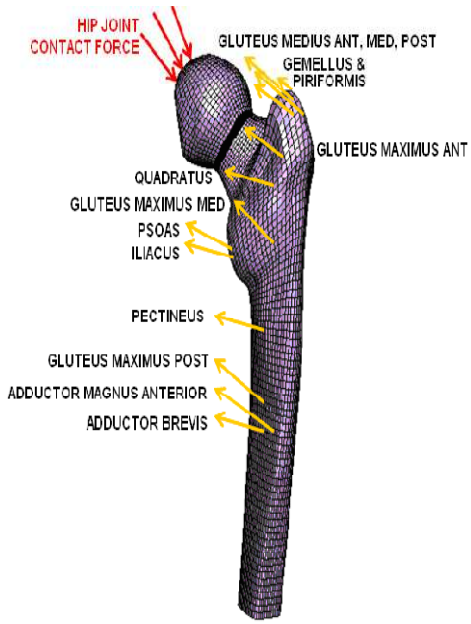


Figure 1- 28. Finite element model of the proximal femur. (© Shefelbine 2010)

b) Joints

Human articulations, defining the position of one segment to another, are generally assumed to be ideal mechanical joints. The hip joint, connecting the pelvic and femoral segment, is modeled as a ball-in-socket joint allowing rotation around the hip joint center. The knee joint, connecting the femoral, tibial and patellar segment, is modeled as a sliding hinge joint, allowing rotation in the sagittal plane accompanied with a small translation through the knee joint anatomical axis. The ankle joint, connecting the tibia and the foot segments, is modeled as a hinge joint allowing flexion/extension movement in the sagittal plane.

Kinematic functions, relating the rotations and translations to the generalized coordinate, are attributed for each joint. Generalized coordinates consists of a set of angles that describe the joint position relative to the neutral upright posture (Figure 1.29).

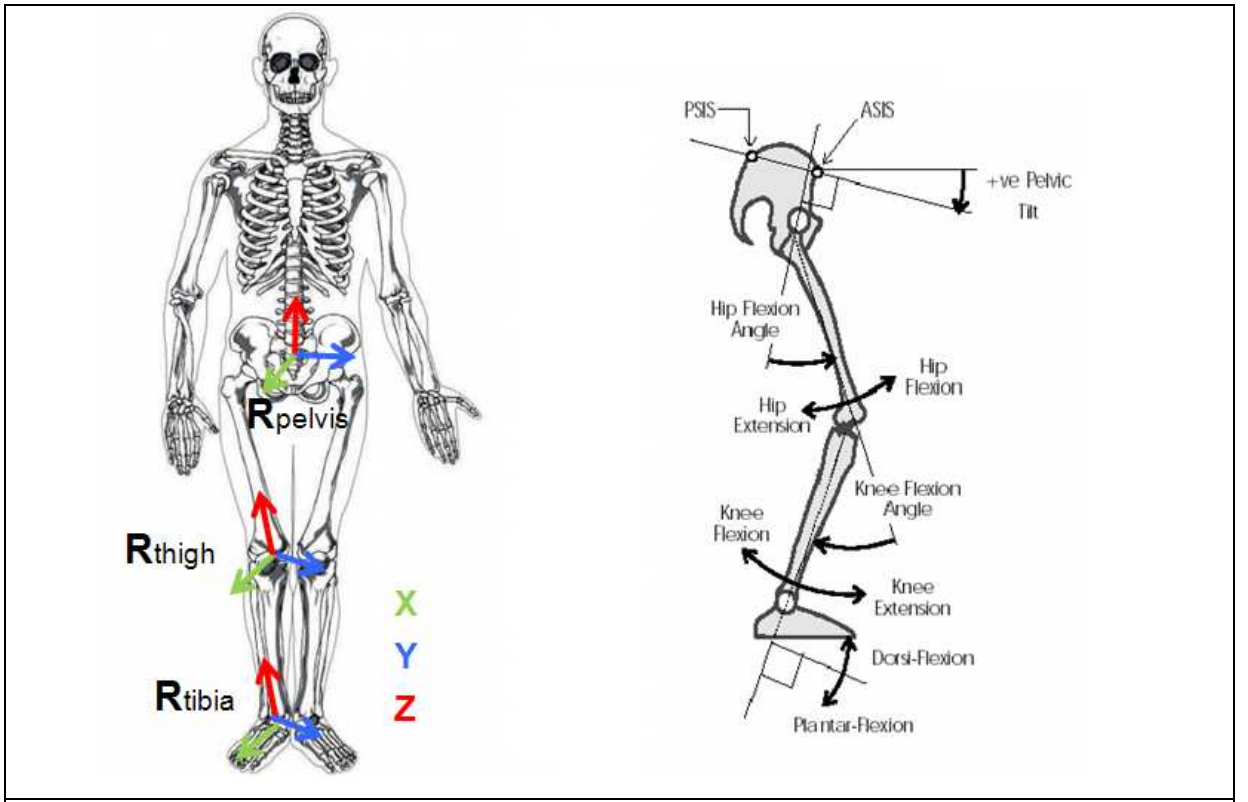


Figure 1- 29. Conventional Joint Angles Definition

i. Joint Center Localization

The location of the joint centers and axis of rotation is required when defining location and orientation of adjacent bones, and calculating then joint torques, muscle lengths and lever arms of muscular forces.

The accuracy of joint center location is crucial and challenging. Sensitivity studies have shown that a small mislocation of joint centre of rotation can lead to errors in joint kinematics and kinetic calculations and error propagation to the entire lower limb [*Kadaba 1990, Holden 1998, DellaCroce 1999, Kirkwood 1999, Stagani 2000, Piazza 2000, Piazza 2001, Donati 2006, Begon 2007, Harrington 2007, Cereatti 2007*] and may remarkably alter the estimation of muscle moment arms and muscle moments and forces [*Delp 1992, Pierce 2005, Lenaerts, 2009, Oberhofer 2009*].

Several methods have been developed for estimating joint center positions, which can be classified into two categories: the predictive and the functional methods.

- **Predictive Methods**

The predictive method estimates the joint center location based on regression equations derived from anthropometric measurements. Regression functions have been obtained by direct cadaveric measurements (*Seidel 1995*) or by using imaging techniques (*Bell 1989, Davis 1991, Kirkwood 1997*) obtained from healthy adult populations. Several regression equations have been developed to estimate the hip joint center combining an initial morphological estimation of joint center position with kinematic data gathered from gait trials to improve the accuracy of regression coefficients through iterative optimization techniques [*Frigo 1998, Shea 1997, Bruening 2008*].

Even if predictive methods gave acceptable joint center estimation for healthy adults (*Bell 1990, Leardini 1999*), they reported an error about 40 mm when estimating the hip joint center (HJC) for normal children and 85 mm for CP Children when compared to measurements obtained from imaging techniques [*Fieser 2000, Jenkins 2001*].

- **Functional methods**

The inaccurate determination of the hip joint center (HJC) leads to erroneous gait analysis results [*Cappozzo 1984, Delp 1992, Stagni 2000*]. For that reason, the majority of studies have focused on estimating the hip joint center using functional approach [*Halvorsen 1999, Gamage 2002, Halvorsen 2003, Piazza 2004, Schwartz 2005, Camomilla 2006, Begon 2005, Hicks 2005, Ehrig 2006*]. Some of them focused on determining the accurate axis of rotation of knee [*Holzreiter, 1991, Cheze 1998, Piazza 2000, Marin 2003, Most 2004, Rivest 2005, Schwartz 2005, Siston 2006, Ehrig 2007, MacWilliams 2008*] and ankle [*Siston 2005, Lewis 2009*].

Originally, Cappozzo (1984) assumed that the HJC is the pivot point of a relative movement between the femur and pelvis. From this assumption, the functional methods represent the process of fitting spherical movements of a set of markers related to two adjacent bones rotating around a specific joint center.

The functional approaches have been developed through experimental studies using mechanical analog of socket mechanical joint (*Piazza 2001, Marin 2003, Siston 2006, Camomilla 2006, MacWilliams 2008*), mathematical approach developing several new and more accurate algorithms (*Halvorsen 1999, Gamage 2002, Halvorsen 2003,* Cereatti *2004, Begon 2005, Camomilla 2006, Ehrig 2006, Desailly 2008*).

The accuracy of joint center estimated by the use of functional methods requires specific cluster of markers placed on adjacent bones and the analysis of several movements performed around the joints (flexion-extension, adduction-abduction and circomduction) (*Piazza 2001, Camomilla 2006, Begon 2007*) with specific range of motion and velocity (*Holden 1998, Begon 2007*) during standing posture, seated posture and walking trials.

Several comparative studies between predictive and functional methods have been presented and showed that functional approach provides more accurate estimation even with a limited range of motions [*McGibbon 1997, Leardini 1999, Besier 2003, Christopher 2003, Lopomo 2010*]. However, that performance could be strongly related to the implemented optimization procedure [*Piazza 2004, Camomilla 2006, Donati 2006, Ehrig 2006*].

Functional methods still have some limitations because they are highly affected by the skin artifacts (*Cereatti et al. 2004*) and because they require significant joint motion which may not be easily performed by pathological patients. But, they remain easy to use compared to imaging techniques [*Harrington 2007, Scheys 2008, Lanearts 2009, Peters 2010*]. Nowadays, the predictive methods derived from Bell et al. (1990) and Davis et al. (1991) are the most widely used in clinical applications and surgical planning.

ii. Elastic Joint Parameters

The joints have spring like behavior and can be modeled as torsional spring elements [*Davis 1996*]. The dynamic joint stiffness has been detected as an important parameter of joint's clinical evaluation.

According to Hooke's law, the dynamic joint stiffness is defined as the gradient of the joint torques - joint angles graph (Figure 1.30) and represent the resistance that muscles and other joint structures manifest during intersegmental displacement and as a reaction to an external moment of force [*Davis 1996*].

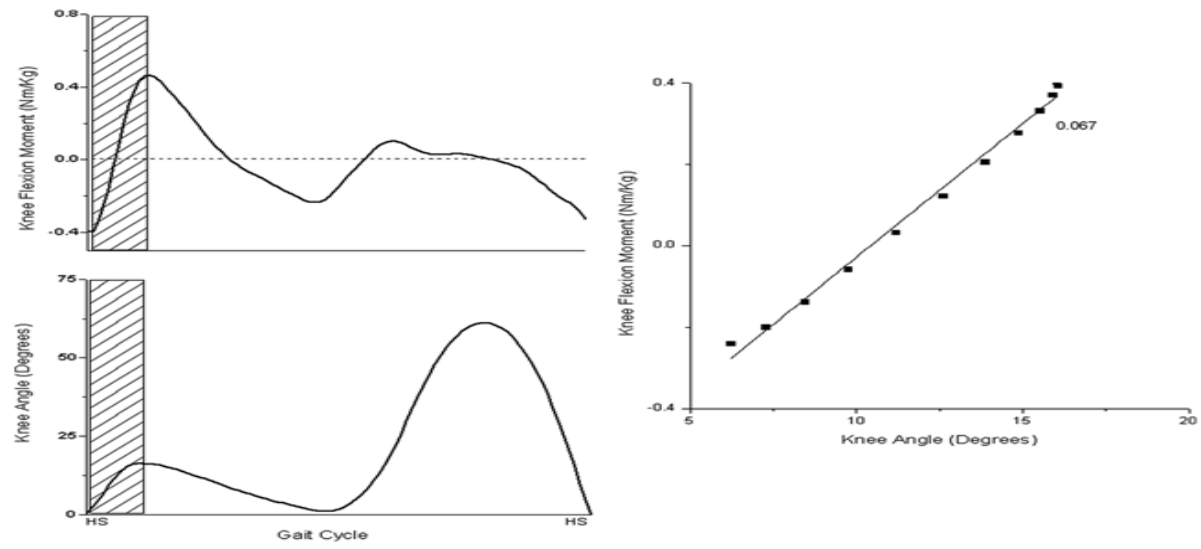


Figure 1- 30. Dynamic knee stiffness calculated as the slope of the linear regression line of the graph representing the knee joint moment as a function of knee joint angle (© Zeni et al. 2009)

The dynamic joint stiffness has been analyzed over linear region during loading response when muscle activities were assumed to be quasi-constant. For the knee, dynamic joint

stiffness was analyzed over the linear region during loading response, corresponding to 3%–15% of the gait cycle. The hip and ankle joint stiffness have been determined during the second rocker, corresponding to 10% - 30% of gait cycle [*Davis 1996, Frigo 1996*]. The joint damping is usually fixed to 10% of stiffness values.

The dynamic joint stiffness have been explored as a clinical parameter to understand pathological cases such Cerebral palsy (*Davis 1996*), Down syndrome (*Galli 2008*), knee osteoarthritis (Zeni 2009), total ankle replacement (*Houdijk 2008*).

c) Muscle model

Muscles are the actuators of skeletal system transferring neural signal into mechanical forces through chemical transformation. Developing a realistic muscle model is still challenging purpose. A complete model, Cross-bridge or Huxley Model (Huxley 1957), taking into consideration these specificities has been developed. Such a model, described through several differential equations, is generally computationally prohibitive and difficult to be integrated in musculoskeletal models.

The muscle's model basically used is the Hill-model (Hill 1938) represented by three components acting together in a manner that describes the viscoelastic behavior of a whole muscle (Figure 1.31). The contractile component is the element of the muscle model that converts the stimulation of the nervous system into a force and reflects the shortening/ lengthening of the muscle. The parallel elastic element (PEE) represents the passive properties of the muscle and the series elastic element (SEE), is a highly nonlinearly elastic structure, represents primarily tendon.

Force characteristics of a muscle depend on both its architecture and its intrinsic properties (force length, force-velocity relationships, architecture). Even if the Hill-muscle model is not a very detailed description, it requires only a few parameters to give an accurate prediction of the mechanical nature of muscle. Personalizing required muscle properties (peak isometric force, corresponding optimal muscle fiber length, pennation angle, tendon slack length, and maximum shortening velocity) is challenging. Parameters values are generally taken from cadaveric studies of elder healthy subjects [*Winter 1990, Delp 1990*] and then scaling

techniques are used to match subject-specific measurements. Alternatively, ultrasonography and magnetic resonance imaging techniques can also be used to collect patient-specific muscle-tendon properties: muscle origins/insertions sites and moment arms [*Schyes 2008, Dao 2009, Oberhofer 2010*].

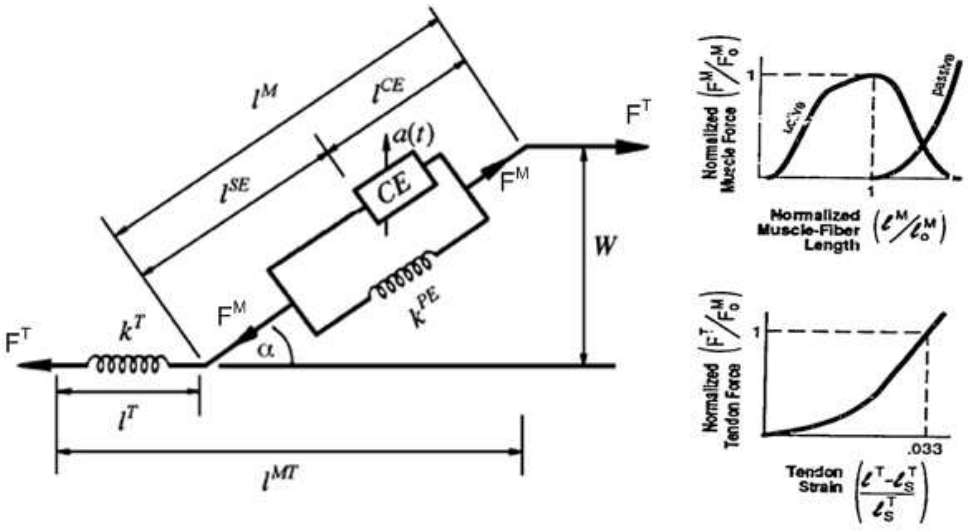


Figure 1- 31. The musculo-tendon actuator model - the Hill Model (© Delp 1990)

More recent studies have been focused on modeling the dynamic of muscle by determining the relationship between muscle forces and EMG measurements [Cao 2009, Staudenmann 2010].

1.4.1.2. Simulation procedure

From a mathematical point of view, deriving a musculoskeletal model consists of solving a set of dynamic equations of motions (Newton, Lagrange) described below. The acceleration, the joint angle, the center of gravity, the foot force, the joint moment, the muscular force, the transmitted force at the joint, the electric activity of the muscle, the power generated by the leg and energy expenditure in walking are calculated from these dynamic equations.

The Newton second law is represented in the following equation $\sum \vec{F}_{external} = M * \vec{a}$ in which $\vec{F}_{external}$ is the external force, M is the Mass and $\vec{a}$ is the acceleration vector.
This last equation can be extended as follows:

$$M(q).\ddot{q} + C(q,\dot{q}).\dot{q} + G(q) + R(q).F_{MT} + E = 0 \quad \text{(Equation 1)}$$

where :

$(q, \dot{q}, \ddot{q})$	vectors of the generalized coordinates (angular positions), velocities, and accelerations, respectively		
$M(q)$	the mass matrix	F_{MT}	vector of musculotendon forces
$C(q,\dot{q})$	the centrifugal and Coriolis loading	$R(q)$	the matrix of muscle moment arms
$G(q)$	the gravitational loading	$E(q,\dot{q})$	vector of External forces

Because of redundancy and since the number of unknown variables (muscle forces and moments) exceeds the number of available mechanical equilibrium equations for most human joints, this mathematical representation is considered as an indeterminate problem [*Pandy and Anderson 1998, Ren 2007*]. The most common approach to solve this problem is the use of optimization techniques, which represent the human motor control strategies and assume that human movement occurs when minimizing some performance criteria. To define a normal human walking, a variety of performance criteria have been suggested; minimizing the energy expenditure, joint torques, sum of squares of muscle activations, the risk of damaging the muscle, the oxygen consumption [*Crowninshield 1981, Patriarco 1981, Marin 2000, Pandy 2003,Thelen 2003, Li 2006, Deluca 2009*]. The most frequently optimization criterion describing the human locomotion is the minimum energy expenditure per unit distance traveled [*Anderson 2003, Ren 2007*].

a) Methods

There are two major methods used to estimate the muscle forces: static optimization and dynamic optimization or optimal control theory (Figure 1.32) [*Erdemir 2007*].

In the static optimization approach, the dynamic equations are solved first to calculate the muscle forces, the net forces and torques at the joints from experimental kinematics measurement, called the inverse dynamic simulations. An optimization problem is then applied to resolve the muscle force redundancy at each time step along the movement trajectory [*Anderson 2001, Marin 2000, Pandy 2003*].

In contrast, dynamic optimization is a more powerful approach for estimating muscle forces during movement. In this approach, called the forward dynamic simulation, experimental motion or load inputs are not required. Muscle activations have been used as the inputs to calculate the corresponding motions and because the number of muscles crossing a joint is greater than the number of degrees of freedom specifying joint movement, Muscle forces and associated motion are predicted by solving a single optimization problem for one complete cycle of the movement. The most important inconvenient of dynamic optimization methods is the heavy computational cost comparing to static optimization methods [*Anderson 2001*].

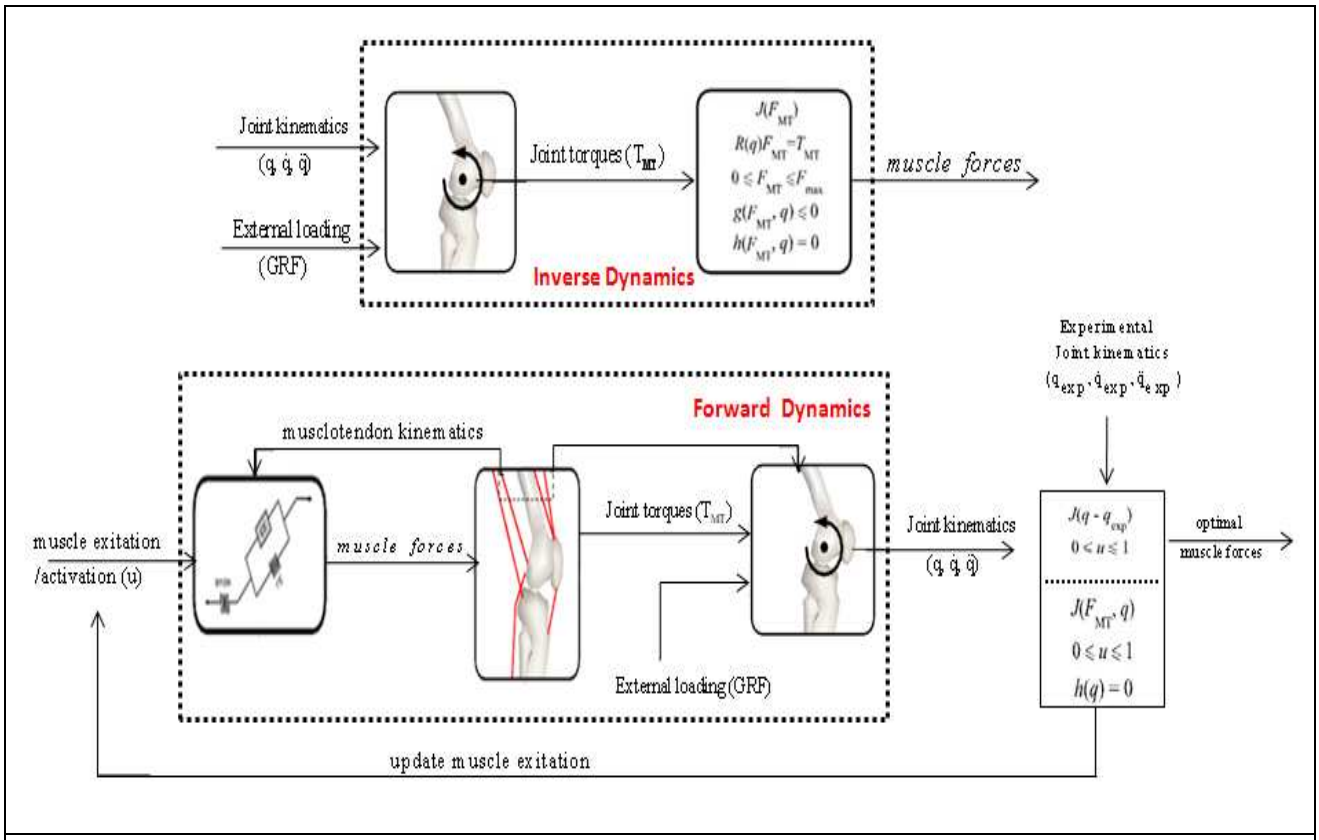


Figure 1- 32. Strategies to estimate the muscle forces: static optimization (inverse dynamics) and dynamic optimization (forward dynamics) or optimal control theory. (© from *Erdemir 2007*)

b) Boundary conditions

To solve dynamic equations of motions and to deal with redundancies, boundary conditions are required to ensure realistic solutions.

i. Physiological Boundary Conditions

A realistic musculoskeletal model is constrained by several physiological constraints that have been incorporated in the dynamic formulation, such as the maximum range of motion allowed by the joint and the intrinsic muscle characteristics (the maximum allowed lengthening and the maximum muscle forces [*Delp 1990*].

ii. Dynamic Boundary Conditions

The numerical solution of the human dynamics, described above, may be constrained to follow a given kinematics. The external force measurements (specially the ground reaction forces) obtained from a motion analysis experiment can also be used as inputs of model and then considered as additional dynamic constraints.

EMG measurements can be used as additional experimental inputs to derive the musculoskeletal model using the patient-specific neural control strategies [*Jonkers 2002, Lloyd 2003, Shao 2009, Bisi 2011*]. The principle drawbacks of this technique (called EMG-driven methods) are:

- Non invasive EMG measurements are limited to the important surfacic muscles
- Deep muscles activities are not take into consideration in modeling
- EMG crosstalk still problematic [*Barr 2010*]

1.4.1.3. Post Processing

a) Expected results

Musculoskeletal modeling allows estimating several parameters such as joint kinematics (joint positions, velocities, accelerations, joint angles, the center of gravity's trajectory), kinetics (estimated ground reaction forces, if not introduced as inputs, joint moments), the muscular activation, lengthening/ shortening histories and quantified forces, the power generated by the joint movements and energy expenditure during the performed motion.

b) Validation

Musculoskeletal modeling attempts to be applied in the clinical routine applications and play a basic role to predict the outcomes of diagnosis and treatment and to investigate the if-then scenarios. Regardless of such application, accuracy, robustness and precision of simulation results are crucial. The greatest obstacle, encountered in musculoskeletal modeling given the number of associated assumptions, is the ability to validate the estimated results.

Muscle force, joint contact forces, loading distribution predictions during a motion are difficult to validate [Heller 2010]. A direct validation can be carried out through in-vivo measurements by using instrumented joint implant to collect experimental joint contact forces (*Stansfield 2003, Kim 2009*), joint implants with built-in load sensors and telemetry (Bergmann 2004), or in vivo instrumentation to measure muscle-tendon forces during human movement (*Finni 2001*). But, these techniques are still considered as invasive [*Ravary 2004, Fleming 2004*].

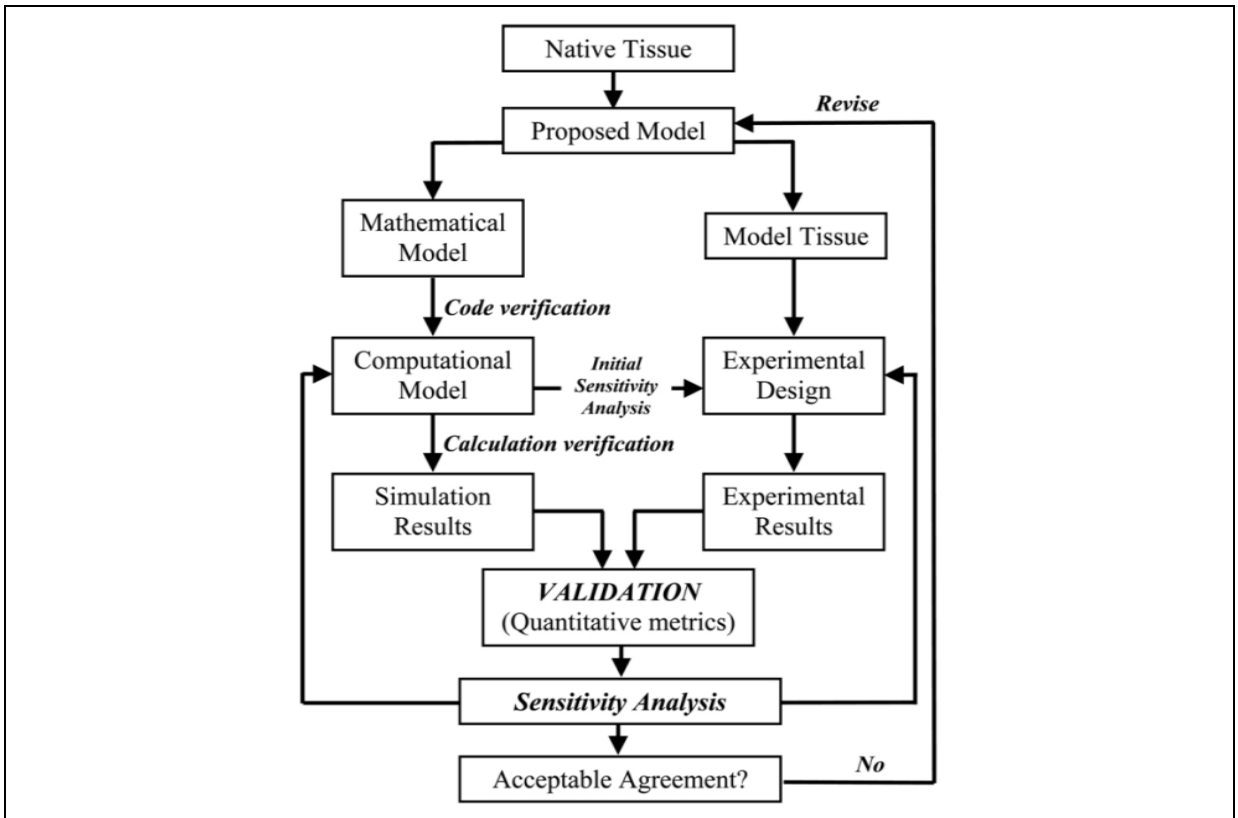


Figure 1- 33. Overview of the Verification and Validation process
(© Anderson 2007)

In musculoskeletal modeling, the validation has been assessed indirectly by quantifying the differences between the experimental (measured) and predicted (simulated) results (Figure 1.33) [*Anderson 2007, de Zee 2010*]. Most often predicted muscle forces are evaluated only qualitatively by comparing predicted muscle activation and measured EMG activation patterns. The Ground reaction forces estimated using a mechanical model for the contact with the ground can be compared to those measured with forces plates. However, the validation procedure is also depending on the accuracy of the data recorded during a motion analysis experiment.

1.4.2. Application of the musculoskeletal modeling

Computer models and simulation of the musculoskeletal system have been introduced to biomechanical, medical research, and sport training [*Delp 2007, Seth 2011, Reinbolt 2011*]. A wide variety of models have been developed to study human movement going from simple two-dimensional models to complete three dimensional musculoskeletal models [*Delp 1990, Pandy 2003, Arnold 2005, Hicks 2008*]. Many software packages have been developed to enhance graphical interfaces and simulation codes of musculoskeletal modeling, such as Software for Interactive Musculoskeletal Modeling (SIMM), Anybody Modeling System (Anybody Technology A/S, Aalborg, Denmark) and LifeModeler (ADAMS package).

Musculoskeletal modeling aims to enhance knowledge about normal walking and pathological gaits , to study the biomechanical consequences of surgical reconstruction such as joint replacements [*Fregly 2007, Fregly 2009*], or to validate surgical treatment strategies [*Delp 1994, Delp 1996 , Arnold 1997, Arnold 2000, Arnold 2001 , Zajac 2003, Arnold 2006*].

1.4.2.1. Normal Gait understanding

Several musculoskeletal models have been developed to study the normal walking in order to enhance our understanding about muscle coordination and to explain how they contribute to move the body segments and joints.

Some studies characterized the contributions of individual muscles to forward progression, vertical support and balance during walking [*Anderson 2003, Neptune 2004, Liu 2008*]. They pointed out that only five muscles, non sagittal muscles, are the primary movers of the leg and the contributors to the forward progression and support needed for normal walking at all speeds; the vasti and gluteus maximus decelerate the body's center of mass during the first half of the stance phase, the soleus and gastrocnemius propel the body forward during the terminal phase of stance, and the gluteus medius during single-limb stance. The same muscles (vasti, soleus, and gastrocnemius) contributed laterally to the body's balance throughout stance and the hip abductors, anterior and posterior gluteus medius controlled balance medially [Pandy 2010].

Another study determined the principle muscles contributing to the increase of the knee flexion during double support, which are Vasti, rectus femoris, gastrocnemius, and iliopsoas (Goldberg 2004). Recent studies focused on defining the main muscle contributors for knee and hip joint contact forces. Shelburne and al (2006) showed that quadriceps and gastrocnemius are most contributors to body's support and forward propulsion during normal walking and the most contributors to knee stability in the frontal plane in addition to the knee ligaments. Whereas, Correa and al (2011) proved that muscles spanning the hip (gluteus medius, gluteus maximus, iliopsoas, and hamstrings) are the major contributors to the hip contact force and the vasti, soleus, and gastrocnemius are the major contributors to hip joint loading.

Muscle mechanical work requirements during normal walking have also been studied. Using a parameterized generic musculoskeletal model, Neptune et al (2004) revealed the importance of mechanical energy costs when the center of mass upward in early single-limb support and founded that the mechanical energetic cost cannot be estimated from external mechanical power.

Generic Musculoskeletal models have been used to understand factors influencing a normal joint motion. Arnold et al. (2007) studied the potential muscles contributing on the knee movement during the swing phase of normal walking, in order to understand the decrease in knee extension in terminal swing observed with cerebral palsy children. Other generic

musculoskeletal models have been developed to study the stress's behavior of a deformable tibia during walking [Al. *Nazer 2008*].

1.4.2.2. CP gait understanding

The management of gait abnormalities in cerebral palsy children is a challenging issue because of the variety of the degree of neurological and orthopedic impairments with variety of biomechanical consequences. The gait analysis examination is insufficient to understand the pathology's causes or to predict how the patient's abnormal gait patterns will progress after treatment. Only the identification of biomechanical factors that contribute to those abnormal movements can predict suitable planning treatment. Together with gait analysis, the musculoskeletal modeling and simulations are a powerful tool for quantifying muscle function and understanding muscle coordination during pathological gait pattern [*Delp 1990, Arnold 2005*].

Several musculoskeletal models have been developed to help clinicians in investing the causes of crouch gait [*Arnold 2005, Arnold 2006, Hicks 2008, Steele 2010, Hicks 2011*], excessive internal hip rotation [*Arnold 2000*] and stiff knee gait [*Goldberg 2004, Goldberg 2006, Jonkers 2006, Reinbolt 2008*]. Others are developed to take part of the clinical decision making process, to assist orthopeadic surgery, or to evaluate the outcome of a treatment [*Delp 1998a, Delp 1998b, Arnold 2001, Desailly 2008*].

Because of the specificities of cerebral palsy children, taking into consideration the deformed bone geometry and specific muscle properties is crucial to better interpret musculoskeletal modeling results and guide treatment decisions [*Arnold 2000, Arnold 2001*].

The first personalized musculoskeletal model was developed by Arnold et al. (2000) and was performed on three patients with cerebral palsy, aged between 7 and 27 years old. The MRI techniques were used to specify the subject-specific geometry and muscle-wrapping surfaces. This study aimed to determine whether medial hamstrings or hip adductors are responsible for excessive internal rotation of the hip and suggested that other factors are more likely the major causes of internally-rotated gait.

Furthermore, constructing a patient-specific model for every child with a gait abnormality would be costly and labor-intensive. Given these difficulties, diverse studies focused on determining the accuracy of the generic musculoskeletal models, or graphic –based musculoskeletal models, to evaluate cerebral palsy treatments.

Arnold et al (2001) developed a deformable generic model in SIMM software, by deforming bone structure to take into consideration bony deformities (e.g. femoral anteversion, torsional tibia) for four CP patients (Figure 1-34). Compared to personalized models, the deformable generic model can provide accurate estimation of muscle-tendon lengths, errors about 3 -5 mm, of the hamstrings and psoas muscles. This study validated the deformable generic model to study deformities of the bony structures observed in CP patients.

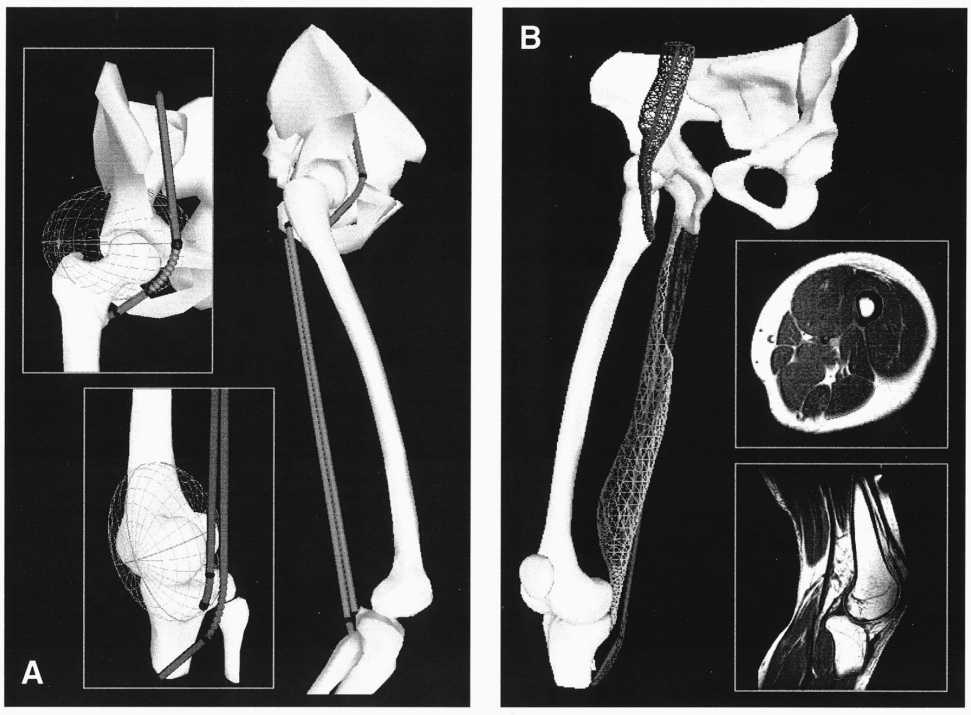


Figure 1- 34. The deformed generic musculoskeletal model in SIMM software (A) vs the personalized musculoskeletal model (B) (© Arnold et al. 2001)

Once validated, the deformable generic model was used to determine the rotational moment arms of several muscles (hamstrings, semimembranosus, semitendinosus, gracilis, adductor brevis, adductor longus, pectineus, adductor magnus, rectus femoris) in CP patients with crouch gait [*Arnold 2001*a], internally rotated gait [*Arnold 2001*b] and stiff knee gait [*Jonkers 2006*]. Retrospective studies have been then developed to guide surgeon's decision by

examining the outcomes of surgeries comparing the predictive (simulated) results of the tendon-muscle lengthening surgery to post-operative ones [Arnold 2006] . These studies identified surgery consequences for knee extension and hamstring lengthening in CP patients, with crouch gait.

Hicks et al. (2008), with a study of 316 Cerebral Palsy patients with crouch gait, aimed to determine biomechanical factors explaining the increased energy requirement with such posture. This can be explained by remarkable reduced capacities of the hip and knee extensors in a crouched gait posture, with a maintained full extension capacity of the hamstrings muscle group and also by the increase of flexion accelerations induced by gravity at the hip and knee throughout single-limb stance.

Desailly and al. (2008) proposed a 3D musculoskeletal model, based on CT-scan data collected from normal subjects, and then deformed to fit well the cerebral palsy geometry of ten children. These authors demonstrated that the Rectus Femoris transfer, on CP children with Stiff knee gait, had an effect for both swing and stance phases of gait, by determining the relationship between the lengthening of Rectus Femoris and its velocity, consequences of the spasticity. They showed that the premature timing of the Original Rectus Femoris path peak length could be identified as a prognostic factor of a successful surgical outcome.

Validating musculoskeletal modeling's results is a controversy issue. Many comparative studies have been developed to validate the generic musculoskeletal modeling when treating cerebral palsy cases. On one hand, Scheys, et al (2008) developed a 25 years old-personalized musculoskeletal model to assess muscle function in lower limbs. Authors focused on calculating the length of lever arm muscle during motion using a deformable generic SIMM musculoskeletal model and a personalized one. Results showed that the generic models of the SIMM software overestimated the length of the lever arm muscles for flexion, extension, abduction, adduction, external rotation of the hip and underestimated the internal rotation of the hip. On the other hand, Correa et al. (2011) evaluated the accuracy of scaled-generic musculoskeletal models relative to MRI-based models in calculating the potential contributions of the lower-limb muscles to the acceleration of the centre of mass during gait, for four children with spastic diplegic cerebral palsy. Authors demonstrated that, for CP children with anatomical abnormalities and mild gait

deviations, both scaled-generic musculoskeletal models and personalized MRI- based models yield to comparable results concerning the muscle's potential contributions to the acceleration of the centre of mass during the single-leg stance phase of gait.

1.5. Thesis objectives

On the basis of this literature review, treating Cerebral palsy pathology seems very difficult, because musculoskeletal troubles may vary from one child to another, depending on the severity and the nature of the neurologic abnormalities. Consequently, clinicians would like to predict post-treatment clinical outcome on an individual-patient basis and musculoskeletal models have to be personalized to each case.

Even if musculoskeletal modeling can be useful in clinical routine for children with cerebral palsy, results have to be easily subjected to experimental verification and should not be overly sensitive to the musculoskeletal model input parameters. Results from comparative studies, presented above, are contradictory when determining the accuracy of generic musculoskeletal model. But, in clinical routine practice, the personalized musculoskeletal modeling is very costly and expensive time processing on daily working.

Since the clinical examination of Cerebral Palsy children does not systematically incorporate a magnetic resonance imaging procedure, the investigation of the accuracy of rescaled-generic musculoskeletal model will represent the main objective of our work.

Based on existing data (e.g gait analysis and clinical examination), we will opt first to study the accuracy of standard rescaled generic musculoskeletal models, developed using LifeModeller software. In a second step, we will study musculoskeletal modeling with healthy population, adults and children, with normal gait and gaits imitating CP characteristics (crouch and jump gaits). The last step will aim to develop a musculoskeletal model specific to CP population by developing a calibration procedure of the musculoskeletal model's parameter values from subject's collected data.

Conclusion of the chapter 1

This chapter dealt with the pathology of cerebral palsy which affects the quality of gait patterns as consequences of bone growth and the abnormal motor control and spasticity. To understand the pathological gaits, characteristics of children with CP, an overview of the normal human locomotion activity is detailed. Different methods used to make it more and more understandable, namely gait analysis examination and the musculoskeletal modeling, are described.

The next chapter will present the method we developed to deal with our objectives in improving existing musculoskeletal models to make them more suitable for the clinical management decision.

Chapter 2.
Materials and Methods

Developing a musculoskeletal model for Cerebral Palsy children requires many steps, detailed in the following chapter.

The first part presents the hypothesis, taken into consideration to develop a musculoskeletal model answering thesis's objectives (§2.1). The second part extends the capture motion and the experiment procedure to collect gait data. These experimental data are used as boundary conditions for the model and as elements of validation of a given model. The third part explains the musculoskeletal modeling procedure of LifeMod Software (Adams plug-in) and the proposed calibration procedure to improve the model's parameters (§2.3). The musculoskeletal modeling's methodology, developed in this thesis, (§2.4) and the data processing procedures (§2.5) are reported in the last part of this chapter.

2.1. Studies and Hypothesis

Studying Cerebral palsy using musculoskeletal modeling requires to take into consideration the specificities of the pathology (e.g. the child skeleton, the altered muscle activities and bone deformities) and then careful interpretations of the obtained results.

According to the literature review, dealing with specificities of CP requires very complex procedure. As a retrospective study, we have been constrained to respect the current protocol of the clinical gait lab and to use existing clinical data.

Several hypotheses are necessary to simplify the musculoskeletal modeling procedure:

H1: Child Skeleton geometry could be deduced from rescaled adult skeleton

H2: Walking with pathological posture alters the muscle function and the motion control strategy

H3: The nervous system is not modeled. The motion control could be represented by optimization function representative of gait.

To investigate these hypotheses, we proposed to simplify the problem when dealing with cerebral palsy and to treat each characteristic apart, as shown in figure 2.1 and explained below. Three studies are then necessary:

- The first study investigates the accuracy of the skeleton's geometry when using a rescaled generic model. In this case, two healthy populations' adults and children have been analyzed when performing a normal gait.
- The second study aims to introduce the abnormal motion control to musculoskeletal model. The altered muscle function was defined when healthy subjects (adults and children) imitated pathological postures observed in cerebral palsy children. Firstly we aimed to study several pathological postures. We limited our study to investigate the jump gait, considered as the most common pattern seen in young patients in earlier independent walking stage (Rodda 2004) and the crouch gait because it becomes the characteristic gait of patients with diplegic cerebral palsy in older age (Gage 2010, Wren 2005).
- In the third study, we developed the musculoskeletal models for three typical cerebral palsy groups with the crouch gait, the jump gait and the stiff knee/ recurvatum gait.

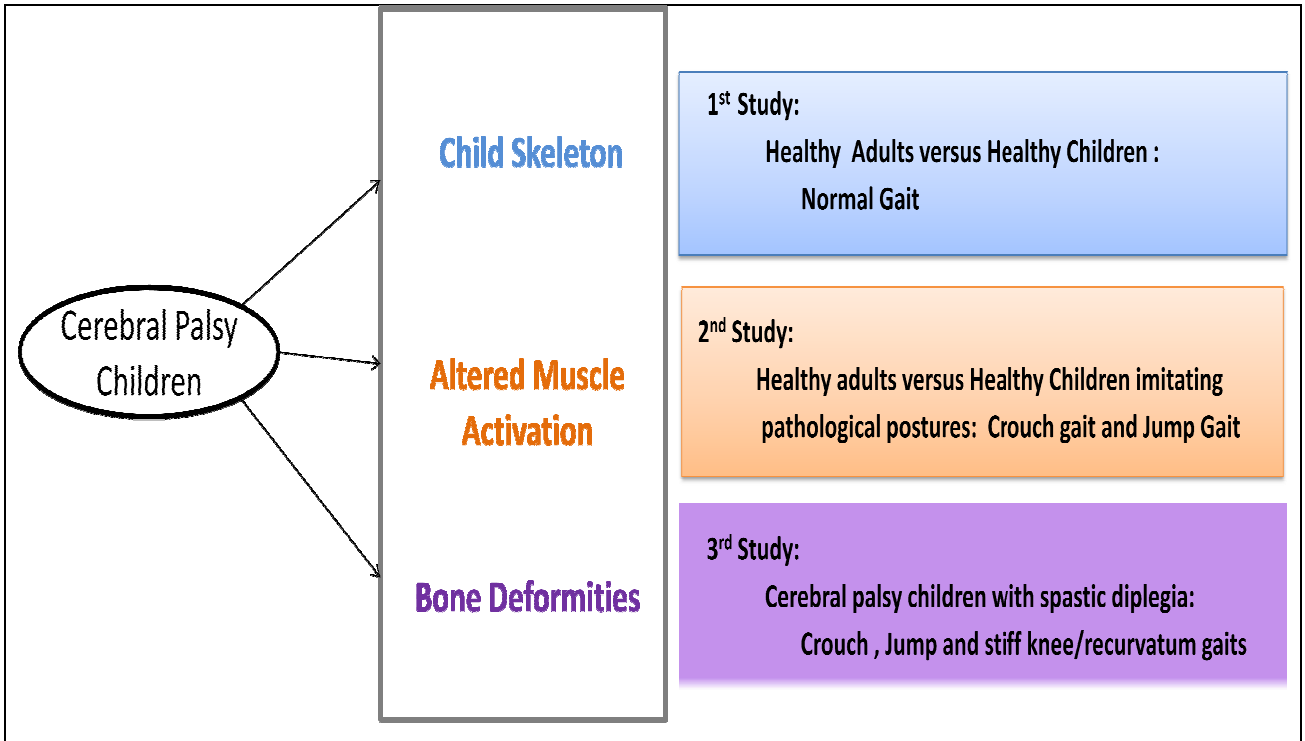


Figure 2-1: Different studies developed in this project

Musculoskeletal modeling requires two essential steps: gathering gait data through capture motion procedure and then developing a musculoskeletal model using LifeMod Software, as explained in the following paragraphs.

2.2. Experimental Data: Motion capture

In the last decade, capture motion technology (e.g gait analysis technology) has been calibrated significantly, resulting in a potential for wider clinical application. Gait laboratories have developed a standard methodology and consistent protocols to measure body motion, ground forces and muscular activation patterns, respectively, in a non-invasive manner, helping them describing natural gait and gait pathologies.

The capture motion was performed in the Capture motion Laboratory (CinDyAh) in University of Technologies in Compiegne and in the Gait analysis laboratory of the rehabilitation center of the French Red Cross (Dr F. Megrot-UCAMM). Healthy subjects and children with cerebral palsy were investigated. For healthy subject, written consents were given. For CP population, a retrospective analysis based on UCAMM gait analysis database is performed.

2.2.1. Equipments

Since June 2009, the "biomécanique et bio-ingénierie" Lab (BMBI- UMR6600) in the University of Technologies in Compiegne has been equipped with complete and newest capture motion systems. Today, in our laboratory, required capture motion systems are available (Figure 2.2), and consist of:

- ✓ **An optoelectronic system**

An optoelectronic system tracks the 3D movement of a set of reflective spherical skin markers placed at palpable anatomical bony of the limb, allowing a complete description of the gait kinematics. The capture motion system is composed of six MX3-Vicon cameras with seven T160–Vicon cameras with a high resolution and high accuracy (Vicon, Oxford Metrics, Oxford, UK). The resolution is about 659 x 493 pixels for the MX-cameras and about 16 megapixels for the T160 –cameras. The acquisition frequency is fixed to 100 Hz.

- ✓ **Force Plates**

Two force platforms AMTI (Advanced Mechanical Technology, Inc., Boston, MA, USA), with a 1000 Hz acquisition frequency, have been used to measure the three components of

the ground reaction forces, the ground reaction torques and trajectories of centers of pressures during walking.

- ✓ **An electromyography system**

A Telemyo 2400T EMG system (Noraxon, Scottsdale, AZ, U.S.A), with eight recording channels, have been used to record the activity of muscles during gait. The acquisition frequency is fixed to the 1000 Hz, to be synchronous with AMTI system.

This set of materials system, completed with two Basler GigaE digital reference video cameras (Basler, Vision Technologies, USA), are connected and synchronized to a computer to acquire and to record data. The Nexus software is then used for calculating the various gait parameters (Vicon, Oxford Metrics, Oxford, UK).

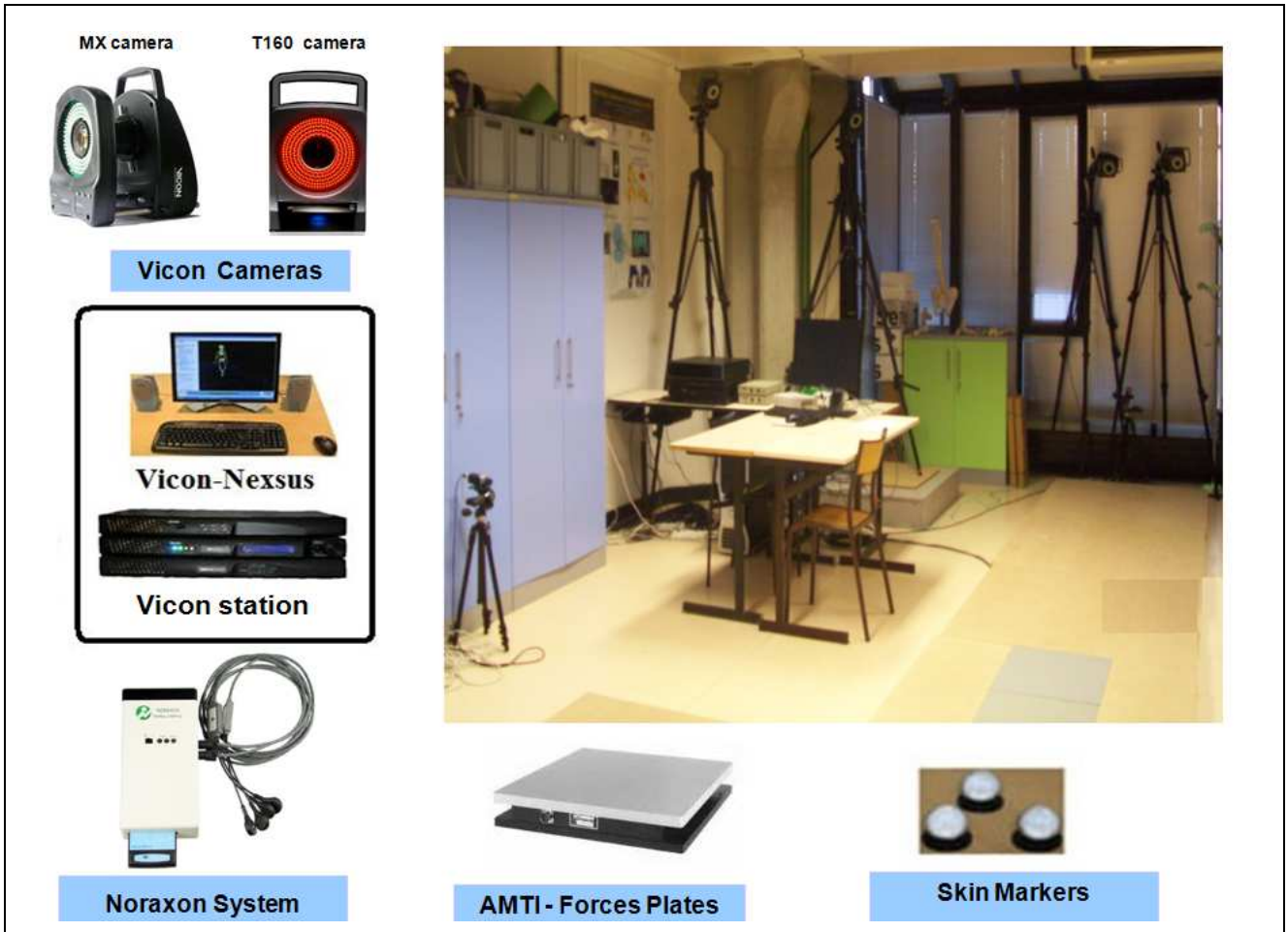


Figure 2-2: The complete capture movement system at the University of Technology in Compiegne: Vicon Cameras, EMG, Force plates and Camera video.

2.2.2. Clinical Protocol

The protocol measurement used in this thesis is the standard Helene Hayes clinical protocol (Davis 1991), frequently used in clinical practices (Figure 2.3).

The first step consists of measuring the anthropometric data, which include height, weight, the distance between the anterior and posterior superior iliac spines (ASIS – PSIS) and the leg lengths of lower limbs measured from the greater trochanter to the knee joint center and from the knee joint center to the lateral malleous. These measurements are important for the calculation of the thigh, calf and foot centers of mass, estimated using regression equations developed by Winter et al. (1990) and the calculation of the inertial properties.

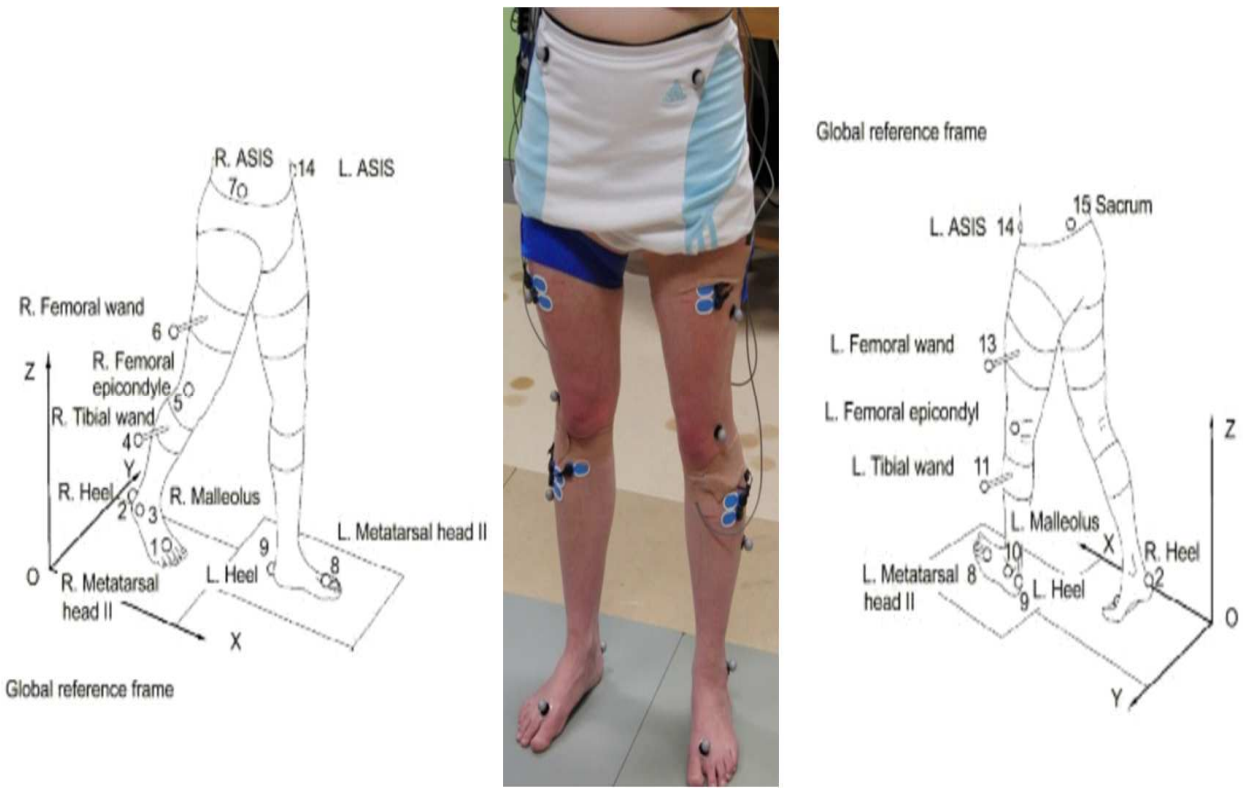


Figure 2-3: Experimental protocol - Skin marker placements according to Helene-Hayes Protocol (Davis et al. 1991) (© www.lifemodeler.com 2008)

Reflective markers and electromyographic electrodes are fixed by double face tape. The placement of reflective markers is performed according to the protocol of Helene Hayes / Davis (1991). Fifteen reflective spherical markers, 12.5mm of diameter, are positioned on the anatomical points of the lower extremities and pelvis (Figure 2-3). These include Sacrum, Anterior Superior Iliac Spine (ASIS), lower lateral 1/3 surface of the thigh (THI), lateral

epicondyle of the knee (KNE), lower 1/3 of the shank (TIB), lateral malleolus (ANK), second metatarsal head (TOE), calcaneous at the same plantar surface of the foot as the toe marker (HEE).

After degreasing the skin with abrasive cleaners, the surface EMG electrodes are placed on the principles muscles of the lower limbs (Figure 2-4), the Tibialis Anterior, Rectus Femoris, Biceps Femoris and Gastrocnemius, according to the SENIAM recommendations [*Hermens 2000*].

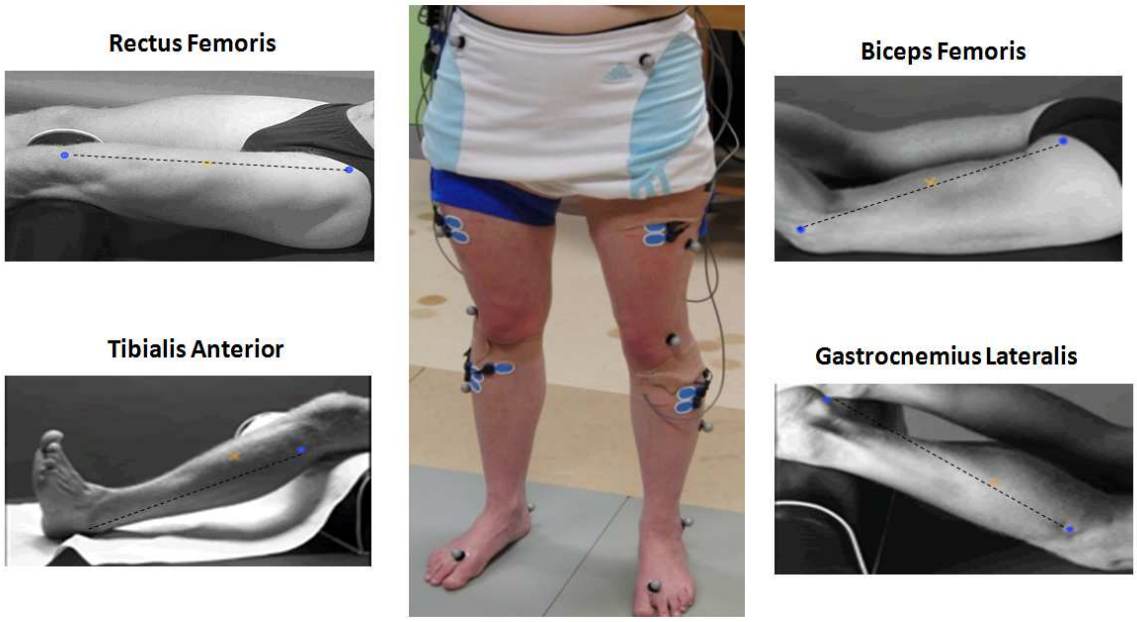


Figure 2-4 : Experimental protocol - sEMG electrodes placements according to SENIAM recommendations (© www.seniam.org)

2.2.3. Clinical Gait Exam Results - Plugin gait modeling

Gait analysis offers a unique tool to deduce the mechanical factors of joint loading, orientation, and muscle activation during daily living activities such as walking. By tracking targets on each limb segment (reflective markers) of the lower limb, joint angles at the knee hip and ankle and the ground reaction forces are computed. Then, the joint loading, net reaction moments and forces, are evaluated via inverse dynamics. Calculations of kinematical and kinetical parameters are largely documented in the literature (Kadaba 1989). Various commercial software are available as post processing of gait measurements. In this project, we used the biomechanical model implemented in Vicon software, namely plug-in gait or

conventional Gait model, developed by Kadaba et al. (1990) and Davis et al. (1991), detailed in the Annex 1.

The results from gait analysis are:

- spatio-temporal parameters: stride length, width, cadence, velocity, time of double limb support and the ratio of swing and stance times.
- the kinematic variables: displacement, velocity and linear acceleration and angular position of the various segments of the body in different anatomical planes to indicate hip, knee and ankle flexion, adduction-abduction and rotation states.
- kinetic variables: moments in the ankle, knee and hip as well as the power consumed or generated at these joints during motion
- Electromyographic strokes: the timing of beginning and duration of contractions of the muscles recorded.

The calculation principle is resumed in the following figure (Figure 2.5).

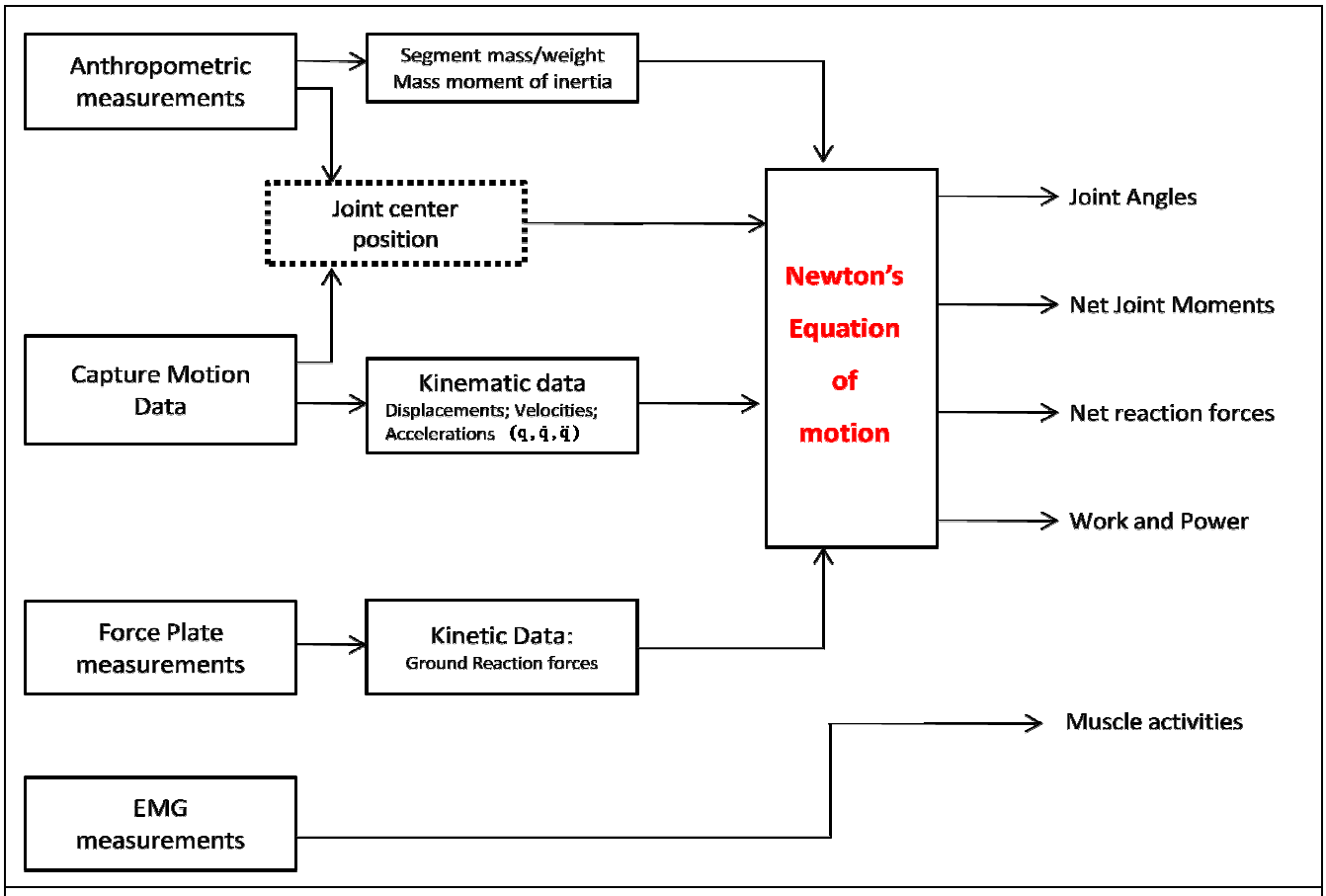


Figure 2-5 : the calculation of the kinematic, kinetic entities using plugin- gait

2.3. Musculoskeletal Modeling

The three-dimensional lower body musculoskeletal model, presented in this study, has been developed using the commercial software BRG.LifeMODE. The software is based on the commercial multi-body software ADAMS (ADAMS, Biomechanics Research Group, Inc., USA).

Different steps of modeling are summarized in the following figure (Figure 2.6).

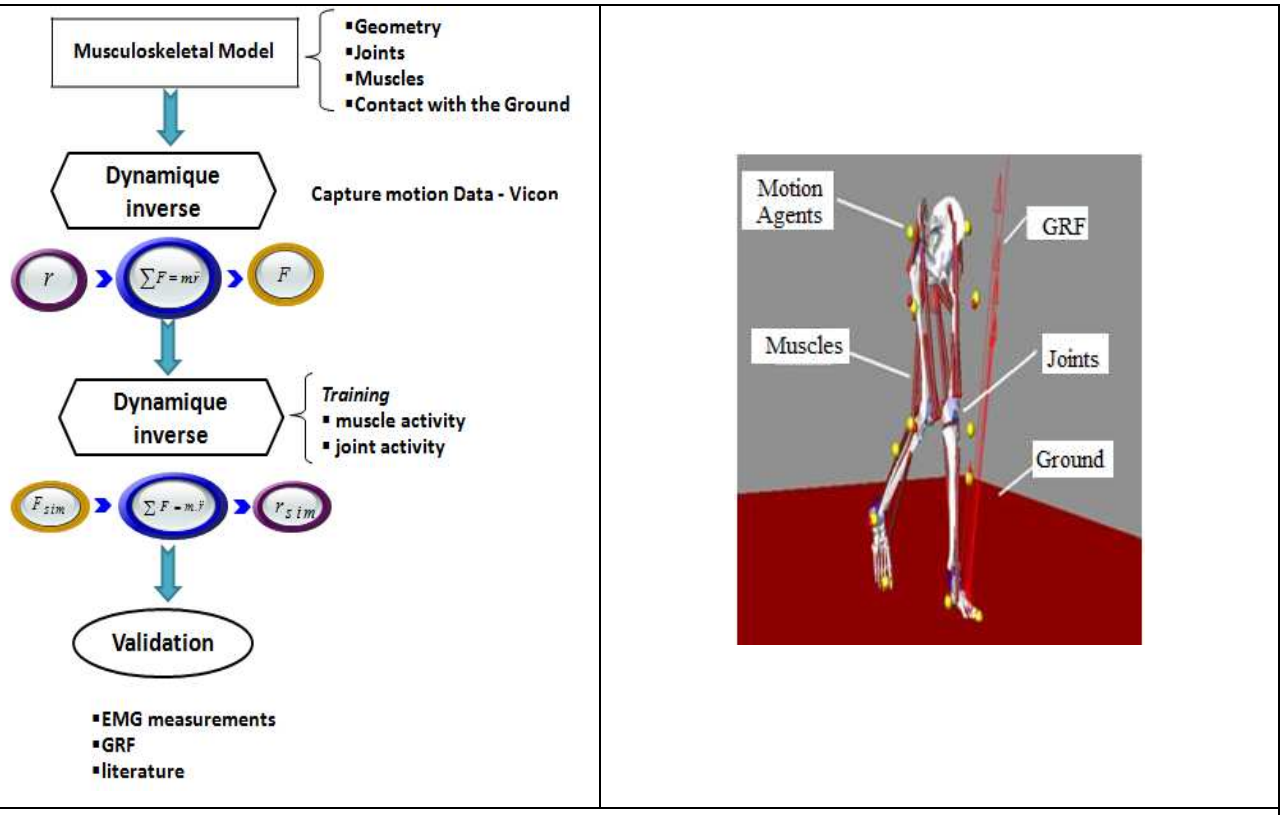


Figure 2-6 : The musculoskeletal modeling using LifeMod software

2.3.1. Rescaled Generic Musculoskeletal Model

The musculoskeletal model is a rescaled generic one, generated from the anthropometric database accessible through the software, based on the experimental subject's height, weight, age, ethnicity, and gender [*Lifemodeler 2008*].

2.3.1.1. Musculoskeletal Modeling

a) Skeletal model

The skeleton of the musculoskeletal models consists of seven rigid body segments: the pelvis, thighs, shanks and foots. The upper body is assumed to be a mass point at the pelvis. The model is generated from the GeBOD anthropometric database accessible through the software based on the experimental subject's measurement [*Lifemodeler 2008*]. The GeBod database is developed by the Air Force Space Medicine Research Lab and the Daton University, United States. The People Size database is measured by the American Health Center and Loughborough University [*Cheng 1994*]. These database include segment dimensions, mass and inertia properties. These data are scaled by using three independent parameters, height, weight and the gender.

b) Joints

The joints are modeled as mechanical joints, a revolute joint for the knee and spherical joints for the ankle and the hip. The stiffness of the joint is provided by ligaments and muscles as well as cartilage, capsule and menisci. However, these entities are not taken into consideration into lifeMod^{TM}.

In this model, the hip and the ankle joints are defined by three degrees of freedom on the principle anatomical planes and the knee is represented only by a sagittal degree of freedom. Nonlinear torsional springs and dampers are applied at each constraint degree of freedom in the model. Joints are then considered as passive torque functions, defined by stiffness and damping properties as well as joint angle limits, used primarily to stabilize the body during the inverse-dynamics simulation. The mechanical joint torque is determined as follow:

$$M_{joint}(t) = K_{joint} * \theta(t) + C_{joint} * \dot{\theta}(t) \quad \text{(Equation 2)}$$

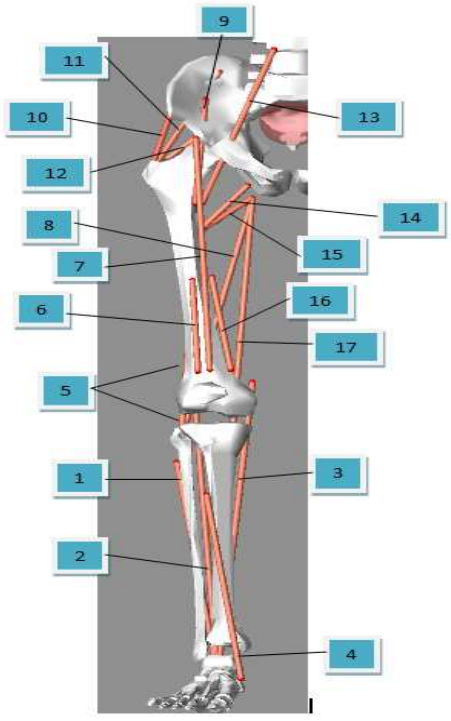
Where $(\mathbf{K_{joint}}, \mathbf{C_{joint}})$ represent respectively the joint stiffness and the damping and the $(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}})$ represent respectively the rotational angle and its velocity, estimated in the inverse dynamic. These parameters are usually fixed to ensure numerical convergence of the musculoskeletal models.

The joint center locations are defined from GeBOD database (*Cheng 1994*), and illustrated in Annex 2.

c) Muscle

The musculoskeletal model of lower limbs is actuated by basic muscle groups including 17 muscles (Figure 2.7). The muscle groups are: the soleus, gastrocnemius, tibialis anterior, biceps femoris, vastus lateralis, rectus femoris, iliacus, gluteus medius, gluteus maximus, adductor magnus, vastus medialis and semitendinosus; and are represented through several physiologically-determined equations in order to produce the necessary forces to track the desired motion of the body, while staying within each muscle's physiological limits.

Muscle parameters such as physiological cross sectional area (PCSA), maximum tissue stress and the muscle paths (origins and insertions) are based on muscle geometry database, generated from Schumacher (1966) and Eycleshymer (1970) studies. These muscle data are scaled based on the anthropometric data of the subject.



ID	Muscle	Attach Proximal	Attach Distal
1	Soleus	lower_leg	foot
2	Gastrocnemius 2	upper_leg	foot
3	Gastrocnemius 1	upper_leg	foot
4	Tibialis Anterior	lower_leg	foot
5	Biceps Femors 1	upper_leg	lower_leg
6	Vastus Lateralis	upper_leg	lower_leg
7	Rectus Femoris	lower_torso	lower_leg
8	Biceps Femors 2	lower_torso	lower_leg
9	Iliacus	lower_torso	upper_leg
10	Gluteus Medius 1	lower_torso	upper_leg
11	Gluteus Medius 2	lower_torso	upper_leg
12	Gluteus Maximus 2	lower_torso	upper_leg
13	Psoas Major	central_torso	upper_leg
14	Gluteus Maximus 1	lower_torso	upper_leg
15	Adductor Magnus	lower_torso	upper_leg
16	Vastus Medialis	upper_leg	lower_leg
17	Semitendinosus	lower_torso	lower_leg

Figure 2-7 : The principle muscles actuating the lower limb in the LifeMod™ (© lifemodeller 2008)

Several muscle models are available in LifeMOD™ library, such as simple closed loop model and Hill-model. In our Study, we will use the standard closed loop model. In such a model,

the muscle is modeled as a line of action passing from an origin to the attachment site and represented as a set of recording elements or trained elements (Figure 2.8). Recording elements, active during the inverse dynamic simulation, record the muscle lengthening history during motion when the model is moved using external drivers such as motion agents. The trained elements, used in the forward dynamics, are represented by linear proportional-integral-differential controllers (PID-Servo) to calculate the muscle activation based on the previous desired muscle contraction trajectory in order to reproduce the motion.

The PID controller is the most common form of feedback and represents a closed loop control system. Each trained muscle has a PID controller that tries to match the instantaneous length of the muscle to the trained curve. These controllers are independent to each other, even though the resulting muscle lengths are coupled through the kinematic model. Under this scheme, without any additional considerations, any arbitrary solutions could match the trained position profile is presented. Because of this, they require an inverse dynamics simulation using passive recording muscles prior to simulation with closed loop muscles [*Lifemodeler 2008*].

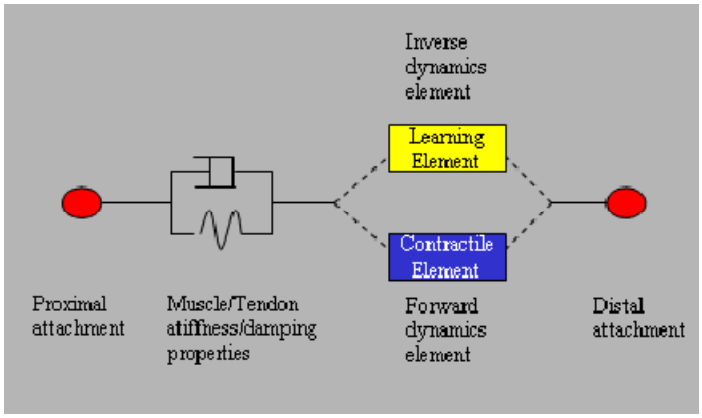


Figure 2-8 :The standard human muscle used in LifeMod™ (© lifemodeller 2008)

Muscle forces, used by the closed loop algorithm, have to respect intrinsic physiological constraints $F_{muscle,max}$. This last entity is the product of the physiological cross sectional area PCSA and maximum isometric muscle stress σ_{max}

$$\mathbf{F_{muscle,max} = PCSA * \sigma_{muscle,max}} \qquad \text{(Equation 3)}$$

The closed loop algorithm is governed by the following formula:

$$u_{muscle}(t) = P_{gain}\left(P_{error}(t)\right) + I_{gain}\left(I_{error}(t)\right) + D_{gain}(D_{error}(t)) \qquad \text{(Equation 4)}$$

and

$$P_{error}(t) = \frac{(\text{target value} - \text{currentvalue})(t)}{\text{Range of Motion}} \; ; \; I_{gain}(t) = \int_{t-1}^{t} P_{error}(t)\, dt \; ; \; D_{gain}(t) = \dot{P}_{error}(t)$$

where, u_{muscle} is the control signal. The target value is determined during the inverse dynamics and the current values represent the muscle lengthening, recalculated during the forward dynamics.

If u_{muscle} satisfies convergence and numerical stability of the model, the current value is took as a good values represent the muscle lengthening history. And then, the muscle activities and forces are calculated as follows:

$$A_{muscle}(t) = \frac{\text{currentvalue}(t) - \text{minimum length}}{\text{maximal length} - \text{minimum length}} \text{ and } F_{muscle}(t) = A_{muscle}(t) * F_{muscle,max}$$

d) Foot-Ground Contact Model

In LifeMOD, the interaction between the musculoskeletal model and the environment is taken into account. The foot-ground contact is a Hertz model, defined as a five spring-damper systems located under each metatarsal head, in addition to one spring-damper system located under the calcaneous, interacting with the ground, represented as rigid plane (Figure 2.9).

Solving this mechanical (physical) problem is based on the ellipsoid-plane algorithm, presented in the following equation, which does not allow any penetration of the foot in ground: $F_{\text{contact point (J)}} = k * g^{e} + C(g) * \frac{dg}{dt}$ (Equation 5)

where,

e and k are predefined parameters; g is the penetration level of the ellipsoid into the solid ground plane; $\frac{dg}{dt}$: The penetration velocity at the contact point; C(g): damping function

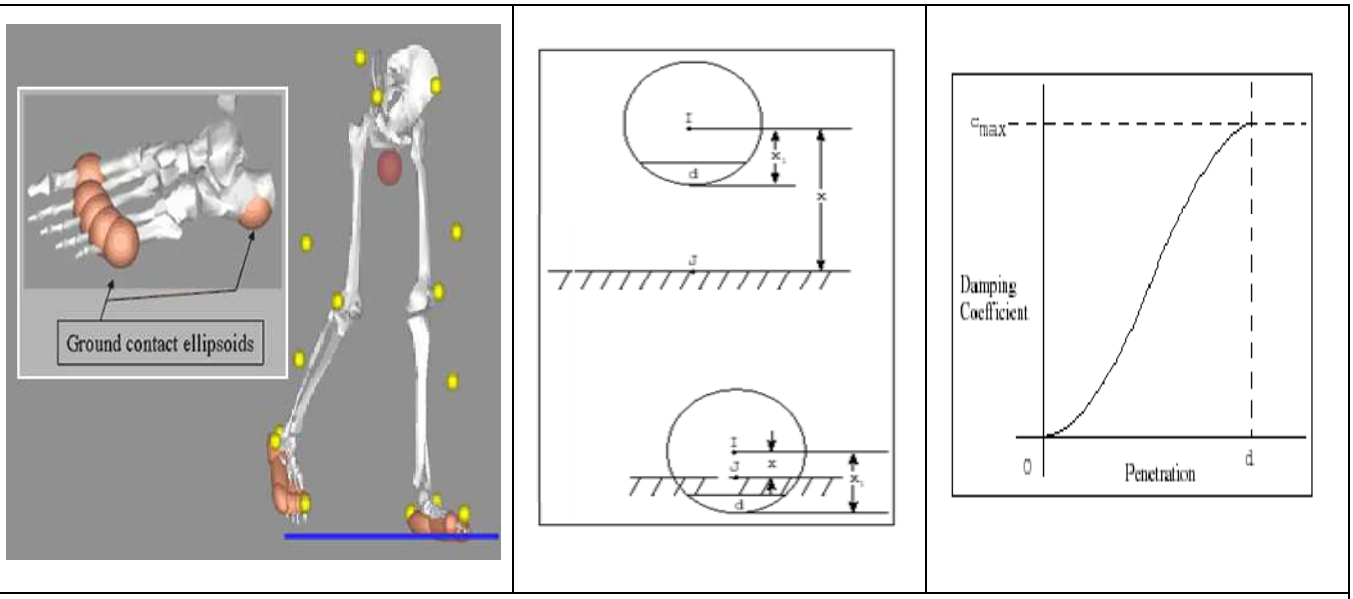


Figure 2-9 : The foot-ground contact model in lifeMod software (© lifemodeller 2008)

The contact model properties are the stiffness, the damping, vertical force exponent coefficient, full damping depth, static friction coefficient, dynamic friction coefficient, friction transition velocity and friction transition velocity. These properties are defined based on the study of Gilchrist and Winter (1996) and Güler (1998).

2.3.1.2. Simulation process

The simulation procedure involves both inverse and forward dynamics simulations. The 3D capture motion data, previously presented in this chapter, is used as input for the inverse dynamics simulation.

a) Inverse Dynamic

In inverse dynamics simulation, the desired muscles shortening/lengthening patterns, required to reproduce the motion described through the markers' trajectories are calculated.

In this step, the data motions, called the motion agents, are imported to drive the musculoskeletal model. These motion agents are modeled as a massless viscoelastic (pushing spring) element between the experimental positions of the skin markers and its corresponding

location attached to the skeleton, displayed respectively as yellow and red spheres (figure 2.10). During the simulation, the distance between the experimental markers and anatomical positions has to keep fixed values to ensure the reproducibility of the recorded kinematics.

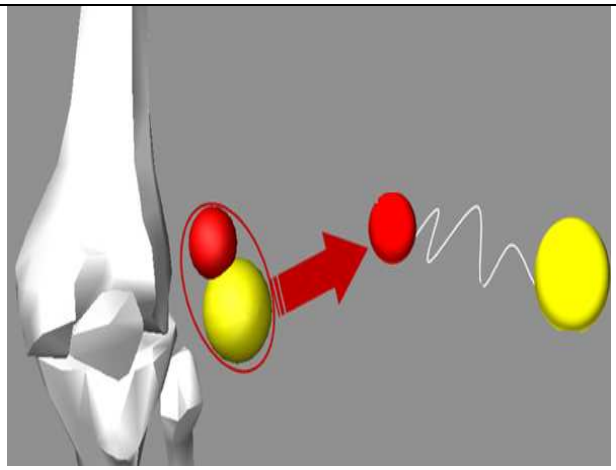


Figure 2-10 : Motion Agent configuration (© Lifemodeller 2008)

b) Forward Dynamic

In the following step, the muscle activation history, obtained in the inverse dynamic simulation, are used to drive the model in the forward dynamics simulation. Each muscle attempts to replicate the desired shortening/lengthening pattern by using a proportional derivative servo controller. The PD controller tries to minimize the error between the desired/recorded kinematics the instantaneous calculated one of each muscle obtained from the forward dynamics simulation at each simulation time step.

In short, the inverse/ forward simulation procedure recalculates the kinematic data, such as 3D markers trajectories, velocities, accelerations and joint angles in the three anatomical plans and estimates the ground reaction forces, the internal joint forces and torques, muscles activation, muscle lengthening history and quantify muscle forces.

2.3.2. Calibration procedure

The standard musculoskeletal model represents a healthy male adult. Model's parameters are compiled through several experimental studies for a normal gait. When studying children population and when studying pathological gaits, parameters have to be calibrated.

LifeMod software offers the possibility of an entire parameterized system. But, since our study is retrospective and we are limited to use clinical gait analysis data, we restricted the calibration procedure to the following set of parameters: contact with the ground, joint center locations, the stiffness joint parameter and some intrinsic parameters required in Lifemod muscusloskeletal model (MSM) such as motion agent and training parameters.

✓ Joint center position

The joint center position was estimated using predictive methods and functional methods. Without any information about the real positions, the use of only the walking data cannot allow to determine the best method to predict the accurate joint center position. We have been limited in this study to determine the joint center positions using the regression equation of Davis et al. (1991), the standard model used in clinical routine and incorporated in the Workstation/ Vicon platforms. According to this method, the joint center positions are defined as follows (Figure 2.11):

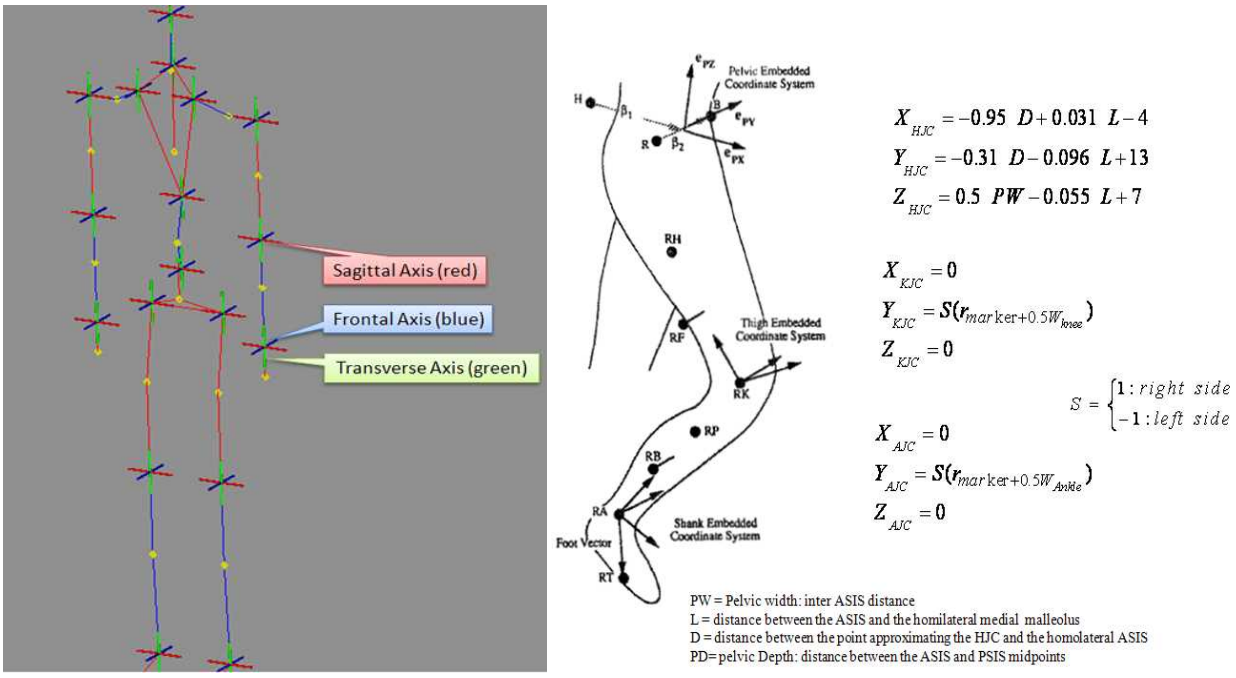


Figure 2-11: Joint axis and center localization in LifeMod MSM (left) and Determination of joint center position according to Davis 1991 (right)

The joint center locations are calculated and then manually placed in the same global reference frame of the musculoskeletal models.

✓ Joint stiffness

In the standard model, the joint stiffness is identical to different joints and for different ages and gender. The accurate choice of joint stiffness is essential. In our study, the joint stiffness, in the sagittal plane, is defined as a clinical parameter to evaluate the joint's functioning. In this study, the joint stiffness was determined using the graph of joint angles-moments, according to Davis et al.1996 (Figure 2.12). The joint angles and moments are calculated using the biomechanical model, plugin-gait, used in clinical practice. The joint stiffness, in the other planes, stayed at their standard values.

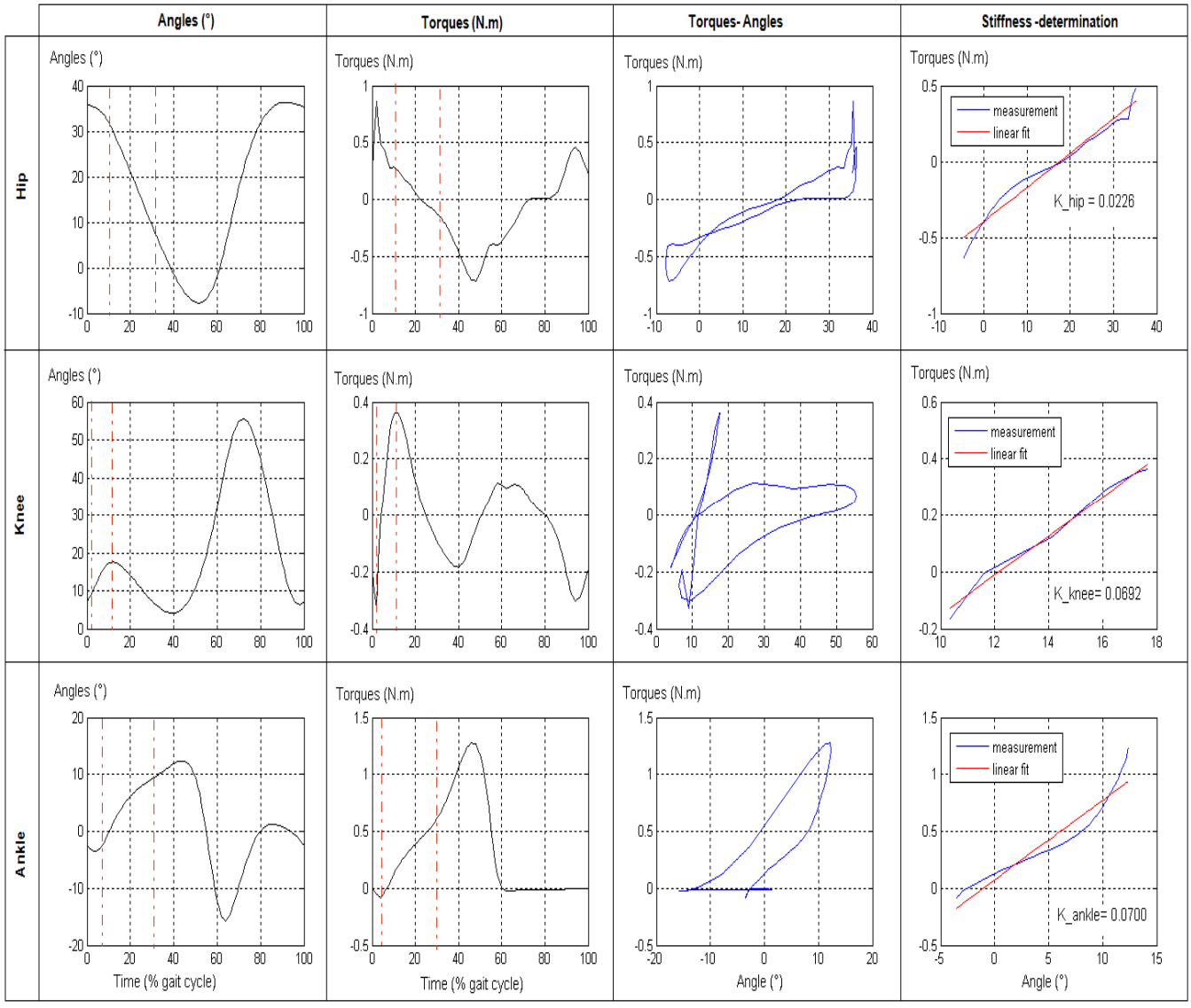


Figure 2-12: Joint stiffness determination

The joint damping is fixed to be 10% of the joint stiffness values.

✓ Contact with the ground

The stiffness and damping parameters of the contact greatly influence the estimated ground reaction forces of the musculoskeletal model. As illustration, figure 2.13 shows the effect of the contact model's parameters (stiffness and damping), when comparing the standard values of the model (Winter 1996) and when multiplying theses parameters with a coefficient of 10.

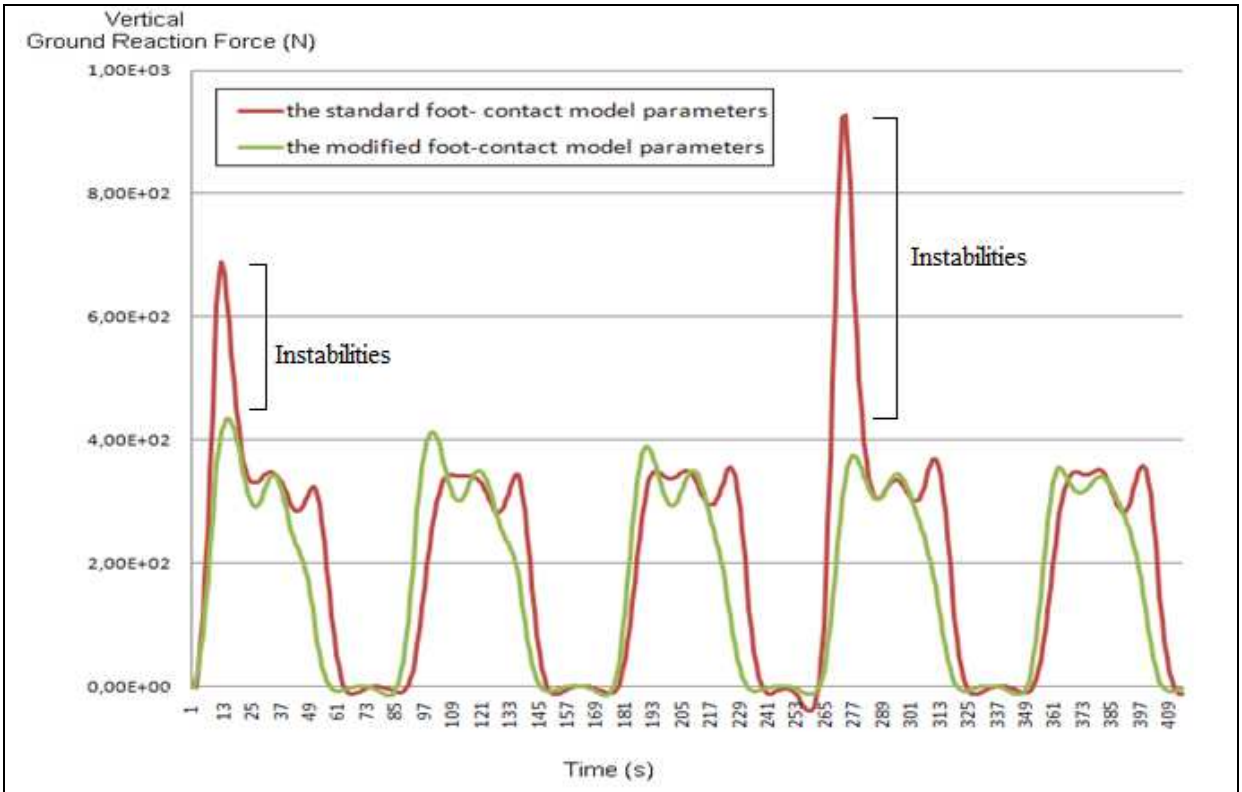


Figure 2-13: An illustration of the effect of the contact model's parameters on estimating the Ground Reaction forces:
the red curve corresponds to the original hypothesis of the Lifemod Model (Winter et al. 1996: stiffness =200) and the green curve (stiffness =2000)

In our study, we opted to derive the model from the ground reaction force measurements. But, introducing these measurements as an input of the musculoskeletal model greatly depends on the quality of the measurements. As an alternative, if these measurements are not available, which is the case of some children with cerebral palsy, a parametric study is performed to minimize the contact instabilities, by increasing manually the rigidity of the contact's model.
In general, the contact does not exceed 120% Body Weight (BW) in normal gait and 160% BW for cerebral palsy children with jump gait. According to such observations, the contact

instability is defined as an overload of 50% BW of the known range of maximum GRF for each population, based on literature or GRF measurements.

✓ Intrinsic Model Parameters

Our preliminary studies showed that the stability of the musculoskeletal model and the accuracy of its results are highly sensitive to some intrinsic parameters of the computational code, such as the motion agent's parameters and the training parameters.

▪ Motion Agents Parameters

As shown previously, the motion capture data are represented in the musculoskeletal model by motion agents or pushing spring elements. In addition to global translational stiffness and damping properties, a constant weight is attributed to each motion agent, which represents a multiplier on the stiffness of the springs between the agent and its rigidly-attached point on the segment (figure 2.10). Preliminary studies showed that the standard parameters defined in the standard model may lead to contact instabilities. For this reason, the weights are fixed at 90% for the motion agents placed at the pelvis, the knee and the foot and at least 50% for those placed at the thighs and shanks.

▪ Training Parameters

The proportional derivative controllers are defined through two essential parameters: proportional and derivative time gains (P_{Gain} and D_{Gain}), defined in Equation 4.

There is no physiological analogy to these parameters. These parameters serve essentially for numerical convergence of the mathematical equations, modeling gait activity. These values modulate results by decreasing the oscillation tendency when tracking the motion [*Lifemodeller 2008*].

Nevertheless, preliminary studies showed that modifying these parameter's values had a great effect on muscle force determination. The accuracy of the musculoskeletal modeling' results greatly depend on these parameters. The choice of the suitable set of values has been defined

when referring to muscle forces gathered from literature during normal gait [*Delp 1990, Kromer 1993, Pederson 1997, Marin 2001, Fraysse 2009*].

2.4. Trials and sessions

All musculoskeletal modeling simulations are based on realistic gait data, to take into account the variability of human motion. Two principal phases are then required: a motion capture phase and then the musculoskeletal modeling and simulations.

2.4.1. Population of study

Two populations are investigated in this project, healthy adult, healthy children and CP children.

2.4.1.1. Healthy population

Ten healthy voluntary adults, aged between 22 and 37 years (27,8 yrs ± 5), and five healthy children, aged between 6 and 9 years (7,2 yrs ± 2), with any gait abnormalities, participated in the study. Children's parents gave written consent for participation. As presented in §2.2.2, reflective markers are placed according to the standard Helene Hayes clinical protocol (*Davis 1991*) and the surfacic EMG electrodes are placed on principles muscles of the lower limbs (Tibialis Anterior, Rectus Femoris, Biceps Femoris and Gastrocnemius) according to the SENIAM recommendations [*Hermens 2000*].

Healthy subjects performed a series of normal gaits and imitated the crouch gait and the jump gait (Figure 2.14). For each group of gait, six trials had been recorded. At least, ten cycles are used to ensure the reproducibility of imitated gait. The good trial measurements, or high-quality of makers tracking, EMG signals and GRF measurements, had been used to derived the musculoskeletal model and validate its results.

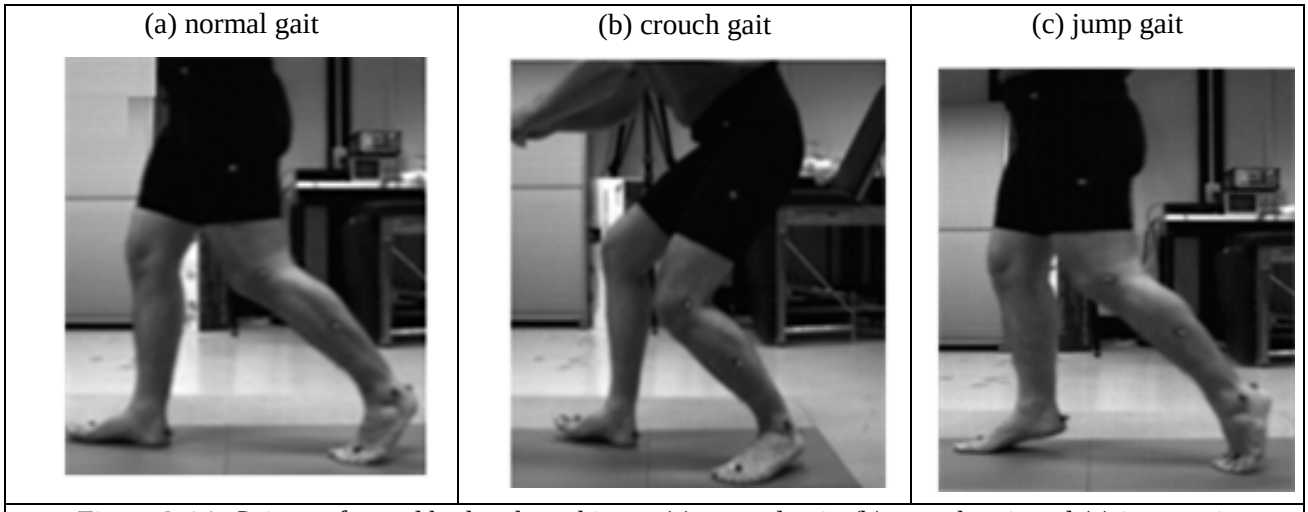


Figure 2-14: Gaits performed by heathy subjects: (a) normal gait, (b) crouch gait and (c) jump gait

2.4.1.2. Cerebral Palsy population

In this project, we will focus on children affected with spastic diplegia. All subjects with spastic diplegia are ambulatory patients, in which the degree of spasticity does not prevent them from walking.

As a retrospective analysis, Cerebral palsy patients have been carefully chosen, from Bois Larris database, to be representative of each group of gait abnormalities: crouch gait, jump gait and recurvatum /stiff knee gait. Six CP children of each group have been studied, with a mean aged about ten years old (10,8 ± 3.3). The preoperative gait analysis exams were performed in the Gait analysis laboratory of the rehabilitation center of the French Red Cross with the collaboration of Dr F. Megrot (Figure 2.15). Reflective markers are placed according to the standard Helene Hayes clinical protocol (Davis 1991) and the surfacic EMG electrodes are placed on principles muscles of the lower limbs (Tibialis Anterior, Rectus Femoris, Biceps Femoris, Gastrocnemius, Semitendinous, Vastus medialis and Peroneus).

Table 2.1: Subject group characteristics

	Healthy Population		**Cerebral Palsy Population**		
characteristics	Adults	Children	CP with Crouch Gait	CP with Jump Gait	CP with stiff knee/ recurvatum Gait
Number of subjects	10 (5 F ; 5M)	5 (1 F ; 4M)	6	6	6
Age (years)	27.5 (4.4)	6.8 (1.3)	13,8 (2.3)	7.6 (1.8)	10,5 (2.3)
Body weight (Kg)	71.2 (13.1)	23.8 (4.2)	35.8 (4.2)	25.2 (3.4)	26.1 (2.1)
Height (m)	1.704 (0.08)	1.25 (0.07)	1.26 (0.021)	1.23 (0.035)	1.25 (0.028)

Figure 2-15: The gait analysis exam in the rehabilitation center of the French Red Cross –Bois Larris. © ucamm-boislarris.megrot.com

2.4.2. Sessions of study

In this project, two models have been developed. First, the biomechanical model has been defined through the Plug-in gait software (Nexus software) and considered as the clinical reference in this study. Lately, it will be named kinematic model. Second, musculoskeletal models have been developed using Adams/LifeMOD software and derived by experimental data.

In our study, to develop a child skeleton, the rescaling process of an adult musculoskeletal model of the lower limb is applied to fit the size of each young subject. For cerebral Palsy cases, the spasticity of the muscles was not taken into account on the modeling process.

The development of the musculoskeletal models has been performed in two steps. In the first part, the musculoskeletal model relied on the original / standard values of several parameters, defined through literature or derived from several experimental essays [*LifeModeller 2008*], we named it the standard rescaled generic musculoskeletal model. In the second part, parameters are specific-subject calibrated, as defined in the paragraph 2.3.2 and named the calibrated rescaled generic musculoskeletal model.

Table 2.2: Parameters setting: the rescaled generic standard MSM versus Calibrated MSM

Parameters	Rescaled generic Standard MSM	Rescaled generic Calibrated MSM
Joint Stiffness	Standard Values fixed to ensure numerical convergence (LifeMod)	Calibrated in sagittal plane based on Davis et al. 1996
Joint center of rotation	Predictive equation of GeBOD (Cheng 1994)	Predictive equations of Davis 1991
Contact with the ground	Parameters defined by Gilchrist et Winter 1996	Measured GRF used as input of the MSM
Motion agent parameters	Standard Values (LideMod)	Calibrated based on minimizing the contact instabilities and erroneous bone deformities
Training parameters	Standard Values (LideMod)	Calibrated to ensure numerical convergence and satisfy literature data of muscle forces *[Delp 1990, Kromer 1993, Pederson 1997, Marin 2001, Fraysse 2009]*

Musculoskeletal modeling's limitations are studied by comparing the estimated results from the musculoskeletal models and those from the kinematic model. Figure 2.16 below summarizes the procedure adopted for this resolution.

2.5. Data processing

Results from musculoskeletal models and kinematic model were balanced. Variations of the determinants of gait, from different models, have been compared [*Shutte 1999, Romei 2003*]. In kinematics, we had focused on the hip flexion, Knee flexion and dorsi-flexion at initial contact, Maximum of hip extension, Maximum knee flexion at mid stance phase and during the swing phase, Minimum and maximum dorsi-flexion and finally maximum plantar-flexion. For the Ground Reaction Force, we had focused on the maximum vertical force during heel-strike, minimum vertical force during mid-stance and maximum vertical force during push-off [*Shutte 1999, Romei 200*3].

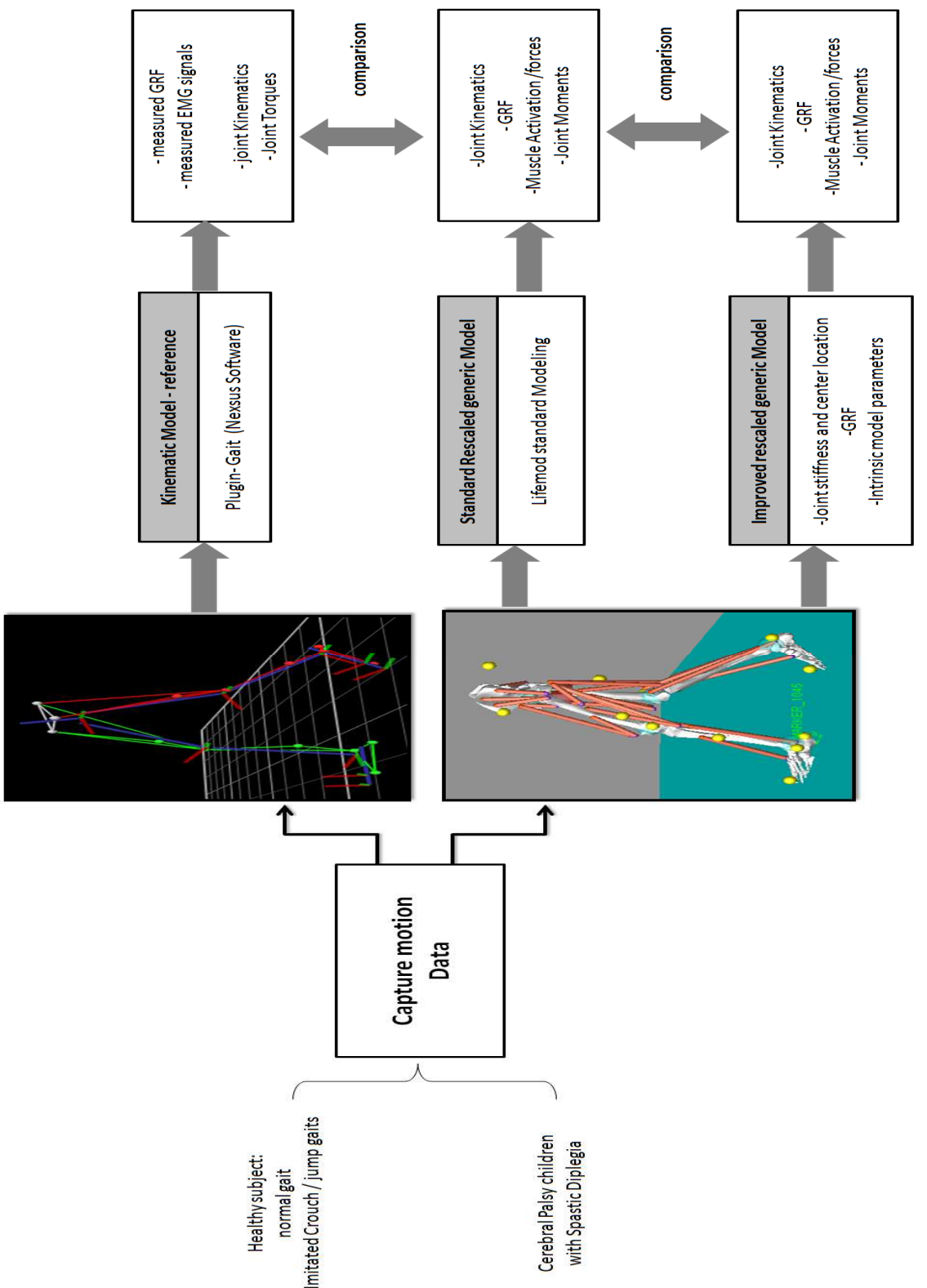


Figure 2-16: Schematic diagram of the procedure developed in this project.

The estimated muscle forces are compared to the envelop of measured EMG signals. According to Perry et al. (1993), “the onset and the cessation times from the three gait cycles have proved to be representative of the average muscle activation function”. These two parameters have been used to determine whether the musculoskeletal modeling may predict accurate muscle activations.

Finally, the means and percentage of variation (SE), between the MSM and the plugin-gait (kinematic) model, were calculated for each subject. Correlation coefficients, CMC coefficients and statistical t-tests were calculated. Level of significance was set to $p<0.05$. The MATLAB software package (MathWorks, USA) has been used for all calculations. For each population, results are presented through the mean of correlation coefficients with the standard deviation.

Regarding the simulated pathological posture, the inter-subject reproducibility was calculated using the CMC coefficients. The kinematics, moment joints and EMG results are compared to the following literature references [*Gage 1994, Ganley et al. 2005, Schwarz et al. 2008, Rodda et al. 2004, Rozumalski et al. 2008, Lin 2000, Vaughan 1996, Schache 2007, Steele 2010, Romeks 2007, Thomas 1996*].

Conclusion of the chapter

This chapter dealt with the procedure of developing a musculoskeletal model for a healthy adult and improvement required to study Cerebral Palsy children. A capture motion analysis had been performed for each subject. The 3D tracked trajectories during the motion capture were used to monitor the developed musculoskeletal models. A calibration procedure was presented to improve the musculoskeletal model’s parameters.

In this project, three models have been compared: kinematic model (a plug-in gait), a standard rescaled generic MSM and a calibrated MSM.

The results of simulations using rescaled generic model and calibrated models will be presented in the following chapter.

Chapter 3.
Results

This chapter details results of comparison between the Plug-in gait model (kinematic model), the standard rescaled musculoskeletal model and the calibrated musculoskeletal model developed in different studies in this project.

In the first part, the preliminary study was done for normal gait to establish the suitable set of parameters of the calibrated musculoskeletal model (§3.1). The second part aims to determine the effect of using musculoskeletal models with varying levels of refinement (calibration) on the accuracy of biomechanical results: kinematics, kinetics and muscle forces during gait in healthy and CP populations (§3.2). Finally, the clinical outcomes of the influence of the imitated pathological gait in healthy population's kinetics and muscle activities are reported in the last paragraph (§3.3).

3.1. Preliminary study: Evidence of parameters influence on musculoskeletal modeling's results

Focusing on normal gait, the preliminary study was performed to study the influence of the model's parameters on the predicted results and then propose appropriate set parameters to calibrate the standard musculoskeletal model.

3.1.1. Motion Agent Parameters

The influence of the motion agent parameters on the MS results are presented in the following figures. To avoid contact instabilities (Figure 2.13) and unrealistic bone deformities, sensitivity studies have been performed to determine the required parameters of motion agents. Table 3.1 represents the standard value used in the generic musculoskeletal model and the values we determined. The stiffness and damping parameters of the spring related to motion agents are fixed by default in LifeMod, as following: transitional stiffness= 500 N/mm and damping = 50 N.s/mm.

Table 3. 1: Standard and specific motion agent weights in MSM en %

Joints	Standard MA weights LifeMod	Specific MA weights in current study
RASIS	3	90
LASIS	3	90
SACRUM	3	90
L/ R THIGH	1	50
L / R KNEE	3	90
L / R SHANK	1	50
L / R ANKLE	10	90
L / R HEEL	10	90
L / R TOE	10	90

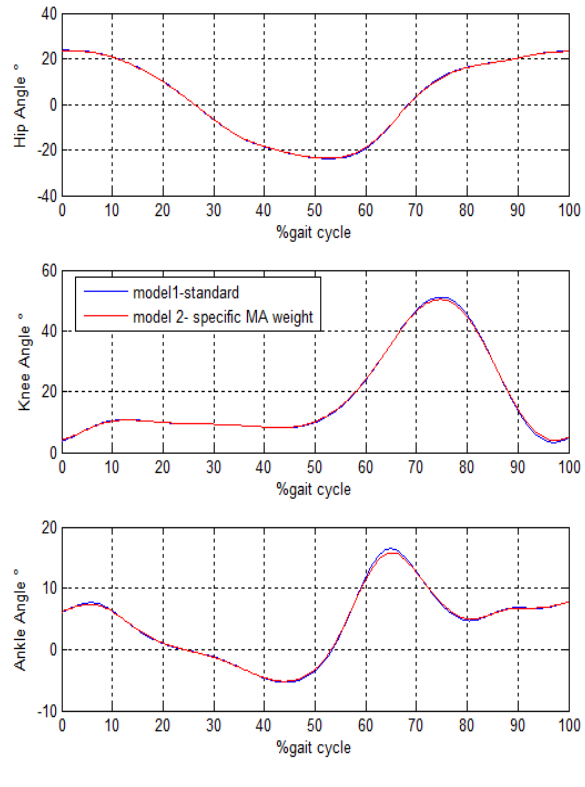
The comparison between the generic model with standard motion agent parameters and specific motion agents parameters are shown through kinematics, kinetics, GRF and muscle forces. Even if the variations founded in kinematics are negligible, less than 1°, the errors in moments and muscle forces are important (Figure 3.1-3.2).

Table 3. 2: An example of the influence of motion agent weight values in MSM results: mean differences and relative errors for a normal gait with a healthy adult.

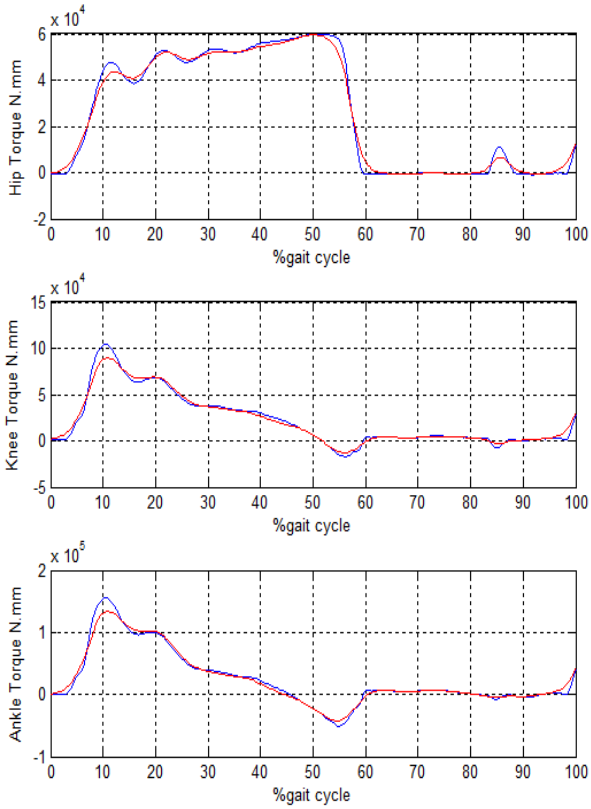
	Kinematics (°)			Torques (N.m)			GRF (N)		
	Hip	**Knee**	**Ankle**	**Hip**	**Knee**	**Ankle**	**GRF x**	**GRFx**	**GRFz**
Variation	0.4985 °	0.7975°	0.6812 °	1.6074	2.5981	1.7689	17.6 N	30.5 N	82.2 N
% variation	<0.01	<0.01	<0.01	0.15	0.17	0.12	0.1	0.3	0.2

Table 3. 3: An example of the influence of motion agent weight values in MSM results: mean differences and relative errors on predictive muscle forces for a normal gait with a healthy adult.

	Muscle forces (N)			
	Gastrocnemius	**Biceps femoris**	**Tibialis Anterior**	**Rectus femoris**
Variation	0.8	0.05	0.17	27
% variation	0.12	0.6	0.5	0.7

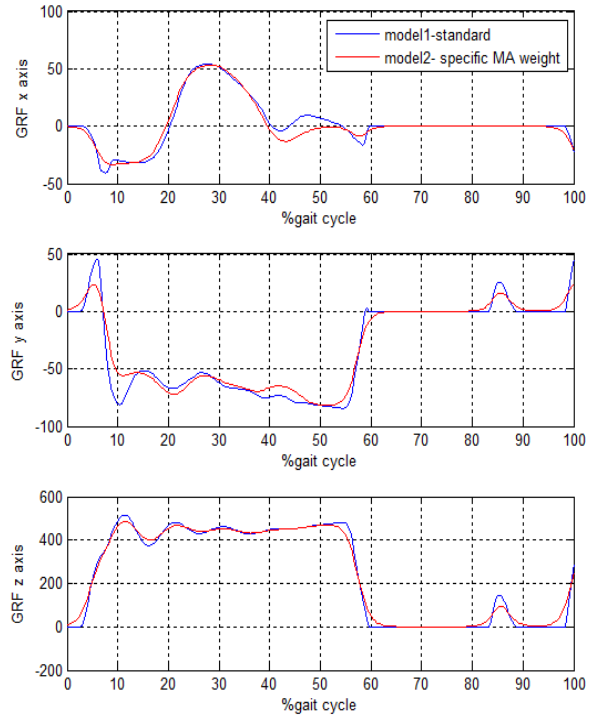


(a) Joint Kinematics

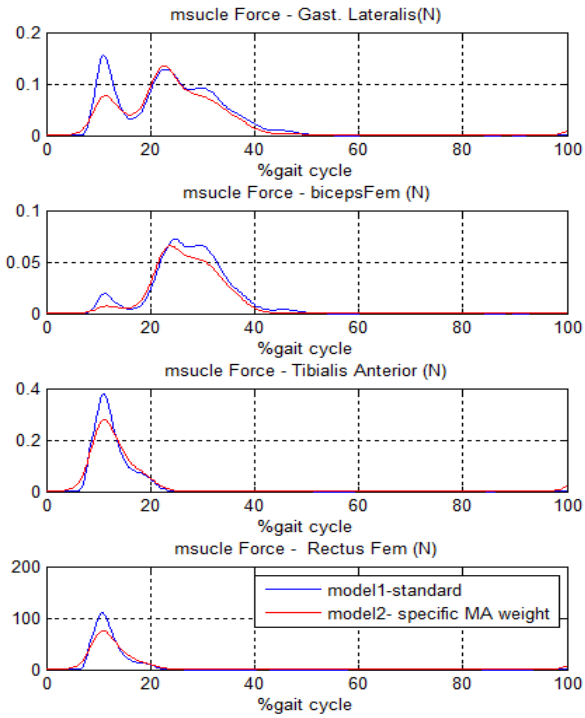


(b) Joint torques

Figure 3. 1: Impact of the Motion agent parameter on musculoskeletal results: joint kinematics and joint torques



(a) Ground Reaction Forces



(b) Muscle forces

Figure 3. 2: Impact of the Motion agent parameter on musculoskeletal results: the GRF and estimated muscle forces.

3.1.2. Joint stiffness parameters in sagittal plane

In the standard model, the joint stiffness is identical to different joints and for different ages and gender. The influence of the change of the joint stiffness has been studied. Three set of parameters are used in a musculoskeletal model of a healthy population when performing a normal gait, respectively $10e^6$ N.mm/° (A), $10e^5$ N.mm/° (B) and $10e^4$ N.mm/° (C). The second parameter represents the standard model's hypothesis, the other represent a random values. The results of hip joint angles, in sagittal plane, are shown in Figure 3.3. We might conclude that the change in joint stiffness greatly affects the results and would possibly provide wrong results, as reported in literature (*Ho and al, 2008*).

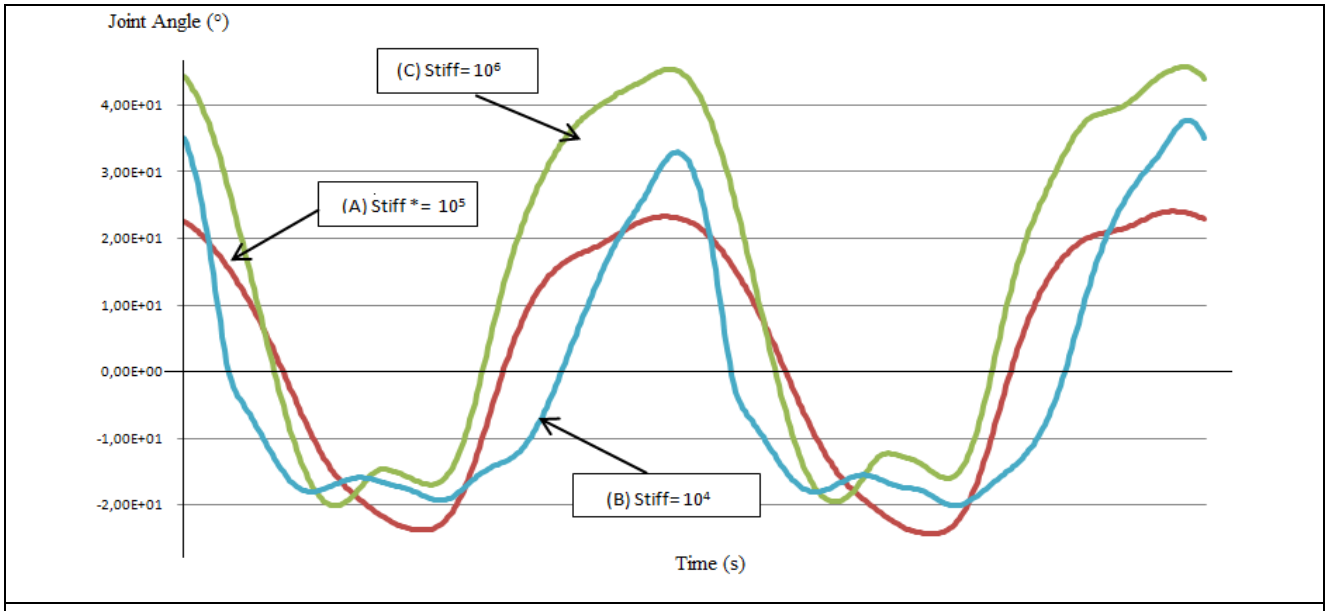


Figure 3. 3: Example of the effect of the joint stiffness parameter on the hip joint kinematics: the red curve corresponds to the original hypothesis of the lifemod Model (Stiffness= $10e^5$ N.mm/°), the blue curve corresponds to (Stiffness= $10e^4$ N.mm/°) and the green curve corresponds to (Stiffness= $10e^6$ N.mm/°)

As detailed in previous chapter, the torsional stiffness joints, in sagittal plane, were determined as the slop the net joint torque-angle graphs (figure3.4). The stiffness values in other planes remain on standard values.

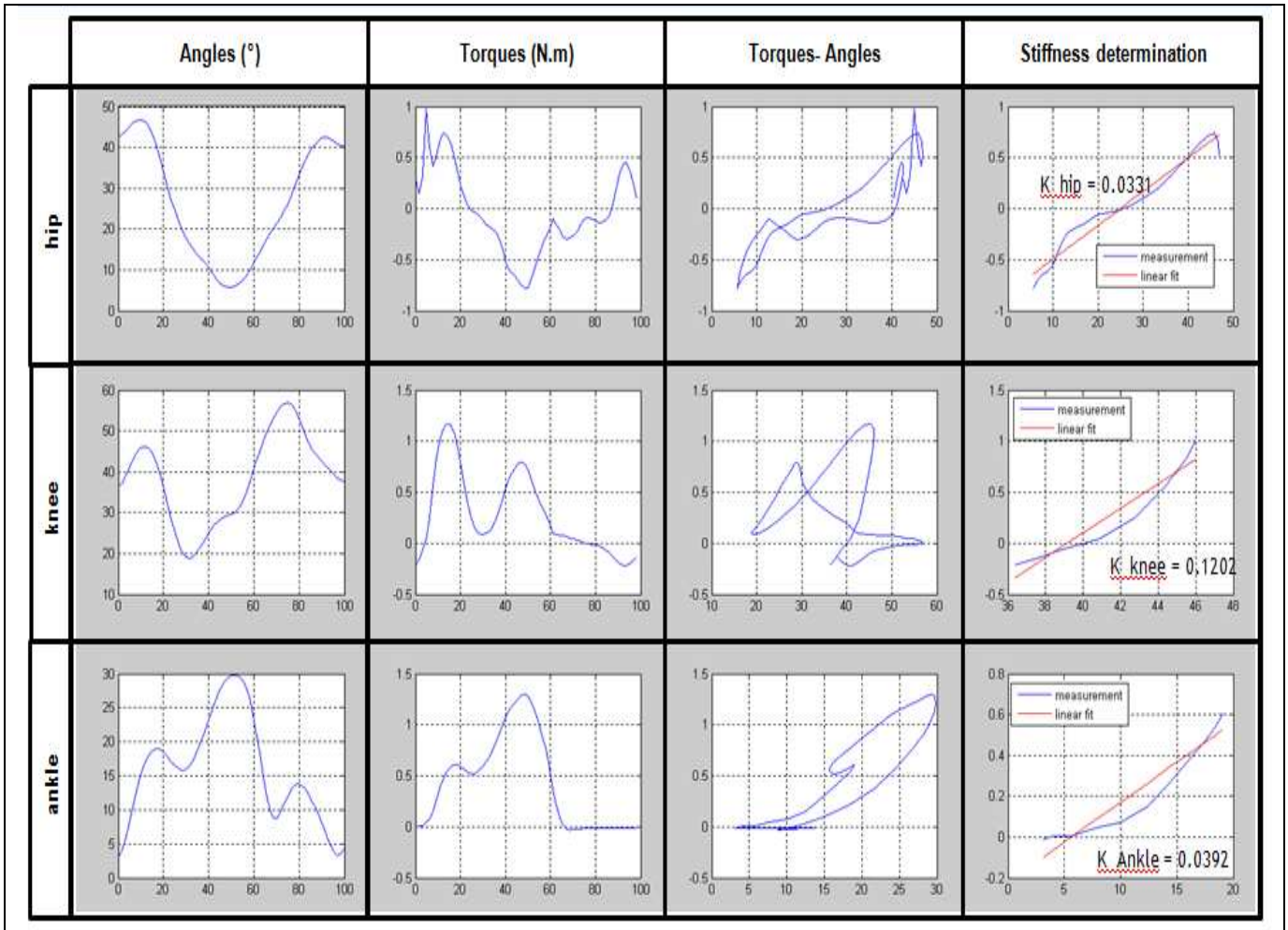


Figure 3. 4: Example of the determination of joint stiffness from experimental data of joint angles and torques - case of cerebral palsy child with crouch gait

Table 3. 4: Experimental values of joint stiffness (N.m/°.kg) in sagittal plane – Healthy Adult Population

		Healthy Adult Population : Flexion – extension – stiffness [N.m/°.kg]		
	Joints	**Hip**	**Knee**	**Ankle)**
Experimental value (N.m/°.kg)	Sujet1	0.0592	0. 0741	0.0482
	Sujet2	0.0522	0.0789	0.0535
	Sujet3	0.0528	0.0695	0.0627
	Sujet4	0.0377	0.0755	0.416
	Sujet5	0.0408	0.0658	0.0578
	Sujet6	0.0573	0.0703	0.0519
	Sujet7	0.0444	0.0775	0.0489
	Sujet8	0,0399	0.0845	0.0408
	Sujet9	0.0514	0.0694	0.0412
	Sujet10	0.0559	0.0837	0.0438
Litterature (N m/kg.°)		0.067 (0.016)	0.0596 (0.015) 0.092 (0.022)	-

Table 3. 5: Experimental values of joint stiffness in sagittal plane – Healthy Children Population

		Healthy Children Population : Flexion – extension – stiffness [N.m/°.kg]		
Joints		**Hip**	**Knee**	**Ankle)**
Experimental value (N.m/°.kg)	Sujet1	0.0239	0.059	0.0623
	Sujet2	0.0226	0.0667	0,0678
	Sujet3	0.0199	0.0571	0.693
	Sujet4	0.0227	0.0494	0.0472
	Sujet5	0.0275	0.0635	0.0763
Litterature (N m/kg.°)		0.028 (0.007))	-	0.0598 (0.016) 0.103 (0.014)

Table 3. 6: Experimental values of joint stiffness (N.m/°.kg) in sagittal plane – Cerebral Palsy Population

		CP Children Population : Flexion – extension – stiffness [N.m/°.kg]		
Joints		**Hip**	**Knee**	**Ankle**
Experimental value (N.m/°.kg)	**CP - Crouch Gait**	0.0331 (0.015)	0.1421 (0.063)	0.521 (0.155)
	CP - Jump Gait	0.0429 (0.029)	0.3822 (0.086)	0.859 (0.094)
	CP - Recurvatum Gait	0.0487 (0.074)	0.7233 (0.068)	0,0763 (0.058)

Tables 3.4-3.6 show differences of experimental joint stiffness between healthy adults, healthy children and Cerebral Palsy cases. Results presented considerable inter-subject variability for different joints, because of the anthropometry of each subject and the associated pathology's severity for cerebral palsy population.

For an adult of 70 kg and Children of 20kg, the stiffness values of the joints used in the standard rescaled generic model have been compared to experimental values for different populations, s shown in Figure 3.5 and Table 3.7. On one hand, defining the same stiffness values in a musculoskeletal model, for different joint, different ages and genders, doesn't seem realistic. On the other hand, results show that these standard values, in sagittal plane, are over-estimated in different joints and different population. The experimental values does not exceed 5% of that standard value for healthy adults, 1.5% for healthy children and may attaints 17% for Cerebral Palsy cases, because spasticity and bone deformities limits the joint range of motion and increase their stiffness.

Table 3. 7: Experimental values of joint stiffness (N.mm/°) compared to standard LifeMod proposed values for an adult (70kg) and children (20kg)

		Experimental value (N mm/°)			
Joints	**Standard values – LifeMod**	**Healthy Adult 70kg**	**Healthy Child 20 kg**	**CP Child - 20 kg Crouch Gait**	**CP child - 20 kg Jump Gait**
Hip	10^5	$4.144 * 10^3$	$0.46 * 10^3$	$0.858 * 10^3$	$0.662 * 10^3$
Knee	10^5	$5.187 * 10^3$	$1.18 * 10^3$	$7.644 * 10^3$	$2.842 * 10^3$
Ankle	10^5	$3.36* 10^3$	$1.34 * 10^3$	$17.18 * 10^3$	$10.42 * 10^3$

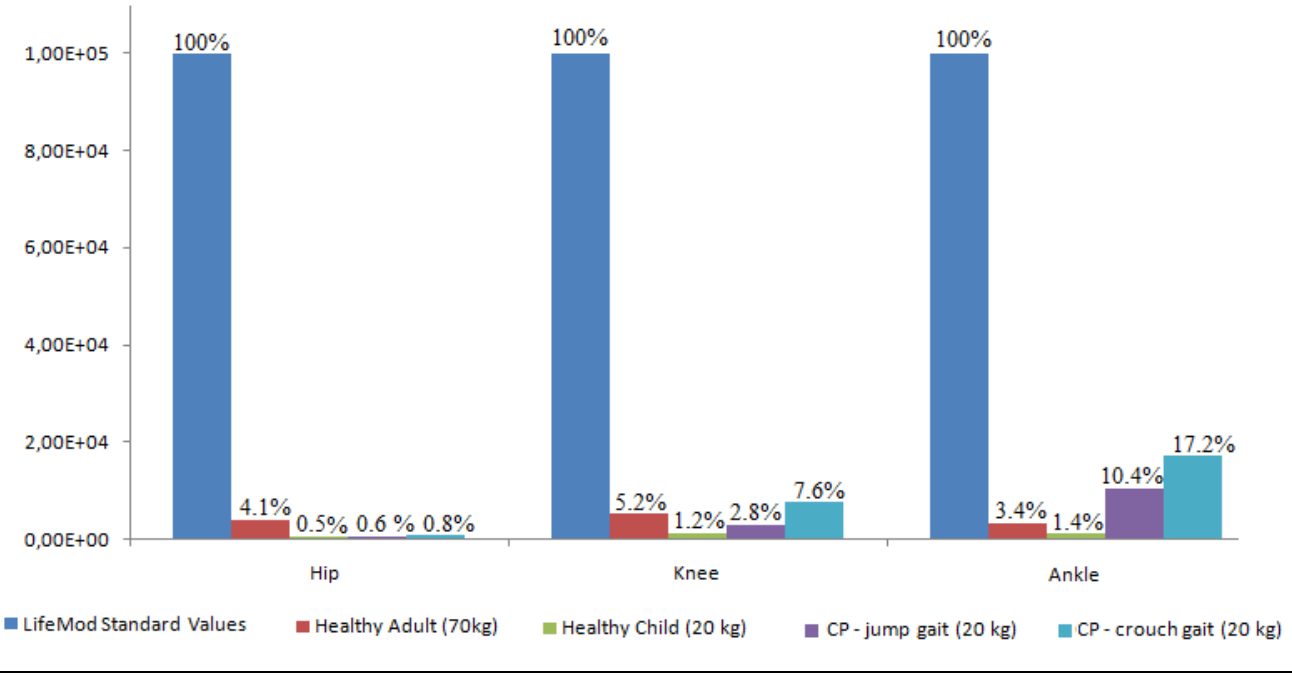


Figure 3. 5: Experimental values of joint stiffness compared to standard LifeMod proposed values

Such significant differences may lead to erroneous results of musculoskeletal model. Specific-subject joint stiffness values have been introduced in the calibrated musculoskeletal modeling.

3.1.3. Training Parameters

Our primary studies showed that modifying Proportional and Derivative gains (P_{Gain} and D_{Gain}), defined in training process of forward dynamic, had a great effect on muscle force determination, as shown in this following figure (Figure 3.6).

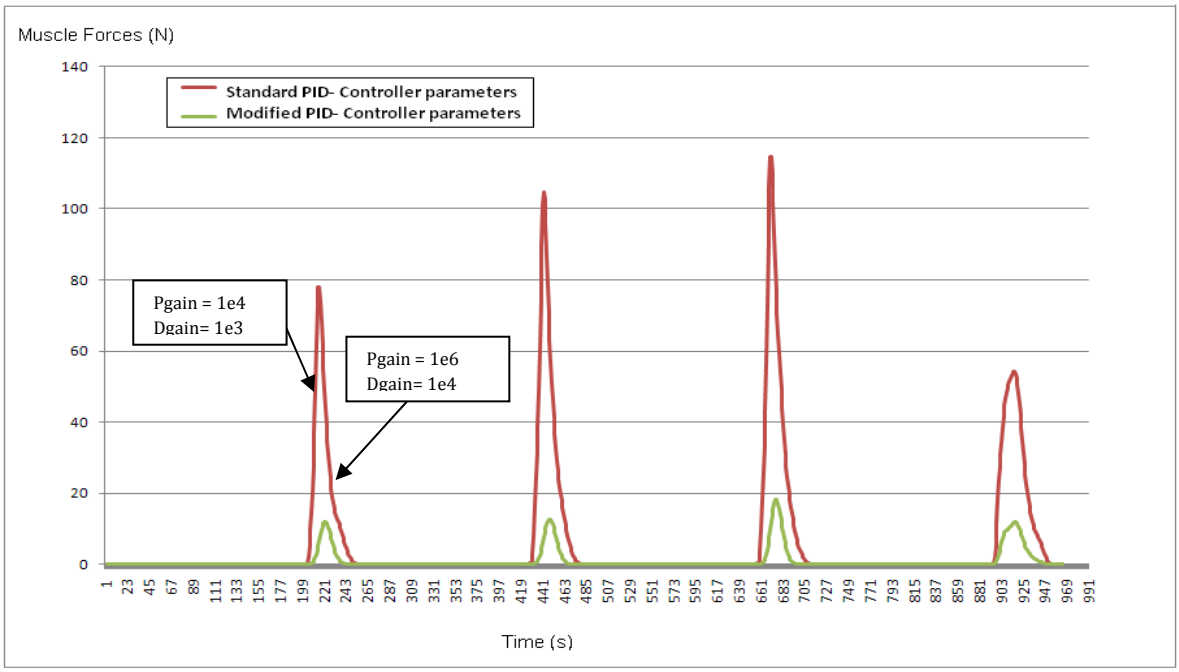


Figure 3. 6: An illustration of the effect on the estimated muscle force when changing the Pgain and Dgain parameters. The forward dynamics simulation, of a cerebral palsy child, is performed using two sets of these parameters values: the green curve corresponds to (Pgain = $1e^6$; Dgain= $1e^4$) and the red curve corresponds to (Pgain = $1e^4$; Dgain= $1e^3$)

The PID controller's parameters serve usually to decrease numerical errors, oscillations tendency in tracking the motion and ensure computational convergence [Lifemodeller 2008]. Consequently the accuracy of the musculoskeletal modeling' results greatly depend on these parameters. The choice of the suitable set of values has been defined when referring to muscle forces gathered from literature during normal gait [*Delp 1990, Kromer 1993, Pederson 1997, Marin 2001, Fraysse 2009*], presented in Table 3.8.

Table 3. 8: Predicted maximal muscle forces during normal gait

	Delp 1990	Frayess 2009	Kromer 1993	Peaderson 1997	Marin 2001
Tibialis Anterior	-	-	300	-	-
Biceps Femoris	1120	960	600	-	672
Rectus Femoris	800	477	500	370	
Gastrocnemius Lateralis	1115	781	600	240	892
Gastrocnemus Medialis	490			-	196
Vastus medialis	-	-	-	-	-

3.1.3. Calibration procedure: MSM Parameter's evaluation

To sum up, the calibration procedure can be recapitulated as follows (Table 2.2):

- **Motion agents:** the weight of markers placed a bony surface is fixed at 90% and weight of markers placed at bone parts with highly skin movement artifact (thigh and shank) is fixed at 50%.

- **Joint center positions:** calculated and placed manually according to predictive method of Davis (1991).

- **Stiffness joint parameter:** determined using the graph of Angle-Torques in the sagittal plane. In the other planes, the stiffness joints remains to the standard values

- **GRF:** the GRF measurements are introduced as inputs of the MSM. If not, contact parameters will be manually determined after several essays to avoid contact instabilities.

- **Training parameters:** Pgain is fixed at $1e^4$ and Dgain is fixed to $1e^3$ ($Time^{-1}$ unit)

3.2. Musculoskeletal Modeling results

In this section, results from the standard musculoskeletal model, the calibrated musculoskeletal model and the kinematic model are balanced for different populations in this project. In order to verify the accuracy of the introduced MS models, the kinematics, the ground reaction force and EMG activation patterns obtained from experiments are compared with their correspondent results obtained from the models.

3.2.1. Normal Gait: Healthy Adults vs Healthy Children

Figures below balanced results of the rescaled generic standard and the generic calibrated musculoskeletal models of healthy adult and healthy children (Figure 3.7- 3.13). This section deals will the normal gait of a healthy adult to test whether the standard musculoskeletal model give good agreements when studying healthy adults.

❖ **Joint Kinematics**

For the normal gait, results from musculoskeletal models of healthy subjects, joint kinematics GRF and muscle forces are compared with the kinematic model (the reference data in the current study). For healthy adult population, a minimum mean correlation coefficient of 0.79 for the standard generic musculoskeletal model and a minimum mean correlation coefficient of 0.89 for the calibrated musculoskeletal model were founded. Whereas, minimum mean correlation coefficients of 0.76 and 0.89 are respectively founded for standard and calibrated musculoskeletal models for healthy children population (Table 3.9-3.12).

Table 3. 9: Correlation coefficient values comparing standard and calibrated MSM results vs. kinematic model as reference

	Sagittal kinematics (Angles)						**Ground reaction Forces**	
	Adults			**Children**				
	Hip	**Knee**	**Ankle**	**Hip**	**Knee**	**Ankle**	**Adults**	**Children**
Standard MSM vs Kinematic model	0. 86 (0. 04)	0. 81 (0.05)	0.79 (0. 08)	0.83 (0.08)	0.84 (0.05)	0.76 (0.11)	0.85 (0.12)	0.79 (0.07)
Calibrated MSM vs Kinematic model	0. 94 (0. 02)	0. 92 (0.03)	0.89 (0.09)	0.90 (0.02)	0.93 (0.04)	0.89 (0.02)	1	1

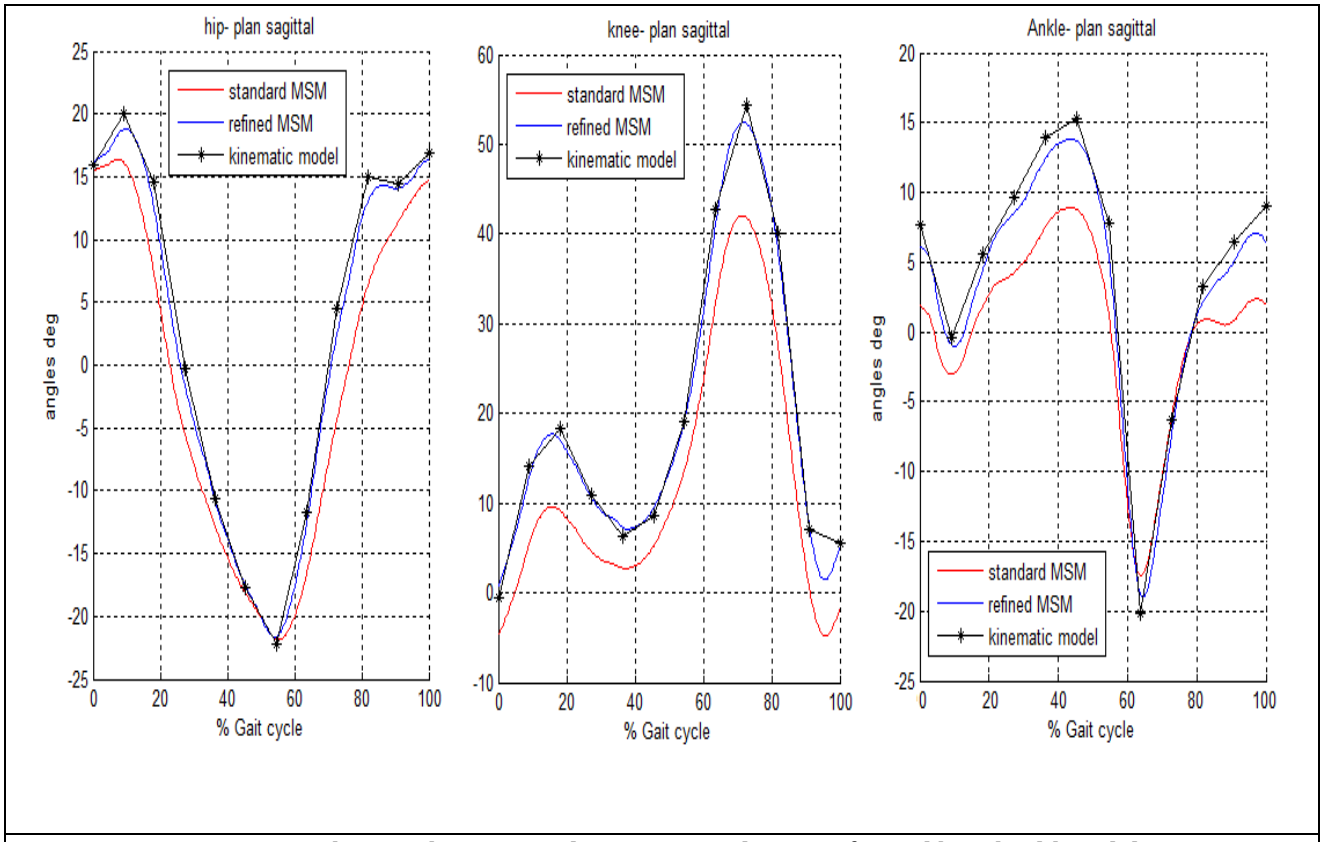


Figure 3. 7**:** The joint kinematics during a normal gait performed by a healthy adult. Comparison between the three models developed in the current study: the red curve corresponds to the standard lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model (the reference)

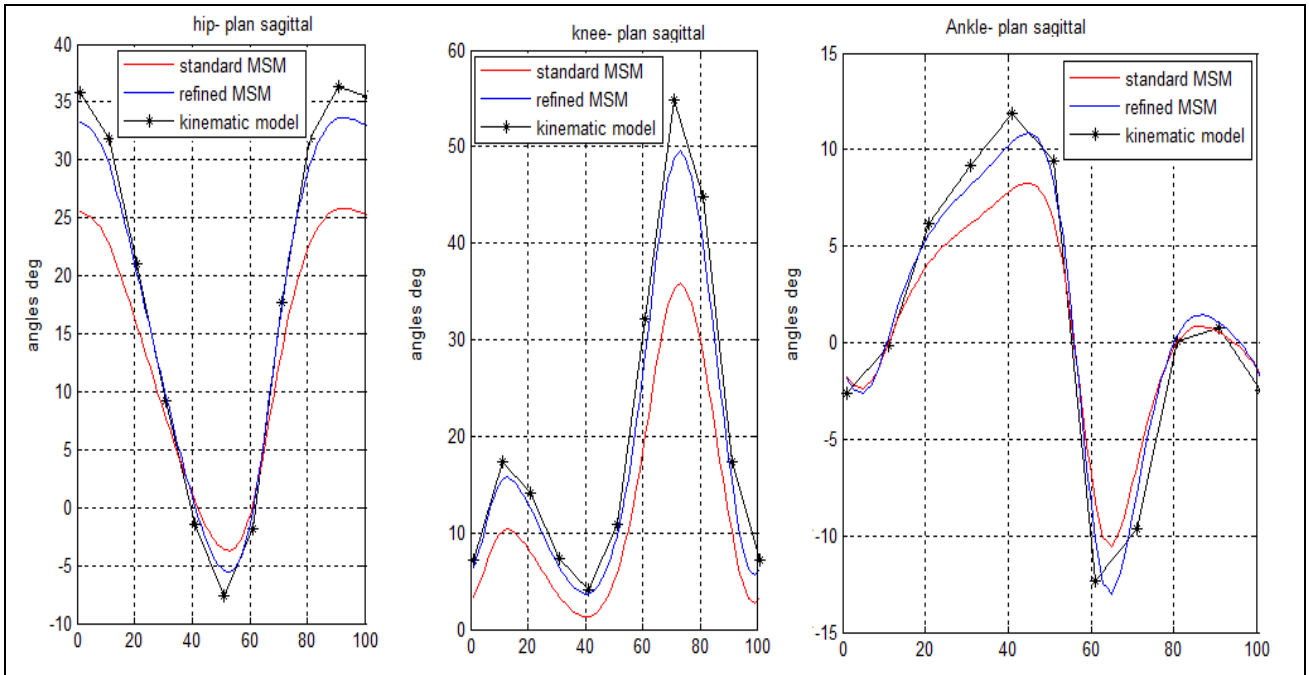


Figure 3. 8**:** The joint kinematics during a normal gait performed by a healthy child. Comparison between the three models developed in the current study: the red curve corresponds to the standard lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model (the reference)

❖ **The Ground Reaction Forces**

In general, using the standard parameters, MSM may lead to contact instabilities and inaccuracy during the first/two first gait cycles, as shown in figure below (Figure 3.9). For this reason, the third gait cycle was used in this comparison.

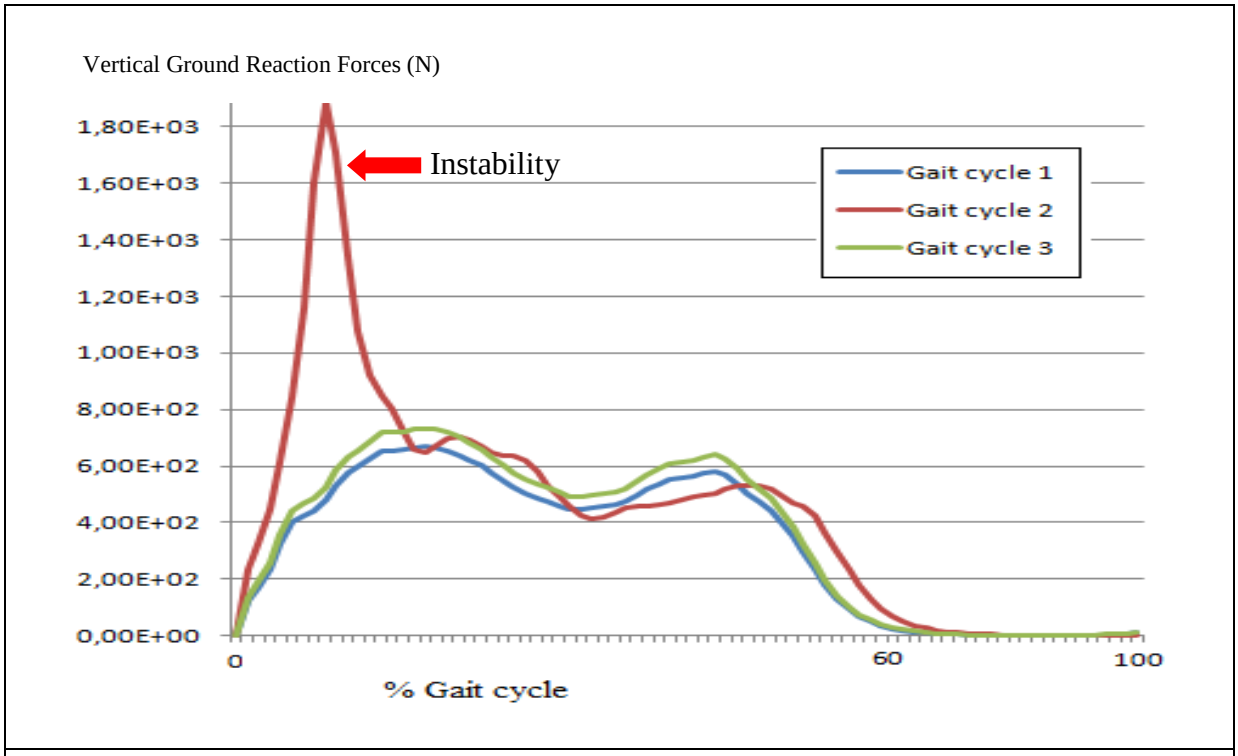
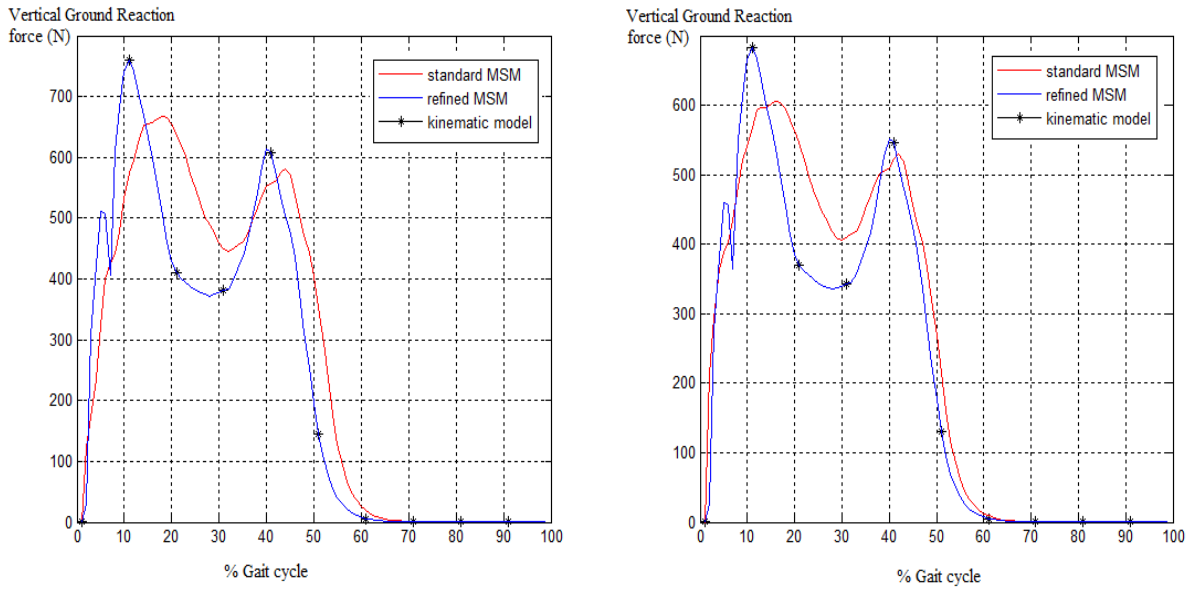


Figure 3. 9**:** The vertical ground reaction forces during three gait cycles, estimated using the standard musculoskeletal lifemod model.

Concerning the ground reaction forces, even if we did not introduce the measured data, as in the calibrated MSM model, the estimated results shows that the standard contact-model introduced in the LifeMOD may slightly predict acceptable contact forces during normal gait. Correlation coefficients of 0.85 and 0.79 are respectively achieved for healthy adults and children population (table 3.9).



(a) Healthy adult (b) Healthy child

Figure 3. 10**:** Vertical ground reaction forces– healthy adult (a) and healthy children (b). Comparison between the current studies (models): the red curve corresponds to the standard Lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model (the reference)

Table 3. 10: Variation between the kinematic model and Standard and Calibrated MSM results in normal gait (** represent a significance level <1% and * represents a significance level <5%)

	Healthy Adults				Healthy Children			
	Kinematic model vs Standard MSM		**Kinematic model vs Calibrated MSM**		**Kinematic model vs Standard MSM**		**Kinematic model vs Calibrated MSM**	
Kinematic determinants	Max var.	% of var.	Max var.	% of var.	Max var.	% of var.	Max var.	% of var.
P1:Hip flexion at Initial Contact	15 (2.6)**	15.2%	4 (2.1)	6.1%	12 (3.2)**	10.3%	4 (1.8)	3.1%
P2: Hip extension Maximum	13 (3.5)**	16.4%	6 (3.4)*	8.1%	5 (1.2)**	8.4%	2 (0.4)*	2.1%
P3: Knee flexion at Initial Contact	5 (4.1)*	8.5%	3(2.3)	4.6%	4 (2.0)*	6.8%	1(0.3)	4.2%
P5: Knee flexion Max	9 (5.3)**	12.1%	5 (4.6)*	7.5%	7 (2.3)**	8.1 %	5 (3.4)*	3.5%
P4: Knee flexion at mid stance phase	24 (6.1) **	22.3%	8 (5.3)*	11.6%	19(3.4) **	20.1%	8 (2.3)*	9.3%
P6: Dorsiflexion at Initial Contact	8 (2.8)	10.8%	4 (2.4)*	4.4%	1.6 (0.8)	4.2%	1.2 (0.4)*	2.4%
P7: Dorsiflexion Minimum	4 (3.1)	5.4%	2 (1.5)	3.4%	5 (2.1)	4.4%	2 (1.3)	1.4%
P8: Dorsiflexion Maximum	17 (2.4)**	18.6%	6 (2.6)	7.2%	6(2.2)**	6.8%	3 (1.6)	2.2%
P9: Plantarflexion Maximum	7 (3.4)	5.3%	4 (1.4)	4.6%	2.1 (1.3)	3.9%	2 (1.4)	1.9%
GRF Parameters (N/Kg)								
P1: 1st peak GRF	4 (2.3)	5.1%	-	-	6 (2.8)	4.8%	-	-
P2: second peak GRF	8 (3.9)*	9.1%	-	-	9.2 (4.5)*	10.2%	-	-
P3: min valley GRF	12 (2.08)*	9.8%	-	-	10.5 (3.8)*	12.5%	-	-

❖ **Muscle Activation and Forces**

Focusing on predicted muscle activities, figure 3.11 compares the estimated muscle activities and corresponding measured EMG signals. Muscular forces obtained from the MS models are evaluated based on a qualitative comparison with the corresponding EMG patterns obtained from experimental measurements. Table 3.11 shows the correlation coefficient between envelop of measured EMG and simulated values, regarding the on-off times of activations.

This comparison shows that the predicted muscle activities present a little inter-muscle variability. But, results from both musculoskeletal models predict confident muscle activation, especially to detect the beginning of activation. However, the standard model does not predict exact activation times. The off time is slightly earlier than the activation estimated by the calibrated model and EMG measurements (figure 3.11-3.12). For young population, results are different. The correlations coefficients present an amount inferior than 0.7 (Table 3.11). It is important to revealed/ precise that these comparisons and interpretations are highly dependent on the quality of EMG measurements.

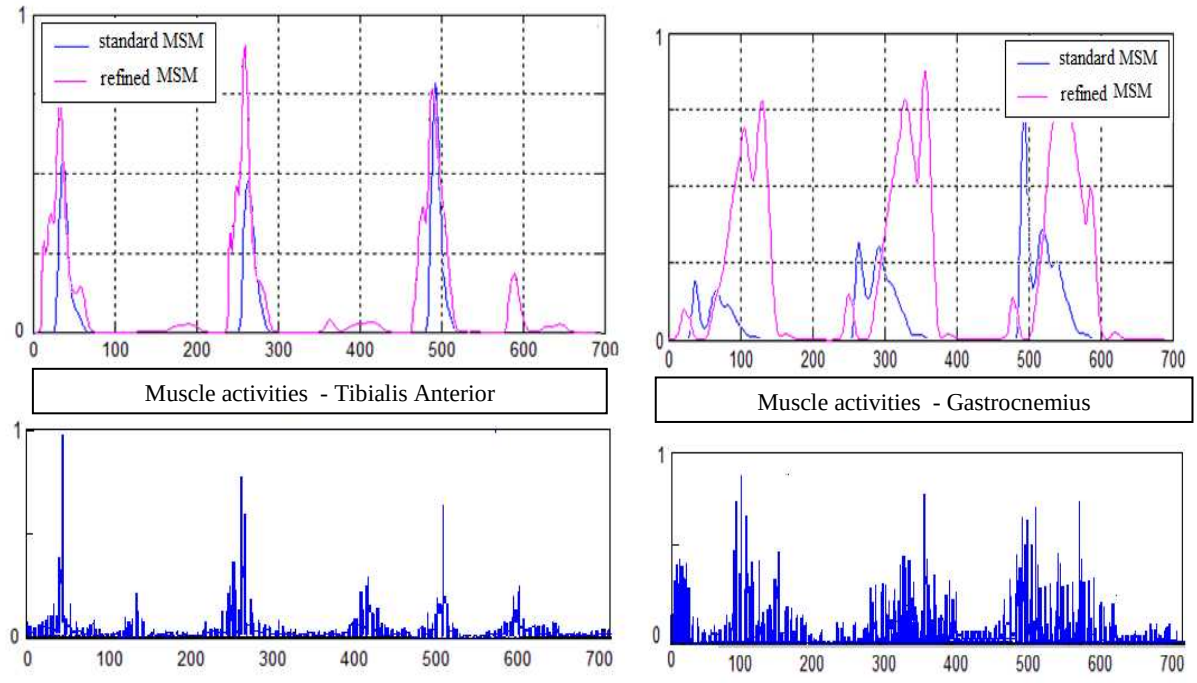


Figure 3. 11**:** comparison between the EMG measurements and predicted muscle activities using the standard model (blue curve) and the calibrated MSM (magenta curve) for healthy adult

Table 3. 11: Correlation coefficient values comparing EMG measurements to predictive muscle activities using the standard and calibrated MSM results.

	The Standard model		The calibrated model	
	Healthy adults	Healthy children	Healthy adults	Healthy children
Tibialis anterior	0.75	0.72	0.82	0.77
Biceps femoris	0.81	0.68	0.89	0.72
Rectus femoris	0.62	0.47	0.78	0.62
Gastrocnemius Lateralis	0.68	0.58	0.75	0.74

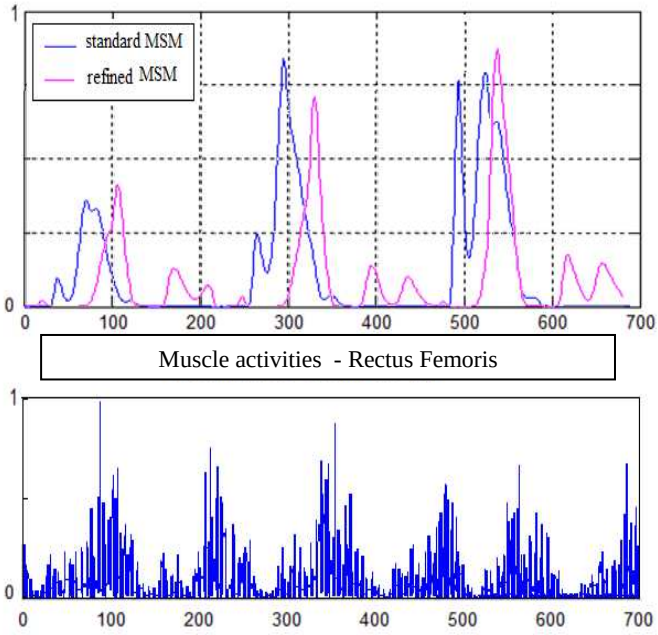


Figure 3. 12**:** illustration of the comparison between the EMG measurements and predicted muscle activities using the standard model (blue curve) and the calibrated MSM (magenta curve) for healthy children

The calibrated musculoskeletal model gives more acceptable amount of muscle forces, compared to literature data (Table 3.12).

Table 3. 12: Recapitulative results of predicted muscle forces using the standard and calibrated musculoskeletal model of normal gait, compared to literature data

	Healthy adults (N)		Healthy children (N)		Litterature data
	The standard model	The calibrated model	The standard model	The calibrated model	(Adult-normal gait)
Tibialis anterior	0.32	77	0.08	94	**300N ***
Biceps femoris	0. 17	186	0.12	122	**600N -960N***
Rectus femoris	46.2	380	1.2	240	**300N- 477N***
Gastrocnemius Lateralis	0.28	220	0.49	153	**420N –781N***

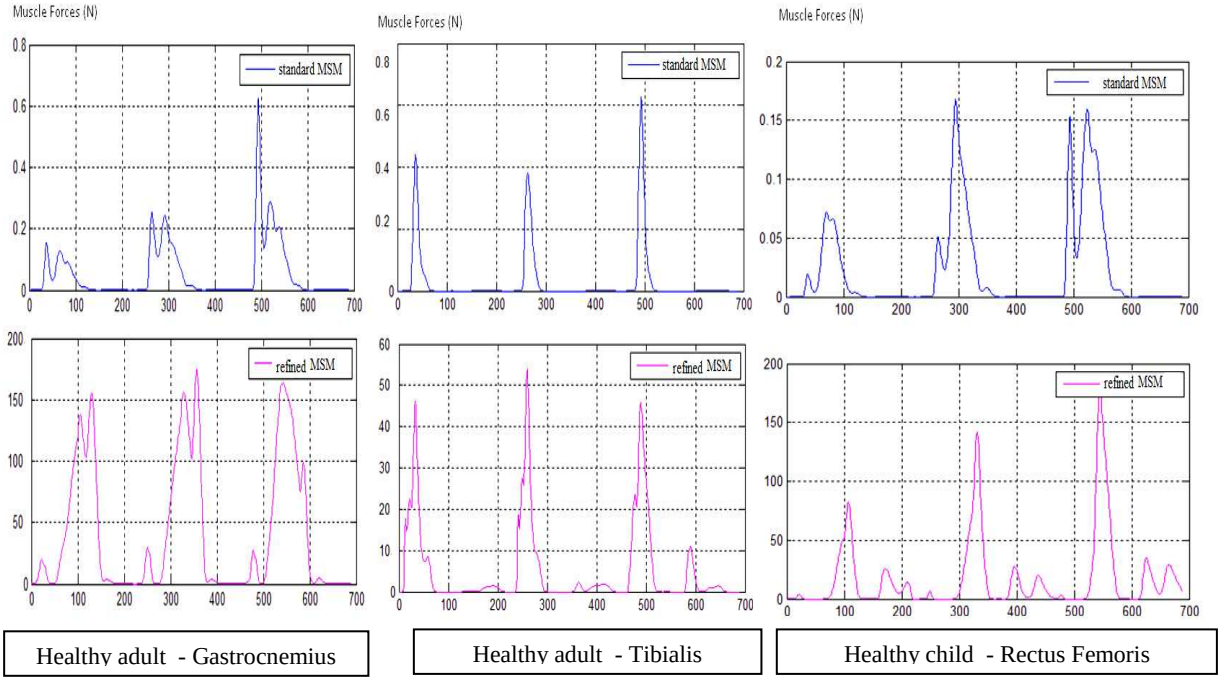


Figure 3.13**:** Example of predicted muscle forces with the standard model (blue curve) and the improved MSM (magenta curve) for healthy adults and children

3.2.2. Musculoskeletal modeling for simulated gaits

In this part, we aimed to test the performance of a musculoskeletal modeling when introducing an altered motion control. For this, the alteration of muscle activities are defined when healthy subjects performed imitated pathological postures observed in CP children. Results from the standard musculoskeletal model, the calibrated musculoskeletal model and the kinematic model are balanced for healthy population and for each simulated pathological posture.

3.2.1.2. Pertinence of simulated pathological gait patterns

Figures 3.14-15 shows that healthy subjects have performed similar gaits when imitating pathological crouched and jump gait patterns. Healthy adults and children have achieved a repeatable posture during different trials; the mean CMC values are higher than 0.8, except a slight inter-subject variability of the sagittal rotation of the knee during the simulated crouch gait. However, in comparison to reference literature data (*Gage 1994, Ganley et al. 2005, Schwarz et al. 2008, Rodda et al. 2004, Rozumalski et al. 2008*), the results were contrasted, the mean CMC values vary between 0.66 and 0.99.

According to Rozumalski et al. (2008), healthy subjects simulated a severe form of crouch gait, with large amplitudes, confirmed by a CMC value of 0.66 and 0.72. For simulated jump gait, knee kinematics presented a delay of knee flexion during loading response of the stance phase and a less knee flexion amplitude during the terminal swing. These differences were quantified by CMC values of 0.67 and 0.68 (table 3.14).

As conclusion, healthy subjects had achieved extreme characteristics of pathological gait patterns in terms of range of motion and posture and that simulated gait patterns are representative of the cerebral palsy pathology.

Table 3.13: Inter-subject CMC values for joint kinematics for normal and imitated pathological gait patterns

	Sagittal joint kinematics : Inter-subject variability					
	Ankle		**Knee**		**Hip**	
	Adults	**Children**	**Adults**	**Children**	**Adults**	**Children**
Normal gait	0.95	0.92	0.98	0.94	0.99	0.90
Simulated jump gait	0.90	0.93	0.93	0.89	0.93	0.87
Simulated Crouch gait	0.95	0.93	0.76	0.82	0.80	0.84

Table 3.14: CMC values of joint kinematics comparing current study data vs. literature data

		Validity of the simulated gaits current study data vs. literature data					
		Ankle		**Knee**		**Hip**	
		Adults	**Children**	**Adults**	**Children**	**Adults**	**Children**
Normal gait	Gage (1994)	0.84	0.90	0.99	0.96	0.95	0.92
	Ganley et al. (2005)	0.73	0.84	0.98	0.94	0.97	0.94
	Schwarz et al. (2008)	0.89	0.83	0.99	0.92	0.94	0.90
Simulated jump gait	Lin et al. (2000)	0.80	0.79	0.67	0.72	0.97	0.95
	Rodda et al. (2004)	0.84	0.89	0.66	0.67	0.91	0.94
Simulated Crouch gait	Lin et al. (2000)	0.83	0.78	0.85	0.79	0.90	0.88
	Rodda et al. (2004)	0.93	0.91	0.59	0.65	0.96	0.90
	Rozumalski et al. (2008) (severe)	0.93	0.90	0.60	0.77	0.98	0.94

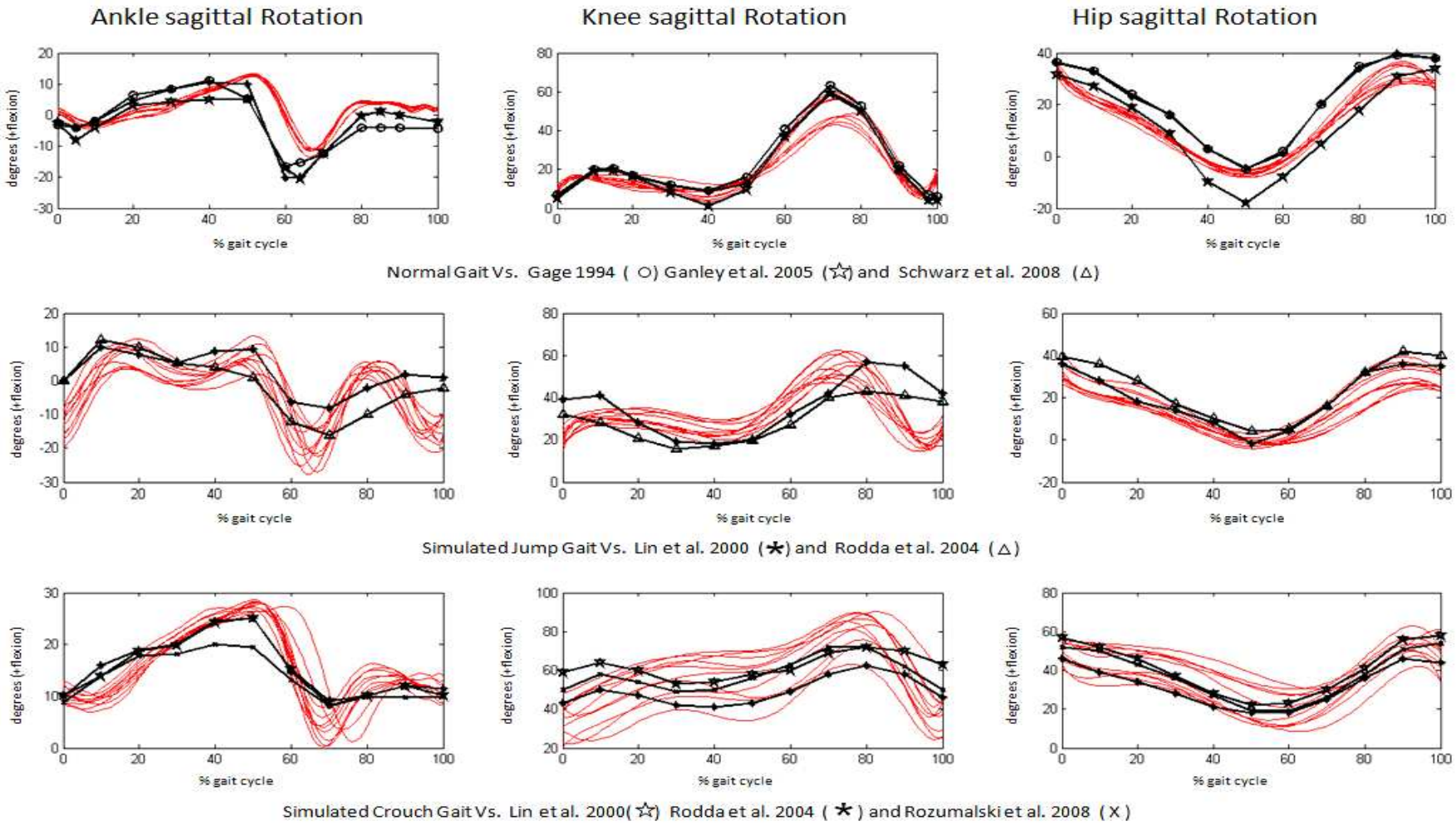


Figure 3. 14**:** Joint Sagittal kinematics for the normal and simulated jump and crouch gaits for healthy adult population

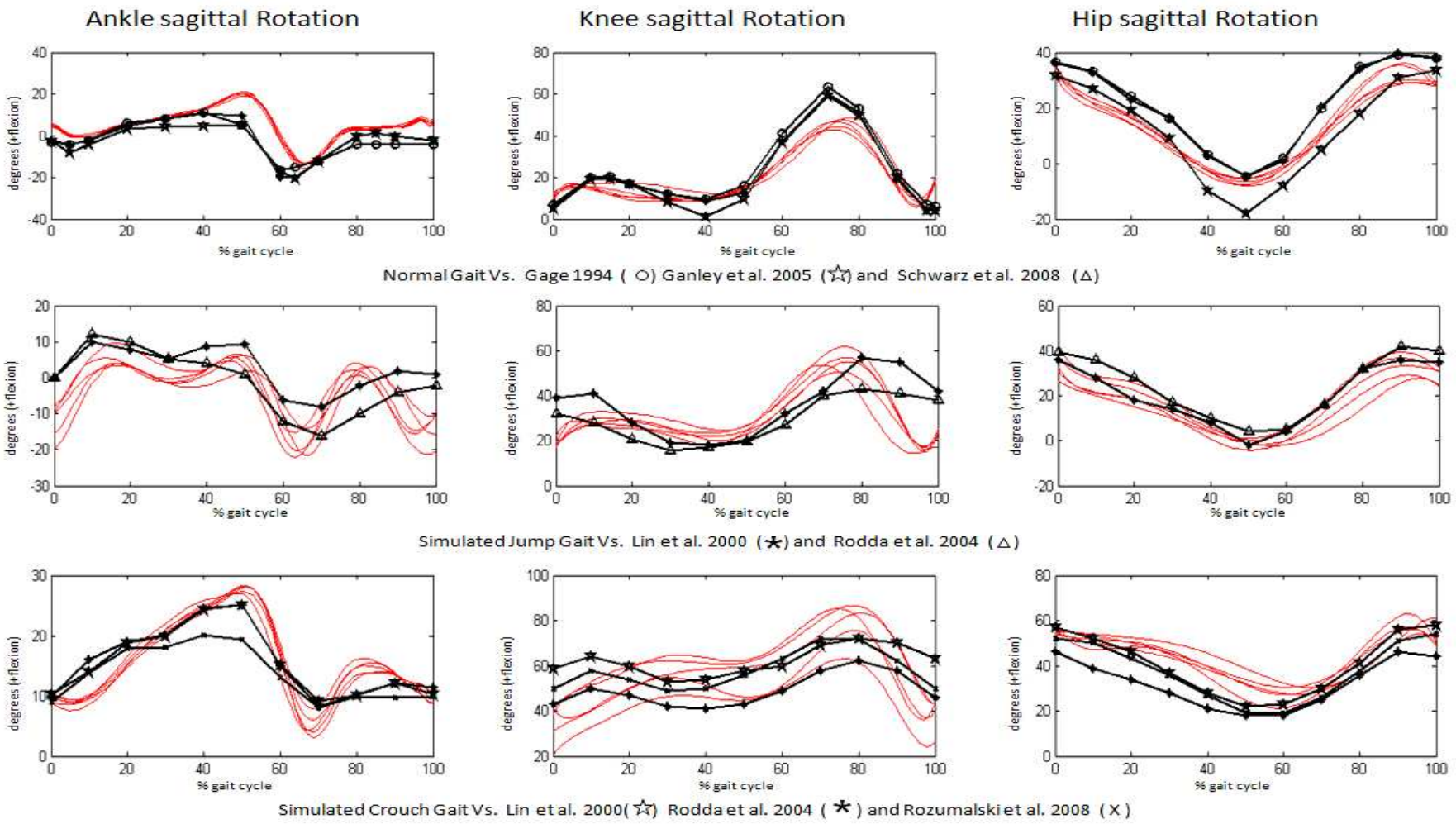


Figure 3. 15**:** Joint Sagittal kinematics for the normal and simulated jump and crouch gaits for healthy children population

3.2.1.2. Healthy Adults: standard model versus calibrated MSM

The two musculoskeletal models, the standard and calibrated rescaled generic models, have been developed for healthy populations and for jump and crouch gaits. Figures below compared the results from these studies (Figure 3.16- 3.23).

- **Joint Kinematics**

From kinematic point of view, compared to the plugin-gait model, the calibrated musculoskeletal model can predict more accurately the joint kinematics then the standard musculoskeletal model, independently from the population of study and the imitated pathological postures (table 3.15-16). Concerning standard musculoskeletal models, correlation coefficients around 0.75 are founded for each joint and for both healthy adults and children. But, correlation coefficients of 0.67 are founded for predicted knee kinematics in crouch gait and of 0.69 for predicted ankle joint in simulated jump gait. As regards to the calibrated MSM, the correlation coefficients exceed 0.9 for jump gaits, and 0.88 for crouch gaits. The correlation coefficients of ankle joint are enhanced but still with low values. Consequently, for children, the curves do not fit exactly those from kinematic models.

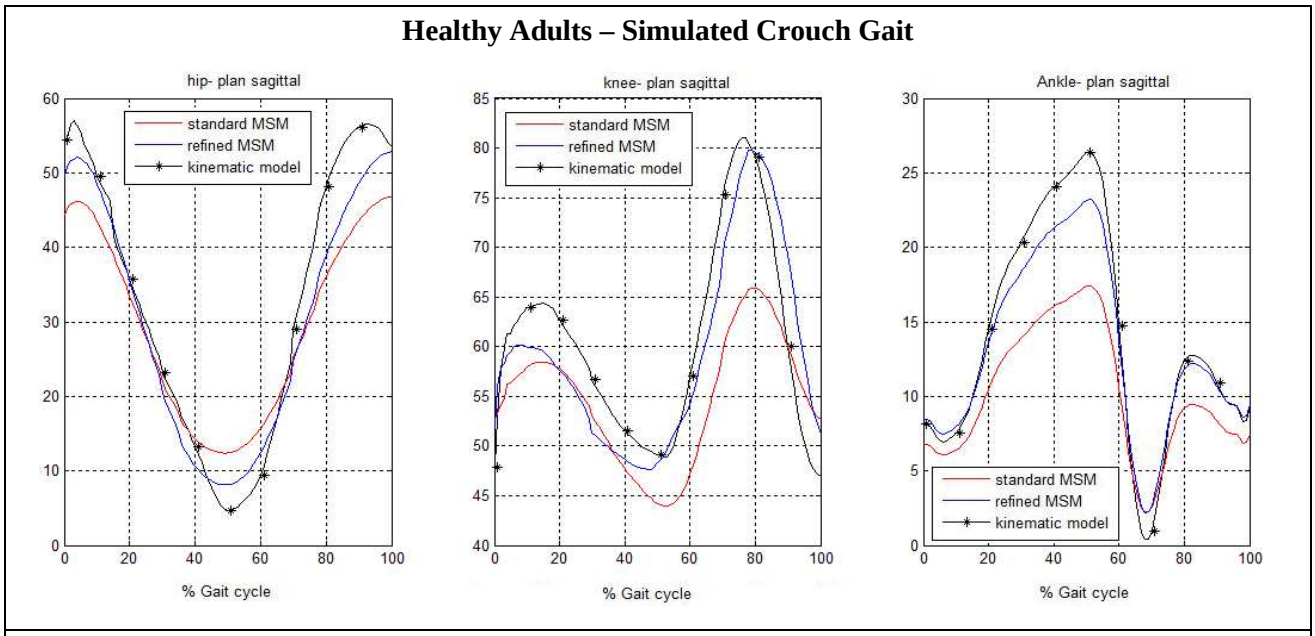


Figure 3. 16**:** comparison between Joint angles kinematics estimated through musculoskeletal models and the kinematic model for healthy adults imitating crouch gait:
the red curve corresponds to the standard Lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model.

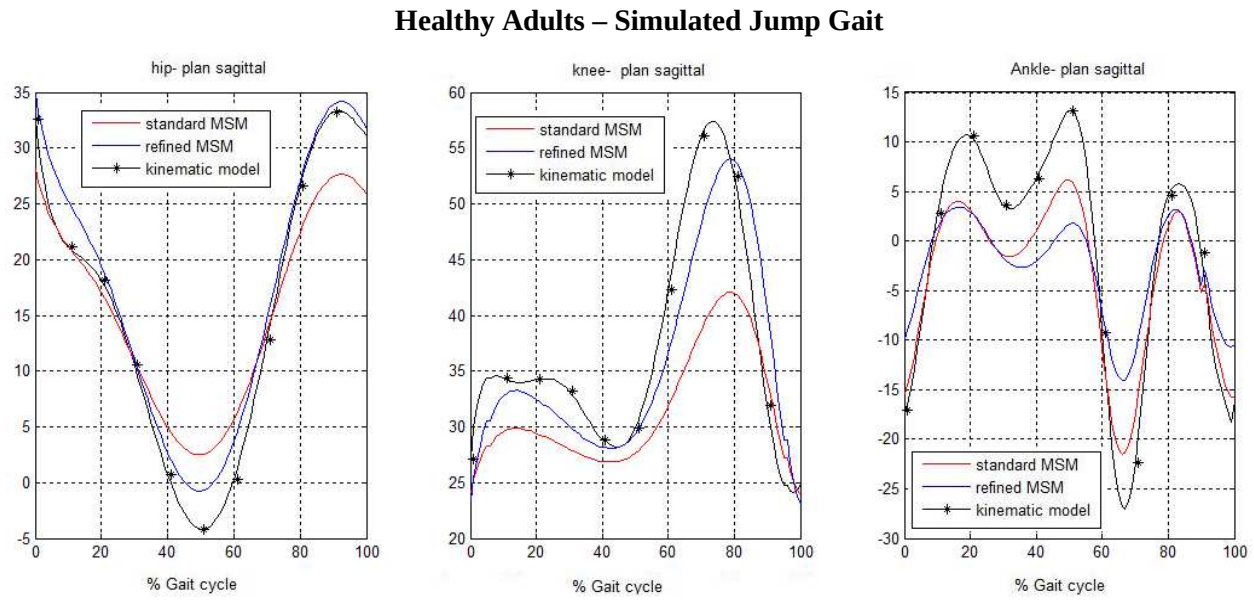


Figure 3. 17**:** comparison between Joint angles kinematics estimated through musculoskeletal models and the kinematic model for healthy adults imitating jump gait:
the red curve corresponds to the standard Lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model.

Table 3. 15: Correlation coefficient values comparing standard and calibrated MSM results vs. kinematic model in the case of simulated jump gait

	Sagittal kinematics (Angles)						**Ground reaction Forces**	
	Adults			**Children**			**Adults**	**Children**
	Hip	**Knee**	**Ankle**	**Hip**	**Knee**	**Ankle**		
Standard MSM vs Kinematic model	0.84	0. 75	0.72	0.80	0.78	0.79	0.82	0.78
Calibrated MSM vs Kinematic model	0. 98	0. 92	0.88	0.93	0.94	0.74	1	1

Table 3. 16: Correlation coefficient values comparing standard and calibrated MSM results vs. kinematic model in the case of simulated crouch gait

	Sagittal kinematics (Angles)						**Ground reaction Forces**	
	Adults			**Children**			**Adults**	**Children**
	Hip	**Knee**	**Ankle**	**Hip**	**Knee**	**Ankle**		
Standard MSM vs Kinematic model	0. 78	0.68	0.72	0.75	0.67	0.68	0.89	0.84
Calibrated MSM vs Kinematic model	0. 94	0.89	0.91	0.89	0.91	0.72	1	1

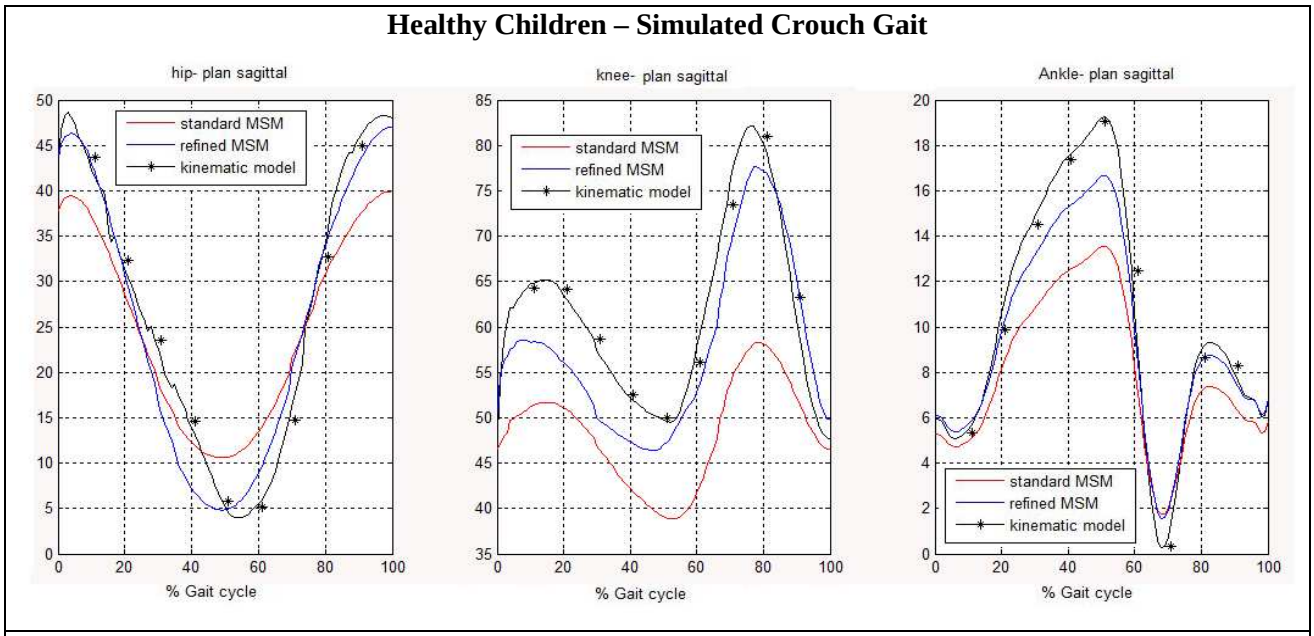


Figure 3. 18**:** comparison between Joint angles kinematics estimated through musculoskeletal models and the kinematic model for healthy children imitating crouch gait:
the red curve corresponds to the standard Lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model.

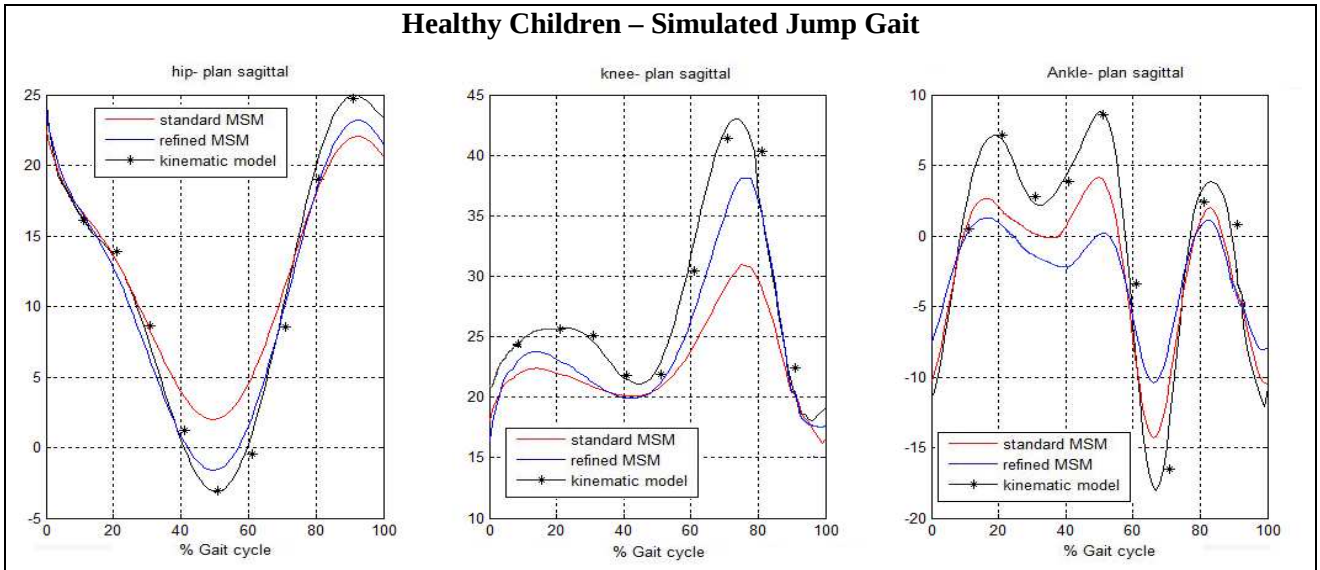
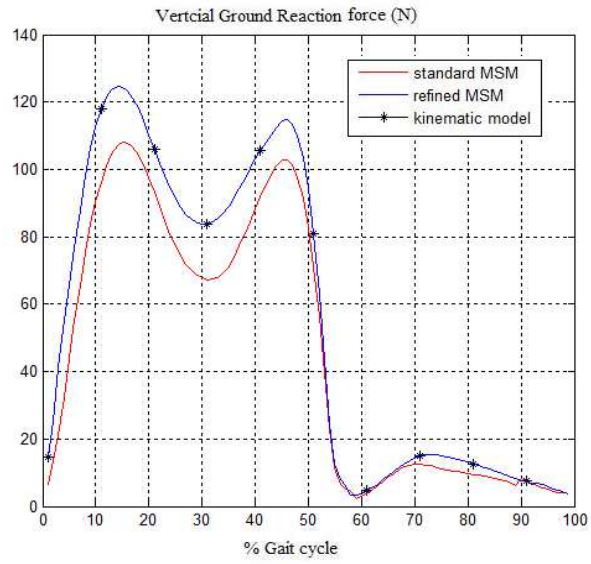


Figure 3. 19**:** comparison between Joint angles kinematics estimated through musculoskeletal models and the kinematic model for healthy children imitating jump gait:
the red curve corresponds to the standard Lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to the kinematic model.

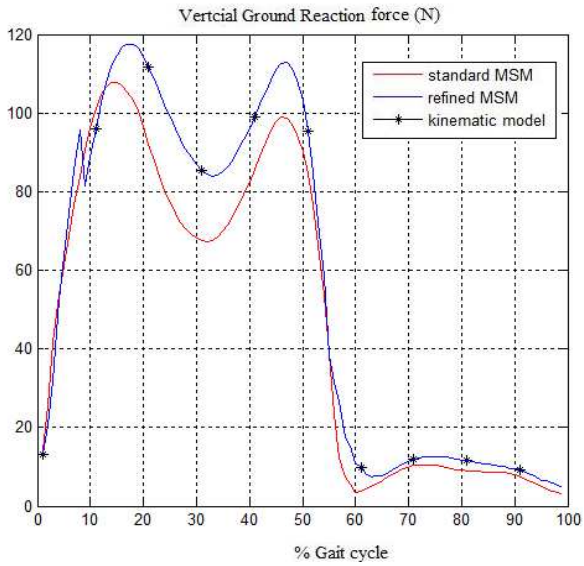
❖ The Ground Reaction Forces

Focusing on ground reaction forces, the estimated forces during simulated pathological postures are correlated to the measured ground reaction forces. Correlation coefficients of 0.89 and 0.84 are respectively achieved for healthy adults and children population in

simulated crouch gait. For jump gaits, correlation results are lower with 0.82 and 0.78, respectively for healthy adults and children.

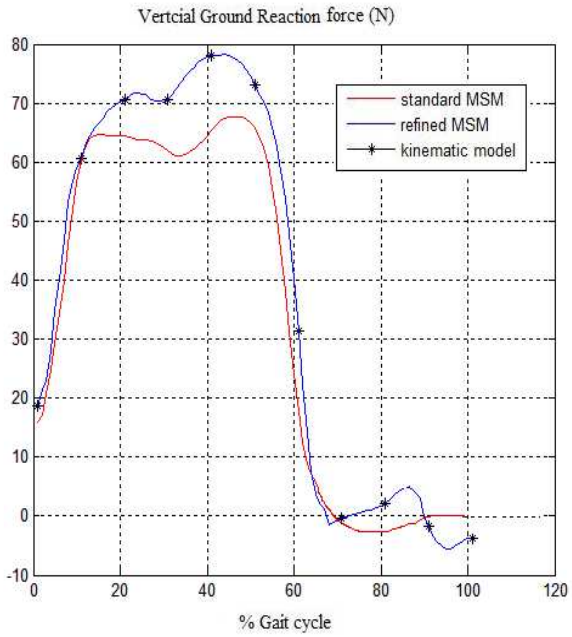


(a) healthy adults

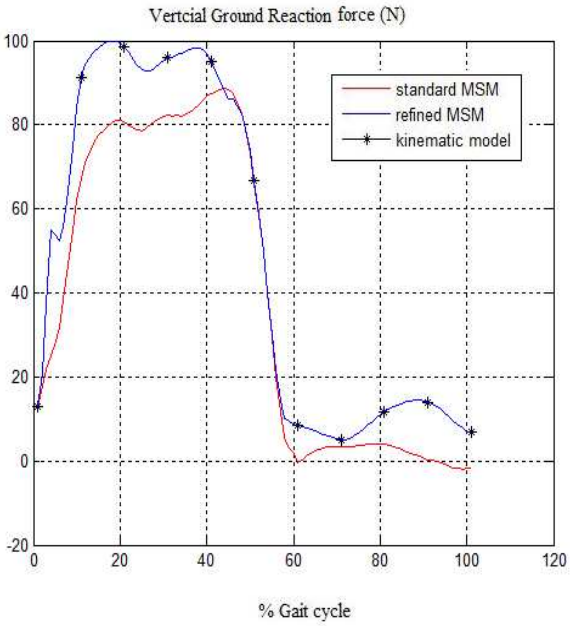


(b) healthy children

Figure 3. 20**:** Vertical ground reaction forces during simulated jump gaits– healthy adult (a) and healthy children (b). Comparison between the current studies (models): the red curve corresponds to the standard Lifemod model, the blue curve corresponds to GRF introduced as input in the calibrated MSM and the black curve corresponds to GRF measurements.



(a) healthy adults



(b) healthy children

Figure 3. 21**:** Vertical ground reaction forces during simulated crouch gaits– healthy adult (a) and healthy children (b). Comparison between the current studies (models): the red curve corresponds to the standard Lifemod model, the blue curve corresponds to GRF introduced as input in the calibrated MSM and the black curve corresponds to GRF measurements.

❖ **Muscle activation and forces**

Figure 3.22 and figure 3.23 are illustrations of results from musculoskeletal models and represent the predicted muscle activation for the principle muscles on the lower limbs compared to the rectified EMG measurements, respectively, when healthy adults have performed a jump gaits and healthy children had performed a crouch gaits.

Muscle activations, obtained from the MS models, are compared qualitatively with the corresponding EMG patterns obtained from experimental measurements. Results of correlation tests, comparing the envelop of measured EMG and simulated muscle activations, are presented in tables 3.17-18.

In the case of simulated jump gaits, poor correlations of 0.3 have been founded for the Rectus Femoris and Tibalis Anterior on the standard MSM. The calibrated MSM has not a big influence in predicting muscle activation, the correlation insignificantly increased to 0.38. For the Biceps Femoris and Gastrocnemius, the results from the standard and calibrated MSM are likely correlated with an amount of 0.61 and 0.64 respectively.

Table 3. 17: Correlation coefficient values comparing EMG measurements to predictive muscle activities using the standard and calibrated MSM results in case of simulated jump gaits

	The Standard model		**The calibrated model**	
	Healthy adults	**Healthy children**	**Healthy adults**	**Healthy children**
Tibialis anterior	0.33	0.35	0.41	0.38
Biceps femoris	0.65	0.55	0.69	0.61
Rectus femoris	0.22	0.32	0.34	0.42
Gastrocnemius Lateralis	0.64	0.58	0.67	0.63

Table 3. 18: Correlation coefficient values comparing EMG measurements to predictive muscle activities using the standard and calibrated MSM results in case of simulated crouch gaits

	The Standard model		**The calibrated model**	
	Healthy adults	**Healthy children**	**Healthy adults**	**Healthy children**
Tibialis anterior	0.40	0.37	0.46	0.43
Biceps femoris	0.57	0.62	0.63	0.67
Rectus femoris	0.12	0.17	0.29	0.22
Gastrocnemius Lateralis	0.23	0.19	0.33	0.39

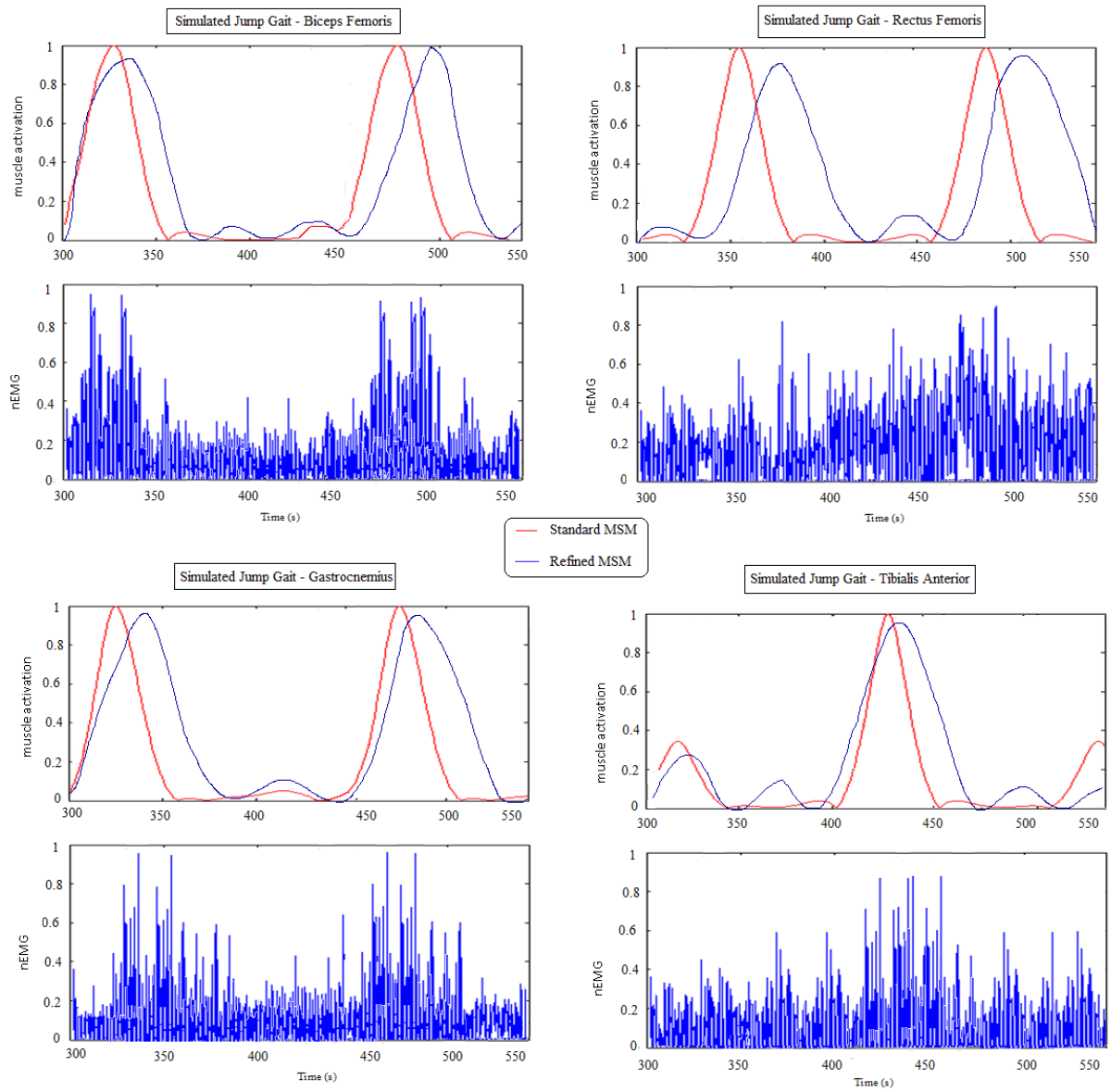


Figure 3. 22**:** comparison between the EMG measurements and predicted muscle activities using the standard model (blue curve) and the calibrated MSM (magenta curve) for healthy adult performing jump gaits

In the case of simulated crouch gait, correlations values of the Rectus Femoris and Gastrocnemius of the standard MSM are lower than 0.2. These values increased to 0.3 with the calibrated MSM. For the Tibialis Anterior, the correlation coefficients are respectively of 0.37 and 0.43 for the two MS models. For the Biceps Femoris and Gastrocnemius, correlation results present coefficients of 0.60 and 0.64 for the standard and calibrated MS respectively as a mean of all healthy populations.

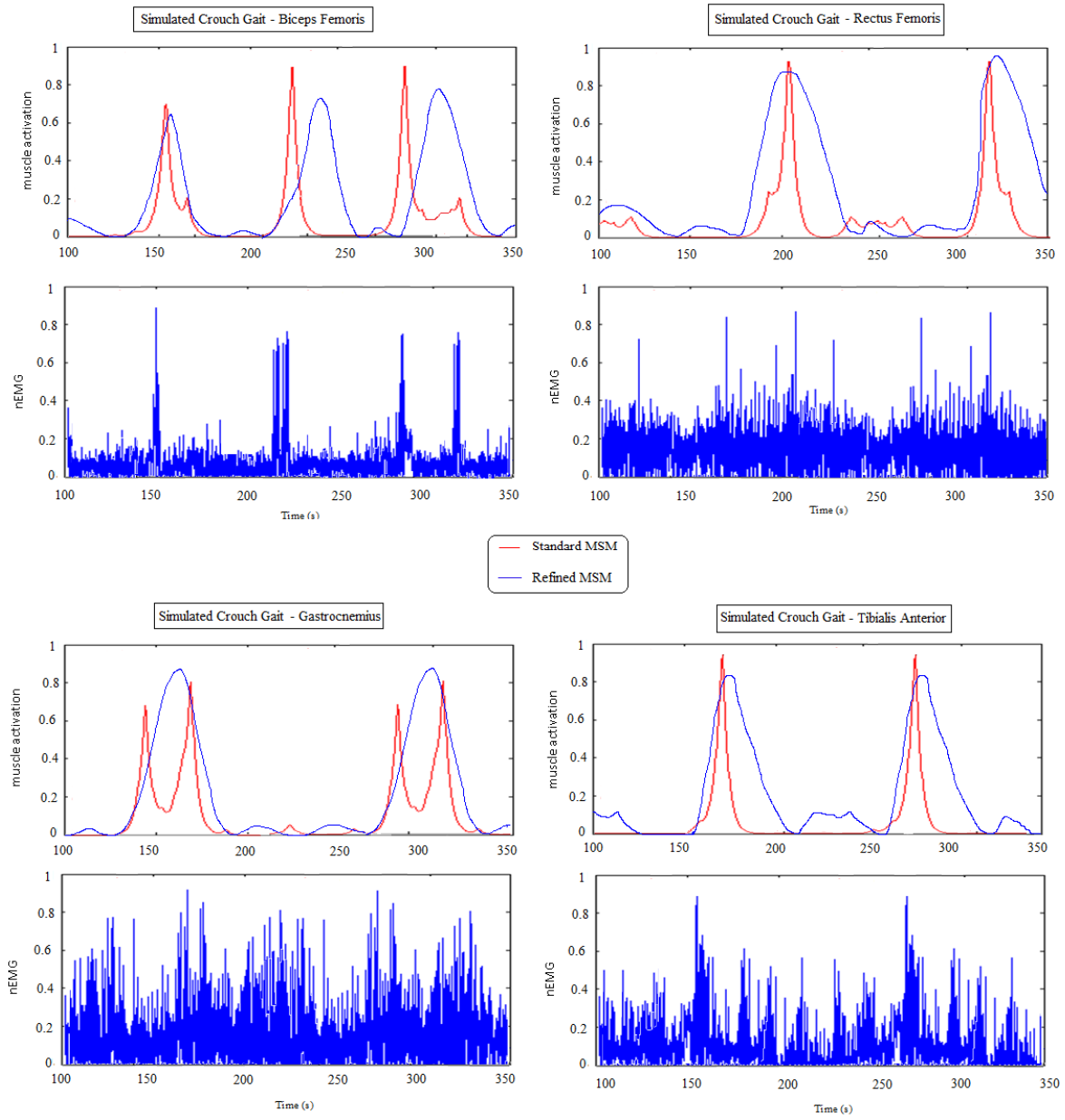


Figure 3. 23**:** illustration of the comparison between the EMG measurements and predicted muscle activities using the standard model (blue curve) and the calibrated MSM (the magenta curve) for healthy children, performing crouch gaits.

About the predicted muscle forces, for both healthy populations, the standard rescaled musculoskeletal model gives a very low amount of muscle forces. The calibrated musculoskeletal model gives more acceptable amount of muscle forces. The maximal muscle forces are presented in table 3.19.

Table 3. 19: Maximum muscle forces (N) predicted using the standard and calibrated MSM results in case of simulated jump and crouch gaits

	Healthy adults				Healthy children			
	The standard model		The calibrated model		The standard model		The calibrated model	
	Jump	Crouch	Jump	Crouch	Jump	Crouch	Jump	Crouch
Tibialis anterior	0.15	0.18	120	97	$0.35*10^{-3}$	0.09*10-3	68	79
Biceps femoris	0.42	0.38	170	215	$1.20*10^{-3}$	0.16*10-3	84	125
Rectus femoris	0.25	0.42	250	380	$2.84*10^{-3}$	0.87*10-3	158	234
Gastrocnemius Lateralis	0.47	0.37	290	355	$1.44*10^{-3}$	0.56*10-3	124	184

3.2.2. Musculoskeletal Modeling of CP children

The current section will be dedicated to present results from musculoskeletal modeling of cerebral palsy children. It is essential to point out that muscle spasticity and bone deformities have not been included in this study. Results from the standard musculoskeletal model, the calibrated musculoskeletal model and the kinematic model are balanced for Cerebral palsy groups with crouch, recurvatum and jump gaits.

- **Joint Kinematics**

Results of the comparison between the plugin-gait model, the standard and the calibrated musculoskeletal models, presented in figure 3.24, are relatively contrasted (table 3.20).

Table 3.20: Correlation coefficient values comparing standard and calibrated MSM results vs. kinematic model in the case of CP children

	Sagittal kinematics (Angles)								
	CP recurvatum gait			**CP jump gait**			**CP crouch gait**		
	Hip	**Knee**	**Ankle**	**Hip**	**Knee**	**Ankle**	**Hip**	**Knee**	**Ankle**
Standard MSM vs Kinematic model	0. 81	0.68	0.55	0.74	0.72	0.49	0.71	0.43	0.64
Calibrated MSM vs Kinematic model	0. 87	0.72	0.68	0.79	0.88	0.82	0.84	0.65	0.78

In comparison to kinematic model, the correlation tests show that results from the standard MSM are qualitatively correlated for different CP gait groups. Relative mean correlation

coefficients exceed 0.7. However, it is noticeable that the correlation corresponding to the knee joint in CP crouch gait, the ankle in CP jump and recurvatum gaits represent a low amount, about 0.43, 0.49 and 0.55 respectively for the standard MSM. In fact, regarding motion agent's trajectories during simulations, errors between tracked trajectories and those recalculated in the forward dynamic may exceed 2 cm. When looking on results from calibrated musculoskeletal models, the correlation coefficients increase for all CP population and for each joint. Nevertheless, the ankle and the knee joint in CP recurvatum and crouch populations represent a lower value of the correlation tests of 0.67.

Table 3.21 reveals that the standard MSM induced a high error level in several determinants of gaits, compared to those obtained from the plugin-gait biomechanical model. Correlation coefficients around 0.75 are founded for each joint and for both healthy adults and children. But, correlation coefficients of 0.67 are founded for predicted knee kinematics in crouch gait and of 0.69 for predicted ankle joint in simulated jump gait. The correlation coefficients exceed 0.9 for jump gaits, and 0.88 for crouch gaits when dealing with calibrated MSM.

Table 3. 21: Variation between the kinematic model and Standard and Calibrated MSM results in the case of CP population**.**

		CP recurvatum gait		CP jump gait		CP crouch gait	
		Kinematic model vs Standard MSM	Kinematic model vs Calibrated MSM	Kinematic model vs Standard MSM	Kinematic model vs Calibrated MSM	Kinematic model vs Standard MSM	Kinematic model vs Calibrated MSM
Kinematics determinants (°)	P1:Hip flexion at Initial Contact	26 (3.6)	2 .4 (1.3)	23 (2.1)	8 (3.1)	29 (5.8)*	10 (3.8)
	P2: Hip extension Maximum	8.5 (2.9)*	5 (3.1)*	10 (1.5)	4 (2.8)	36 (7.6)*	8(2.4)
	P3: Knee flexion at Initial Contact	23 (2.4)	5.6 (2.7)	19 (1.8)	9 (5.2)	30 (4.4)*	9 (5.3)
	P5: Knee flexion Max	28 (5.5)	18 .4(3.4)	16 (4.2)*	3 (2.2)	-	-
	P4: Knee flexion at mid stance phase	41 (7.1)	6 .3(3.9)	22 (3.7)*	5 (3.4)	32 (2.5)	7 (5.4)
	P6: Dorsiflexion at Initial Contact	24 (3.3)	5.1(2.5)	5.5 (1.8)	2 (1.5)	28 (4.8)	3 (1.6)
	P7: Dorsiflexion Minimum	26 (1.9)	2.8 (1.6)	-	-	-	-
	P8: Dorsiflexion Maximum	22 (3.1)	6 ,5(4.1)	12 (3.2)*	8 (2.3)	30 (2.6)	8 (4.5)
	P9: Plantarflexion Maximum	46 (5.8)*	7.2 (3.4)	1 (0.6)	3 (1.8)	21 (3.2)	9 (5.1)
GRF Parameters (N/Kg)	P1: 1st peak GRF	1 (1.8)	-	1 (2.0) *	-	30 (5.9) *	-
	P2: second peak GRF	13 (2.6)	-	4 (3.4)	-	10 (2.2)	-
	P3: min valley GRF	35 (3.1)*	-	19 (1.8)*	-	15 (1.8)	-

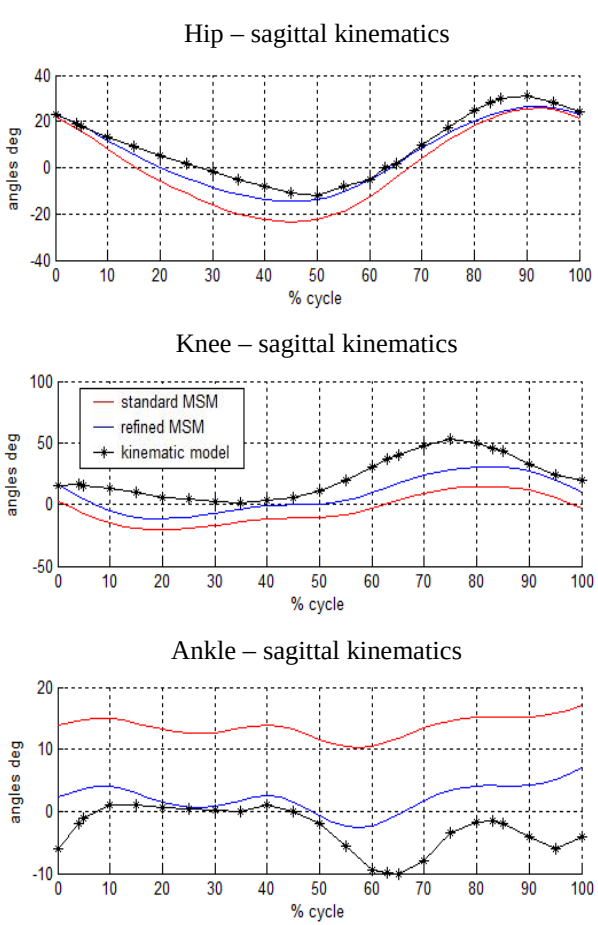
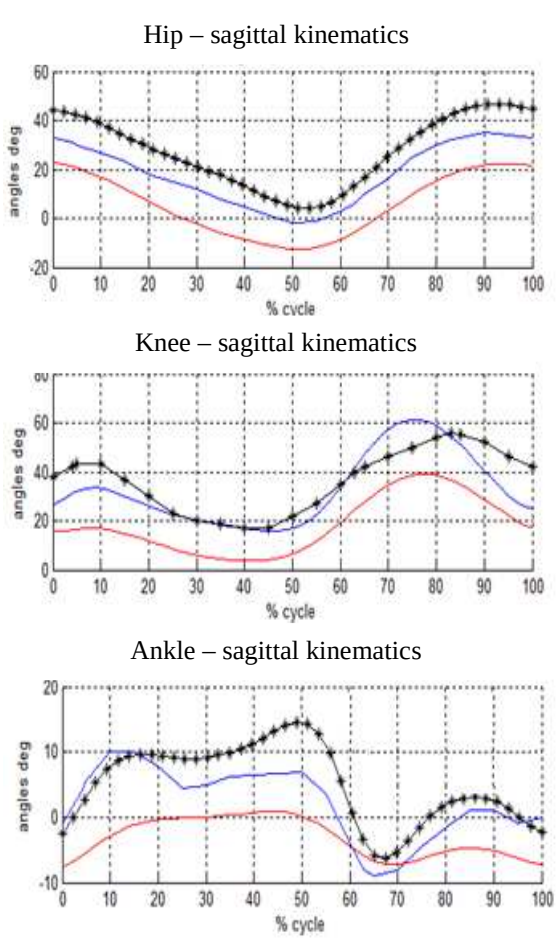
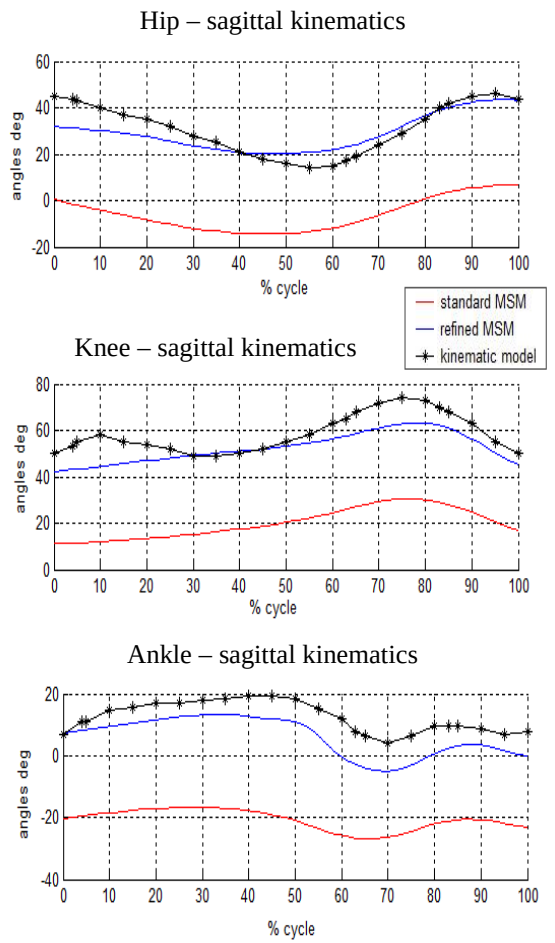
Recurvatum Gait	Jump gait	Crouch gait
		

Figure 3. 24**:** comparison between Joint angles kinematics estimated through musculoskeletal models and the kinematic model for Cerebral Palsy population: the red curve corresponds to the standard lifemod model, the blue curve corresponds to the calibrated MSM and the black curve corresponds to Kinematic model.

❖ The Ground Reaction Forces

Concerning the standard MSM, comparative analysis reported that estimated ground reaction forces are well correlated to measurements through force plates for different CP groups (Figure 3.25). The correlation coefficients exceed 0.7 for different CP subjects. When looking more closely at GRF parameters, Table 3.22 confirmed that the range of difference between GRF simulation and experimental results varied depending of the CP gait characteristics.

Table 3.22: Correlation coefficient values of estimated GRF comparing standard and calibrated MSM results in cerebral palsy populations

	GRF values		
	CP recurvatum gait	**CP jump gait**	**CP crouch gait**
Standard MSM vs Kinematic model	0.79 (0.11)	0.68 (0.12)	0.87 (0.07)
Imporved MSM vs Kinematic model	1	1	1

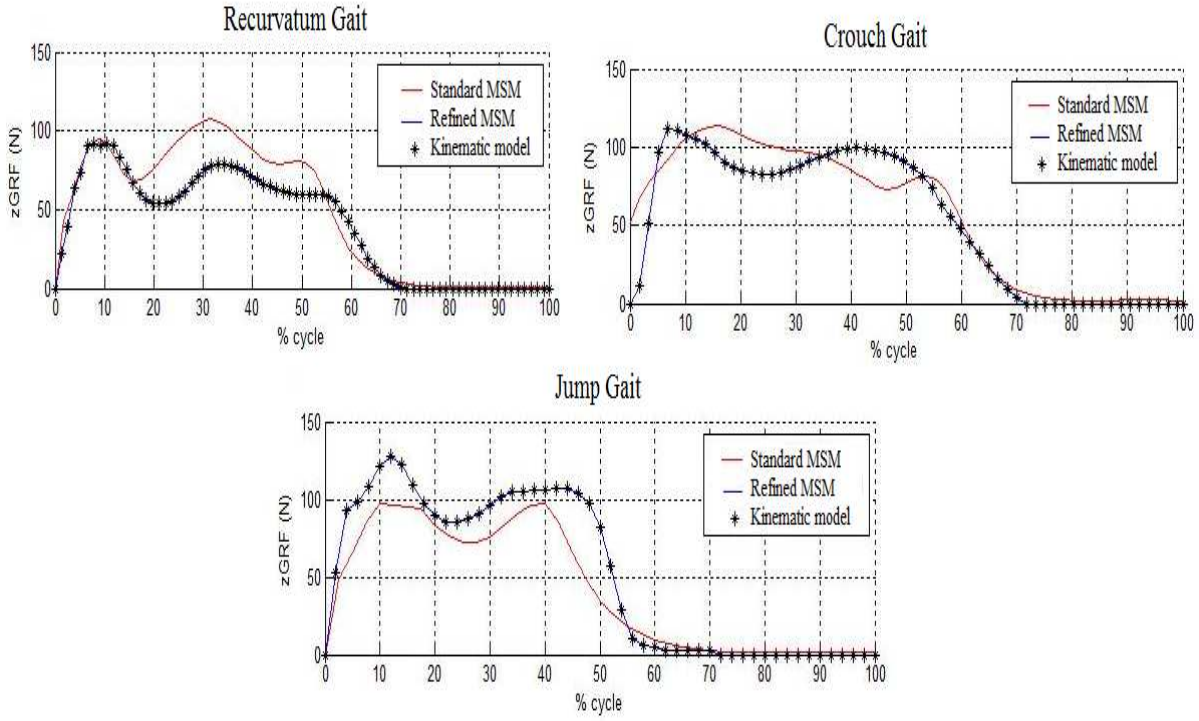


Figure 3. 25**:** Vertical ground reaction forces (N/kg) of CP with recurvatum gait (a), crouch gaits (b) and jump gaits (c). Comparison between the current studies (models): the red curve corresponds to the standard Lifemod model, the blue curve corresponds to GRF introduced as input in the calibrated MSM and the black curve corresponds to GRF measurements (the reference).

❖ **Muscle activation and forces**

Figures 3.26-3.28 represent an illustration of the comparison between the estimated muscle activation of the principle muscles on the lower limbs using the musculoskeletal models and the rectified EMG measurement's envelops, for CP with jump gaits, CP with crouch gait and CP with recurvatum gaits respectively.

Results from correlation tests are presented in Table 3.23. Globally, it is observed that a high level of correlation is obtained in several CP cases independently from their gait specificities. With the standard MSM, the correlation coefficients are greater than 0.70 and, with the calibrated MSM, they exceed 0.87 with a small dispersion. It can be showed that the standard MSM can predict likely expected muscle activity patterns. Whereas, in CP groups with jump gait, the correlation analysis of the Tibialis Anterior reported a poor correlation level of 0.32; in CP groups with crouch gait, the correlation analysis of the Rectus Femoris as well as the Tibialis Anterior reported also a poor correlation level of 0.52 and 0.41 respectively. With the calibrated MSM, the correlation results insignificantly increased.

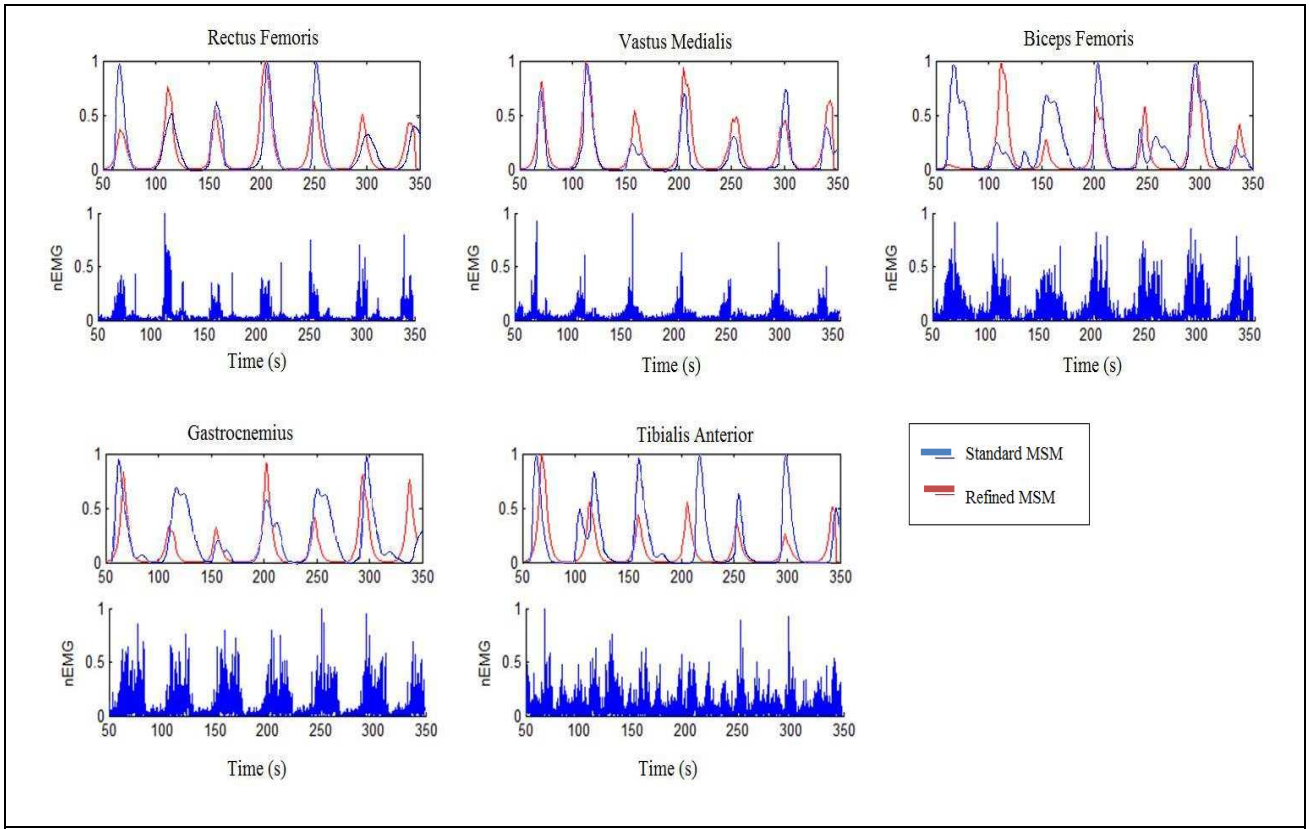


Figure 3. 26**:** comparison between the EMG measurements and predicted muscle activities using the standard model (red curve) and the calibrated MSM (blue curve) for CP with Jump Gait.

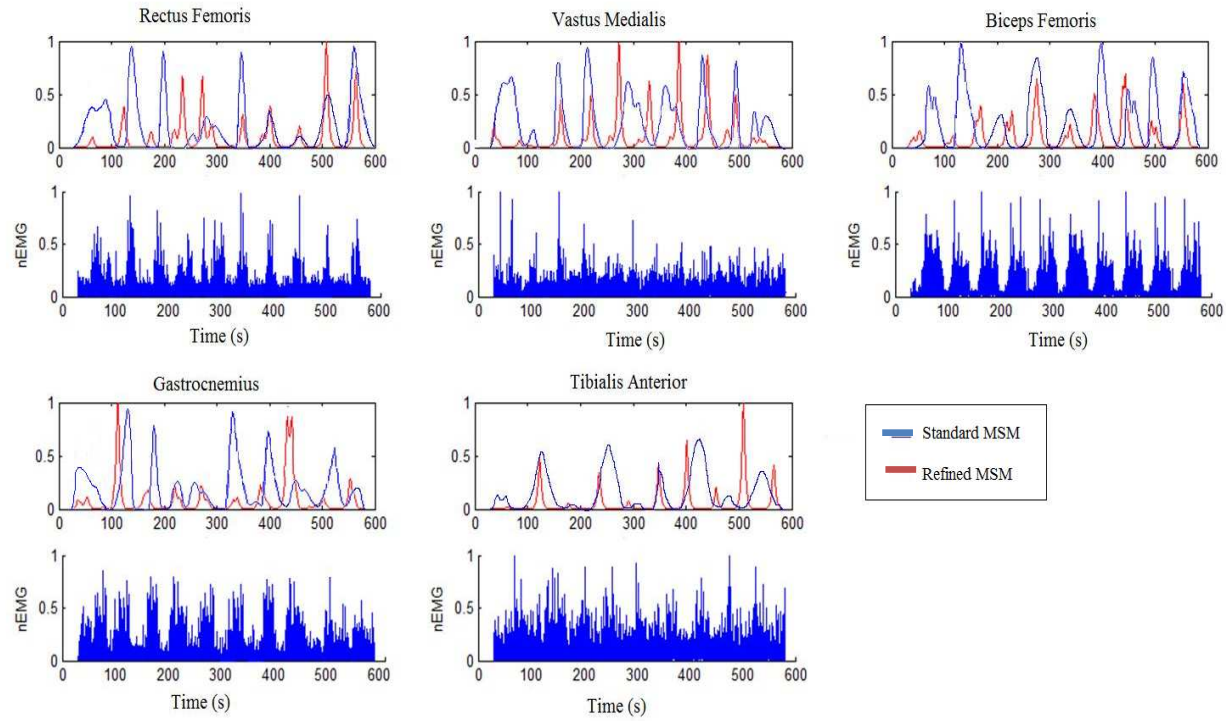


Figure 3. 27**:** comparison between the EMG measurements and predicted muscle activities using the standard model (red curve) and the calibrated MSM (blue curve) for CP with Crouch Gait.

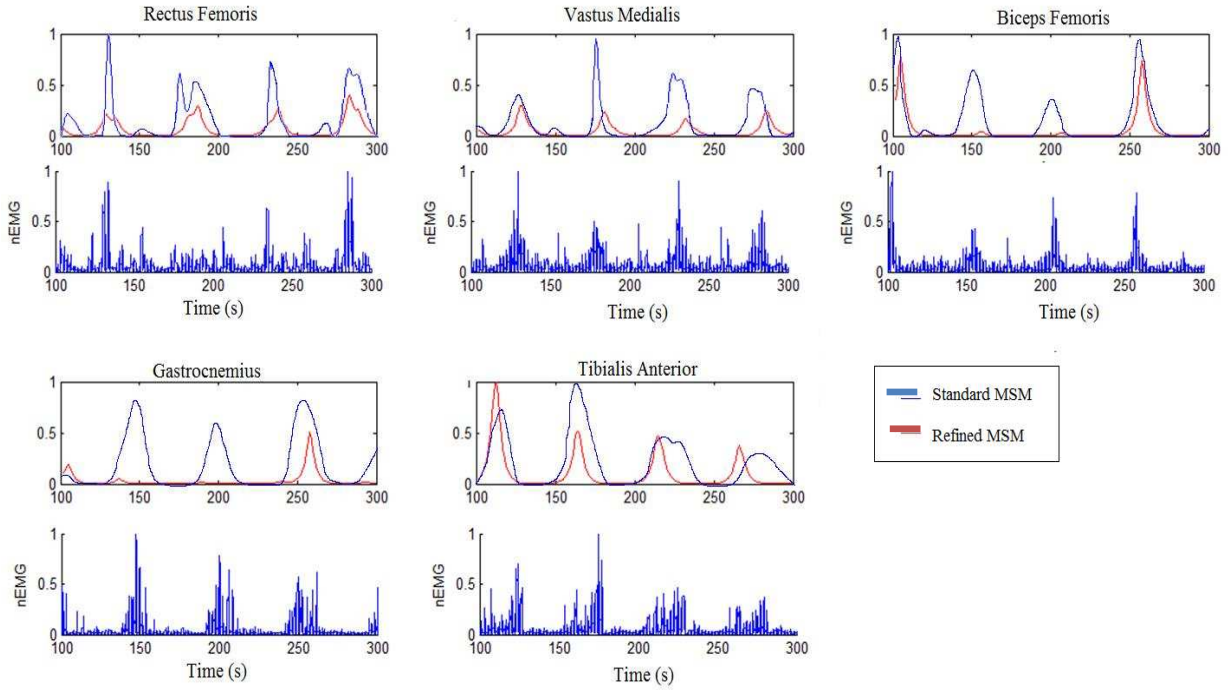


Figure 3. 28**:** comparison between the EMG measurements and predicted muscle activities using the standard model (red curve) and the calibrated MSM (blue curve) for CP with Recurvatum Gait.

Table 3.24 describes the predicted muscle forces from the two MS models for each CP groups. Results figure out that the standard musculoskeletal modeling failed to predict muscle forces. With the calibrated MSM, predicted muscle forces are more realistic.

Table 3. 23: Correlation coefficient values comparing EMG measurements to predictive muscle activities using the standard and calibrated MSM results in case of cerebral palsy Population

	CP recurvatum gait		**CP jump gait**		CP crouch gait	
	The standard model	The calibrated model	The standard model	The calibrated model	The standard model	The calibrated model
Tibialis anterior	0.70	0.89	0.32	0.41	0.41	0.52
Biceps femoris	0.74	0.85	0.71	0.75	0.64	0.75
Rectus femoris	0.666	0.73	0.89	0.91	0.52	0.58
Gastrocnemius L.	0.75	0.83	0.58	0.73	0.73	0.87
Vastus medialis	0.74	0.84	0.46	0.53	0.87	0.89

Table 3. 24: Maximum muscle forces (N) predicted using the standard and calibrated MSM results in case of cerebral palsy Population

	CP recurvatum gait		**CP jump gait**		**CP crouch gait**	
	The standard model	The calibrated model	The standard model	The calibrated model	The standard model	The calibrated model
Tibialis anterior	$0.12*10^{-6}$	54	$0.25*10^{-6}$	96	$1.52*10^{-7}$	46
Biceps femoris	$0.95*10^{-6}$	120	$1.14*10^{-6}$	146	$1.98*10^{-6}$	70
Rectus femoris	$1.62*10^{-6}$	180	$1.24*10^{-6}$	198	$4.65*10^{-6}$	229
Gastrocnemius L.	$1.20*10^{-6}$	150	$0.98*10^{-6}$	124	$2.45*10^{-6}$	168
Vastus medialis	$0.64*10^{-6}$	98	$0.22*10^{-6}$	68	$0.41*10^{-6}$	84

To end with Musculoskeletal modeling, the table below summarizes the principle results obtained through several studies in this project. The (+) signify positive results regarding the used model, the (±) represent acceptable results with some improvement needs and the (-) symbolizes that the model failed to predict realistic results.

		Kinematics		**GRF**		**Muscle activation**		**Muscle forces**	
		Standard MSM	Calibrated MSM	Standard MSM	Calibrated MSM	Standard MSM	Calibrated MSM	Standard MSM	Calibrated MSM
Normal Gait	**Healthy Adults**	±	+	+	GRF measurements are used as inputs of the MSM	±	+	-	+
	Healthy Children	±	+	+		±	+	-	+
Simulated pathological postures	**Healthy Adults – crouch**	-	+	+		±	+	-	+
	Healthy Adults – Jump	-	+	+		-	-	-	+
	Healthy Children - Crouch	-	+	+		-	-	-	+
	Healthy Children – jump	-	+	+		-	-	-	+
CP population	**Crouch CP**	-	+	±		± Except spastic muscle	±	-	+
	Jump CP	-	+	±		± Except spastic muscle	±	-	+
	Recurvatum CP	-	+	±		± Except spastic muscle	±	-	+

3.3. Simulated gaits: clinical results

Analyzing the gait parameters of healthy subjects imitating characteristic CP patterns may indicate the effects of voluntary crouched and toe-walking postures on kinetics and muscle activation. The objective is to reveal whether the biomechanical constraints induced by pathological posture are sufficient to explain sources of detected abnormalities on the gait parameters, kinetics and the muscle activities on CP population.

❖ Inter-subject Reproducibility

As mentioned previously, kinematic graphs (figure 3.14-15) showed that the inter-subject reproducibility has been achieved. The joint sagittal moments, as well as normalized and rectified EMG of Gastrocnemius, Rectus Femoris, Bicep Femoris and Tibialis Anterior, presented an excellent intersubject reproducibility, with a mean CMC value of 0.90 for all types of gait (Table 3.25-26).

Table 3. 25: Inter-subject CMC values for ankle, knee and hip sagittal moments

	Sagittal Moment					
	Ankle		Knee		Hip	
	Adults	Children	Adults	Children	Adults	Children
Normal gait	0.93	0.96	0.97	0.94	0.95	0.92
Simulated jump gait	0.98	0.91	0.94	0.89	0.95	0.97
Simulated Crouch gait	0.97	0.93	0.89	0.91	0.94	0.90

Table 3.26: Inter-subject CMC values for normalized rectified EMG of Gastrocnemius, Rectus Femoris, Biceps Femoris, and Tibialis anterior

	Normalized rectified EMG							
	Biceps Femoris		Gastrocnemius		Rectus Femoris		Tibialis Anterior	
	Adults	Children	Adults	Children	Adults	Children	Adults	Children
Normal gait	0.86	0.92	0.91	0.94	0.96	0.91	0.93	0.91
Simulated jump gait	0.94	0.90	0.93	0.89	0.90	0.97	0.84	0.89
Simulated Crouch gait	0.86	0.87	0.89	0.92	0.93	0.88	0.84	0.94

❖ Simulated pathological gaits vs CP gaits

According to Lin et al. (2000), the joint kinetics of simulated jump and crouch gait pattern are correlated (Figures 3.29-30). The minimum CMC value obtained are about 0.72 for knee moments and 0.62 for sagittal hip moment, respectively for simulated jump and crouch gaits (Table 3.27).

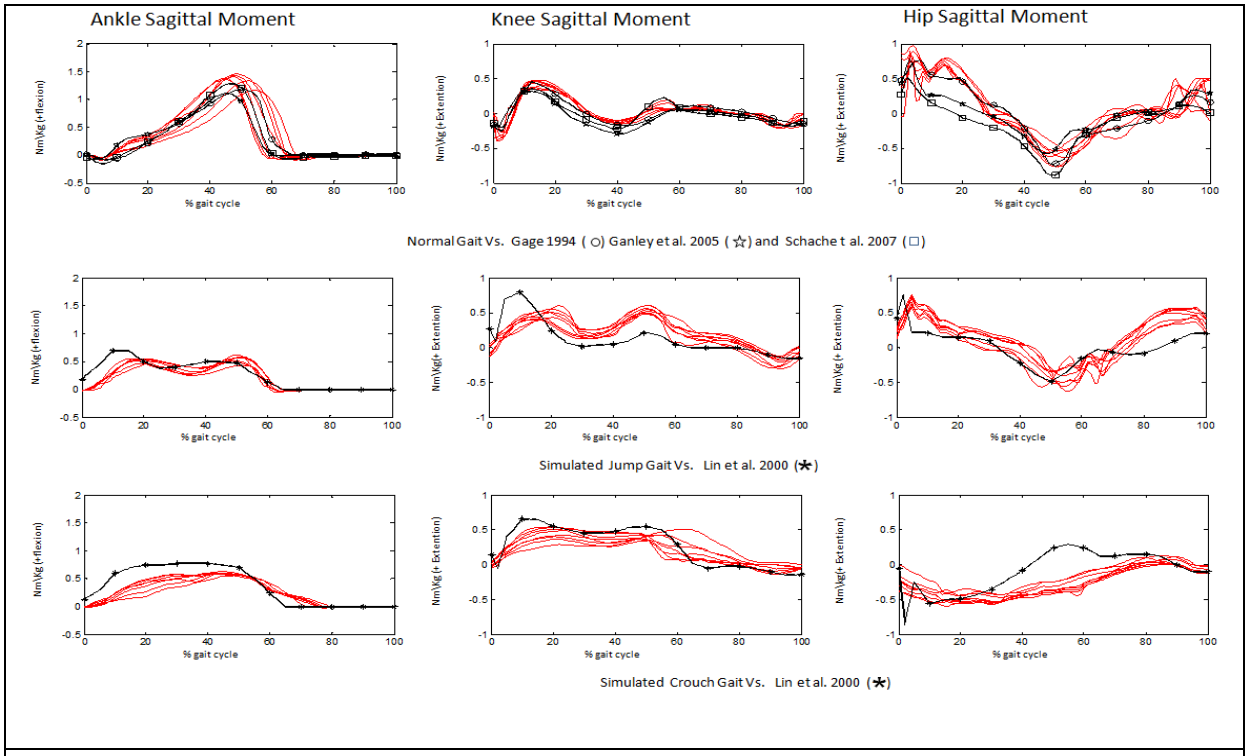


Figure 3. 29**:** Ankle, knee and hip sagittal moments for the normal gait and simulated jump and crouch gaits, performed by healthy adults.

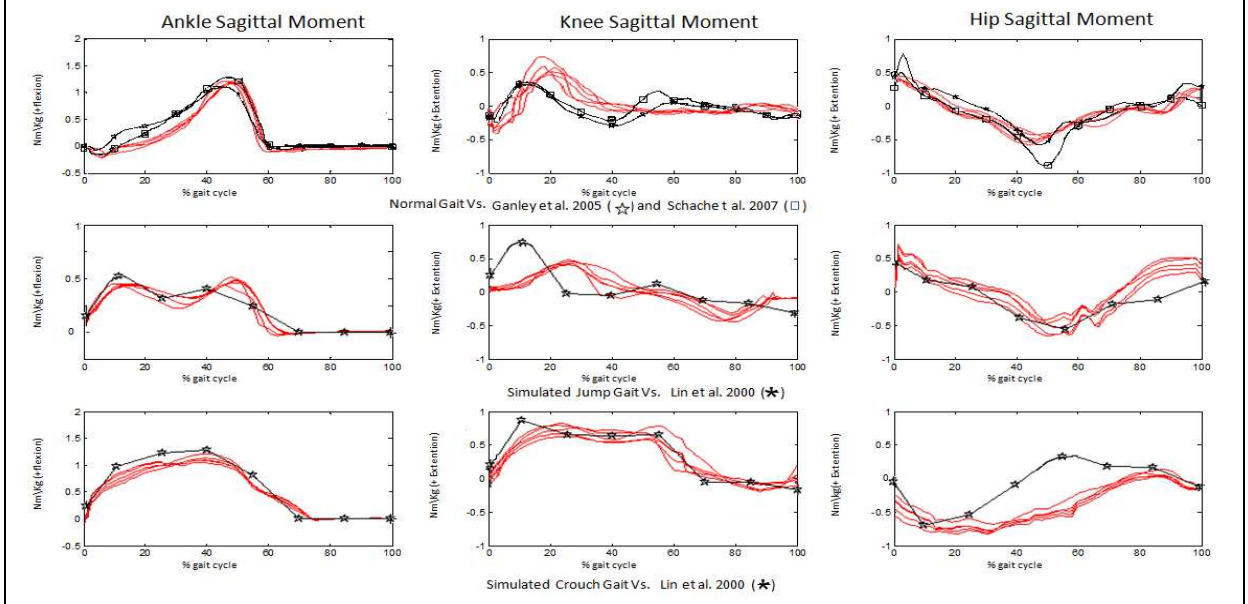


Figure 3. 30**:** Ankle, knee and hip sagittal moments for the normal gait and simulated jump and crouch gaits, performed by healthy children.

Table 3. 27: Joint sagittal moments: CMC values comparing current study data vs. literature data

versus		Sagittal Moment					
		Ankle		Knee		Hip	
		Adults	Children	Adults	Children	Adults	Children
Normal gait	Gage (1994)	0.99	0.94	0.97	0.91	0.98	0.92
	Vaughan (1996)	0.96	0.92	0.87	0.87	0.90	0.86
	Ganley et al. (2005)	0.96	0.97	0.87	0.89	0.93	0.88
	Schache et al. (2007)	0.99	0.94	0.91	0.92	0.85	0.91
Simulated jump gait	Lin et al. (2000)	0.90	0.87	0.72	0.78	0.82	0.75
Simulated crouch gait	Lin et al. (2000)	0.86	0.78	0.90	0.84	0.62	0.69

Obviously, compared to normal gait, differences in kinematics and kinetics observed in simulated gaits contribute to an alteration of muscle activations, as observed in figures 3.31-32. When comparing to common characteristic of CP, as presented in table 3.28, simulated crouch gait induced similar modification on muscle activities in biceps femoris, Gastrocnemius and tibialis Anterior. But, low CMC values are reported in the rectus femoris activation. Moreover, the alteration in muscle activities when simulating jump gait are considerably not correlated to those from CP.

Table 3. 28: Rectified and normalized EMG: CMC values comparing current study data vs. literature data

versus		Normalized rectified EMG							
		Biceps Femoris		Gastrocnemius		Rectus Femoris		Tibialis Anterior	
		Adults	Children	Adults	Children	Adults	Children	Adults	Children
Normal gait	Romkes et al. (2007)	0.87	0.78	0.97	0.95	0.93	0.89	0.92	0.98
Simulated jump gait	Lin et al. (2000)	0.30	0.42	0.67	0.88	0.26	0.48	-	-
Simulated crouch gait	Lin et al. (2000)	0.69	0.72	0.72	0.79	0.53	0.64	-	-
	Steele et al. (2010)	0.89	0.84	0.80	0.86	0.32	0.41	0.61	0.70

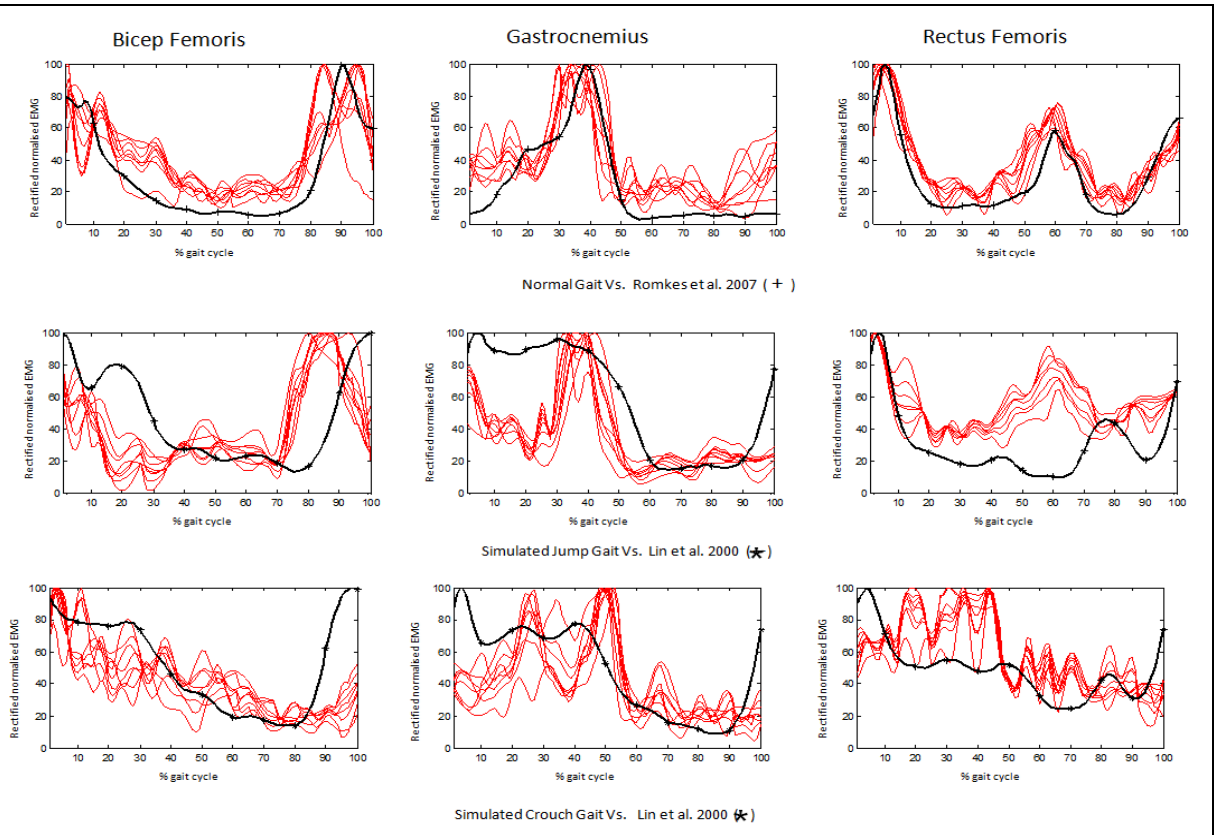


Figure 3. 31**:** Normalized rectified EMG for the normal gait and simulated jump and crouch gaits performed by heathy adults.

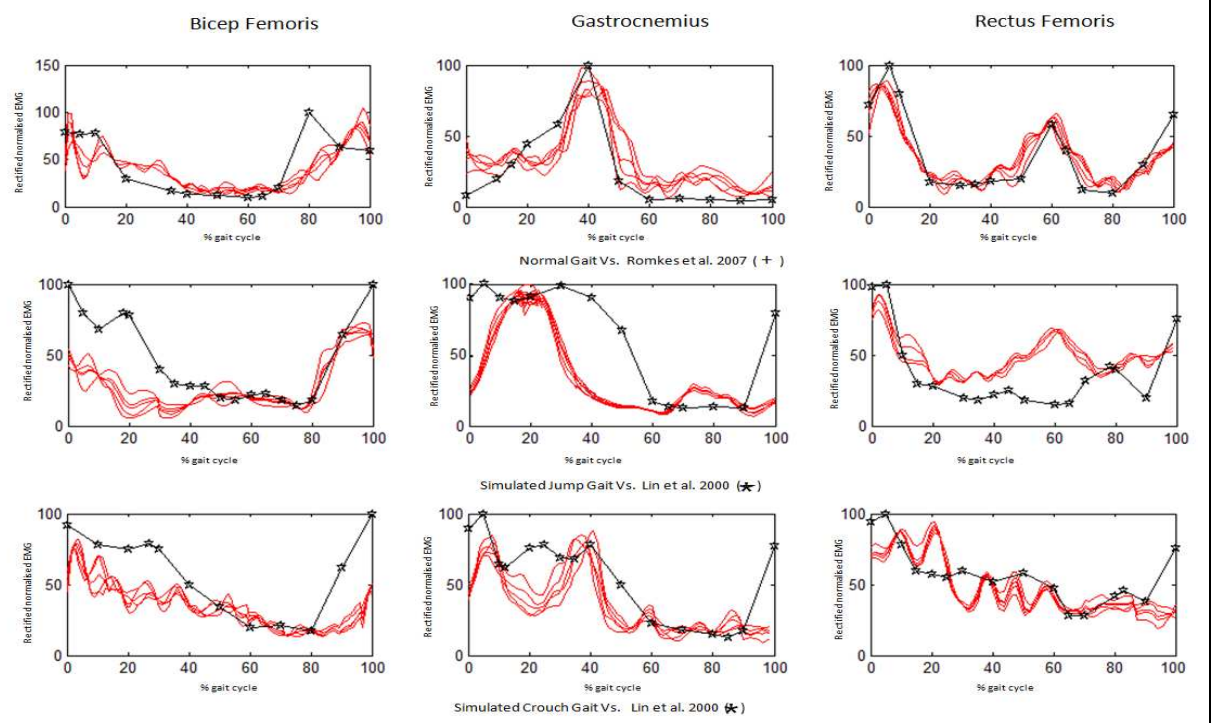


Figure 3. 32**:** Normalized rectified EMG for the normal gait and simulated jump and crouch gaits performed by healthy children.

Conclusion of the chapter

This study investigated the effect of using musculoskeletal models with varying levels of refinement on the accuracy of biomechanical results. Results from plug-in gait (Gait analysis) were compared to simulation results from the rescaled generic standard and calibrated musculoskeletal models.

When compared to specific-subject plug-in gait model's results, biomechanical analysis revealed that even for a healthy adult, the standard rescaled generic model led to significant differences in the calculated kinetics and muscle activities. Larger discrepancies are expected in a CP population. The analysis of inter-model differences revealed that calibration procedures seem to diminish the substantial differences compared to the generic model. But when studying CP children, the calibrated musculoskeletal models still show differences. All these results will be discussed in the next chapter.

Chapter 4.
General Discussion

4.1. Musculoskeletal Modeling

Normal walking is a complex activity. Gait abnormalities present additional challenges when developing realistic MSMs and reliable numerical simulations. Musculoskeletal modeling becomes essential to quantify biomechanical data required in managing the pathology of cerebral palsy and it becomes more interesting in clinical practice to evaluate the impact of a possible proposed surgical treatment on the patient's gait. Even if musculoskeletal modeling is a promising tool to improve clinical outcomes, clinicians have to be aware of its limits. The biomechanical entities, resulting from the MSM, are very sensitive to several parameters introduced in the model. We tried in this project to present a calibration process to refine such parameters from subject's experimental data and studied the effect of varying levels of calibration details on the accuracy of the obtained results.

The objective of this thesis work was the development of an adaptable musculoskeletal model of lower limbs of the Cerebral palsy with spastic Diplegia. It represented a retrospective study aiming to provide a suitable tool, easily implemented in clinical routine practice, to evaluate the if-then scenario with respect to gait disabilities and available treatment. This goal was analyzed through two principle point of view, regarding the efficiency of a standard generic musculoskeletal modeling (Model1: Standard Model) to study cerebral palsy gaits, and the efficiency of the calibration process of musculoskeletal modeling parameters using only gait examination data (Model2: Calibrated Model). Two principle studies were developed during this project. The first concerned healthy adults and children performing normal and imitated crouch and jump gaits to evaluate the influence of the accuracy of the geometry and the altered muscle activities on the musculoskeletal models predicted results. The second evaluated the accuracy of both developed musculoskeletal models to study the Cerebral Palsy cases.

4.1.1. MSM components

The present work has compared the kinematics, GRF, muscle activation and muscle forces parameters, estimated using the two developed musculoskeletal models and their respective measured parameters or entities which are defined using plug-in gait biomechanical model.

In general, the musculoskeletal modeling is used to determine muscle lengthening, muscle moment arms, muscle activation and muscle forces occurring during motions, that are difficult to obtain by direct measurement in vivo or from gait analysis experiments. In our study, we focused on predicted muscle forces. Moreover, different approaches and techniques are used to estimate muscle forces during motion [*Delp 1990, Kromer 1993, Pederson 1997, Marin 2001, Frayess 2009*]. The comparison with predicted muscle forces in literature is tricky but usually used to validate results of musculoskeletal modeling.

4.1.1.1. Muscle forces

As regards to quantitative muscle forces, summarized in tables 4.1-4.2, for both healthy populations, the standard rescaled musculoskeletal model gives a very low amount of muscle forces, compared with other literature references (Table 4.1). The estimated muscle forces present large variation according to literature data during normal gait. It can be observed, in our study, that for healthy children and children with cerebral palsy, all muscles are almost inactive.

When calibrating the musculoskeletal model's parameters, the estimated muscle forces have expected to increase. In one hand, the dynamic joint stiffness, fixed in the standard musculoskeletal model, are extremely higher than those determined using the calibration process, as shown in figure 3.5, which contribute to increase the torques around the joint, and as a consequence, to reduce significantly the muscle contribution during motion. On the other hand, the training parameters of PD and PID controllers are carefully fixed to ensure the convergence of the mathematical model and to avoid integration and derivatives noises and perturbations.

Compared to literature, the calibrated musculoskeletal models still underestimated the predicted muscle forces during normal gait. Figure 4.1 summarizes the findings of the predicted muscle forces in healthy population during normal gait.

Table 4. 1: Recapitulative results of predicted maximal muscle forces (N) using the standard musculoskeletal model in different studies, compared to literature data

	Our study									Litterature				
	Normal Gait		simulated crouch		simulated jump		Cerebral palsy study							
Muscle forces (N)	**Healthy adults**	**Healthy children**	**Healthy adults**	**Healthy children**	**Healthy adults**	**Healthy children**	**CP crouch**	**CP jump**	**CP recurvatum/ stiff knee**	**Delp 1990**	**Frayess 2009**	**Kromer 1993**	**Peaderson 1997**	**Marin 2001**
Tibialis Anterior	0,32	0,08	0,18	0,00009	8.45	0,00035	$1.52*10^{-7}$	$0.25*10^{-6}$	$0.12*10^{-6}$	-	-	300	-	-
Biceps Femoris	0,17	0,12	0,38	0,00016	0,42	0,0012	$1.98*10^{-6}$	$1.14*10^{-6}$	$0.95*10^{-6}$	1120	960	600	-	672
Rectus Femoris	46,2	1,2	75.36	0,00087	23,25	0,00284	$4.65*10^{-6}$	$1.24*10^{-6}$	$1.62*10^{-6}$	800	477	500	370	
Gastrocnemius L.	27,95	0,49	37,12	0,00056	17,47	0,00144	$2.45*10^{-6}$	$0.98*10^{-6}$	$1.20*10^{-6}$	1115	781	600	240	892
Gastrocnemius M.	-	-	-	-	-	-	-	-	-	490			-	196
Vastus medialis	-	-	-	-	-	-	$0.41*10^{-6}$	$0.22*10^{-6}$	$0.64*10^{-6}$	-	-	-	-	-

Table 4. 2: Recapitulative results of predicted muscle forces using the calibrated musculoskeletal model in different studies, compared to literature data

	Our study									Literature				
	Normal Gait		simulated crouch		simulated jump		Cerebral palsy study							
Muscle forces (N)	**Healthy adults**	**Healthy children**	**Healthy adults**	**Healthy children**	**Healthy adults**	**Healthy children**	**CP crouch**	**CP jump**	**CP recurvatum/ stiff knee**	**Delp 1990**	**Frayess 2009**	**Kromer 1993**	**Peaderson 1997**	**Marin 2001**
Tibialis Anterior	177	94	97	79	120	68	46	96	54	-	-	300	-	-
Biceps Femoris	286	122	215	125	170	84	70	146	120	1120	960	600	-	672
Rectus Femoris	380	240	480	324	250	158	229	198	180	800	477	500	370	-
Gastrocnemius L.	220	153	355	284	290	124	168	124	150	1115	781	600	240	892
Gastrocnemius M.	-	-	-	-	-	-	-	-	-	490			-	196
Vastus medialis	-	-	-	-	-	-	84	68	98	-	-	-	-	-

First, literature data, dealing with the quantified muscle forces, are estimated only for healthy adults. Moreover, it can be noticed that a large variations are founded in literature. This can be explained by the diversity of muscle modeling strategies and the diversity of the optimization process used to predict muscle forces.

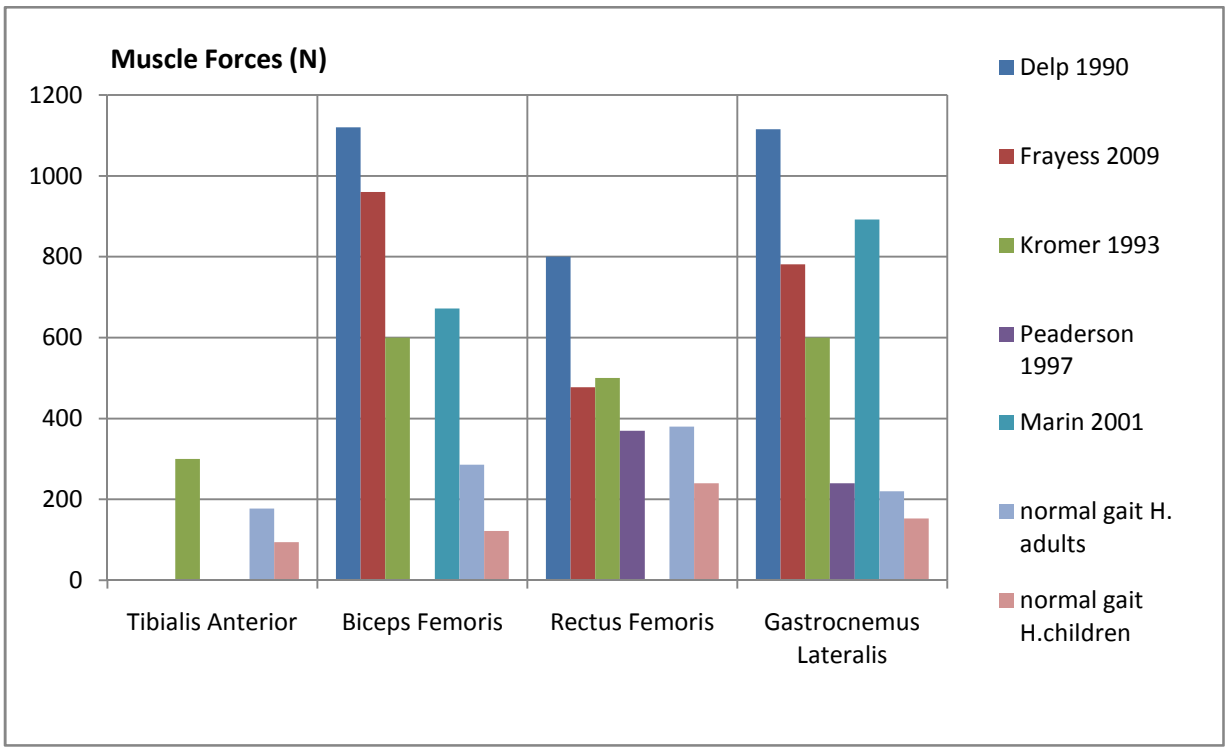


Figure 4. 1: Recapitulative results of predicted muscle forces
using the calibrated musculoskeletal model in different studies, compared to literature data

The direct comparison between our results and literature data is difficult because musculoskeletal modeling results are highly dependents on the modeling methodology [*Dao 2009*]. This finding confirms that there is no consensus on the design of the musculoskeletal modeling, which makes the clinical applicability of the MSM controversy. Second, the calculation of muscle forces is primarily based on the muscle geometry (muscle volume, attachment points) and its intrinsic parameters. Several studies pointed out that lengthening muscle and estimated muscle forces are very sensitive to these parameters [*Dao 2009, Scheys 2008*].

In our study, muscles are defined as lines of action between the insertion and the attachment positions. Since the musculoskeletal modeling is based on the rescaling process, erroneous estimations of anatomical muscle attachment may occurred, mainly for children with tinny skeleton and cerebral palsy children with bone deformities, leading to new anatomical muscle

configurations (Figure 4.2). These inaccuracies and inconsistencies are unfortunately not quantifiable, but lead consequently to mistaken estimated muscle activation and forces.

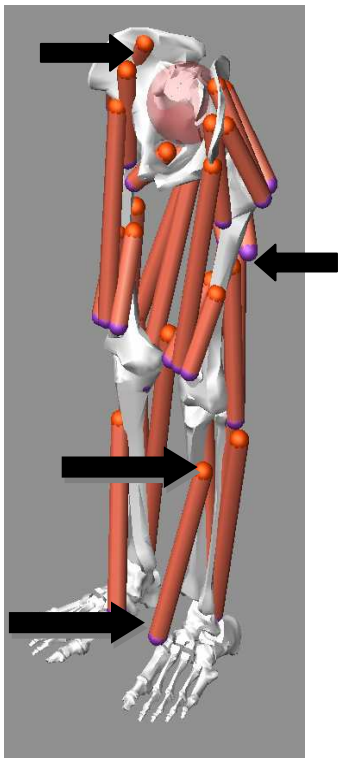


Figure 4. 2: muscle attachment sites for a LifeMod MSM of a healthy child

Using a simple viscoelastic model for muscle is very simple description. Several studies relied on more sophisticated muscle models, such as the Hill or Zajac muscle models. It may lead to improve the accuracy of muscle forces evaluation, but these models require a large number of parameters, difficult to be subject-estimated for pathological cases [*Reinbolt 2007, Dao 2009, Desailly 2009*]. The limit of such improvement is the fact that, even for a healthy adult, the existing anatomical datasets (for example optimal fiber lengths) are incompleteness and the missing parameters are guessed *[Delp 1990*]. Today, in vivo measurement of a complete and accurate muscle dataset remains challenging. The personalized musculoskeletal models using the MRI techniques can determine precisely the muscle volume and the attachment sites on the bones, but cannot predict the mechanical properties of muscles. For example sarcomere length, optimal fiber length, fiber directions are currently not measurable in standard MRI routine. When dealing with pathology with deformable bones, muscle attachment sites may be difficult to be identified. Also, since the MRI techniques (acquisition and proceeding) are time-expensive, basing the treatment's decisions on personalized MSM is unrealistic in clinical routine practice.

4.1.1.2. EMG vs predicted muscle activation

The validation of MSM results is generally based on the comparison between the estimated muscle activation and the measured EMG signals. This qualitative validation showed controversy results.

In normal gait, the mean correlation coefficient, for healthy adult population, is around 0.72 and 0.82, respectively for the standard and the calibrated models (Table 3.11). For healthy children, these correlation values present respectively 0.62 and 0.71. It was observed that, for standard MSM, the correlations of Rectus Femoris and Gastrocnemius Lateralis were mediocre, about 0.47 and 0.58 respectively. In the calibrated MSM, the correlation coefficient of the Gatsrocnemius Lateralis increased to 0.74 because the contact instabilities are avoided. But the Rectus Femoris‘s correlation coefficient remained mediocre, about 0.62 (Table 3.11). These differences can be explained by the fact that recorded EMG signals may not represent the muscle activity of a single muscle but can represent also surrounded muscles activities. The EMG electrodes placements on the limbs of small child morphology, presented in Seniam recommendations [*Hermens 2000*], are not necessary fixed in the minimal diaphonic zones and could not avoid crosstalk [Sussman 1992]. However, it’s important to precise that EMG interpretations are highly dependent on the quality of EMG measurements and equipments [*Sussman 1992, Perry 1998, Barr 2010*]. Consequently, the EMG signals are not always enough reliable and accurate.

In cerebral palsy populations, Table 3.23 showed a high correlation level between the EMG measurements and the predicted muscle activation for both models, with a small increase in the calibrated model. This result demonstrated that the standard model, as well as the calibrated one, can predict expected muscle activity patterns. Nevertheless, both models failed to predict the Tibialis Anterior activation in CP jump gait and CP crouch gait. Correlation analysis reported a poor correlation level of 0.32 and 0.41. The Rectus Femoris presents a correlation coefficient of 0.51 in the CP crouch gait, with an insignificant increase in the calibrated MSM. These results can be explained, first, by the simplicity of the contact with the ground and secondly by the fact that the spasticity, an important factor in cerebral palsy affecting muscle activities and growth, is not taken into account in our musculoskeletal modeling.

Lastly, it can be depicted, from tables 3.17-18, that both MSM models failed to accurately estimate the muscle activities during imitated pathological postures because of the high level of muscle co-contraction which is not taken into account neither in muscle modeling, nor in the optimization procedure in our work.

Meanwhile, the improved correlation tests values in the calibrated model, confirm that the estimation of muscle activation and forces is the result of several interactions in the modeling process.

4.1.1.3. Ground Reaction Forces

Focusing on the GRF results, presented in figures 3.10, 3.20 and 3.21, the contact model with the ground in LifeMOD may predict acceptable contact forces during normal gait. We also found that the important factors, interacting with the contact stability, were the motion agent weight parameters. These last factors ensure that the recorded foot markers fitted well the bony anatomical positions, avoid the unrealistic foot deformities on healthy populations and bring more contact stability during simulation.

Looking into CP population, the standard musculoskeletal model leads to good correlated GRF with measurements; correlation coefficients exceed 0.70. But, the results may vary depending on the CP gait characteristics. Results from CP jump gaits are worsen then those from recurvatum and crouch gaits.

In fact, differences, obtained on estimated GRFs, can be also explained by the simplicity of the foot model, which is represented as a single rigid bone. Such models can not represent foot deformities, neither the secondary foot joints which play an important role on the cerebral palsy patient's dynamics. This foot model may give some erroneous results of the ground reaction forces (Figure 4.3). Gait analysis exam has not included motion measurements that distinguish the movements occurring in the hind-foot and the forefoot, which explains the lack of pertinent information regarding the contact characteristics, especially in cerebral palsy cases [*Sussman 1992*], and may then underestimate the numerical evaluation of the contact forces.

The foot in cerebral palsy with spastic diplegia is the most affected bone. Its secondary joints played an important role in the investigation of cerebral palsy [Sussman 1992, Leardini 2007].

Developing more accurate biomechanical models of the foot and ankle functioning, integrating CP specificities of the deformed foot, may increase the accuracy of the contact results.

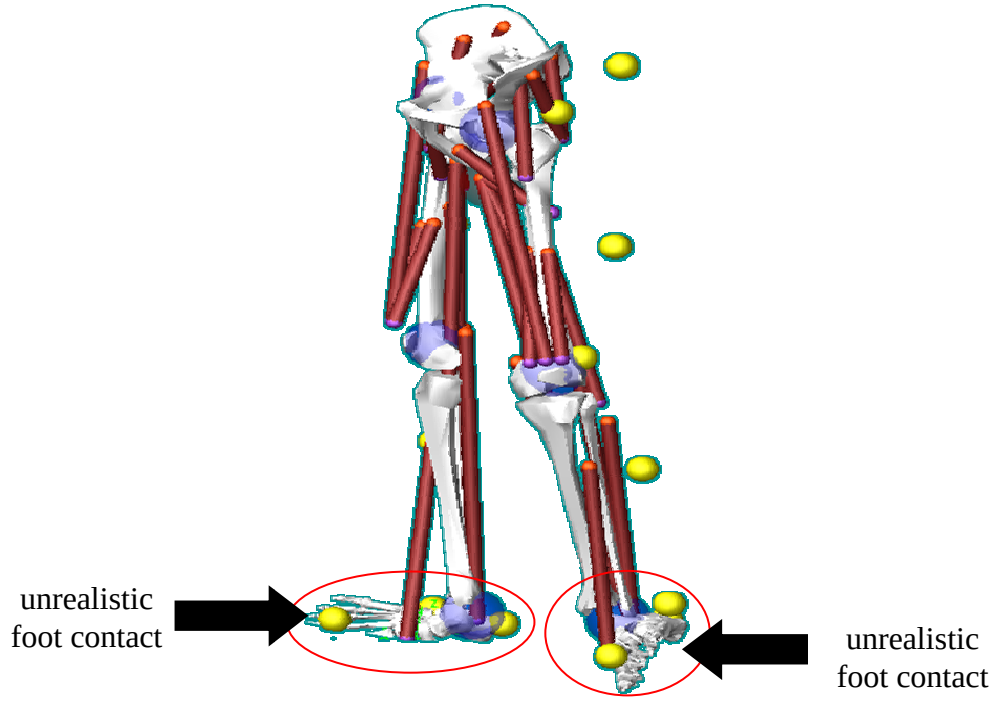


Figure 4. 3: Erroneous configuration of the foot during gait on standard LifeMod MSM of a healthy adult

4.1.1.4. Kinematics

The tracked trajectory of an anatomical marker is represented in the MSM as a motion agent, attached to a fixed anatomical point on the bone with an elastic system, to which is attributed a weight. In standard MSM, weights are very low, which allows large displacements of these motion agents. Errors between tracked trajectories and those recalculated in the forward dynamic may exceed 2 cm. These numerical inaccuracies influence the kinematic results, which may explain the large differences observed between the kinematics of the MSM and those from the kinematic model, in addition to marker misplacements in rescaled musculoskeletal models.

Additionally, the most important limitation is our study was the use of the Helene Hayes protocol of marker placements. It is widely used in the clinical routine practice, but recent

studies showed that such a model has to be adjusted and completed [*Ferrari 2010, Sangeux 2010, Zijden van der 2010*].

4.1.2. Limits and contributions

This project investigated the effects of using musculoskeletal models with varying levels of refinement on the accuracy of biomechanical results. The analysis of inter-model differences revealed that even for a normal gait in healthy population, the standard generic MSM leads to significant differences in the calculated kinetics and muscle activities. Larger discrepancies are founded in a CP population. The calibration procedure, proposed in our project, would seem to diminish the substantial differences compared to the generic model but when studying CP children, the refined MS models still show differences.

Three important issues may explain our results:

- The use of Plug-in gait biomechanical model as a reference to evaluate the kinematical musculoskeletal model results
- The general limitations of using the rescaled generic musculoskeletal models to study the Cerebral palsy pathological gaits and the effect of intrinsic MS parameters
- The strategy of calibration adopted in this project.

4.1.2.1. Choice of reference

Our study was retrospective, based only on the clinical gait data collected during gait analysis examination and the physical examination: the 3D trajectories of the markers placed at the principle anatomical positions of the lower limbs, the EMG signals of principal muscles, the measured ground reaction forces. As a retrospective study, the Plugin-gait biomechanical model, called also a kinematic model, has been used as a reference to evaluate the predicted results of the musculoskeletal model.

In this project, we limited our study to represent comparison of different results in sagittal plan. First, Mackey et al. (2005) and McGinley et al. (2009) pointed out that, in gait analysis examination, results from the coronal and axial axis are not very reliable, as those from sagittal plane. Second, most of clinical studies concerning biomechanical parameters of gaits are limited to sagittal plane.

To ensure a consistent comparison, the 3D trajectories of anatomical markers, recalculated by the musculoskeletal model in the forward dynamics, were introduced as new inputs in the clinical plug-in biomechanical model. In this strategy, the kinematic data were evaluated in the same anatomical references [*Wu 2005*]. The joint torques and powers, estimated by the MSM are not evaluated and compared to those from the Plug-in Gait. In the musculoskeletal model, these estimated quantities represent joint torques as the summation of individual joint torques and soft tissues contributions. But, in the kinematic model, presented in Plug-in gait, it represents only the net joint torques calculated from the measured ground reaction forces. Several studies [*Deluca 1997, Ferrari 2008, Sangeux 2010, Zijden van der 2010*] showed that such a model has many limitations when studying pathological gait. But, nowadays, it is the universal clinical model and several surgical treatments are based on it.

4.1.2.2. General modeling

Developing a musculoskeletal model requires a large number of information about the anatomy of the subject, joint and muscle descriptions. The clinical use of musculoskeletal modeling involves certain level of accuracy on model description which is still a challenging issue. Despite the realistic behavior of simulations when deriving the musculoskeletal models using measured gait data, the accuracy of musculoskeletal modeling result's prediction is highly dependent on the modeling limitations and assumptions. Even if most of the studies relied on the rescaled generic musculoskeletal models, there are several important limitations of such models. The common assumptions are related to the geometry, the simplified muscle modeling and the simplified control motor.

- **Scaling techniques**

The effect of the scaling techniques is uncertain and inaccuracies are inevitable [*Scheys 2008, Dao 2009, Klets 2010*]. Despite the large anthropometric database provided in LifeMod (male, female, adults, children, Asian population), the accuracy of the musculoskeletal geometry is still debatable [*Cheng 1994*]. The generic model presents the geometry as linked rigid bodies. This simplified hypothesis might be available for adults, but it isnot applicable for children especially for those with bone deformities which evolute with growing process and could not be predicted with rescaling anatomy techniques. The bone deformities, which are frequently accompany neurological damage, are not modeled. Rescaling an adult

geometry to fit small CP skeleton with bone deformities is not representative of the real subject morphology. This hypothesis may limit the accuracy of the model and our result's interpretations. The scaling process may give some inaccuracy bone's positioning and erroneous muscle insertion sites for child skeleton (Figure 4.4). This point is a clear limitation in the present study and demonstrates the limited use of the retrospective gait data.

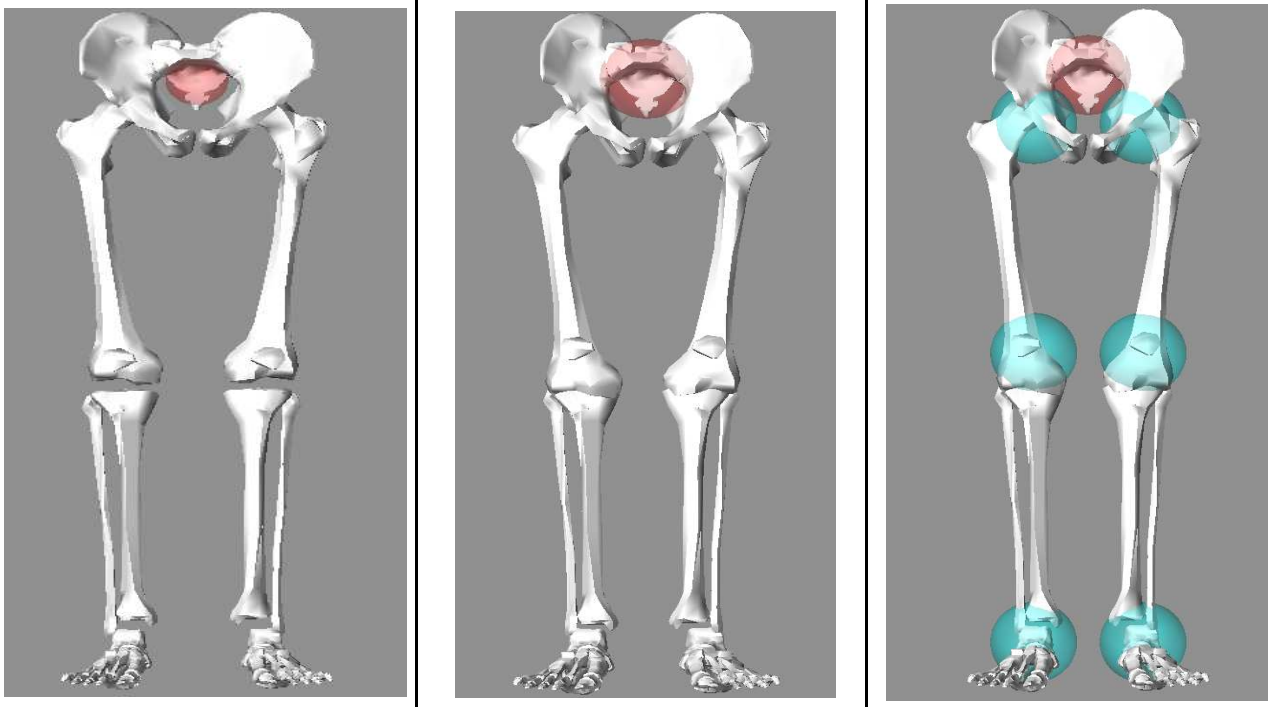


Figure 4. 4: adult (a) vs child (b) lower limb skeleton generated by lifemod software

When dealing with subjects with bones deformities on the lower limbs, the accuracy of muscle attachment sites remains problematic. The MRI techniques are potentially powerful, but expensive and time consuming for routine clinical use [*Scheys 2010*]. It is suggested to develop efficient palpation techniques or primary physical examination to objectively evaluate the degree of bone deformities and define new parameters to be introduced in the calibration procedure of the musculoskeletal modeling [*Alexander 2001, Jenkins 2003*].

- **Foot modeling**

As mentioned previously, the most important limitation of the skeleton modeling is considering the foot as a single rigid body. It is important to understand the biomechanics of the foot when dealing with pathologies [*Abboud 2002*]. Some experimental foot models have

been developed (Figure 4.5) [*Leardini 1999, MacWilliams 2003, Stebbins 2006, Baker 2006, Wright 2011, Carbes 20111*]. The Oxford Foot Model is proposed in clinical practice but not already routinely used.

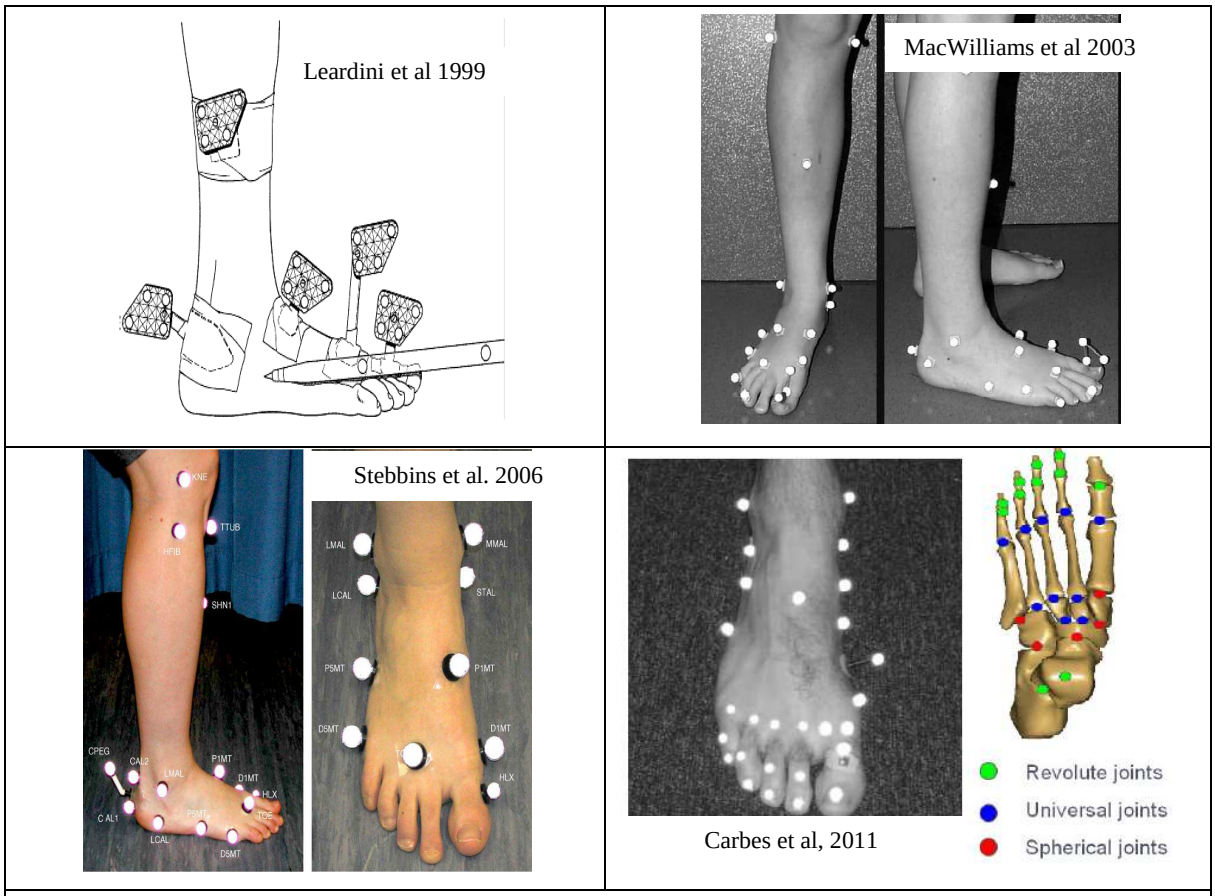


Figure 4. 5:.Experimental foot models

The Oxford Foot Model, a multi-segment model for calculating hindfoot, forefoot and hallus motion, is developed to describe both normal and pathological foot and ankle kinematics and dynamics in children. This model is important to monitor the progression of foot deformity [*Stebbins 2008*] and to clarify controversial findings from conventional lower limb kinematics (Plugin- gait model). It is becoming increasingly required for clinical routine and the decision-making process to plan management and assess its outcomes. This foot model is clinically tested and validated [*Stebbins 2006, McCahill 2008, Morris 2008, Curtis 2009, Wright 2011]*

We are convinced that a better description of the foot's dynamics during gait, provided from the Oxford Foot Model, for example, will add crucial information of foot deformities and then

enhance the evaluation of the CP using MSM. For this reason, the gait analysis protocol based on Helene Hayes could not be so efficient in our retrospective study.

- **Joint modeling**

Joints are represented by mechanical joints with only a torsional spring models in the three anatomical plans. The hip and ankle joint have three degrees of freedom, whereas the knee joint is limited only to the sagittal degree of freedom. With bone deformities, all joints are additionally constrained in terms of angular limits and degrees of freedom. In our study, when performing a retrospective study based on clinical available data, the lack of reliable information doesn't allow us to take into consideration such specific conditions. More realistic description of the joint models that reveal the complex behavior of the joints, especially in the cerebral palsy subjects, which might improve the consistency of the kinematic and kinetic data, has to be developed.

- **Muscle modeling and motor control**

The accuracy of muscle forces, predicted with a musculoskeletal model is highly dependent on both the muscle modeling and the optimization procedure.

Firstly, muscles are modeled as a primitive elastic system with a line of action between the insertion and the attachment positions. In addition to inaccuracies introduced by the rescaling anatomy process, mistaken estimated muscle activation and forces are attributed to muscle function parameters, the authorized maximum muscle lengthening and forces gathered from literature. And, when dealing with CP, it is important that the correct amount of force is attributed to the muscle to investigate its activation function during motion.

Secondly, redundancy problem of modeling human motion is mostly solved by optimization function to estimate the muscle forces. Many models minimize mechanical cost functions which are based on energy expenditure or muscle forces [*Erdemir 2007*]. However, relationship between these cost functions and the actual distribution of forces over the different muscles has never been demonstrated. With these mechanical cost functions, muscles with large moment arms and large PCSAs are preferred. Small muscles with short fibers and moment arm do not contribute, resulting in unrealistic synergies [*Praagman 2006*].

In our project, currently redundant muscles are countered by the use of closed loop muscle modeling. Each trained muscle has a PID controller that tries to match the instantaneous length of the muscle to the trained curve. These controllers are independent to each other, even though the resulting muscle lengths are coupled through the kinematic model. Under this scheme, without any additional considerations, it is arbitrary to get a solution, in which muscle activities match the trained position profile, which means that the motor control process is not represented. Studying healthy subjects with simulated pathological posture revealed that co-contraction has a great influence on musculoskeletal model‘s results. Predicted amount of muscles forces, in these studies, show the limitation of using closed loop muscle models to describe altered motor control. The same muscle function and control function give acceptable results when studying cerebral palsy cases. These controversy results admit that minimizing total muscle efforts could not be the accurate optimization law governing the pathological gaits [*Davids 2004, Fraysse 2009, Carrier 2011*]. Thus, the altered neurological control has to be reformulated through significant optimization laws rather than those representative of the natural walking, because they may not represent a pathological gait with spastic muscles and jerky movements [*Waters 1999, Davids 2004*].

Finally, the muscle spasticity, the principal characteristic of the cerebral palsy pathology was not introduced in our models. This hypothesis is reducing the clinical representation of such neurological pathology. In our knowledge, there is no research dealing with muscle spastic modeling from a mechanical point of view or introduced in the optimization law governing this kind of altered motor control. The lack of objective clinical measurements and representation of spasticity limits increasingly the development of the MSM of cerebral palsy. We are here at the border of technological and clinical knowledge.

In summary, it is likely that a physiologically more realistic optimization approach in combination with accurate and consistent muscle model parameters will outcome more accurate estimation of muscle forces.

- **Validation**

The validation of a musculoskeletal model is important to evaluate how accurate model’s outputs are. Up till now, the available sources for validation are very limited and the direct validation of optimized muscle forces is impossible. Consequently, there is an agreement in

the scientific community about the validation of the outputs of a contact model and the measured GRF and also the quantitative validation of the muscle forces based on the relation between the EMG measurements and the corresponding predicted muscle force [*Anderson 2001, Lloyd 2003*]. Our studies pointed out that this quantitative validation cannot be considered as a suitable technique when studying spastic muscles.

4.1.2.3. Calibration procedure

Despite significant advances in MS modeling, clinicians obviously need a clinically applicable model with accurate results and a realistic evaluation of the uncertainties and errors. Given the number of assumptions into a musculoskeletal modeling, a better understanding of hypothesis, the contribution of diverse parameters and a sensitivity analysis become crucial. In our project, we focused on several mechanical parameters, defining the MSM, as follows: the joint stiffness parameters, the joint center position, the motion agent's weights and the Training parameters. It was observable from previous chapter, that all these parameters have a big influence in MSM results. Despites its originality, the methodology we developed, still has many limits.

- **Intrinsic model parameters**

The Inverse and forward dynamic simulations are classical tools to calculate joint torques and estimate muscle activation and forces deriving from a specific movement. The accuracy of the MSM results is a certainly the consequence of several modeling assumptions. It is also highly dependent on performance of the inverse and forward dynamics simulations, such as the numerical calculation errors, noises and its amplification, numerical convergence conditions. The numerical training parameters (PID and PD controllers) in a musculoskeletal model are defined specially by numerical experimentation to end up with compromise between realistic results and numerical convergence of the mathematical models. Usually, these parameters are determined when studying normal gaits with healthy adult population. These parameters, training parameters, have a great influence in MSM outputs, as presented in preliminary sensitivity analysis (chapter 3), and managing them is essential to ensure realistic results.

In clinical practice, this sensitivity has to be reduced and quantified in order to better understand and correctly interpret the predicted results. For this reason, in order to reduce perturbations and ensure convergence, a set of PID and PD controllers' parameters (training parameters) has been carefully fixed is this study by ensuring the numerical convergence of the model and ensuring acceptable estimated muscle forces compared to literature. In addition, a set of the motion agent weight parameter was proposed that then avoids the unrealistic foot deformities on healthy populations and brings more contact stability during simulation and reducing noisy results.

- **Joint stiffness**

The joint is described as a visco-elastic model, in which the joint stiffness parameter is fixed arbitrary. In our study, the Dynamic joint stiffness is subject – specific determined using the collected gait analysis data. This entity represents the resistance that muscles and other joint structures manifest during motion as a reaction to an external moment of forces. Recently, it is widely explored as a clinical parameter to understand the effect of pathological gaits on joint functions. Introducing this parameter in joint modeling gives more realistic behavior to the joint activity and consequently to the muscles contributions.

Nevertheless, in this project, the dynamic joint parameter is only defined in the sagittal plan, as developed in clinical studies. There are no complementary studies that explored the dynamic joint stiffness in the coronal and axial axis. For that reason, the standard values proposed in LifeMOD model are attributed to the joint stiffness in the coronal and axial planes. But, pathological gaits doe not respect this condition and motions occurring in other planes can give an idea about compensatory gait strategies. It will be interesting to study the dynamic joint stiffness in different planes. But, when developing a MSM we have to be aware of the algorithm convergence and stability of the numerical simulations [Al *Nazer 2008*].

- **Joint center position**

The joint center positions are defined using predictive method of Davis et al (1991). Since our MS models have been compared to their representative kinematic models (plugin gait), we have limited our study to compare the influence of the joint center positions on MSM

outcomes, by comparing the Davis and Lifemod joint center positions. In the calibrated MSM, we placed the joint center position as recommended by Davis 1991, because it is widely used in clinical interpretations. But, recent studies about functional methods pointed out that Davis model is not accurate [*Besier 2003, Christopher 2003, Piazza 2004, Camomilla 2006, Ehrig 2006, Sangeux 2010*]. It is noticed that we tried to predict joint center positions with functional methods. However, determining precise joint center positions of pathological subjects by only using the walking data cannot give as any useful information for the musculoskeletal modeling.

To conclude about the calibration procedure, we are conscious that, in addition to relying on the rescaled generic musculoskeletal modeling, the great limit in our project is taking the plug-in gait biomechanical model as reference to evaluate the accuracy of the MSM results [*Schwartz 2004*]. Nevertheless, today, it's the only sophisticated clinical tool to interpret pathology and gait deviation indexing from normal gait standards. Recent studies are working on improving this clinical/ biomechanical model by increasing the marker's set protocol with the instantaneous determination of the accurate joint centers and axis positions [*Sangeux 2010, van der Zijden 2010]*. The development of such techniques to daily clinical use may improve clinical outcomes as well as musculoskeletal modeling.

4.2. Clinical relevance

4.2.1. Practical use of the MSM: which model for which use?

In our project, the CP subjects, from several gait pattern groups, with high quality gait analysis data acquisition and GRF measurements, are carefully chosen to facilitate the validation of the contact modeling in the MSM. Results, presented in chapter3, have shown that the standard rescaled generic model can correctly estimate the contact with the ground (GRF) and the muscle activation in most cases of Cerebral Palsy. GRF results concerning CP with jump gaits are fairly different and have to be carefully interpreted.

Hence, the standard musculoskeletal model may be used as an alternative to estimate the ground reaction forces, especially for cerebral palsy. Recording good GRF measurements cannot be satisfied for all CP population. In some cases, the foot step is so small that the

second feet cannot attain the following force plates. The measurements represent the resultant contact forces rather that the GRF attributed to each feet contact. Mathematically, deducing the GRF of each foot from the resultant forces is difficult, but may be feasible [*Wong 2010*]. But, in our knowledge, there are no studies dealing with this problematic. For these reason, a standard musculoskeletal modeling can give a general view of the shape and the values of the GRF during motion. However, according to previous remarks, it will be recommended to take into consideration the proposed set of weights parameters of the motion agents, to minimize contact instabilities and unrealistic bone deformities and foot positioning.

According to the muscle activation results with CP patients, the standard MSM may simplify the gait analysis procedure without using sophisticated equipment. It may help relieving the required EMG electrode placements. In clinical practice, the gait analysis exam may be reduced only to record kinematic and kinetic data, and then muscle activation could be determine and predicted relying to standard MSM. But, generic and standard musculoskeletal modeling is still qualitative approach to study pathological gaits.

The calibrated model is based on simple calibration of some MSM parameters, but gives more accurate results, compared to the kinematic model. Refining parameters attempts to give a physical meaning to some values, such as stiffness joints, and are easily to be determined from subject-collected anthropometric data and gait performances. Comparing to Plug-in gait biomechanical model, results from our calibrated MSM could be easily understood and correctly interpreted by clinicians. This musculoskeletal model reduces the human complexity, but gives the clinician a useful tool to study the surgical outcomes from retrospective review of patient with reasonably cost. However, as the validation is still the challenge of musculoskeletal modeling, the only use of the quantitative observation can be performed with the present procedure of musculoskeletal modeling.

4.2.2. Simulated pathological postures

In this research project, we also analyzed the effect of imitating pathological postures during gaits, compared to those observed in diplegic CP patients. In our study, healthy adults and children have performed gaits with pathological postures, such as crouch and jump gaits in order to investigate sagittal joint rotations, sagittal moments and EMG parameters of such gait

patterns. The objective was to analyze whether biomechanical constraints, induced by a a complex kinematical postures, were sufficient to explain sources of abnormalities on the gait parameters, kinetics and the muscle activities observed in CP patients.

4.2.2.1. Reproducibility of Simulated Pathological Postures

Focusing on kinematics, it has been demonstrated that healthy subject could imitate CP pathological gait patterns in a reproducible and homogeneous manner [*Thomas 1996, Romkes 2007*].

When comparing to literature results from CP population, It is important to precise that literature CP studies showed a large inter-variability references data, because of the large heterogeneity CP gait patterns, depending on location of the damaged brain [*Lin 2000, Rodda l994, Rozumalski 2009*]. The homogeneity of the gait patterns performed by healthy subjects, observed in our study, suggests that only main extreme characteristics of the CP gait patterns could be imitated.

We limited our study to jump and crouch gaits, because the inter-subject reproducibility was fulfilled and because modifying a gait pattern needs a learning phase, and the more complex the modifications are, the longer the necessary training time is. The indications given to the subjects, to imitate pathological gait patterns, concern only respecting the equines of the ankle during jump gait and the excessive knee flexion during crouch gait.

4.2.2.2. Simulated pathological posture vs CP gaits

The variations on kinematics, induced by imitating pathological gait patterns, lead consequently to variations on kinetics and EMG parameters.

In imitated jump gait, based on results presented in figures 3.30-3.32, required gait stabilities tend the subject to flex less the knee which explains differences in the muscle activation of the Bicep Femoris, and the Gastrocnemius and differences in the sagittal knee rotation during the stance phase in comparison with the reference data [*Lin 2000, Rodda l994, Rozumalski 2009*].

In imitated crouch gait, based on results presented in figures 3.29-3.31, the subjects tend naturally to compensate the excessive knee flexion by an excessive ankle dorsiflexion. The upper body, in this case, stayed behind the feet with the consequence of no extension moment for the hip and an early activation of the Rectus Femoris. Nevertheless, healthy subjects cannot perform a "perfect" imitation, because bone deformities and joint dislocation cannot be reproduced.

4.2.2.3. Clinical contribution

Despite these contrasted results, this study may help clinicians to understand how the pathological posture, defined especially as biomechanical constraints at joints, could explain sources of abnormalities observed in muscle activities of CP patients.

- **Gait Deviation Index**

Imitating pathological gait can be considered as sufficiently reproducible and comparable to real pathological gait. It could be helpful on calibrating the gait deviation index for a specific controlled alteration of the gait in relation with postures. To deal with CP children, the gait deviation index, a composite value based on various gait parameters used to globally quantify the gait quality [*Schwartz 2008, Baker 2009, Rozumalski 2011*] , is compared with those of normal walking children. Using the gait deviation index derived from imitated gait patterns might better enlighten the deviations of the CP children.

In this study, results of imitated gait patterns by healthy adults and children are similar. Clinical interpretation, based on healthy adult population, is preferred. First, the imitated CP gaits, performed by adults, are more reproducible, than those performed by children. Second, the mature gait patterns is established by seven years of age [*Sutherland 1988*] and the gait data remains similar to those of normal adults [*Ounpuu 1991*].

- **Musculoskeletal Modeling**

Based on musculoskeletal modeling results, the altered motor control introduced when imitating the crouch and jump gaits showed that the optimization laws defining a normal gait,

fail to predict suitable muscle forces in this case. Consequently, the optimization paradigm of the musculoskeletal coordination is still challenging when studying required compensation strategies to achieve a natural progression.

The analysis of the simulated pathological gait can improve musculoskeletal modeling to better understand cerebral palsy by exploring alternative criteria to solve the mathematical redundancy of musculoskeletal modelling for CP patients [1, 2] or developing more complex muscle function taking into account spasticity.

To sum up**,** the imitation of CP gait patterns by healthy subjects appears to be an interesting additional tool to analyze CP gait patterns. Our study demonstrated that healthy subjects are able to reproduce modified gait pattern in order to simulate a pathological attitude. The last two decades were dedicated to prove to the medical staff and society the benefit of gait analysis for CP patients, the next step will be to have reliable reference data based on large populations and multicentre sources. Therefore, imitating pathological gaits may point out that the altered muscle responses were not only the consequences of central nervous system impairments, but also the consequences of biomechanical restrictions at joints. Further studies may help clinicians to better understand the causes of CP gait disorders and thus to differentiate the consequences of biomechanical constraints of those caused by brain damage. With this knowledge, the effectiveness of surgical intervention in the management of CP will be better appreciated.

General Conclusion

The goal of this research project was to develop a musculoskeletal model for cerebral palsy children as a promising clinical tool to enhance surgical decision making, based on a retrospective study using the existing clinical gait data. To achieve this goal, we first explored the accuracy of a standard rescaled generic musculoskeletal modeling and then developed a calibration procedure of musculoskeletal model's parameter values, based on gait data collected from patients. The parameters taken into consideration are the visco-elatsic parameters of the joints, parameters of the contact with the ground and intrinsic parameters of the model.

Today, gait analysis has become an indispensable tool in the clinical management of patients suffering from a wide variety of medical conditions, such as Cerebral Palsy. Together with a musculoskeletal modeling, clinical exams aim to calculate relevant biomechanical parameters to guarantee clinical decision making without a profound technical background. In the absence of an optimized work-flow methodology and optimized time consuming to develop a personalized musculoskeletal modeling, we have to rely on rescaled generic models. Therefore, as a retrospective study, we focused on rescaled generic musculoskeletal. Several studies have been developed. Firstly, normal gait for healthy populations (adults and children) have been studied to determine the impact of rescaling on child skeleton model. Secondly, the influence of altered muscle activation on MSM results has been performed when healthy subject imitated representative CP pathological gait, crouch and jump gaits. Lastly, musculoskeletal models of cerebral palsy children with spastic diplegia have been developed.

Major results of our project may be grouped into three categories. These include the limits of the standard rescaled generic musculoskeletal models, the contribution of the proposed calibration procedure and the clinical contribution of simulated pathological gait patterns.

As concern the imitation of CP gait patterns by healthy subjects, it appears to be an interesting tool to better understand CP pathology using clinical gait analysis and to enhance defining limitations of musculoskeletal modeling. Our study demonstrated that healthy subjects are only able to reproduce extreme pathological attitudes. In addition, imitating pathological gaits may point out that the altered muscle responses were not only the consequences of central nervous system impairments, but also the consequences of biomechanical restrictions at joints. Further studies may help clinicians to better understand the long-term consequences of

CP gait disorders and thus to differentiate the consequences of biomechanical constraints of those caused by brain damage. With this knowledge, having a reliable reference data based on large populations and multicentre sources, will become clinically interest to better appreciate therapeutic management of cerebral palsy.

As concern musculoskeletal modelling, compared to gait analysis data, chosen as a reference in this project, musculoskeletal modeling results showed that even if the model outputs gave correct results with healthy adults, the standard rescaled generic musculoskeletal modeling showed limits on predicted kinematics and muscle forces for healthy and CP children. These can be explained, first, by the fact that, despite the large anthropometric LifeMOD database, skeleton's rescaling procedure gives not quantifiable inaccuracies, which should be taken into account when interpreting MSM results. Second, joints and muscles are modeled as simple visco-elastic model usually do not represent, neither healthy child characteristics nor CP ones. The standard musculoskeletal model's parameters does not represent child description, it represents only healthy adult population. Finally, the motor control functioning used in LifeMOD, was chosen as a compromise to have computational efficiency at the expense of a complex biomechanical considerations. This last limitation was emphasized when considering predicted muscle of imitated pathological crouch and jump gaits. Imitating pathological gait patterns raised the issue of the current optimization approach to study muscle co-contractions, compensatory strategies and non-optimized gait patterns.

The calibration procedure proposed to refine some musculoskeletal model's parameters based on the clinical gait analysis data (e.g. stiffness joint, motion agent and intrinsic model's parameters). Additional to this simple procedure of calibration, driving the model with the experimental Ground Reaction Forces data have a huge influence in model's outputs and it improved quantitatively the predicted muscle activations and estimated forces.

The results of our studies showed how rescaled generic models fail to accurately predict muscle activation and forces of children populations. Whereas, the resulting differences between the rescaled generic and the calibrated models reflected the effect of the aberrant choice of musculoskeletal model's parameter when studying healthy children and pediatric CP subjects. This work pointed out that, in addition to geometrical specificities of subjects (healthy or CP ones), the parameters of a rescaled generic musculoskeletal model play an

important role in model's outcomes and can be refined and calibrated to improve model's outcomes.

The main limitation of our work was the use of the kinematic plug-in gait model as reference to evaluate the kinematical musculoskeletal model's results. Despite this limitation, our retrospective study may represent a new perspective in clinical applications.

In the absence of sophisticated gait analysis equipments, the standard rescaled generic musculoskeletal model may be used as an alternative to estimate the ground reaction forces and muscle activations. It will be helpful, especially for post-surgical gait analysis, which is not routinely performed. Predicted ground reaction forces of CP with jump gaits have to be carefully interpreted. The calibrated musculoskeletal model, in turn, uses subject-collected anthropometric data and gait performances and serves to refine some MSM parameters. The calibration procedure attempts to give a physical meaning to several parameters. Correlated to the plugin-gait biomechanical model, results from the calibrated MSM could be easily understood and correctly interpreted by clinicians. Even if this model reduces the human complexity, but gives the clinician a useful tool to study the surgical outcomes from retrospective review of patient with a reasonable cost.

Possible perspectives of this work include gait experimentation and musculoskeletal modeling improvements.

At first, performing additional experiments and developing powerful daily clinical techniques of gait analysis are crucial to give the possibility to calibrate more musculoskeletal parameters by increasing the marker's set protocol, including additional foot markers, determining instantaneous accurate joint centers and axis positions. The development of such techniques may improve clinical outcomes, as well as musculoskeletal modeling.

Second, improvements on musculoskeletal modeling are required:

- Improving joint models, to represent abnormal joint functioning due for example to bone deformities and developing new joints to take into consideration secondary joints of the foot, in the case of equinus foot.
- Improving muscle and motor control process model, to represent the muscle co-contractions and spasticity.

The challenge of musculoskeletal modeling is the validation. Until today, the validation is based only on qualitative observation. Because of that, relying on musculoskeletal modeling to patient's management decision is still not appropriate.

During this PhD, we mainly determined the limits of the standard rescaled generic musculoskeletal modeling by testing several hypotheses, which include the young healthy and CP population, and the altered muscle activation by imitating pathological gait patterns, and then we developed a calibration procedure of the model's parameters, base on gait analysis data. Nevertheless, the presented solutions for calibrated musculoskeletal modeling are flexible and general enough to test several MSM parameters and can be willingly applied to other patient populations where musculoskeletal models can provide relevant information concerning their motor disorders.

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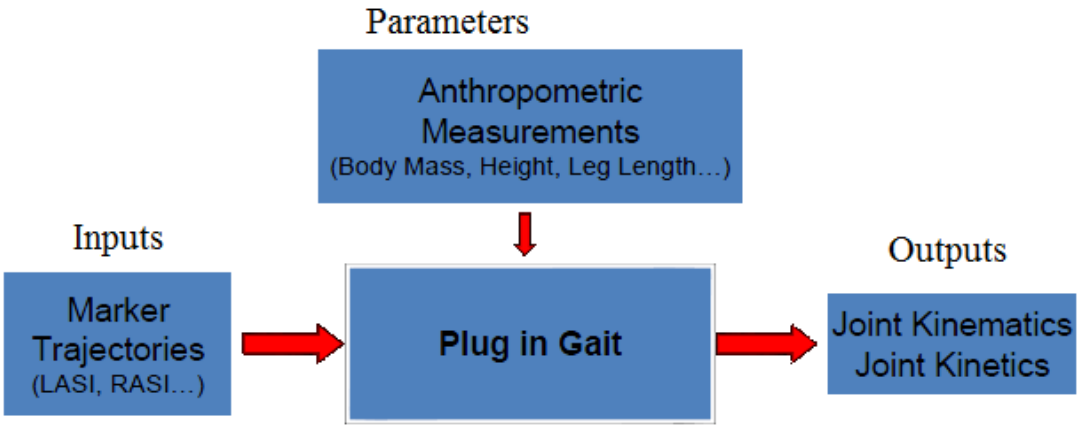
Annex 1:
Conventional Gait Model: Plug-in Gait

Plug-in Gait is a biomechanical model for the lower limbs developed by Kadaba et al. (*1989*) and Davis et al. (1991), implemented in the commercial software of Vicon.

The biomechanical model requires anthropometric measurements of the subject and different markers' trajectories to calculate joint kinematics and kinetics.



1. **Required anthropometric measurements**

The anthropometric data measurements include height, body weight, the distance between the anterior and posterior superior iliac spines (ASIS – PSIS) and the leg lengths of lower limbs measured from the greater trochanter to the knee joint center and from the knee joint center to the lateral malleous. These measurements are important for the calculation of the thigh, calf and foot centers of mass, estimated using regression equations developed by Winter et al. (1990), the calculation of the inertial properties and the center joint location (Davis 1991).

2. **Marker set: Helene Hayes clinical model**

The standard Helene Hayes clinical protocol (*Davis 1991*), is frequently used in clinical practices (figure1).

Figure 1: Marker Label Definition Position on Subject

The marker placements are defined as follows.

Marker's name		Marker's position
SACRUM	SACR	On the skin mid-way between the posterior superior iliac spines (PSIS) and positioned to lie in the plane formed by the ASIS and PSIS points
Left / Right ASIS	LASI RASI	Left / Right anterior superior iliac spine
Left / Right PSIS	LPSI RPSI	Left/ Right posterior superior iliac spine (immediately below the sacro-iliac joints, at the point where the spine joins the pelvis)
Left / Right thigh	LTHI RTHI	Over the lower lateral 1/3 surface of the left/right thigh
Left / Right knee	LKNE RKNE	On the flexion-extension axis of the knee
Left / Right tibia	LTIB RTIB	Over the lower 1/3 surface of the shank
Left / Right ankle	LANK RANK	On the lateral malleolus along an imaginary line that passes through the transmalleolar axis
Left / Right heel	LHEE RHEE	On the calcaneous at the same height above the plantar surface of the foot as the toe marker
Left / Right toe	LTOE RTOE	Over the second metatarsal head, on the mid-foot side of the equinus break between fore-foot and mid-foot

3. Local anatomical frames and joint center Location

In the plug-in-Gait, lower limbs are modeled as 7 rigid segments: Pelvis, Left/Right Thigh, Left/Right Shank, and Left/Right Foot. The markers positions were used, first, to define the local reference systems $(\vec{i}, \vec{j}, \vec{k})$ for each segment to predict the joint centers and segment endpoint, and second, to use these joint center positions and external marker positions for

generating segment global reference frames $(\vec{x},\vec{y},\vec{z})$, which are embedded at the centers of gravity of each segment.

❖ Pelvic anatomical frame and hip joint position

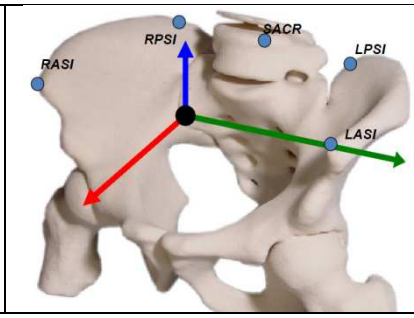
The pelvis segment coordinate system is defined from markers placed at the pelvis, RASIS, LASIS, LPSIS and RPSIS or Simply the Sacrum marker (the midpoint of the two posterior markers). The midpoint of the two ASIS markers defines the origin of the anatomical frame of the pelvic segment. The Yp axis is oriented along the line passing through the ASISs in the direction from the right to the left ASIS marker. The Zp axis is defined as the perpendicular axis to the plane composed by the markers of the pelvis. The Xp-axis is the third axis of the orthogonal frame, is mutually perpendicular to both the Yp-axis and the Zp-axis.

$$O_p = \text{RASI} * \text{LASI}$$

$$\vec{y}_p = \frac{(\text{RASI} - \text{LASI})}{\|\text{RASI} - \text{LASI}\|} = \vec{j}_p$$

$$\vec{z}_p = \frac{(\text{Sacrum} - \text{LASI}) \times (\text{Sacrum} - \text{RASI})}{\|(\text{Sacrum} - \text{LASI}) \times (\text{Sacrum} - \text{RASI})\|} \vec{k}_p$$

$$\vec{x}_p = \vec{y} \times \vec{z} = \vec{i}_p$$

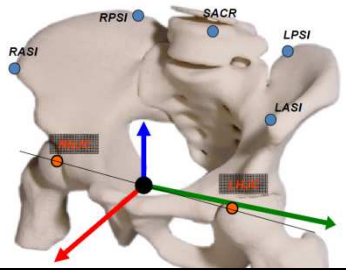


According to Davis et al. (1991), the hip joint center is positioned in the pelvis coordinate system as following:

$$X = C*\cos(\theta)*\sin(\beta) - (\text{AsisTrocDist} + mm) * \cos(\beta)$$

$$Y = -(C*\sin(\theta) - aa)$$

$$Z = -C*\cos(\theta)*\cos(\beta) - (\text{AsisTrocDist} + mm) * \sin(\beta)$$



where

θ= 0.5 radians, β=0.314 radians, aa = $\frac{1}{2} * \|\text{RASI} - \text{LASI}\|$, mm = the marker radius

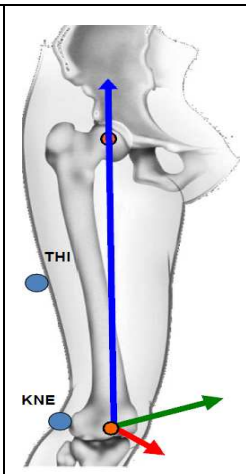
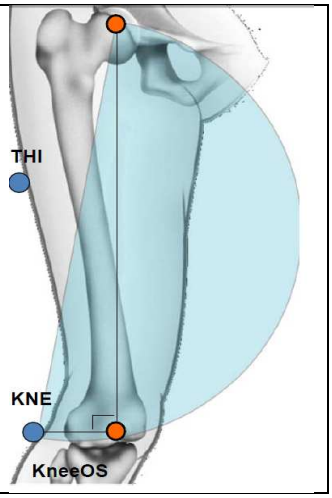
AsisTrocDist = 0.1288 * LegLength - 48.56 ; C = LegLength*0.115 - 15.3

❖ Thigh anatomical frame and knee joint center position

The KJC is defined as the point at distance KneeOs from the Knee marker (KNE) in the plane defined by KNE marker, THI marker and HJC:

$$KneeOS = \frac{Mar\,\text{ker}\,Diameter + KneeWidth}{2}$$

The Thigh Anatomical reference system is then defined as:

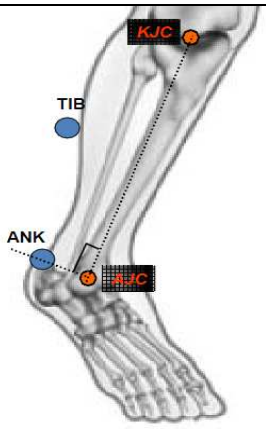
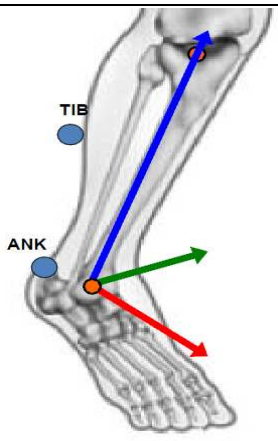
Origin : KJC ***Z axis direction*** : KJC→ HJC ***X axis direction*** : perpendicular to the plane defined by (HJC, KNE, THI) ***Y axis direction:*** Cross product between Z and X unit vector		

❖ Shank anatomical frame and ankle joint center position

The AJC is defined as the point at distance AnkleOs from the Ankle marker (ANK) in the plane defined by ANK marker, TIBmarker and KJC.

$$AnkleOS = \frac{Mar\,\text{ker}\,Diameter + AnkleWidth}{2}$$

The Shank Anatomical reference system is then defined as:

Origin **:** AJC ***Z axis direction*** : AJC→ KJC ***X axis direction*** : perpendicular to the plane defined by (KJC, AJC, TIB) ***Y axis direction***: Cross product between Z and X unit vector		

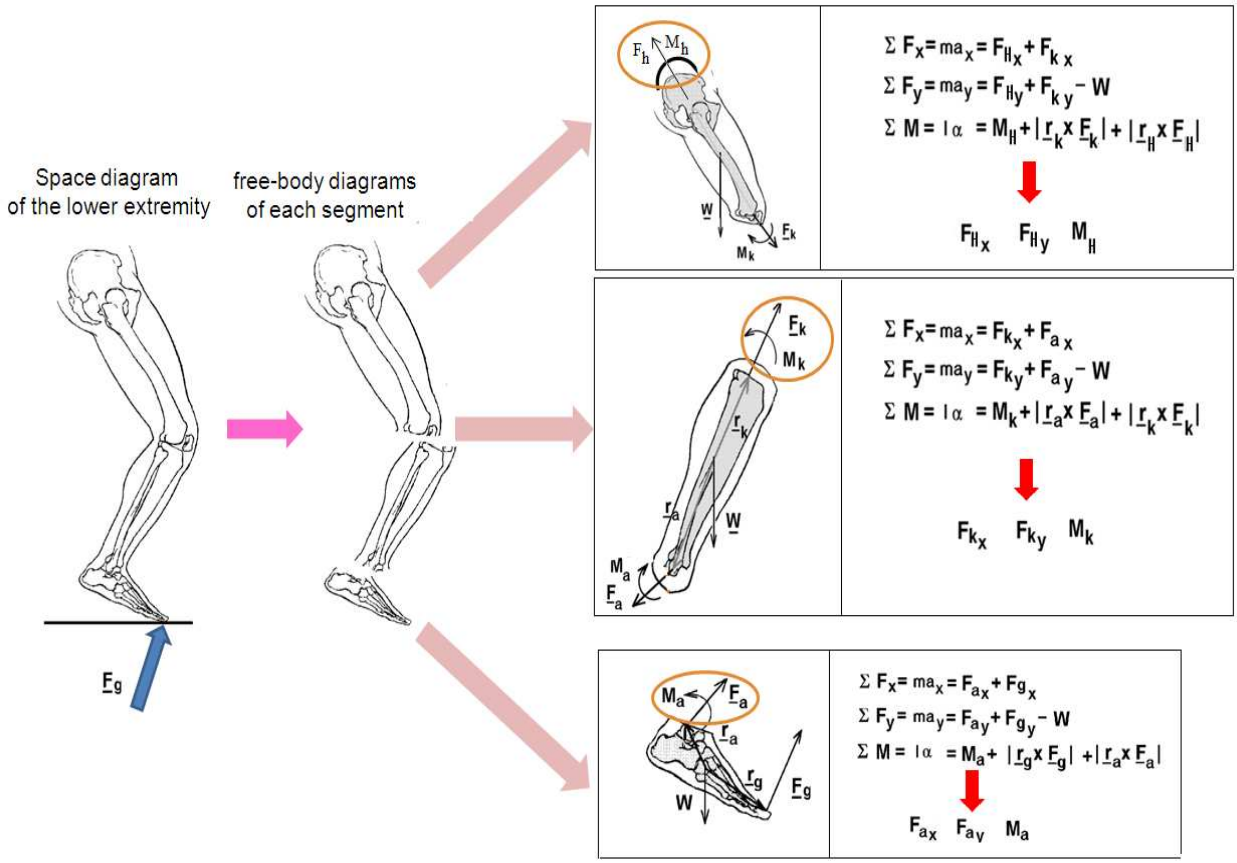
4. Kinematics calculation

Joint angles are the relative angles between two rigid segments, always ordered as flexion-extension angles, adduction-abduction angles and internal and external rotation angles. The

oint angles are determined using both Euler rotation angle definitions [*Ramakrishnan 1989, Kadaba 1989, Kadaba 1990*].

5. Kinetics calculation

By solving the equations of motion for the segments of the lower limbs, in which external forces are the Ground Reaction Forces (GRF), the net joint forces [N/Kg], the joint moments [N.mm/Kg] were estimated. The joint power [W/Kg] is the scalar product between joint moments and joint angular velocities.



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Annex 2: LifeMOD and GEBOD database

The human model in LifeMOD includes 19 rigid segments. Their general properties and dimensions are created using data from the GeBOD anthropometric databases.

The GeBOD database creates a human model based on simple descriptions such as gender, age (for child), height and weight. The gender and the age are required to define which anthropometric database to use (Snyder et al, 1977, McConville et al. 1980, Young et al.1983). The weight and the height are used for rescaling procedure through regression equations, as follows.

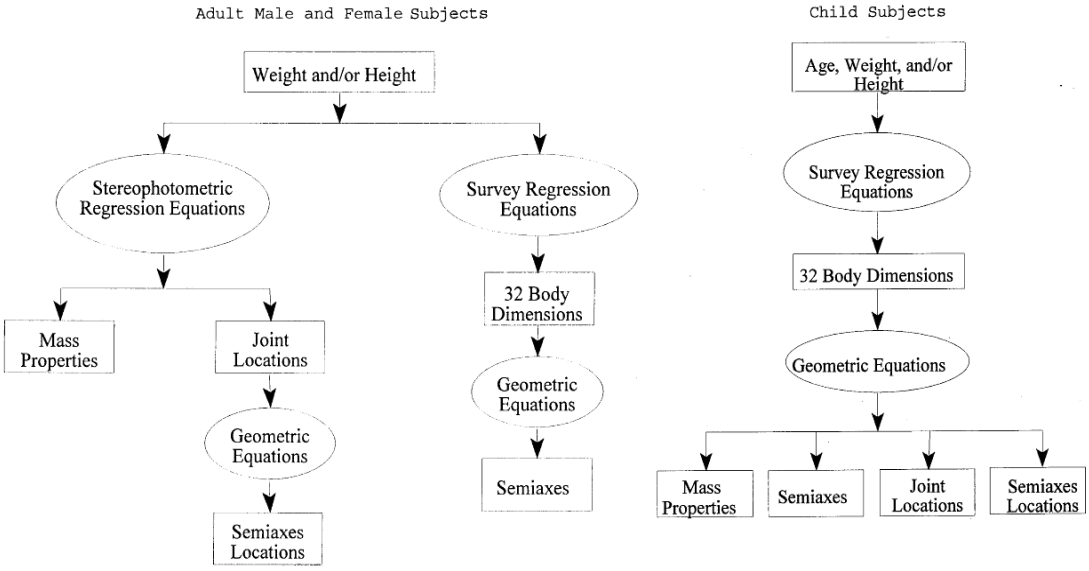

Figure1: Procedure used in generating Adult and Child models

In GEBOD, there are 4 groups of regression equations which are used to determine the body dimensions, the joint location coordinates, the segment volumes and principle moments of inertia, detailed in *Cheng 1994*.

As an example, for the shoulder, the regression equation using both the weight and the height has the best prediction ability:

```
Shoulder Height = 0.07182(Body Weight) + 42.77
Shoulder Height = 0.8751(Standing Height) - 3.936
Shoulder Height = 0.00755(Body Weight) + 0.8469(Standing Height)
                  - 3.096
```

From this description the **body measurement** parameters are created, as shown in figure 2.

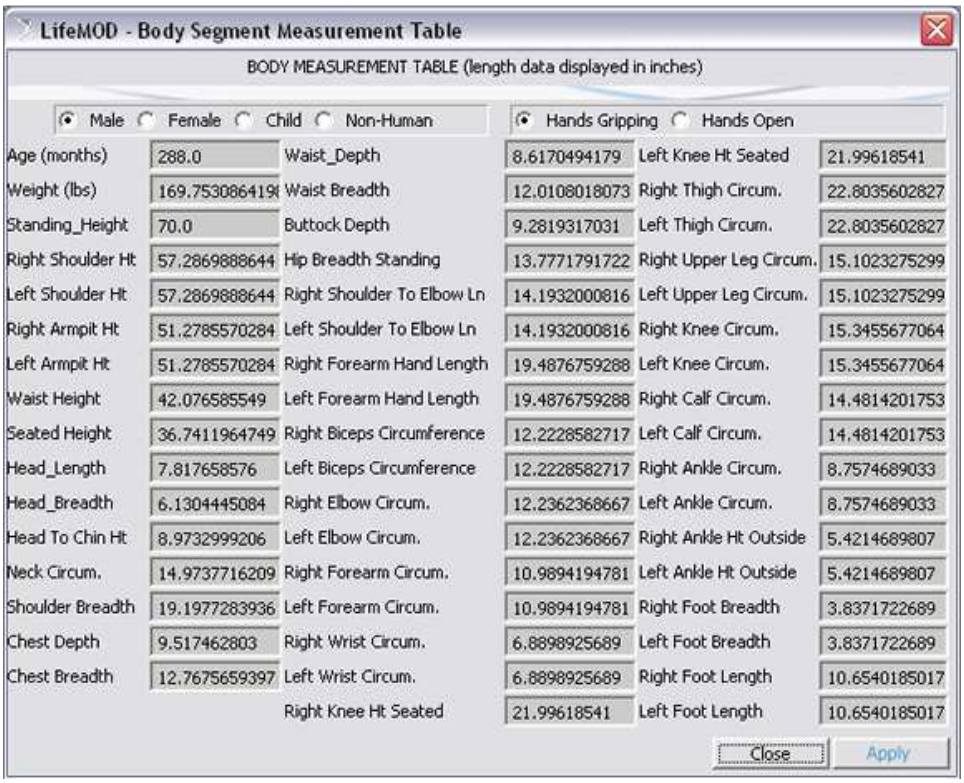

Figure 2: Body Segment Measurement Table created from the GeBOD anthropometric library

The joint locations are determined based on the sterophotometric data from *Snyder et al, 1977, McConville et al. 1980, Young et al.1983,* and defined in the standard anatomical position with the origin on the floor (Figure 4). The rigid segments are then presented as a set of ellipsoid Semiaxes, to give a proper appearance covering between two adjacent joints (Figure3).

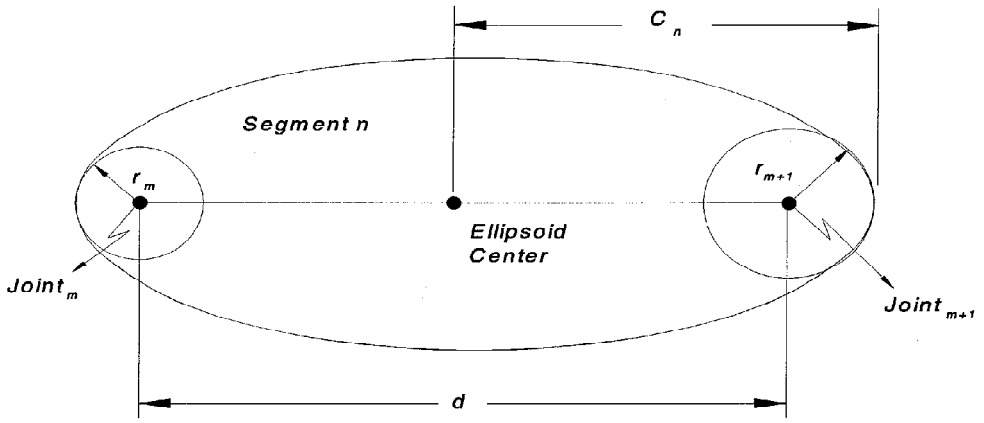

Figure 3: Ellipsoid Semiaxes

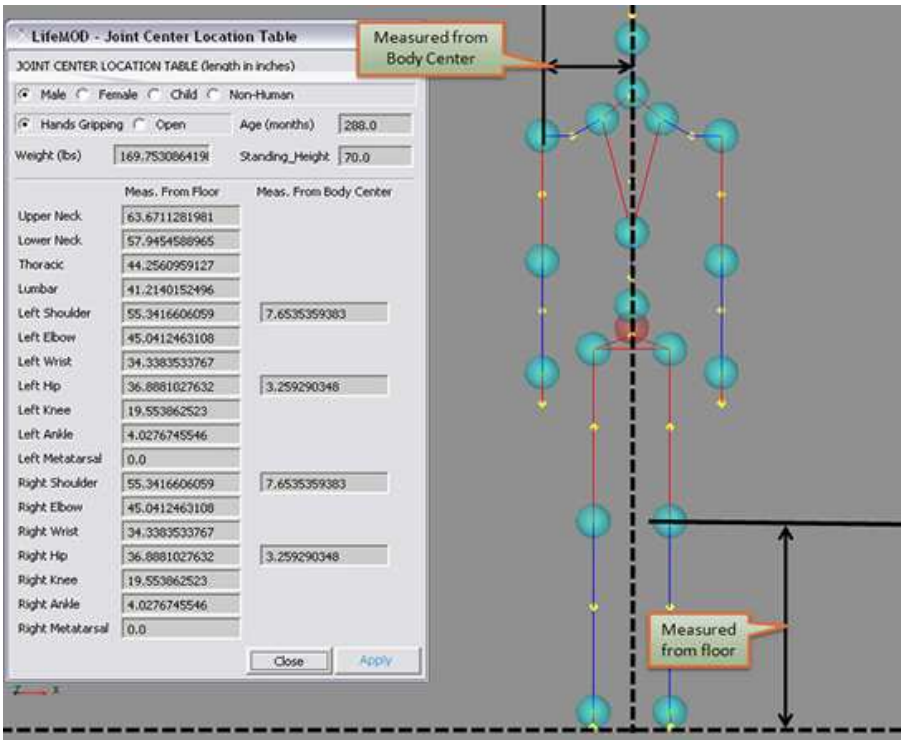

Figure 4: Joint Center Location Table created from the GeBOD anthropometric library

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