Theoretical and computational model of the human gait to simulate pathological patterns.

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About Success

"Nothing can stop the man with the right mental attitude from achieving his goal; nothing on earth can help the man with the wrong mental attitude." - *Thomas Jefferson*

"There is too much emphasis on success and failure and very little on how a person progresses through the effort. Enjoy the journey, enjoy every moment and stop worrying about victory and defeat." -

Matt Biondi

Acknowledgements

To my advisor Eduardo Romero.

To my friend Fabio Martinez.

To my girlfriend Laura Marconi.

To my family.

To CIM&LAB and BioIngenium.

Abstract

The gait analysis is an essential tool in clinical rehabilitation. It facilitates the diagnosis, treatment, monitoring and implementation of methodologies that mitigate the effect of some pathologies associated with the movement. This analysis use a systematic quantification, follow up and interpretation of the temporal sequence of movements that characterize human locomotion. This analysis requires a collection of kinematic and kinetic data that describes the displacements, angles and forces on the lower limbs and its joints during a gait cycle. These data are obtained from three primary sources of information: video, Electromyography (EMG) and force platforms[4], which are analyzed in a clinical report. However, the human gait analysis and the interpretation of its dynamic patterns are completely dependent on the experience of the specialist. To get this experience, it is required an expensive training for several years, coupled with the need to have a high number of patients with different patterns and pathologies that allow the training specialist to identify the characteristics of each movement (Gait Patterns) with certain degree of certainty.
A strategy for reducing training time and the required number of patients is the use of modern computational tools that optimize the interaction and therefore learning. The formulation and implementation of models that describe the dynamics of the normal and pathological gait, are part of this strategy. This thesis aims to develop a computational model of human gait that allow to describe normal and pathological movements using a simple biomechanic representation to describe completely the CoG displacement, and a inverse kinematic approach to obtain the clinical gait description (Kinematic Gait Patterns).

Keywords: Gait Analysis, Biomechanical Modeling, Human Movement, Rehabilitation Engineering

Resumen

El análisis de la marcha es una herramienta fundamental en medicina de rehabilitación. Facilita el diagnóstico, tratamiento, seguimiento e implementación de metodologías que ayuden a mitigar el efecto de patologías asociadas con el movimiento. Este análisis utiliza una cuantificación sistemática, seguimiento e interpretación de la secuencia temporal de movimientos que caracterizan la locomoción humana. Este análisis requiere de una colección de datos cinemáticos y cinéticos que describen los desplazamientos, ángulos y fuerzas ejercidas sobre las extremidades inferiores y sus articulaciones en un ciclo de marcha. Estos datos son obtenidos de tres fuentes principales de información: Video, Electromiografía Dinámica (EMG) y plataformas de fuerza, las cuales son analizadas en un reporte clínico. Sin embargo, el análisis de la marcha humana y la interpretación de sus patrones dinámicos es completamente dependiente de la experiencia del especialista. Para lograr ésta experiencia, se requiere de un extenso entrenamiento por varios años, junto con la necesidad de contar con un número grande de pacientes con diferentes patrones y patologías que permitan al especialista en entrenamiento identificar las características de cada movimiento (Patrón de marcha) con cierto grado de certeza.
Una estrategia para reducir el tiempo de entrenamiento y el número de pacientes requeridos es el uno de herramientas computacionales modernas que optimicen la interacción y el aprendizaje. La formulación e implementación de modelos que describan la dinámica de la marcha normal y patológica hace parte de ésta estrategia. Esta tesis desarrolla un modelo computacional de la marcha humana que permite describir movimientos normales y patológicos usando una representación biomecánica simple para describir el completamente el desplazamiento del CoG, y un sistema de cinemática inversa para obtener una descripción clínica de la marcha (Patrones de marcha).

Palabras clave: Análisis de Marcha, Modelamiento Biomecánico, Movimiento Humano, Ingeniería de Rehabilitación

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1. Introduction

The movement, thought of as the relative change of a system's components, expresses the very basic language of life, from cell or organic scales down to genetic ones. In complex organisms, movements may be classified in function of the degree of upper control, either reflex or voluntary [70]. Reflexes can be understood as the response of living organisms to the necessity of transporting organic material within biological systems, after some interaction with the external environment and with very little internal control, while voluntary means that there exists a kind of control for specific tasks such as search or locomotion. Quantification of these movements is a fundamental step towards the comprehension of the structure and dynamics of every living system [29]. Furthermore, many clinical or diagnosis decisions are based on measurements [67, 68] which are used in the clinical practice as biomarkers, currently defined as particular patterns associated to a certain degree of any disease. The success of many of these potential biomarkers depends on a proper understanding of the system dynamics.

The human gait is the result of complex interactions between several sub-systems: neuromuscular, musculo-tendinous and osteo-articular, which work together to generate the body dynamics that underlies the bipedal desplacement [93, 92, 39]. In despite of the intensive research in biomechanics [25], robotics [11, 21], medicine [24] and computer animation [15, 16], the biological complexity has hindered a proper understanding of the locomotor system. This problem has been overcome in the clinical routine by a gait description, inferred from the traditional gait laboratory and aimed to establish the set of patterns that better describes the system dynamics, i.e., the complex relationships between the neuromuscular, musculo-tendinous and osteo-articular sub-systems [66, 24, 91]. Usually, a physician or rehabilitation expert determines whether there exists pathological gait patterns using statistical tests on the gait laboratory's data, using for doing so her/his expertise [4, 7, 57, 69], a subjective and expert-dependent methodology [26, 47, 93]. Therefore, development of gait models which achieve a quantitative movement description has become a priority to support the physician decisions [26, 24, 74, 93].

Many models have been previously proposed for simulating the human gait in different areas of applications. A first group includes bipedal descriptions built upon structural information, whereby determining the complex relationships between muscles and joint angles by simple mechanical systems as the inverted pendulum [23, 58, 27]. These approximations allow an appropriate locomotion description from an energy standpoint, simulating the change from the kinematic to potential energy phases during the gait cycle and allowing a coarse classification between normal and pathological gaits [27, 57]. However, these approaches are very

limited to describe early stages of any disease. Based on these physical models, McGeer [58] formulated the theory of passive dynamic motion, which describes the movement without a complex control mechanism, that is to say, the important element to understand the body's structure is the gait itself rather than its control or muscular activity [33, 45, 11, 84].
Afterwards, Garcia et.al.[27] introduced a gait model using a double articulated pendulum system, a very simple structure for which the feet are relatively small with respect to the trunk and the heelstrike follows a very restrictive rule. Kuo [48, 49, 50] modified this Garcia's model by adding an element that describes the hip angular moment produced by the heel strike, but not accurate enough as to describe the heel strike in the gait cycle. A main drawback of these approximations is that about a 30 % of the gait cycle that corresponds to the double stance phase is completely eliminated. Overall, these physical models are useful in areas like robotics since they rule out the dependence on a robust control mechanism. However, in medical applications they are very limited because of their strong simplifications, missing relevant gait aspects such as the non-linearities introduced by the heel strike.
Likewise, a complete clinical gait analysis requires a muscular description to identify musculoskeletal disorders like hemiplegic movements [15, 93, 24]. Some models, capable of simulating muscles and tendons, introduce new elements that store and control the energy during the locomotion process. One of the most seminal works was proposed by Hill [35], who described an specific muscular group using three basic components: a contractile element, which represents the muscular fibers and two non-linear spring elements, one in series and another in parallel, which stand for the connective tissue and tendons, respectively. There exist some gait approximations that include muscle relationships using the Hill model [93, 92, 47], but each muscle works independently of a locomotor structure [39]. In these approaches, each model accelerates a specific body segment, allowing to obtain a basic simulation pathological movements. Nevertheless, these models are not accurate enough to describe the complex interaction among different group muscles and also, they require a certain quantity of parameters that need to be tuned. On the other hand, Scott Delp [14, 15] introduces a computational musculoskeletal model that combines the Hill muscle and structural physical models. This models parameterizes specific morphologic relationships to simulate normal and pathological patterns. The drawback is that this model has a high degree of subjectivity and requires expert interventions.
This thesis develops a computational model for describing the human motion using the dynamic patterns. This model is based on a complete and objective description of the CoG for a complete gait cycle, starting from the commonly used approach: *inverted pendulum*, which is modified and extended to represent normal and abnormal movements. This biomechanic representantion allows to simulate both normal and pathological disorders (Parkinson, Diplegia), and represent it using clinical motion description (dynamic patterns), achieving an average correlation of 0.87, that is calculated between the patterns obtained by the model and the patterns of actual patients.

2. Theoretical and Computational Models of Normal and Pathological Gait: A review

The human gait is described as a set of alternating movements of the limbs and trunk, generated by the coordinated action of the locomotor system. This system may be affected by a large set of injuries and diseases that may significanthy alter the natural gait dynamics [93, 24].

In the clinical routine, the physician analyzes the motion patterns (kinetic, kinematic and energy consumption) for each patient using the gait laboratory's data [4, 7, 69, 56]. With the obtained information, the physician starts a treatment and follows up the patient evolution using statistical information obtained from population studies and from their own experience. This methodology is of course highly subjective and depends on the specialist knowledge [24]. Consequently, the development of models that perform a quantitative movement description has become a priority to support the medical act [93, 24].

This chapter presents a review of the different models that have been developed to describe the gait dynamics. A description of human gait is then introduced in terms of the dynamic parameters that characterize it and discusses the models inspired by the inverted pendulum theory. A description of the models that considers also the structure of the body is also presented. Finally, some models that take into account the muscular interaction with the neuromuscular control system are analyzed. The strengths and weaknesses of the models are presented in terms of their potential to describe normal and abnormal gait patterns, and its applicability in the clinical field in the conclusion section.

2.1. The Clinical Gait Analysis

In the clinical routine, the gait analysis allows to identify certain normal or pathological movements, using a simplified model of the human body structure. This analysis is performed at different levels and with complementary techniques that assess different aspects of neuromuscular function, as illustrated in Figure **2-1** and described below [80, 38].

- Kinematic variables describe the body displacement by recording the angular joint variations and the relative body segments motion in space. Similarly, they record the

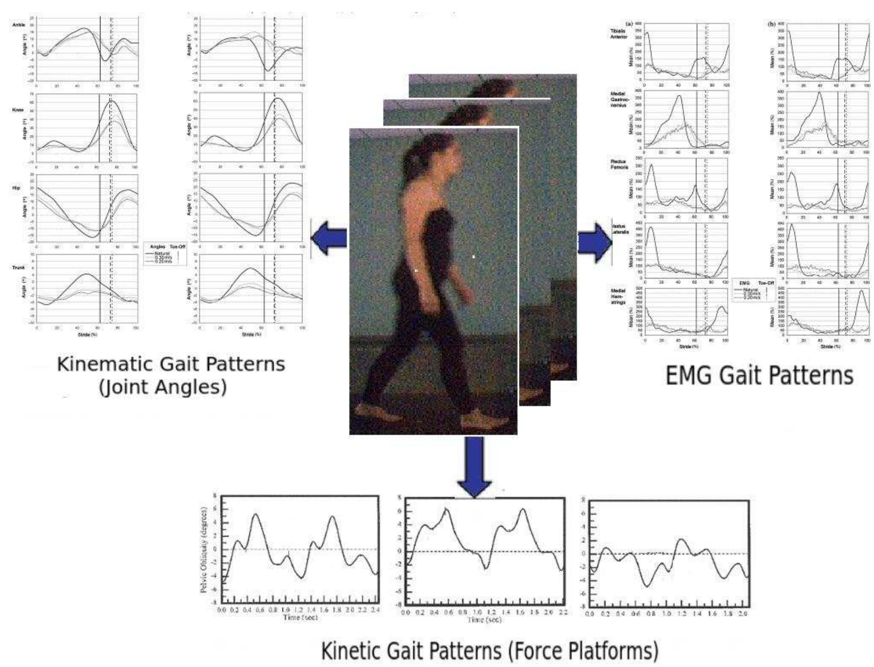

Figura 2-1.: The human gait can be described and analyzed using the dynamic patterns obtained in a Gait Laboratory

step length and movement speed [42].

- Kinetic parameters quantify the relations between action-reaction forces, moments and powers of each body segment [42]. The kinematic variables allow to describe the vertical oscillation of the center of gravity (CoG) and analyze the mechanical energy absorption by the joints.

- Muscle activation variables: evaluate the electrical muscle activity during the gait cycle. These variables are recorded by the EMG (dynamic surface electromyography) [93, 92, 43]. In addition, other variables recorded obtain a ratio of energyconsumption during movement [59, 52].

Based on statistical analysis conducted on large populations groups, physicians defined normal motion ranges, and gait abnormal patterns [52], including:

- *Diplegic Gait:* neuromuscular pathology characterized by a disturbance of the neuromotor commands, resulting in a gait with uncoordinated movements and a high degree of spasticity in lower limbs. During gait, hips and knees are flexed and ankles keep in adduction and an extended internal rotation, while the trunk usually remains tilted [42].

- *Parkinsonian gait:*This movement is caused by a degenerative disorder of the central nervous system. It is mainly characterized by muscle rigidity, tremor and slow movements at each of the body segment. During the gait patient lean forward, the progress is slow at start, the step length is small and often shuffled [42].

In despite of the wide range of diseases and abnormal movements that can be found in a gait laboratory, these descriptions are insufficient in the early diseases stages. For example, in the *Trendelenburg gait*, characterized by trunk inclination towards the supports side, there are patterns that indicate weakness and paralysis of the gluteus medius. These patterns can also describe a displaced hip due to mechanical factors that reduce the lever arm of the abductors [42]. On the other hand, it is important to strength out that the normal gait variability is large enough as to hide the changes that occur in the early stages of Parkinson or any other pathological gait. This has replaced the potential of the gait laboratories as a monitoring tool [42, 93, 92].
These diagnoses allow the practical management of these diseases, but they are not "per-se" entities to analyze. The biggest problem is that the physician does not have any tool that can predict the consequences of their intervention. In this sense, a model that articulates a generic sub-systems of the gait would allow more specific analysis and above all, to estimate the consequences of any treatment [93, 92, 33]. Below are described different models of human motion with different levels of detail and precision.

2.2. Structural Models

A first approach to simulate the human locomotor system consists in studying only the structural relationships: body mass, lengths and angles. Two different schools have used these relations. The first assumes that the CoG [93] describes a linear path, while the second, based on the models that include an inverted pendulum, which promotes the idea of a CoG with a cyclic path (see Figure **2-2**).

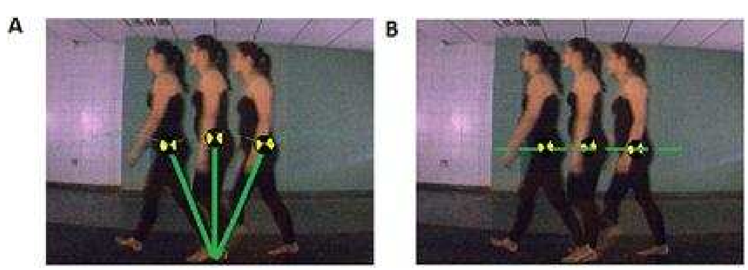

Figura 2-2.: The figure shows a comparision between the two theories that describe the gait using simple structural relationships. A. Inverted pendulum model. B. Six determinants of gait model

This first approach used to describe patterns of normal and pathological gaits was proposed by Saunders et al. [75], using relationships established by *the six determinants of gait* [93], with which it was inferred that any significant change in the trajectory of the CoG indicated an increased energy expenditure, which is directly related to the presence of a pathology [75, 38].
Multiple research studies, mainly in the clinical area [4], have evaluated the importance of these hypotheses. However, Gard and Childress [28] have shown that the determinants associated to the pelvic motion contribute little or nothing to the vertical attenuation of the center of gravity. This fact was later verified by simulating the angular gait patterns [64]. The rectilinear trajectory would be possible if the legs were wheels, a condition which, besides being impossible, would increase the energy consumption required to generate the movement [75].
The second approach proposes a biped model by simplifying the human body structure, using simple mechanical analogies as the inverted pendulum as shown in Figure **2-3**-A: the legs are represented as rigid structures of length l that support a body with mass m.

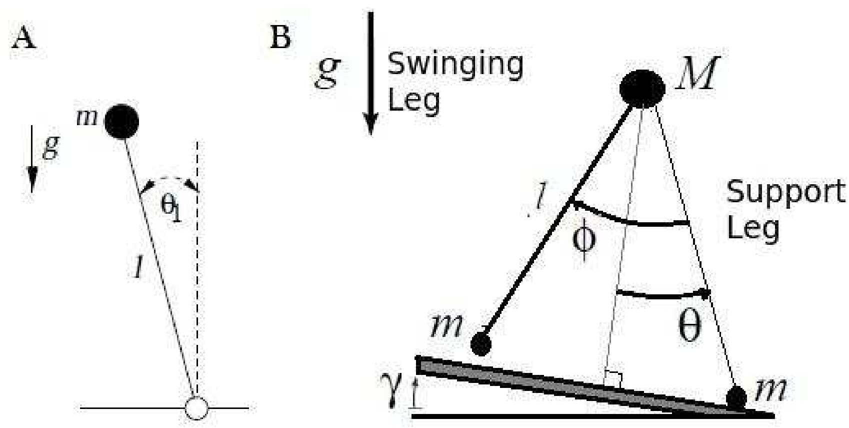

Figura 2-3.: The figure shows two approaches based on the inverted pendulum. A. The diagram of an inverted pendulum system. B. The approach proposed by Garcia et al, using the double inverted pendulum represetantion.

The path described by the inverted pendulum can approximate the global movement of the human body as $\dot{\theta}(t) = \frac{g}{l}cos\theta_1$,generating a semicircular movement that approximates the center of gravity (CoG) pattern observed in gait laboratories [1], but no-linear human movements are impossible to describe using this mechanical representation. This hypothesis introduces a direct relationship between the kinetic and potential energies during movement, making possible the discrimination between normal and pathological gaits [27, 58, 45, 48, 12, 23, 31].
Based on this hypothesis, McGeer [58] proposed the theory of the passive dynamics, which states that the gait dynamics can be described using mainly structural relationships [33, 45, 31, 12, 84], showing that these structural relationships are more important than muscle

strength and the control exerted by the nervous system to produce the efficient movement observed in the normal gait [33, 45, 12, 84].

Based on this approach, Garcia et al. [27] proposed a simple model that represents the human body as a double inverted pendulum system (Figure **2-3**-B). In this model the trunk is represented by a mass M and the feet by two masses m, separated by a rigid body which represents the legs(whose mass is negligible and length l), and a factor that relates the slope of the ramp. A parameter $\beta = \frac{m}{M} \approx 0$ is defined to simplify the classical equations of the inverted pendulum system,

$$\begin{bmatrix} 1 + 2\beta(1 - \cos\phi) & -\beta(1 - \cos\phi) \\ \beta(1 - \cos\phi) & -\beta \end{bmatrix} \begin{bmatrix} \ddot{\theta} \\ \ddot{\phi} \end{bmatrix} + \begin{bmatrix} -\beta \sin\phi(\phi^2 - 2\dot{\theta}\dot{\phi}) \\ \beta\theta^2 \sin\theta \end{bmatrix} +$$

$$\begin{bmatrix} (\beta g/l)[\sin(\theta - \phi - \lambda) - \sin(\theta - \lambda)] - (g/l)\sin(\theta - \lambda) \\ (\beta g/l)\sin(\theta - \phi - \lambda) \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \end{bmatrix} \tag{2-1}$$

and a reduced set of dynamic patterns that characterize human gait, leading to the following relationship:

$$\ddot{\theta(t)} - \sin(\theta(t) - \lambda) = 0 \tag{2-2}$$

$$\ddot{\theta(t)} - \ddot{\phi(t)} + \dot{\theta(t)}^2 \sin\phi(t) - \cos(\theta(t) - \lambda)\sin\phi(t) = 0 \tag{2-3}$$

Garcia also introduces a restriction rule, thereby simulating the "heel strike" during a gait cycle; at the moment $\phi(t) = 2\theta(t)$,given by:

$$\begin{bmatrix} \theta \\ \dot{\theta} \\ \phi \\ \dot{\phi} \end{bmatrix}^+ = \begin{bmatrix} -1 & 0 & 0 & 0 \\ 0 & \cos 2\theta & 0 & 0 \\ -2 & 0 & 0 & 0 \\ 0 & \cos 2\theta(1 - \cos 2\theta) & 0 & 0 \end{bmatrix} \begin{bmatrix} \theta \\ \dot{\theta} \\ \phi \\ \dot{\phi} \end{bmatrix}^- \tag{2-4}$$

This enables the support gait exchange during the leg balancing [27]. They ignore the mass ratio during movement, thereby reducing the model dimensionality.

In other approach, proposed by Kuo et al.[48, 49, 50, 51] the authors modified the simplified model of Garcia, adding a variable that describes the angular momentum that occurs on the hip when the heel strike happens:

$$\ddot{\theta(t)} - \sin(\theta(t) - \lambda) = 0 \tag{2-5}$$

$$\ddot{\theta(t)} - \ddot{\phi(t)} + \dot{\theta(t)}^2 \sin\phi(t) - \cos(\theta(t) - \lambda)\sin\phi(t) = -k\phi(t) \tag{2-6}$$

Where k is a stretching constant related to the torque generated at the hip. It also adds an impulse to simulate the toe-off, given by:

$$\begin{bmatrix} \theta \\ \dot{\theta} \\ \phi \\ \dot{\phi} \end{bmatrix}^+ = \begin{bmatrix} -1 & 0 & 0 & 0 \\ 0 & \cos 2\theta & 0 & 0 \\ -2 & 0 & 0 & 0 \\ 0 & \cos 2\theta(1-\cos 2\theta) & 0 & 0 \end{bmatrix} \begin{bmatrix} \theta \\ \dot{\theta} \\ \phi \\ \dot{\phi} \end{bmatrix}^- + \begin{bmatrix} 0 \\ \sin 2\theta \\ 0 \\ (1-\cos 2\theta)\sin 2\theta \end{bmatrix} P \tag{2-7}$$

Being P the impulse generated at the instant that the heel strike occurs. In this way they evaluated certain mass ratios and speeds to improve the energy conservation during the movement. These modifications allow to smooth the collisions produced by the heel strike, obtaining a minimum energy expenditure for the CoG displacement [60, 27].

In clinical applications, for which it is needed a detailed and accurate description of the gait dynamics, this type of models are insufficient, since they can not simulate the dynamic patterns associated with the leg joints and the ground reaction forces. In addition, the CoG trajectory resulting from these models, is far from the ”soft” path that characterizes the normal human gait, as shown in Figure **2-4**. This kind of representation is useful for the development of humanoid robots, but this is very limited to follow abnormal patterns.

In general, these models do not consider fundamental aspects of robotics related with the stability of a biped system, as the zero moment point (ZMP). Using this concept Kudoh and Komura [47] proposed an improved model of inverted pendulum by which angular momentum can be generated around the center of gravity, achieving to represent a larger number of dynamic patterns of the gait. This improvement of the inverted pendulum model allows that the ZMP simulate the reaction force produced by contact with the ground, such that the reaction force vector is not parallel to the vector that connect the ZMP with the CoG, which ensures continuity of the trajectory throughout of the movement. The model calculates separately the motion in the sagittal and frontal, plane, using inverse kinematics to a set of positions obtained after calculating the trajectory of CoG with respect to the ZMP.

Mochon and MacMahon [60] have shown (using the ballistic gait approach as a paradigm) [60, 86], that the progression of the body during the swing phase or single support can be performed in the absence of any muscle strength, using a mechanical model of three links that represents the body (one for the supporting leg and two for the shank and thigh). Other studies have developed anthropomorphic gait models, with and without knee, which have shown that the periodic gait stability can be achieved without any muscular action because gravity compensates the kinetic energy that is lost during the heel strike [27, 58].

On the other hand, Hurmuzlu et al. [40] proposed a relatively simple model of five links (trunk and legs jointed by the knee), with which it represents the kinetic and kinematic

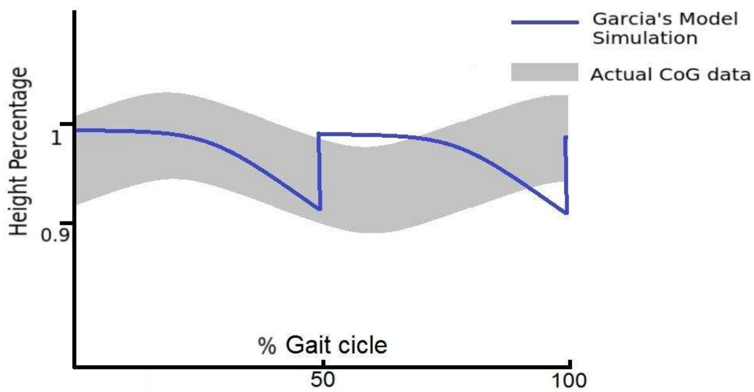

Figura 2-4.: The figure shows the result of a simulated model using the inverted pendulum theory [27] (thick line), compared with the smooth CoG pattern obtained from actual data.

patterns of the normal and pathological human gait. The model considers only the single support gait phase, and reads as:

$$M(q)\ddot{q} + C(q, \dot{q})\dot{q} + G(q) = T \tag{2-8}$$

Where $M(q)$ is a symmetric 5x5 matrix that defines the system inertia, $C(q, \dot{q})\ddot{q}$ is a vector that represents the centripetal and Coriolis forces, $G(q)$ is a vector that stands for the force of gravity acting on each link, and T is a force vector applied to each joint of the model, while $q, \dot{q}, \ddot{q}$ are angles, angular velocities and accelerations, respectively. Other models, based on the inverted pendulum [81], described the dynamic characteristics of pathological gaits that affect directly the osteo-articular sub-system, such as injuries and amputations of lower limbs, case in which prostheses are considered as rigid bodies and variable masses [67, 68, 73]. However, these models are restricted to engineering applications, since hay are unable to aaccurately reproduce the gait cycle.

2.3. Musculoskeletal Models

For the analysis of the human gait it is essential an adequate description of the muscle involved movement generation to identify musculoskeletal disorders [15]. This is why simple models, based only on structural relations are insufficient [93, 87]. In the literature, different models have been proposed to introduce factors that represent the generating force elements needed to accelerate and decelerate the body (muscles and tendons). It has been possible to simulate, using computational tools, pathological gaits such as the prosthetic gait or hemiplegia [4, 1]. These simulations have allowed to optimize the rehabilitation procedures used in these cases [93, 1, 41]. This set of models are here after described.

2.3.1. Models Based on Electromyographic Registers

One of the techniques used for recording and evaluating the muscle activation pattern, is the dynamic electromyography: neuro-physiological study performed by the use of surface electrodes that capture the muscular bioelectrical activity in different gait phases [43, 6]. Based on this information and some biomechanical relationships of the movement, different representations have been proposed using muscle and joint models that approximate the complex interaction of the musculoskeletal system [62, 30, 46, 77]. These models simulate muscle activation patterns and identify major muscle groups involved in the generation of motion. These approximations estimate the muscle strength and the joint moments generated during the gait [86, 54]. According to the complexity introduced by these models, they can be divided into models based on mechanical systems and computational models.

2.3.2. Models Based on Mechanical Systems

The pattern of the contraction of muscle groups can be modeled by subsystems, using simple mechanical models composed of springs, shock absorbers and resistors, under the constraint that each of these sub-systems emulates a very small part of the muscle and that the global contraction is a linear combination of these sub-systems. One of the most influential models in the literature was proposed by Hill [35], who describes a muscle group using three basic elements: a contractile element (CE), which amounts to the muscle fibers, a parallel elastic element (PEE) which stands for the connective tissues, and an elastic element connected in series, which represents all elastic elements such as tendons, as illustrated in Figure **2-5**.

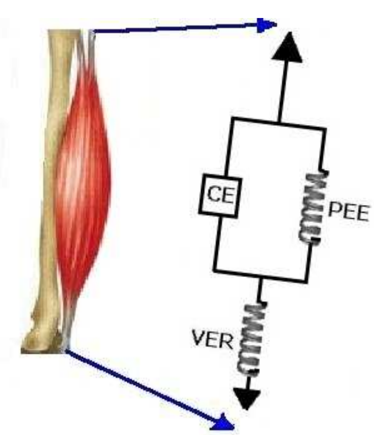

Figura 2-5.: The musculo-tendinous model proposed by Hill in 1938, emulates the visco-elastic characteristics of the musculo-tendinous elements of the human body.

Many models have used the Hill model to represent each of the involved muscle joints during the gait (usually seven groups) [53], including musculoskeletal models that independently emulate the action of each muscle group on the structural mechanisms of the locomotor system [39]. Using these models, it has been proven how the force generated by each muscle group affects the acceleration of each body segment [93, 92]. However, these models are not accurate enough to describe the complex interaction between different muscle groups, using experimental parameters for continual adjustment.

2.3.3. Musculoskeletal Computer Models

In the literature it has been reported many computational models, aimed to describe the interaction of each component of the locomotor system, using either the information provided by electromyographic recordings or based on specific biomechanical relations [93, 15]. These models have allowed to simulate the gait dynamics considering non-linear patterns of the locomotor system and describing different relationships between each of its components. A proposal devised to illustrate the musculoskeletal interaction with a dynamics adjusted using simple actuators, was introduced by Shin et al. [78]. The interaction is simulated with a model that estimates the joint torques generated during movement. To do so it uses information of the particular anatomy and physiology of the muscle groups responsible for these movements. This model considers physical aspects such as muscle tension and certain parameters that represent muscle properties (length, strength), obtained experimentally from dynamic electromyography and the measured torque. Although this model allows to describe certain abnormal situations, the model is highly dependent of the user knowledge because the activation sequence and muscle properties are decision of this model user (based on his experience), whereby, its use has been restricted to some clinical environments [93, 92]. On the other hand, Anderson et al. propose a model whose fundamental assumption is the conservation of energy typical of human gait, using a musculoskeletal representation, which consist in a computational analysis of six muscle groups, which generate almost 70 % of the total energy required to the gait [2]. This model allows a quantitative description of the interaction between components and the work done by each muscle group [93, 80, 22, 21].

Another well known model, proposed by Delp et al.[16, 15, 14] have introduced a musculoskeletal model, which has had a large impact in the field of the computational biomechanics. This model describes different dynamic relationships using muscle-tendon Hill components [35] and the inverted pendulum approach as a global descriptor. These models are parameterized with experimental data patterns, allowing to simulate normal and pathological gaits (Figure **2-6**). These simulations require parameter tuning, usually performed using a graphical interface where the user determines the parameters to simulate, introducing a high degree of subjectivity to this representation [43, 14, 5, 90].

Other researchers have designed models that allow musculoskeletal descriptions that follow the morpho-physiological limitations of the human body, using actual anthropometric data. Arnold et al. used the registration of 21 corpses, including geometric bone representations and kinematic joints descriptions. This information was used to construct a gait kinetics model with 44 muscles. This model coupled the Hill's model with each structure, achiving a

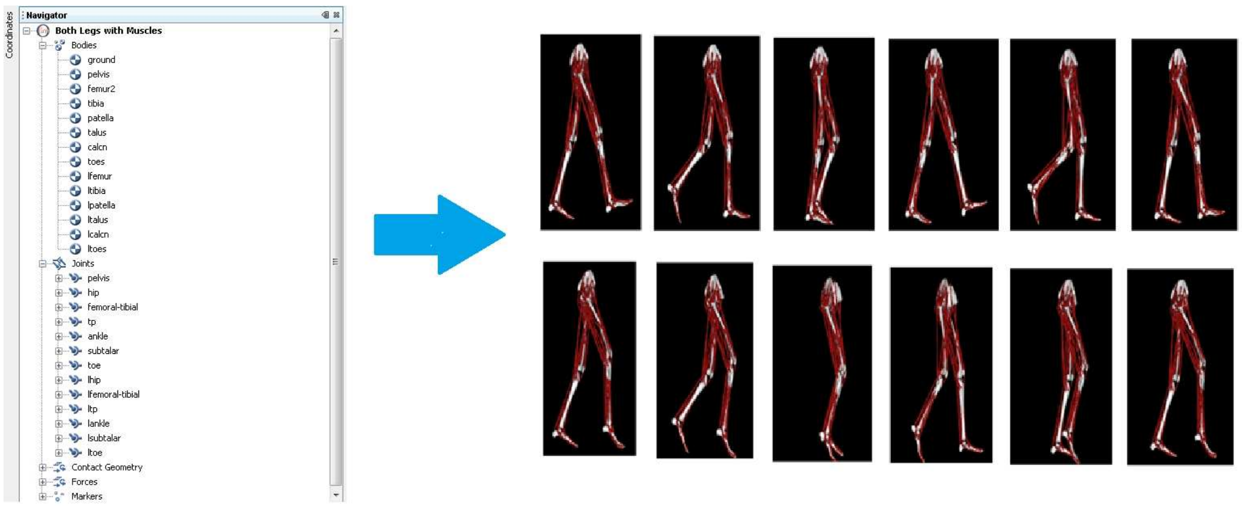

Figura 2-6.: The figure shows the simulation of the computer model proposed by Delp in OpenSim 2.1 (Delp et al. [16, 15, 14]) for a normal (top) and pathological (below) gait. This model is based on the theory of the inverted pendulum and the Hill's muscle-tendon models.

detailed study of the force and moment generation of particular structures such as the ankle, hip or knee [3]. However, the model dependence on experimental data limits its use, because the set of data used to generate the experimental parameters is very reduced, restricting the analysis of functional gait characteristics [3].

In order to develop models that allow to analyze particular relationships, such as the precise location of the joint rotations centers, Scheys [77, 76] adjusted the initial conditions and the geometry domain of Delp's computational model [14, 15], using magnetic resonance images. However, this technique is expensive and dependent on the adequate noise filtering associated with image capture [77, 76].

In rehabilitation engineering it has been important to develop gait models that allow a description, quantification and dynamic simulation of different movement patterns associated with a pathology. Approaches such as the inverted pendulum, seek to describe the dynamics of people's center of mass with some kind of amputation [67, 19], considering the prosthesis as rigid bodies, with no muscular element and variable mass. These strategies have allowed to approximate the dynamics of the prosthetic gait, but its clinical use is still limited because this analysis is very simple [64]. Computer models have allowed a more useful analyses, some of them complementing the information obtained from dynamic biomechanical models [93, 20], making possible to simulate a larger number of dynamic patterns of the gait.

On the other hand, Goujon et al. [32] proposed a dynamic method for calculating the external forces applied to the feet during the gait, approximating the kinetics of a pathological gait, analyzing the behavior of the prosthetic feet and the applied loads during gait. These contributions have significantly impacted in the manufacturing processes of orthopedic and rehabilitation devices, adding new strategies to calculate the charge distribution of existing prosthetic devices [32, 92]. Kuruvilla et al. developed a method that describes the characteristic patterns of the Charcot-Marie-Tooth disease, also known as *hereditary motor and sensory neuropathy, or peroneal muscular atrophy*, based on data recorded from gait laboratories. The obtained results have significantly improved the understanding of this disorder and the extraction of a set of dynamic patterns that characterize this disease, influencing the design of treatments for this condition [88].

The computational biomechanics has introduced new strategies for describing normal and abnormal motion patterns. One of the most representative models has been proposed by Komura et al. [47], proposed a computational model to simulate pathological movements related with weakening of some leg muscle groups [15]. Komura assumes that the Zero Moment Point (ZMP) is not fixed (as is generally considered), generating rotational and angular moments (see Figure **2-7**).

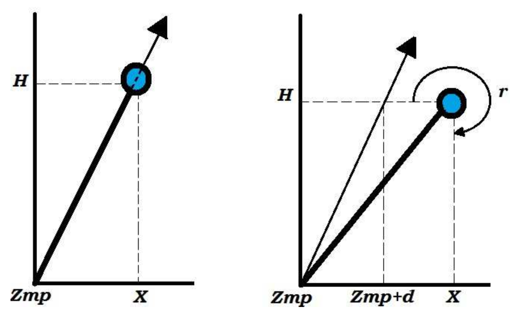

Figura 2-7.: The figure shows the differences between the classical inverted pendulum model and Komura's model (Left). The force vector generated in the ZMP cuts the point given by the CoG (Right). The model proposed by Komura suggested that the force vector generated from the ZMP does not cut the point given by the CoG, allowing the generation of rotational and angular moments

They formulate an inverse kinematics problem, based on delp's structural model [15], calculating the torque generated at each joint as:

$$\tau_j(t) = \sum_i (F_i(t) r_i, j \text{donde} j = 0, \ldots, n_d o f) \tag{2-9}$$

Where r_i is the moment generated by the muscle i over the $j - th$ joint. The force exerted by each muscle groupn is then calculated, given $F_i = f_i(p, t)$. Pathological gaits can be simulated, with any degree of paralysis or muscle weakness. This model allows to modify structural parameters with certain degree of freedom, an attractive issue the clinical area [47].

2.4. Neuromuscular Models

Studies of human locomotion have shown that gait stability and flexibility are caused by interaction between the activity of the nervous system [43] (neuro-motor commands) and the movement of the musculoskeletal system [83]. This shows the importance of describing the dynamics of human movement considering their structural characteristics (skeleton,muscles) and the entity responsible for control it (neuro-muscular system), in order to apply this knowledge in the analysis of the gait patterns to identify early neuromuscular disorders.

One of the most important assumptions for the development of neuro-muscular models, is the existence of a central pattern generator (CPG), responsible for the generation and control of the muscle-articular movement. Discovered in 1985, the CPG may be located in a relatively low level of the central nervous system (spinal cord). Theoretical studies have shown that CPG is a complex adaptive system, where movements emerge by the dynamic interaction between the nervous system, body and environment, in a self-organized way. This concept uses the evidence that living organisms exhibit cyclic patterns of movement, with which there is a high probability of neural circuits that lead to cyclical movements. In a biomechanical model, the term CPG is often approximated by a group of artificial neurons, capable of generating oscillatory output, without external cyclic stimulation [83, 37].

The discovery of the CPG, motivated the development of control systems adapted to systems and existing musculoskeletal models. Buchli [9] made a significant contribution in the robotics area, inspired by the CPG, developing a control scheme with a group of neurological oscillators that induce synchronous activation of certain muscle groups that control a joint, such as illustrated in Figure **2-8**. The output signals of the CPG induced the body movements by the activation of the muscles, so that each neurological oscillator in the CPG controls the muscles around a single joint.

On the other hand, some researchers have attempted to use some classical machine learning approaches, such as neural networks, to represent bipedal gait using the concept of CPG [83, 37, 25]. The simplest models consider only cyclical limb movements in the sagittal plane, assuming gait symmetry. These hypotheses, so simple, constitute a framework for understanding the complex control mechanism required to produce a coordinated movement [13].

In recent years there has been a growing interest in this type of models [84, 83, 94], because of the interest of improving the understanding of the interactions in the neuromuscular systems [84]. Taga et al. [83] have proposed a neuro-musculo-skeletal model that explores the control mechanisms responsible for generating human motion, based on the properties of CPG as the basic neural system. This model allows an approximation to the observed pattern of the human gait, adding a discrete movement generator component [83].

In despite of the increasing number of models that arise, the existence of a CPG in the human locomotor control is still subject of debate, due to the limited current understanding of the physiology of the human neuro-motor system [69, 64, 65]. Other approaches have attempted

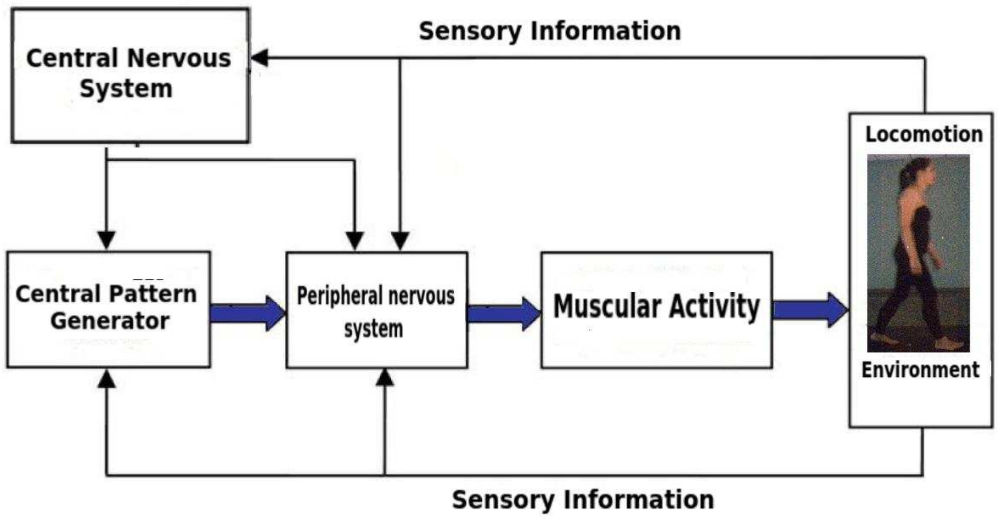

Figura 2-8.: The figure shows an overview of the neuro-muscular model, considering the existence of the CPG. The system is fed by sensory information, processed by the CPG that generates the movement

to model the neuro-muscular system using independent oscillators and autonomous control for each joint. For example, Gentaro Taga [82] formulated a neuro-muscular model composed of eight segments and 20 muscles, without contractile dynamics activation, controlled by a rhythm generator that stimulates seven oscillators distributed throughout the structure, and feedback sensors for the input controllers [71, 83, 94, 34]. Its complex formulation shows good results simulating the neuro-muscular system, but it has not application in the classical dynamic gait analysis.

Finally, in the literature there have been other strategies for the development of neuromuscular gait models, all of them using the hypothesis of independent control for each muscle group. These models require uncoupled oscillators that govern the motion for each set defined, and external sensors that allow feedback to the system [80]. New approaches have defined stochastic parameters introducing variability to the dynamic movement patterns and neural relationships for locomotor control [9, 34, 44], however, their applicability to the clinical field is still in experimentally stage.

2.5. Discussion

This chapter has presented an extensive review of the gait models described in the literature, with the potential to be used in routine clinical applications. These models should provide a quantitative movement description (dynamic patterns) and also should simulate different pathological conditions to improve the analysis that is currently performed in the clinical routine. Overall, these models have been studied from the structural point of view, musculo-

skeletal and neuro-muscular.
The structural models use physical analogies that establish the fundamental relationships between forces and their interactions. The typical example is the inverted pendulum, whose dynamics has shown to be appropriated to the global description of the CoG. These representations take into account only structural relationships, what means that the system (in this case the human anatomy) is simplified, allowing to describe global movements in terms of energy. However, the description achieved with these models is limited, and important factors for the characterization of certain abnormal movements, are not considered. Moreover, given the deterministic nature of these models, it is not possible to represent certain nonlinear patterns of human locomotion. Such essential elements as the *heel strike*,are not adequately described in these models. Similarly, inherent aspects of the gait, as the inter-subject variability, can not be simulated with these approaches, restricting the description to simple trajectories, which are far from natural human gait trajectories.
The second strategy produces more realistic simulations by introducing a rigid structure, which supports the muscular system and allows to determine the basic movement mechanisms. These representations produce a much more accurate emulation of the gait dynamics in terms of forces, moments and positions of the centers of joint rotation . However, this kind of models require a proper knowledge of the muscle physiology as to set the initial conditions and obtain covergent solutions.
As a result, only certain muscle groups can be represented, thereby restricting the adequate description of the muscle activation patterns. and particular kinetic gait relationships . Some of these musculoskeletal models describe, with certain restrictions, dynamic patterns of pathological gaits. These musculoskeletal models describe different degrees of the pathologies using certain parameters that must be tuned, facilitating the analysis of pathological movements. However, the set of conditions that can be represented is still limited, since certain structural and neuro-muscular control deficiencies can not be considered.
Subsequently, this chapter has reviewed a group of models that allow the description of the complex human neuromuscular system by using control methods, using machine learning algorithms. These methods have allowed to simulated the coordinated sequence of muscle activation and the complex feedback control system, which governs the human displacement. The concept of CPG has been defined as a complex adaptive system, inferred from the regulation that requires an autonomous system consisting in neural networks, and from which cyclic movement patterns are generated, as a result of the dynamic interaction between the system and the external environment.
This hypothesis, so attractive, is still controversy in the area, with the little experimental evidence that supports the existence of the CPG and the limited knowledge of the pathophysiological mechanisms involved in gait disturbance.
Finally, due to the complexity of the resulting models, it becomes very difficult a proper gait description and the development of applications for clinical support that describe the human dynamics in normal and pathological movements.

3. Description of Normal and Pathological Gaits Using a Fusion Knowledge Strategy

Gait distortion is the first clinical manifestation of many pathological disorders. Traditionally, the gait laboratory has been the only available tool for supporting both diagnosis and prognosis, but under the limitation that any clinical interpretation depends completely on the physician expertise. This chapter presents a novel human gait model which fusions two important gait information sources: an estimated Center of Gravity (CoG) trajectory and learned heel paths, by that means allowing to reproduce kinematic normal and pathological patterns. The CoG trajectory is approximated with a simulated trajectory of a physical compass pendulum representation that has been extended by introducing some energy accumulator elements between the pendulum ends, thereby emulating the role of the leg joints and then obtaining a complete global gait description. Likewise, learned heel paths captured from actual data are learned to adjust the performance of the physical model, while the remaining joint trajectories are estimated using a classical inverse kinematic rule. We demonstrate the model's utility by accurately simulating two different sorts of neuromuscular diseases: parkinson and cerebral palsy. The predicted trajectories were compared with actual pathological gaits, obtaining a very high correlation (correlation coefficient of 0,95). Finally, a human-like leg structure is animated with the obtained trajectories, allowing the clinician to interact with the model and facilitating the interpretation of an observational analysis. The rest of this chapter is organized as follows: Section *Materials and Methods* introduces our structural model and the fusion strategy used to represent actual gait patterns, Section *Results* demonstrates the effectiveness of the model simulating normal and pathological kinematic. The last section presents a discussion and possible future works.

3.1. Materials and methods

The gait can be thought of as a sequence of complex combinations of several subsystems that help the body to keep the balance while it gains support and propulsion [86]. Structurally, the gait has been modeled using simple mechanical systems like the inverted pendulum, approximating thereby the energy consumption. However these representations are very limited to describe non linear muskulo-skeletical relations, as observed in most diseases. Some of these non linear muscle-tendon interactions have been traditionally modeled by a spring mass system, which are used to structurally represent certain global gait patterns, but lacking of any anatomical meaning. Other approaches have tried to accurately track gait patterns by introducing specific patient trajectories, with the consequent limitation of a required high level of expertise to properly tuning the model parameters. The present thesis simulates normal and pathological kinematic patterns by fusioning two important sources of information: a prior model of the $CoG_{x,y}(t)$ and real data trajectories. The prior knowledge is introduced using a physical gait model, a compass pendulum with springs coupled between both ends, representing the role of the knee and smooth tissues. Inclusion of these non linear elements allow more accurate estimations of the $CoG_{x,y}(t)$ trajectory (section 2.2). A second information source comes from actual heel trajectories that are used to regularize the estimated $CoG_{x,y}(t)$ path and serve to simulate diverse pathological and normal motion (section 2.3). Additionally, this fusion facilitates an accurate estimation of the remaining joint trajectories, using a classical inverse kinematic framework. Finally the set of obtained trajectories animates a human-like leg structure that enables the clinician with a interpretable tool. The method is summarized on Figure **3-1**.

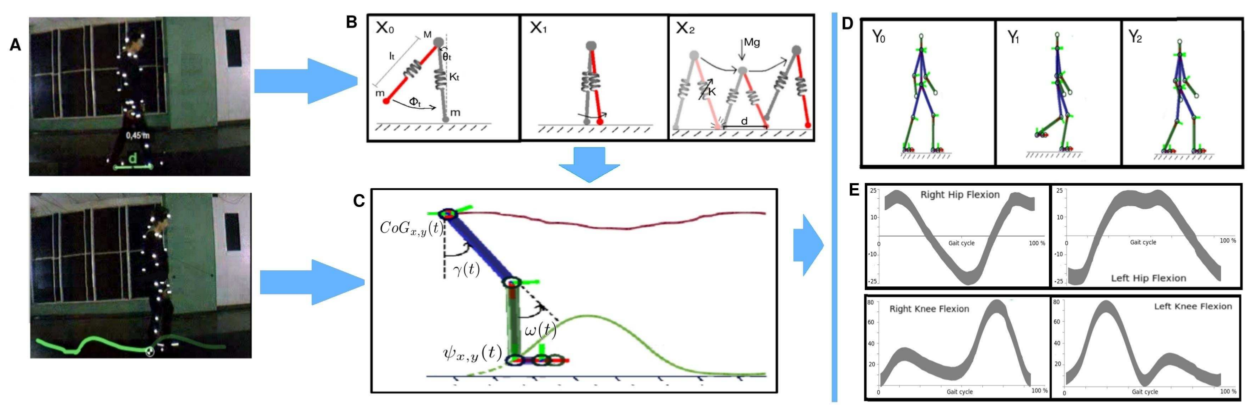

Figura 3-1.: Pipeline of the proposed model. The first panel (A) represents the learning process of the heel trayectory used to regularize the CoG model. The panel(B) shows the proposed CoG model. The panel (C) illustrates the fusion rule used to simulate the human gait (Normal and Pathological) and its patterns (Panels (D) and (E)).

3.1.1. CoG gait representation

In human movement analysis, the gait is divided in cycles, coarsely classified as double and single stance phases [26]. The double stance period accounts for a 25 % and stands for the body movement with both limbs touching the ground, while the single stance represents a 75 % and corresponds to the interval in which a single limb supports the whole body weight. In this work the $CoG_{x,y}(t)$ trajectory for a complete gait cycle is approached using two complementary models: a compass pendulum for the single stance and a spring mass system for the double stance, as follows.

The single support phase

The single support phase conserves a regular periodicity which is properly captured using a compass pendulum representation. This model represents the upper part of the body by a mass M which moves forwards with respect to each fixed point (with mass m), describing a harmonic oscillating trajectory, similar to the inverted pendulum [58, 49]. Likewise, the free foot swings with respect to this mass, establishing a simple pendulum pattern. Provided that these processes are coupled together, the human gait is modeled by a compass pendulum as two coupled non-linear differential equations:

$$\begin{aligned} &\beta(1-\cos\phi)(3\ddot{\theta}-\ddot{\phi})-\beta\sin\phi(\dot{\phi}^2-2\theta\dot{\phi})+(\tfrac{g\sin\theta}{l})(\beta(\sin(\theta-\phi)-1))=0 \\ &\ddot{\theta}(\beta(1-\cos\phi))-\beta\ddot{\phi}+\beta\dot{\theta}^2\sin\phi+(\tfrac{\beta g}{l})\sin(\theta-\phi)=0 \end{aligned} \tag{3-1}$$

where $\beta = m/M$, θ is the angle of the stance leg at particular time t with respect to the slope and ϕ is the angle between the stance leg, and $l_0 = l_r = l_l$. This model also allows to simulate the swing foot when it hits the ground at the heelstrike, a time in the cycle that corresponds to $\phi(t) - 2\theta(t) = 0$ [49], when the double stance start.

Double stance phase

Classical gait models often ignore the double support stance since they have been devised to simplify the gait rather than to accurately follow gait patterns. These model simplifications have ended up by considering the leg structures as rigid segments, a hypothesis that easily leads to conclude for instance that the percentage of gait recovery is inefficient in energy terms, a reason why this phase has been eliminated in most of these models [27, 79, 24, 93]. Additionally, important elastic contributions which produce relevant changes in the $CoG_{x,y}(t)$ trajectory during the double stance, are often neglected. These strong simplifications reduce an appropriate gait understanding and may lead to wrong interpretations when these models are used as supporting tools of clinical decisions.

A more accurate $CoG_{x,y}(t)$ description of the double stance phase was herein achieved by coupling a planar spring-mass system [8] to the compass pendulum mode, previously introduced. This change of the leg length l during the gait stance phase, allows to estimate the reaction force during the whole gait cycle, as illustrated in Figure **3-1** (B). Notice that each leg reaction force points out towards opposite sides, separated by a distance d (the distance between the heelstrike and the other toe-off phase). The coupling of this physical model is obtained as:

$$\begin{aligned} M\ddot{x} &= l_l x - l_r(d-x) \\ M\ddot{y} &= l_l y + l_r y - gM \end{aligned} \tag{3-2}$$

where, g is the gravity, l_l and l_r are the left and right legs, respectively and their length changes as:

$$\begin{aligned} l_l &= k(\tfrac{l_0}{\sqrt{x^2+y^2}} - 1) \\ l_r &= k(\tfrac{l_0}{\sqrt{(d-x)^2+y^2}} - 1) \end{aligned} \tag{3-3}$$

These equations simulate the periodic vertical ground forces, with a period defined by $T = 2\pi\sqrt{\frac{m}{k}}$. This independent formulation of each reaction force allows an independent analysis of each link, whereby gait abnormalities that asymmetrically affect each leg, such as the diplegia, can be simulated. Finally , the $CoG_{x,y}(t)$ trajectory is simulated by the integration of the two gait phases described as follows:

$$CoG_{x,y}(t) = \begin{cases} l_0\left[\sin\theta(t), \cos\theta(t)\right] & \text{if} \quad \phi(t) - 2\theta(t) < 0; \\ \\ \left[\frac{l_l\frac{x^3}{6} - l_r(d-\frac{x^3}{6})}{M}, \frac{l_l\frac{y^3}{6} + l_r\frac{y^3}{6} - gM}{M}\right] & \text{elsewhere.} \end{cases} \tag{3-4}$$

3.1.2. The fusion information strategy

Although the $CoG_{x,y}(t)$ trajectory is a fundamental clinical descriptor [57], a useful identification of a particular disorder also requires a proper gait analysis of other anatomical joint trajectories. Accordingly, a more complete gait description was herein achieved by the fusion of two important sources of information: the physical gait model previously described and the learned heel trajectories.

The learned heel trajectories were modeled as a set of normal distributions with mean μ_i and variance σ_i^2 from three different groups of patients captured in a gait laboratory as:

$$\psi_{x,y}(t) = \sum_{i=1}^{k} p_i N(t|\mu_i, \sigma_i^2)$$

where k represents the total number of learned gait movements (normal, crouch and parkinsonian gait). Each gait distribution was computed from 30 gait cycles belonging to 10 patients (7 men and 3 women). From this multi-gaussian distribution model, we can select a heel trajectory i to regularize the CoG associated to a particular gait movement. Likewise, the normal motion distribution allows a large variety of gait patterns of the same pathology. New relationships are inferred from these two trajectories by assuming the knee joint position as $r_{x,y} = \frac{l_{0_{x,y}}}{2}$ [17]. Afterward, a classical inverse kinematic method is adapted to obtain two main kinematic patterns: the flexion-extension patterns of the hip $\omega(t)$ and knee $\gamma(t)$. For doing so, at each time t of the gait cycle, a CCD method performs an iterative rigid transformation over each couple of joints. The two patterns are defined as:

$$\gamma(t) = acos\left(\frac{CoG_{x,y}(t)^2 - r_{x,y}^2 - r_{x,y}^2}{2}\right) \tag{3-5}$$

$$\omega(t) = atan2(\psi_{x,y}) + atan2(r_{x,y}\sin\gamma, r_{x,y} + r_{x,y}\cos\gamma) \tag{3-6}$$

where r is the distance between the $CoG_{x,y}(t)$ and $\psi_{x,y}(t)$. Unlike other approaches, this model estimates kinematic patterns with medical meaning, but the model can also obtain energy and ground force patterns for normal and pathological cases, obtained from the $CoG_{x,y}(t)$ trajectory.

3.1.3. Building up a human leg structure

Finally a human-like leg structure is animated using the set of kinematic patterns described above. This structure may be used as a clinical interpretability tool. For doing so, we define a human structure composed by a set of 12 rigid elements, connected together, as shown in Figure **3-2**. The lower limbs follow a dynamic established by the proposed model, while the upper limbs are normal trajectories computed from real data in the gait lab.

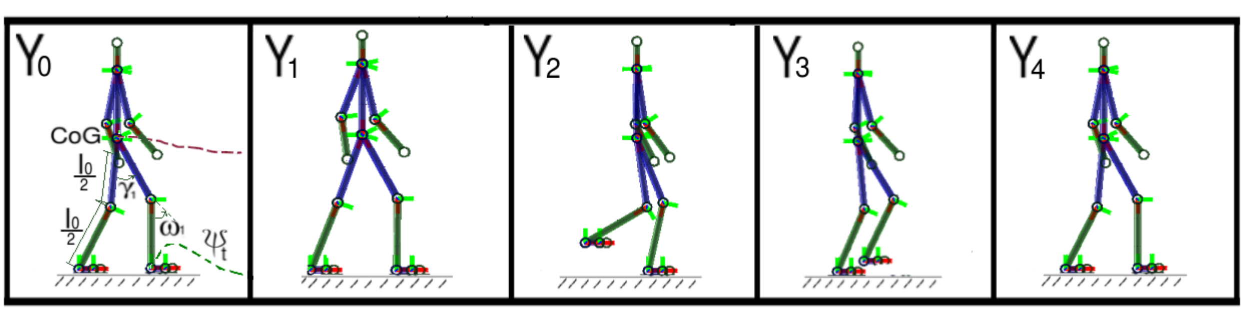

Figura 3-2.: The Figure shows a complete obtained gait cycle animating a Human leg structure, following the $CoG_{x,y}(t)$ and $\psi_{x,y}(t)$ trajectories.

3.1.4. Modeling pathology movements

The proposed model is capable of simulating pathologic patterns, an advantage with respect to any other model, allowing to simulate actual pathologies such as the spastic diplegia (typically represented by a crouch Gait) and Parkinson (See Chapter 4).
In this chapter, the model is used to simulate a crouch gait. This motion is produced by a neuro-muscular disorder known as spastic diplegia that is characterized by the presence of muscle rigidity and loss of muscle force, affecting predominantly the legs, arms and face. The clinical signs include gait pattern distortions of the sagittal plane, like bent-rigid knees, flexed hips and certain anatomical changes like lumbar lordosis. Likewise, this model simulates typical Parkinson gait patterns. In this case, the gait patterns are produced by a degenerative disorder of the central nervous system, characterized by rigidity and slowness of the human movements: this gait is characterized by short steps. This kind of patterns can be modeled and simulated to approximate different phases of this disease, allowing thereby to objectively characterize the pathology. This pathological representation is extended in the chapter 4.
Simulation of these pathologies requires to set the l_0, k and d parameters, using actual patient data. For the crouch gait simulation, the spring constant was fixed within a range of $k = 400 - 350$, and the l_0 length was reduced to $5\,\%$, according to well known biomechanical parameters [61, 68, 67]. For the parkinsonian gait simulation the d parameter was obtained from actual data and fixed to 0,58mts, the k was set to 500. The heel path trajectories fitted a normal distribution and were learned from actual patient data, as previously explained. The simulated trajectory precisely follows the different components of the abnormal gait pattern, in particular the flexed knees, the CoG attenuation and the step length, as is reported in next section.

3.2. Evaluation and results

Evaluation was carried out by comparing the estimated gait kinematic patterns with ground truth trajectories, of normal and pathological patterns, as reported in [26]. A first quantitative evaluation was performed by calculating the correlation coefficient between both trajectories, which are composed of temporal x and y paths, belonging to a single gait cycle. A first part of the evaluation consisted in determining the $CoG_{x,y}(t)$ relation between two models, the physical model herein proposed and a classical compass pendulum model [48]. Figure **3-3** shows the decomposed motion for both models: the x axis representing the percentage of the gait cycle and the y axis the vertical displacement with respect to the body height, also in percentage. Both models follow a CoG periodic sequence, but the classical compass pendulum model systematically misses the discontinuity introduced by the heel strike, much more important in the x axis, while the proposed model accurately predicts this part of the cycle. Notice that the heel strike actually occurs at about a $60\,\%$ of the gait cycle.

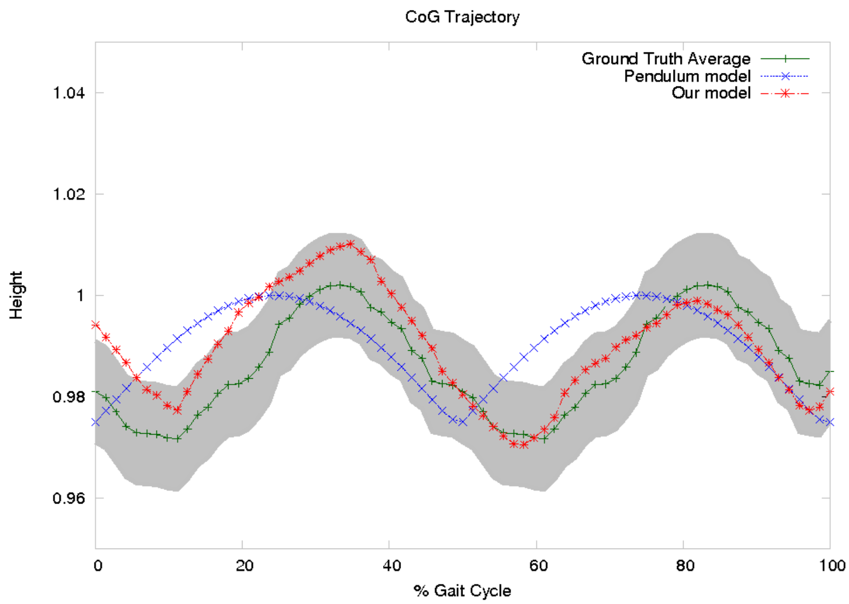

Figura 3-3.: CoG obtained with two different approaches.

A second part of the evaluation compared the hip $\gamma(t)$ and knee $\omega(t)$ joint-angle patterns of simulated normal gaits and ground truth patterns. For this assessment, the fusion strategy used two different $CoG_{x,y}(t)$ estimations: the proposed physical model and the classical compass pendulum previously described. The hip $\gamma(t)$ and knee $\omega(t)$ patterns were expressed as joint-angle variations at the y-axis and plotted against the gait percentages at the x-axis, previously weighted by the entire duration of a cycle. Figure **3-4** shows the ground truth and the predicted gait patterns for a sagittal view (right and left) of a complete cycle, using the two $CoG_{x,y}(t)$ estimations. The joint-angle trajectories computed from the fusion strategy show a very close Correlation Coefficient (CF) w.r.t the ground truth patterns (CF = 0,8 using the compass pendulum, CF = 0,9 using the herein proposed physical approach). During the single stance phase, the angle trajectories computed from both CoG paths have a high correlation, nevertheless the conventional pendulum misses about a 40 : % of the angle variation because of the nonlinearity introduced by the heel strike and therefore the curve correlation also falls down. In contrast, the joint-angle trajectories obtained from the fusion strategy with the proposed physical model follows the actual gait paths and its correlation coefficient remains larger than 0,8. Significant differences were then reported with the conventional compass pendulum, specifically for the part of the cycle dominated by the heel strike.

Table **3-1** shows the correlation coefficient obtained with the temporal differences between both joint angle estimates and the ground truth. This measure was applied only to those gait segments associated with the heel strike since it was previously confirmed that performance

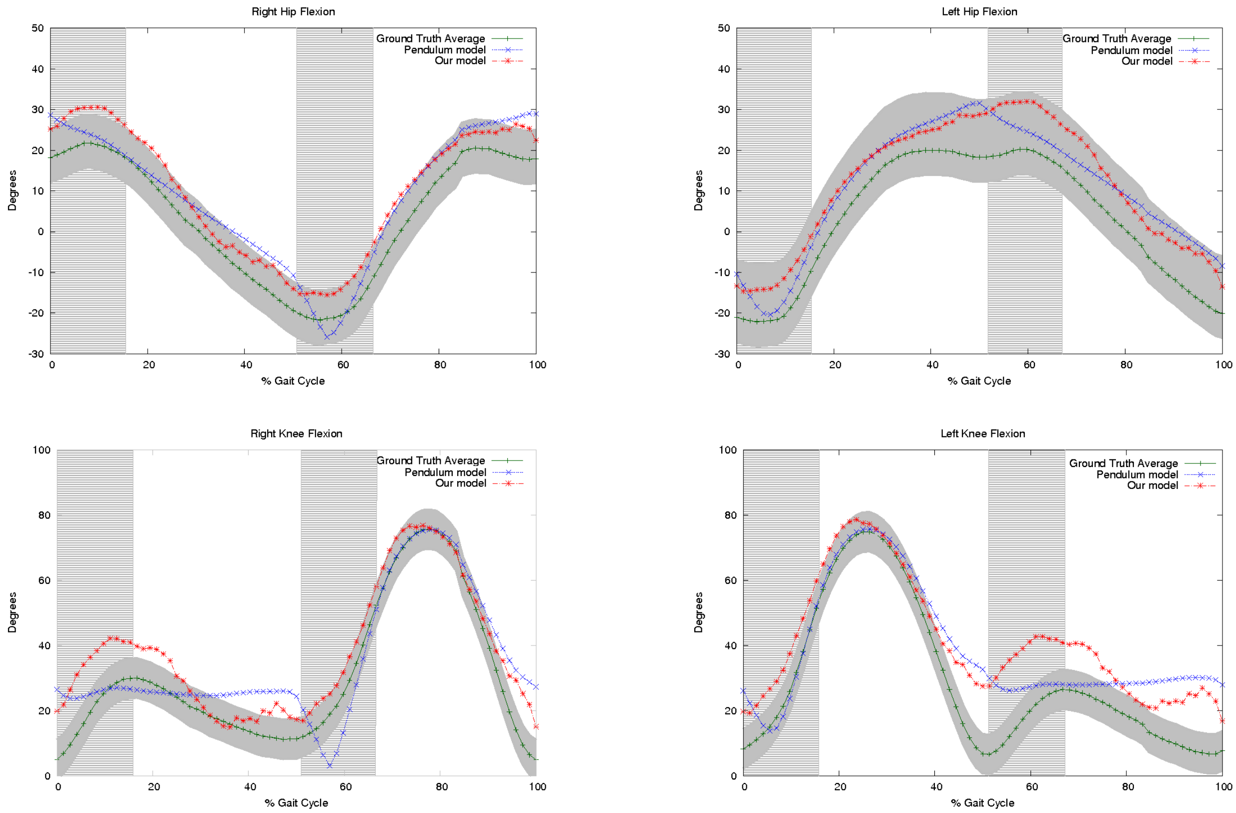

Figura 3-4.: Simulations for different gait cycles. The shadowed zone corresponds to the heel strike phase. Notice that the proposed model tracks better the ground truth, above all on the zones defined by the heel strike which are the most important when assessing pathological patterns.

in the other gait phases are comparable.

Differences were found to be significant during the heel strike phase (student t-test, $p < 0{,}05$) for the joint angle paths computed from the conventional pendulum $CoG_{x,y}(t)$ path, while the joint angle trajectories estimated with our physical model, obtained correlations of about 96 %. In contrast, joint angle paths computed from the conventional pendulum $CoG_{x,y}(t)$ obtained barely correlations of about 46 %, evidencing the weakness of this type of models during this gait phase.
On the other hand, since the whole problem consists in following temporal series which are highly non-linear and whose dynamics is therefore very difficult to establish, evaluation should also include a type of measure that determines a level of agreement between two trajectories. We have then measured this level using the Fréchet distance between two temporal series, i.e., the ground truth and any of the joint-angle trajectories obtained from the

Pattern	R.Hip	R.Knee	L.Hip	L.Knee	CoG
Garcia's Model	0,17 ±0.02	0,38 ±0.05	-0,02 ±0.01	0,17 ±0.05	0,52 ±0,01
Our Model	0,89 ±0.02	0,99 ±0.01	0,95 ±0.003	0,99 ±0.02	0,84 ±0,01

Tabla **3-1**.: Correlation factor for normal Gait simulation using two different physical model in the fusion strategy.

Pattern	R.Hip	R.Knee	L.Hip	L.Knee	CoG
Garcia's Model	0,31 ± 0,025	0,31 ± 0,023	0,38 ± 0,017	0,35± 0,025	0,02± 0,0024
Our Model	0,18 ± 0,012	0,14 ± 0,018	0,27± 0,009	0,22 ± 0,014	0,01 ± 0,0041

Tabla **3-2**.: Fréchet distance measured simulating the classical model and our approach

fusion strategy using the two models. Briefly, the Féchet measure estimates how close two trajectories are during the temporal capture, that is to say it estimates how similar these two curves are. Two trajectories are then similar if this distance is close to zero, the smaller this distance the closer the curves are. The Fréchet distance between two curves is the length of the shortest path between two points that are simultaneously moving through the two curves. The Fréchet metric uses a particular direction of the two curves because the pairs of points whose distance contributes to the Fréchet distance sweep continuously along their respective curves. This makes the Fréchet distance a better measure of similarity for curves than alternatives, such as the Hausdorff distance, for arbitrary point sets. It is possible for two curves to have a small Hausdorff distance but large Fréchet distance.

Table **3-2** shows the Fréchet measure obtained from the temporal differences between estimates and ground truth. Again, the $CoG_{x,y}(t)$ was assessed as well as the hip $\gamma(t)$ and knee $\omega(t)$ joint-angle showing smaller differences. Interestingly, the close curve similarity between the joint angle trajectories estimated from our physical CoG path and the ground truth achieved a gain of 20 % with respect to the classical compass pendulum model, that is to say, joint angle trajectories computed from our physical model was about 20 % more accurate.
Finally, a third part of the evaluation focused on challenging the fusion strategy to simulate pathological patterns. A Chrouch gait was tracked by changing the value of the k constant. Figure **3-5** , it is shown a typical cycle obtained with the proposed model, when tracking this pathological movement. It is observed in this illustration a close relationship between the trajectory described by the proposed model and the pathological pattern. A useful clinical evaluation requires a precise track of the consecutive ups and downs described by this pattern, rather than the magnitude changes.
The crouch gait simulation was also compared with the two similarity metrics previously introduced, i.e., the correlation factor and the Fréchet distance, as illustrated in Table **3-3**.

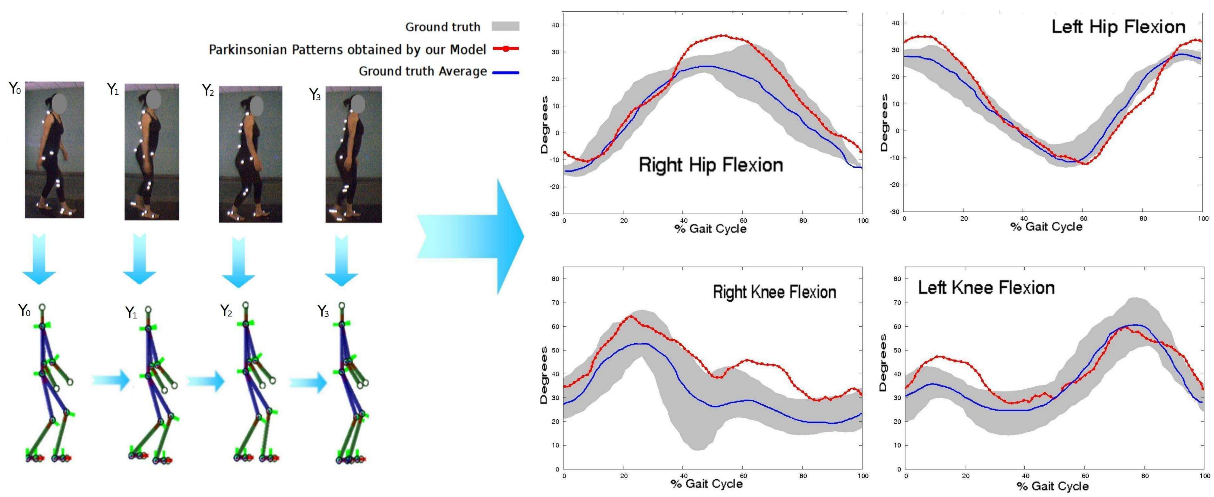

Figura 3-5.: Simulation of pathological gait patterns, in this case the crouch gait

Metric	R.Hip	R.Knee	L.Hip	L.Knee	CoG
Correlation factor	0,96 ± 0,01	0,88 ± 0,04	0,96 ± 0,01	0,93 ± 0,03	0.92 ± 0,02
Féchet distance	0,19 ± 0,006	0,26 ± 0,001	0,13 ± 0,004	0,21 ± 0,002	0,29 ± 0,003

Tabla **3-3**.: Results for the Crouch gait simulation.

The correlation factor is larger for upper joints, as expected since movement is much smaller, but yet correlation is high with joints such as the knees.

The proposed approach was also used to simulate the Parkinsonian gait in different stages of the disease. For each parkinson disease stage it was computed the most probable k and d parameters (See table **3-4**). Then, from these parameters it was generated a $CoG_{x,y}(t)$ from the physical model proposed here (See chapter 4).

Model Parameter	Stage 2	Stage 3	Stage 4
k average	52.3 ± 6.22	67.42 ± 8.25	96.8 ± 2.54
d average	0.76 ± 0.081	0.56 ± 0.0474	0.44 ± 0.0294

Tabla **3-4**.: Model parameters learned from 10 patients in different stages of the Parkinson disease

3.3. Discussion

The gait analysis aims to interpret the complex combination of several motion patterns generated by the interaction of different systems. In the clinical routine, the gait examination is considered as the most important tool for identifying motion disorders, and it is also used as a biomarker of some neuromuscular illnesses like the cerebral palsy or the Parkinson disease, supporting thereby diagnosis and follow up. Interestingly, during this analysis it is possible to evaluate the effectiveness of a specific treatment and the particular response of the multiple gait subsystems. However, these analyses are not actually carried out in the clinical practice, among others because this is very time consuming and examiner dependent. Moreover, measures are contaminated by noise during the capture or, by invasive devices like markers, which inevitably alter the natural motion gestures. Overall, most clinical examinations are devoted to capture a general gait picture which makes that treatments are globally addressed. The actual utility of devising gait models is that they improve the understanding of some subsystem patterns from the captured data, even if they are noisy, and hence they allow to plan more specific treatments.

This work has presented a new fusion scheme that simulates a large set of gait patterns, including pathological conditions and isolating the role of certain subsystems during the simulation. This scheme uses two important information sources, i.e., kinetic and kinematic gait components, maintaining the possibility of an energy consumption analysis. The fusion method assumes that gait kinematic patterns must follow basic physical principles. The underlying trajectories are generated by an adapted version of the compass pendulum representation which has been extended with elements that store energy, thereby describing a larger number of abnormalities than it has been possible so far. These trajectories are then regularized by actual learned paths, completing thereby the rule of fusion by means of an inverse kinematic approximation.

The models reported in the literature are very limited when describing particular variations of a pathological motion. Gait models can be coarsely divided in two large groups: physical based models and musculoskeletal representations. Physical models use an inverted pendulum or a spring mass system that allow an energy consumption analysis and a global dynamic description. These models nevertheless fail to mimic pathological patterns because of the strong simplifications and restrictions, i.e., pendular models represent only the single stance while the double stance is completely omitted. Musculoskeletal representations, typically more complex than physical-based models, are able to simulate muscle patterns at each phase of the gait by adding either a Hill's model or prior information coming from data obtained from actual gait laboratories. Nowadays, musculoskeletal models associate each leg segment to a Hill's model. However, a main drawback of this approach is that there is no interaction between the different segments and hence simulations are quite far from experimental data [15]. Best performances are obtained by combining the prior model with observations coming from actual data [15], but these models completely neglect some kinetic relationships and must be tuned by an experienced observer. In contrast, physics based models are simple and usually tuned with a small number of variables. This simplicity, nonetheless, leads to most physics based models to miss important phases of the gait cycle. The fusion model proposed in this work represents the single stance by an articulated double pendulum system, while the double stance is properly simulated by a spring mass component that stands for the important knee motion contribution. The proposed model in addition emulates a whole skeleton structure by animating this architecture from the CoG trajectory. The whole strategy allows a natural simulation of non-linear gait patterns, representing several kinds of movements. Evaluation demonstrated the fusion model advantages, by comparing several kinematic patterns like the CoG trajectory and the hip and knee flexion-extension movements, considered as the most representative gait patterns for determining whether a motion is pathological or not [79, 67].

The first evaluation compared the adapted physical model with a classical compass pendulum. An appropriate extraction of the CoG is essential since this biomarker is an efficient indicator of the normal/abnormal gait pattern, it constitutes one of the most important markers in pathologies such as hemiplegia, paraplegia or dystonia. A pathological gait can be analyzed in terms of energy using the CoG, which tracks the transfer of potential to kinetic energy (recovery), i.e. normal gait patterns loss 40 % of their energy in this transfer, a higher loss is pathological [18]. The CoG trajectory described by our physical model achieved a CF of = 0,84 while classical pendulum model only achieved a CF of = 0,52, as illustrated in Figure 3.2. In a second evaluation, hip and knee joint-angle paths were compared during the linear part of the cycle. Results showed a close correlation of our fusion strategy with respect to gold standard patterns, a CF of = 0,9 for our strategy and a of CF = 0,8 for trajectories computed from the classical compass pendulum. The same test was repeated exclusively for the nonlinear part of the cycle: a CF of = 0,96 was estimated with our fusion model while a CF of = 0,35 was computed with the classical compass pendulum. Likewise, the Fréchet distance was used as a second metric since it allows to establish how close two trajectories are. Again, the complete approach proposed here (fusion strategy with the physical approach) outperformed the joint trajectories using a fusion strategy with classical compass pendulum: while the complete proposed strategy obtained a 0,1 Fréchet distance, the joint trajectories estimated with the classical pendulum yielded a 0,35 Fréchet distance for the normal gait. Overall, the largest error was obtained during the nonlinear parts of the gait cycle like the heelstrike and the hip moments. This fact illustrates the relevance of obtain a complete CoG global description even to estimate the remain joint angle trajectories

When simulations were performed for pathological patterns, our method maintained in average a CF = 0,90. Two neuromuscular disorders were simualted: the cerebral palsy and the partkinson disease in three different stages. For both abnormal movements, actual patient motion data allowed to adjust parameters and to obtain a closer CoG trajectory.

During the crouch gait simulation the model parameters were set to $k = 400$ and $d = 0{,}45$, according to the data learning process. Simulations achieved a CF of = 0,9, a very close representation of actual patterns (Figure **3-5**). The Fréchet distance maintained a comparable performance when tracking the ground truth. The Parkingson simulation, in three different disease stages, also shows a very alike representation w.r.t actual patterns. The model parameters were fixed according to Table **3-4** for each stage disease. Our model achieved in average a CF = 0,92 and a Fréchet distance average of 0,1, demonstrating the close likeness of the patterns obtained and actual data. These results demonstrated the model ability to accurately follow a different sort of gait paths, either normal or abnormal patterns. The model may be used as training tool for physician and also to predict the performance of a particular gait traitement.

Finally, it is worth to strengthen out the simplicity of the presented model and its ability to represent pathological movements by only tuning two parameters in the physical approach and with a relatively simple fusion strategy. The model may be used as a complementary tool of the classical gait analysis to determine an illness degree of any subsystem, either by correlating the compromise of any leg segment with the clinical data, or by perturbing the model and therefore the resultant trajectory.

3.4. Conclusions and perspectives

This work has presented a model to simulate normal and pathological kinematic gait patterns. Two main contributions are introduced in this work, a fusion strategy of two important information sources which allows the accurate estimation hip and knee joint angle trajectories. Additionally, a physical model is proposed here that describes the COG trajectory, using a pendular motion for the single stance and a spring mass system for the double stance. The model is complemented by an animated structure that allows to visualize and quantify different gait patterns,i.e., the hip and knee flexion-extension. The proposed approach can be easily extended to simulate other pathologies or even to find more dynamic gait relationships that describe a particular movement. Finally the proposed model opens up an actual possibility of understanding more complex gait phenomenon, that also could be used for example in the prothesis alignment.

4. Model Applications - Simulating Pathological Gait

One of the main advantages of the proposed model is the ability to simulate the normal and pathological gait, just varying a parameter set, achieving also clinical representation of motion (dynamic patterns). Thus, it was achived to simulate the gait affected by two neuromuscular diseases (Parkinson and Diplegia), which present a set of characteristic patterns that allow a proper evaluation the proposed model.
The this section was evaluated the Parkinsonian gait. [1] Using a set of real patients data captured in the gait laboratory of the Universidad Nacional de Colombia, the proposed model was adjusted, achieving to simulate the characteristic patterns of this pathological gait.
Subsequently, the proposed model was extended to a 3D version, with which it was possible to simulate normal and pathological gait patterns in both the sagittal and the frontal plane. In this case the dynamics of the Crouch gait [2] was simulated. A better explanations of these applications are shown below.

4.1. Physics-Based model to simulate the Parkinsonian gait

The Parkinson disease is a degenerative disorder of the central nervous system. Two new cases per 1000 persons are worldwide reported, while in Colombia around 75000 persons are affected by this disease [63]. A gait distorsion is one of the early clinical manifestations of this disease (PD), characterized by a forward trunk bending and a CoG shifting with respect of the usual position. Likewise, both legs are bent and the step length is very short and fast, causing a continued and growing imbalance of the vertical CoG. With the disease course, this picture is even aggravated, leading the people to fear normal activities such as walking or writing and resulting in partial or total loss of the movement. There exist different Parkinson classifications, one of the most accepted was established by Hoehn and Yahr [36, 72], signs that can be observed during routine examinations at gait laboratories. However, the early

[1] This motion is characterized by short steps, flexed posture of the body and lack of arm movement.
[2] Characteristic gait pattern in patients with different levels of diplegia.

disease stages are difficult to detect, even for an experienced physician, whereby the disease handling is completely dependent on a physician subjective judgement.

Many models have been reported in the literature, aimed to describe normal gait patterns [39, 93, 10]. These models describe the CoG trajectory, which constitutes a fundamental gait descriptor, i.e., these movements describe both central nervous system and musculoskeletal disorders [4, 26, 57]. Overall, these proposed models are very restrictive since they alter the natural gait and lose important gait features that can be used to classify a motion either as normal or pathological [1, 10]. The classical approach uses an inverted pendulum, whereby the upper part of the body is represented by a mass that moves forward with respect to each fixed point, describing a harmonic oscillating trajectory [58, 50]. At the same time, the free foot swings with respect to this mass, such as a simple pendulum. This simple system can be coupled to built a human gait model by a double- articulated pendulum. Based on this theory, Garcia et. al. [27] formulated a simplified gait model, for which the feet are considered relatively small with respect to the trunk and the heelstrike is subjected to a restriction rule. In despite of the benefits of this approach, the dynamics of inverted pendulum can not fully describe the trajectory of the CoG during a gait cycle, losing the double support phase description (20 % - 30 % of the gait cycle), a fundamental part of the cycle to describe important gait features like the heelstrike or hip moments and crucial towards a proper gait clinical analysis. In the particular case of the Parkinson disease, small changes in the double support phase such as the step length or the vertical/horizontal CoG displacement, are essential to establish the disease stage, as well as the treatment to follow [36, 72]. It is therefore important to introduce in these models a complete description of the double support phase that allows to obtain a full clincal analysis of the CoG movement [57, 55].

In this work we propose a novel model which allows to simulate different stages of any pathology, in particular Parkinson, revealing some important features of each phase. The model couples the dynamics of an inverted pendulum with a spring-mass system whose parameters were learned from 10 Parkinson patients with different degrees of Parkinson (2, 3 and 4) [55, 8]. The simulation yields a complete description of the most important gait patterns, using the heel trajectory and an inverse kinematic approach. The whole method was evaluated by comparing the simulated trajectory with actual Parkinson acquisitions, captured in the gait laboratory of the National University of Colombia, obtaining a correlation coefficient of about 0.85. This paper is organized as follows: section 2 describe the methodology used to simulate the Parkinson disease based on a mechanical gait model. The next section summarizes the evaluation and results and finally we present some conclusions of this approach and possible future work.

4.1.1. Methods and materials

The gait model herein presented fully describes a theoretical CoG trajectory. The path described by a CoG is the most important biomarker in gait analysis since it is altered in early stages of both central nervous system and musculoskeletal disorders [4, 26]. Anatomo-functional correlation is at the base of the present model since elements that introduce non linear patterns were herein considered, that is to say, components that store energy like tendons or cartilages are modeled by a simple spring-mass system, whereby the simulated periodic pattern allows also a cyclic exchange of energy. The model is built upon simple mechanical relationships, approximating the Garcia's model [27], a simple stance movement tracked by a pendular trajectory, while the double stage is herein simulated by a complementary spring-mass system. The parameters were learned from 10 patient videos with different stages of Parkinson. The whole strategy is illustrated in Figure **4-1**.

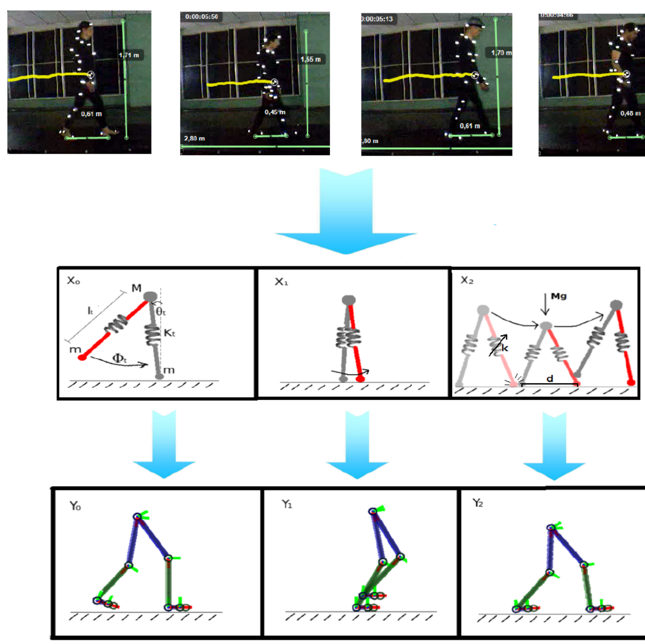

Figura 4-1.: This picture shows the general scheme. Trajectories obtained from 10 patients, with different Parkinson stages (2, 3 and 4) are used to learn the spring-mass model parameters. This trajectory is then used to simulate the Parkinsonian gait in an articulated model.

Gait model

To obtain the $CoG_{x,y}(t)$ trajectory, the global gait motion is represented during the simple stance as a double inverted pendulum. This representation was firstly proposed by Garcia et al. [27], based on the passive dynamic, optimizing the energy comsumption during locomotion. This model used was exposed previouslly in the chapter 3.

Adjusting the model

As we mentioned before, the gait model considers two parameters: k - the spring constant - and d - the step length- that can be modified to simulate normal and pathological gaits [55]. For achiving the Parkinsonian gait, the kinematic parameters of 10 Parkinson patients in different stages of the disease [3] were considered as a Gaussian phenomenon represented as $P(\mu, \sigma)$:P corresponds to a multivariate Gaussian distribution, whose mean was experimentally obtained averaging the 10 gait trajectories and σ describes the inherent gait variability (See table 4.1.2). Then, the CoG and heel trajectories for several gait cycles were so obtained. The simulated gait reflects some important Parkinsonian gait pattern, mainly the step length d. This observation coincides with the accepted clinical meaning of each Parkinson stage, case in which, the more advanced this disease is, the shorter the step length. The constant k of the mass-spring was thus calculated from the real data for each group of patients and used to simulate the CoG and heel trajectory, An articulated model was then animated with this information, making possible a proper description of the Parkinsonian gait kinematics in different stages, i.e., sagittal plane patterns were appropriately simulated, as shown in the figure **4-2**.

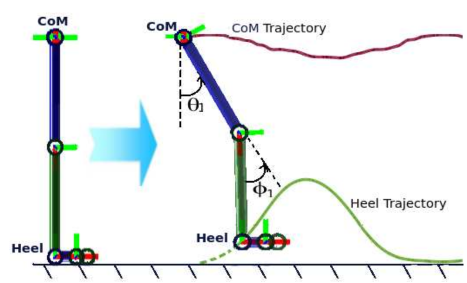

Figura 4-2.: The figure shows the inverse kinematic methodology used to generate the dynamic patterns of Parkinsonian gait using the COG trajectory generated and the mean heel trajectory of each stage of Parkinson.

Dataset description

Experimentation was carried out with sagital video sequences, registered at the gait laboratory of the National University of Colombia, under semicontrolled illumination conditions. This dataset consists of a set of videos captured from 10 patients, each recorded at least at 4 different occasions. The Dataset was divided as follows:

- 3 Patients diagnosed with 2nd grade of the Parkinson disease (No depressive disorder present).

[3] Stages 2, 3 and 4, according to the classification by Hoehn and Yahr[36]

Model Parameter	Stage 2	Stage 3	Stage 4
k average	52.3 ± 6.22	67.42 ± 8.25	96.8 ± 2.54
d average	0.76105 ± 0.087	0.5576 ± 0.0474	0.4356 ± 0.0294

Tabla **4-1**.: Model parameters learned from 10 patients in different stages of the Parkinson disease.

- 4 Patients diagnosed with 3rd grade of the Parkinson disease (No depressive disorder present)
- 3 Patients diagnosed with 4th grade of the Parkinson disease (No depressive disorder present)

4.1.2. Evaluation and results

The evaluation was carried out by comparing the CoG trajectory generated by our model with the actual observed trajectories. For each stage, we obtained a set of parameters k and d (See table **4-1**, used to approximate the CoG movement of each disease stage (Figure **4-1**). The fusion rule, with the computed $CoG_{x,y}(t)$ and the learned heel paths from groups of patients, was used to compute the hip $\gamma(t)$ and knee $\omega(t)$ joint-angle trajectories. The set of these trajectories allows to animate an articulated model as a virtual representation that shows characteristic Parkinsonian signs such as the short step and the slight flexion of the knee and the hip, as illustrated in figure **4-4**. Each parkinson disease stage trajectory was also compared with the two similarity metrics previously introduced, i.e., the correlation, and results are
For each case, we calculated the correlation coefficient to establish the degree of similarity between both, the observed and the simulated paths. As shown in table **4-2**, we achieve a close relationships with respect to the ground truth.

Finally, we simulate the Parkinsonian gait, considering both the CoG and heel trajectories with an articulated model. A virtual representation of the sagittal dynamic patterns, was then generated in Matlab. This animation shows characteristic Parkinsonian signs such as the short step and a slight flexion of the knee and the hip, as illustrated in figure **4-4**:
The CoG,kinematic patterns and heel trajectories of the Parkinson patients were recorded in the gait Laboratory of the National University of Colombia and computed using a free source software of motion analysis [5] which gives a good approximation of body motion. These trajectories were used to fit the model parameters as mentionedm. The model was writed in Matlab.

[5] Kinovea - Video Analysis software for sports-`http://www.kinovea.org/`

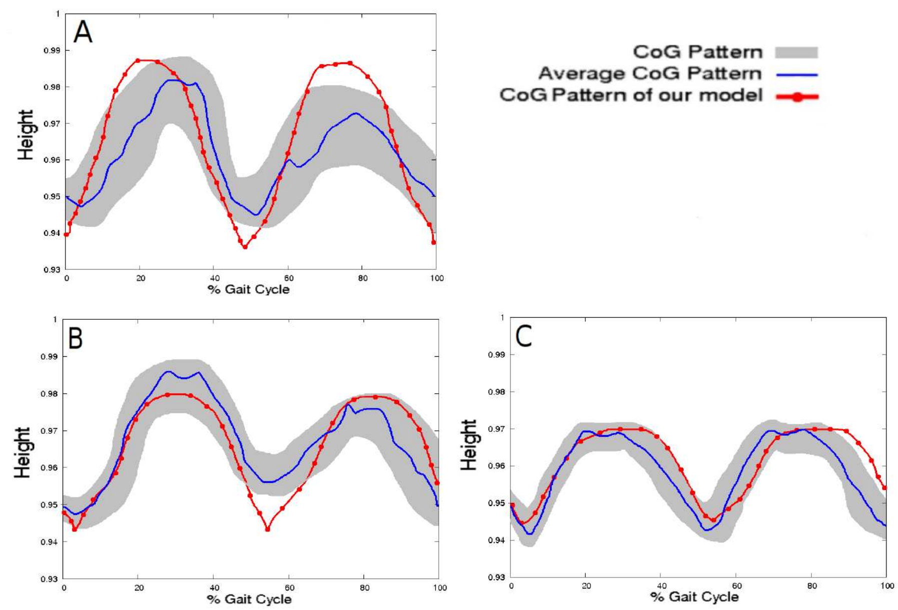

Figura 4-3.: Examples of the proposed method performance when estimating the dynamics of the CoG movement in different stages of the Parkinson Disease: Second Stage (A), Third Stage (B) and Fourth Stage (C).

2nd PS[4]	Metric	R.Hip	R.Knee	L.Hip	L.Knee	CoG
	CF	0.92± 0.01	0.91± 0.01	0.95± 0.01	0.93± 0.01	0.83± 0.004
	Féchet	0.09 ± 0.001	0.11 ± 0.002	0.09 ± 0.001	0.12 ± 0.002	0.12 ± 0.001
3rd PS	Metric	R.Hip	R.Knee	L.Hip	L.Knee	CoG
	CF	0.95± 0.012	0.89± 0.009	0.93± 0.012	0.96± 0.01	0.88 ± 0.003
	Féchet	0.09 ± 0.001	0.13 ± 0.002	0.09 ± 0.001	0.08 ± 0.002	0.13 ± 0.016
4th PS	Metric	R.Hip	R.Knee	L.Hip	L.Knee	CoG
	CF	0.93± 0.008	0.90± 0.008	0.96± 0.008	0.91± 0.008	0.86 ± 0.005
	Féchet	0.09 ± 0.002	0.12 ± 0.001	0.17 ± 0.002	0.13 ± 0.001	0.04 ± 0.001

Tabla **4-2.**: Results for each stage of the Parkinson disease and the trajectory generated by our approach.

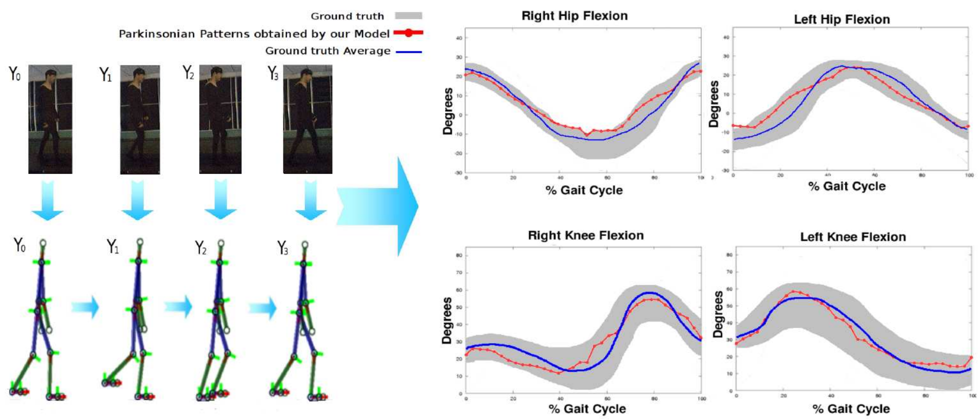

Figura 4-4.: This figure illustrates the simulation of the Parkinsonian gait with an articulated structure and its sagittal dynamic patterns.

4.1.3. Conclusion

This chapter presents a novel approach to simulate different stages of the Parkinsonian gait using a simple bipedal model. The whole model couples the dynamics of inverted pendulum with a spring-mass system whose parameters were learned from 10 Parkinson patients with different levels of the disease (2, 3 and 4). A proper description of the CoG trajectory of the Parkinsonian gait was obtained. The results show that the model simulates properly the main signs of the Parkinsonian gait, such as the knee rotation, the CoG attenuation and the short step length, which are considered as biomarkers of the Parkinson. As future work we propose to improve the estimation of the model parameters of each Parkinson stage analyzing a larger group of patients and develop a diagnostic support tool that allows to reduce the levels of subjectivity of the clinical Parkinsonian gait analysis.

4.1.4. Acknowledgments

This work was partially funded by the project *Anotación Automática y Recuperación por Contenido de Imágenes Radiológicas usando Semántica Latente* number 110152128803 of the Convocatoria Colciencias 521 of 2010, and by the project *Modelo teórico y computacional de marcha humana para simular patrones patológicos causados por trastornos neuro-musculares* number 14510 of 2011 accepted by the DIB (Dirección de Investigación sede Bogotá) of the National University of Colombia.

4.2. A 3D Physics-Based Model to Simulate Normal and Pathological Gait Patterns

The gait is one of the most broadly studied motion in very different domains such as medicine, animation or robotic [89]. This movement is the result of complex interactions between several sub-systems, which work together to generate the body dynamics that underlies the bipedal displacement [26]. This human gait patterns are frequently disturbed in many types of pathologies. In the clinical routine, a physician or rehabilitation expert searches pathological gait patterns (according to her/his expertise), supporting the decision on some statistical tests of the acquired data and introducing thereby an inevitable expert-dependent bias [16].

In addition, traditional methodologies to capture gait patterns are very invasive and alter the natural gait gestures which also are contaminated by capture noise. Therefore, development of gait models achieving an accurate quantitative movement description has become a priority to support the physician decisions [24, 51].

In the literature has been proposed many models that simulate, analyze and represent the human gait. Some of them are physics-based models, which formulation is very restrictive and it consider in general the motion in a planar view [92, 93, 89]. Other approaches, consider optimization and control theories to generate and simulate the human gait, but its representation are in general highly subjective and computationally expensive, which requires a good set of parameters to obtain stable outputs [89].

This work introduces a novel 3D human gait model that generates the dynamic gait patterns in the sagittal and coronal views observed in a clinical gait analysis using a physics-based approach . It provides high flexibility, generating a large number of walking patterns using only a complete CoG trajectory representation and the heel trajectory data obtaining normal and pathological gait patterns, for instance the dynamics of a typical cerebral palsy gait (Crouch gait), changing a few set of parameters. This simple mechanic gait representation simulates the energy accumulation of different anatomical elements, responsible for most non linear gait patterns: it uses a double-inverted pendulum system to simulate the single stance CoG trajectory, while the double stance CoG trajectory is simulated by a double spring-mass system, in both sagittal and coronal views. The obtained trajectory feeds a human like-structure, which is animated using a learned heel trajectory and a classical inverse kinematic to calculate the dynamic patterns. The rest of the paper is organized as follows: Section Related Works review similar approaches related with the model of human gait, section Materials and Methods introduces our model, section Results demostrates the effectiveness of the model. The last section presents the conclusion and future works.

4.2.1. Related Works

Several models have been previously proposed for simulating the human gait, among which the physics-based models obtain realistics and natural representations of the human Center of Gravity (CoG).[89]. Likewise, they properly describe the gait from an energy standpoint, simulating the change from the kinematic to potential energy during the gait cycle and taking into account the movement dependence on some external interactions [89]. One of the well known models was proposed by Garcia et al. [27] based on the *passive dynamic theory* of McGeer [27]: a double articulated pendulum system for which feet are relatively small with respect to the trunk and the heelstrike follows a very restrictive rule. This gait description is still limited because about a 30 % of the gait cycle that corresponds to a double stance phase, is completely eliminated and the CoG displacement in the lateral axis (Y) is not considered at all. Other approaches included the inverted pendulum representation to describe the lateral and frontal CoG motion, but not its vertical displacement, which is assumed constant [47].

Other recent approaches are based on optimization techniques and control-based models. The optimization methods include a large number of degrees-of-freedom for producing optimal motions while joint force profiles remain subjected to a large number of constraints[16, 24, 89]. These methods require relatively few data to simulate simple human structures and predict new motions, very useful in computer graphics, robotics and animation applications.The control-based models have been commonly used on robotics and biomechanics for designing the real-time control in biped walking prototipes [85]. The main advantage of these methods is that they approximate the actual human control systems, allowing to simulate both normal and pathological gaits. The neuromotor system simulation allows the analysis of some neurological pathologies [47], while this is computationally more efficient than the optimization-based models. Nevertheless these methods are computationally expensive and require specific knowlegde of the problem [89] whereby these strategies are highly subjective[89], and also require a large group of experimental data to generate natural motions. This last drawback has highly limited its application in clinical gait analysis because of the specific requirements to obtain a stable and natural motion.

4.2.2. Materials and Methods

The gait model herein proposed fully describes a 3D CoG trajectory of normal and pathological gaits. The whole model is built upon simple mechanical relationships, approximating the 3D CoG displacement with a double-inverted pendulum for the simple stance phase, and a double spring-mass system for the double stance. Using this physics-based representation, we can animate an human leg structure and simulate normal and pathological kinematic patterns. The kinematic patterns for each case was calculated with a classic method of inverse-kinematic, using the CoG trajectory obtained and a learned heel trajectory taken from some gait laboratory's data as illustrated in figure **4-5**.

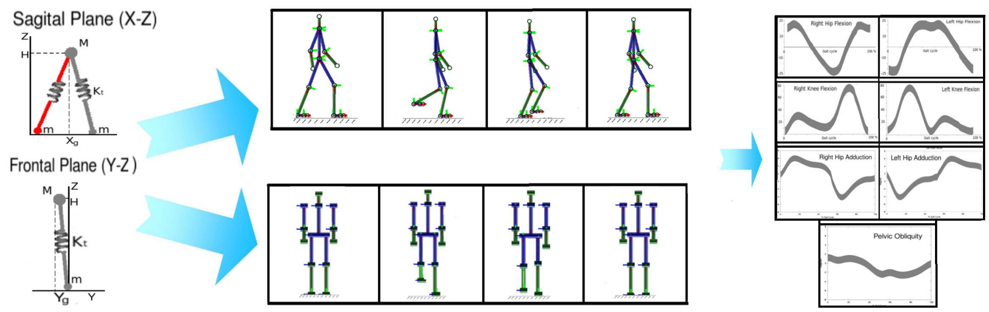

Figura 4-5.: Overview of our proposal. Firstly, it is computed independently the CoG trajectory in both sagittal and coronal views. Then we built up an articulated structure and simulate the human gait using the CoG and a learned heel trajectory using a classical inverse kinematic approach.

Sagital CoG description

As a first step, the CoG sagittal trajectory was computed using a physics-based gait representation. The single stance phase (one foot supporting the body) was represented as a double-inverted pendulum. This representation is based on the passive dynamic [27]. This model is formulated as a pair of coupled nonlinear equations: $\beta(1-\cos\phi)(3\ddot{\theta}-\ddot{\phi})-\beta\sin\phi(\dot{\phi}^2-2\theta\dot{\phi})+(\frac{g\sin\theta}{l})(\beta(\sin(\theta-\phi)-1))=0$ and $\ddot{\theta}(\beta(1-\cos\phi))-\beta\ddot{\phi}+\beta\dot{\theta}^2\sin\phi+(\frac{\beta g}{l})\sin(\theta-\phi)=0$, where $\beta=m/M$, m is the mass of each foot and M is the body mass, θ is the angle of the supporting leg at a particular time t and ϕ is the angle between both legs. When the leg stance has been accomplished, the heelstrike is represented by the non linear rule$\phi(t)-2\theta(t)=0$. This phase is characterized by the hip and knee moments generated within this interval, important biomarkers in many abnormal movements.

On the other hand, the double stance phase, which starts just after the heel strike, was represented as a double spring-mass system. This physical formulation introduces attenuation as it is usually observed in an actual CoG trajectory because of the knee rotation, but it also involves an intrinsic representation of the muscles acting during this phase. This representation is given by the following equations [8]: $m\ddot{x}=Px-Q(d-x)$ and $m\ddot{y}=Py+Qy-mg$, where $P=k(\frac{l_0}{\sqrt{x^2+y^2}}-1)$, $Q=k(\frac{l_0}{\sqrt{(d-x)^2+y^2}}-1)$, m is the body mass, x and y are the sagittal and frontal displacement respectively, d in the step length, k is the spring constant, l_0 is the spring lengh at rest and g is the gravity constant (9,81). This physic-based representation describes completely the CoG trajectory and allows to have the flexibility needed to generate different kind of gaits, i.e., pathological gaits.

Coronal CoG trajectory

A main contribution of this work is a 3D gait representation that allows to accurately mimic pathological motions like the pelvic balancing [27]. The CoG trajectory in the coronal view was simulated using the inverted-pendulum approach. First, we calculated independently the CoG motion in both the $x - z$ and the $y - z$ planes [47]. The CoG trajectory was modeled by a simple inverted pendulum represented by the following equation: $\ddot{y} = \frac{g}{z_i} y$

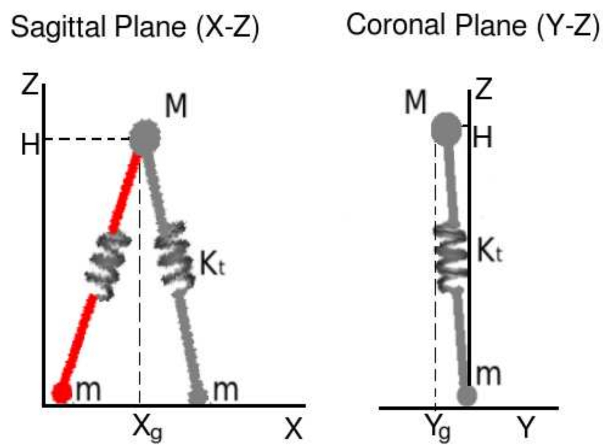

Figura 4-6.: Physical model in sagittal and coronal views. On the right side it is shown the double-inverted pendulum coupled with a spring system used to describe de sagittal CoG trajectory. On the left side it is shown the physical representation in the coronal view which behaves like a simple inverted pendulum.

Where z_i is the CoG heigh at time i (same time for the sagittal view) and g is the gravity constant ($g = 9{,}81$) [47]. Given the initial conditions, the CoG trajectory can be described by the following equation: $y = y_0 \cosh \frac{t}{T} + \dot{y}_0 T \sinh \frac{t}{T}$
Where y_0 and $\dot{y}_0$ are the initial conditions and $T = \sqrt{\frac{z_i}{g}}$.
Finally, a complete 3D CoG trajectory was obtained calculating independently both the sagittal and frontal CoG trajectories.

3D human model representation

This model not only describes the gait in terms of energy, but it also introduces a flexion-extension/adduction-abduction limb description as well as a pelvic obliquity that modify the model dynamics. The kinematic simulation is herein carried out using a classical inverse-kinematic method whose resultant trajectory animates a human-like leg structure, composed of seven articulated rigid segments as illustrated in Figure **4-5**.

Pathological gait simulation

The main advantage of our proposed 3D human representation is that it allows to simulate many kind of movements patterns in both sagittal and coronal views by only tuning the parameters k (the elasticity constant) and d (the step length). With little change in these parameters it is possible to simulate the CoG changes observed in some pathological gaits and to obtain the knee rotation and the CoG attenuation that characterizes some pathological movements. Setting the $k \sim 400$ and $d \sim 0{,}7$, a normal gait is simulated, but if the k value is increased and/or the d constant is decreased, a pathological gait, for instance the *crouch gait*[6] that characterizes the motion of several cerebral palsy patients [26] is simulated.

4.2.3. Results

The evaluation was initially carried out by comparing the CoG trajectory generated by our model with the actual one observed from normal gait trajectories as shown in the figure **4-7**.

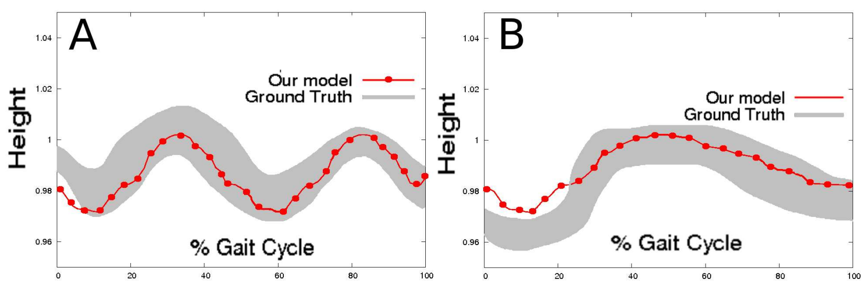

Figura 4-7.: Calculated CoG trajectory in sagittal (A) and coronal view (B).

For both sagittal and coronal views, we calculated the correlation coefficient to establish the degree of similarity between both actual and simulated CoG trajectories. The results shows that the correlation coefficient in the sagittal view is $0{,}85 \pm 0{,}056$ and for the coronal view is $0{,}97 \pm 0{,}013$.

Tabla **4-3**.: This table shows the correlation factor calculated between both actual and simulated CoG trajectories.

Sagittal View	0.85 ± 0.056
Coronal View	0.97 ± 0.013

Simulating normal gait patterns

As a second evaluation, we compared the estimated obtained gait kinematic patterns with ground truth trajectories that were obtained from normal patients, as reported in the literature [26], composed of sagittal and coronal patterns of a gait cycle.
As in the previous subsection we calculated the correlation coefficient to establish the degree of similarity between both trajectories, the patterns obtained by our method and the average of each actual normal pattern(table **4-4**).

[6]This kind of gait is characterized by an exaggerated knee flexion

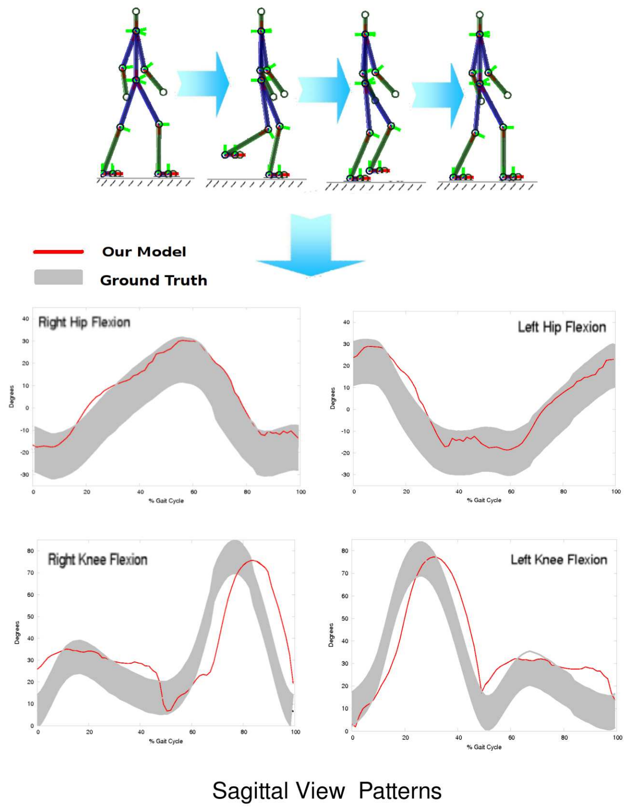

Figura 4-8.: Simulation (top) and the patterns obtained (bottom) for sagittal view.

Evaluation for an actual pathology

As previously mentioned, one of the main advantages of our model is the possibility of representing pathological gaits by only tuning some parameters of the spring-mass model: The elasticity constant k and the step length d. A crouch gait pattern was simulated by simply increasing the k constant and decreasing the d constant for both frontal and coronal views, as is comment in the section 4.2.2 and as shown in figures **4-10** and **4-11**.

The correlation coefficient was calculated to compare the similarity between the patterns obtained and the average of the actual data. The results are shown in the table **4-5**.

4.2.4. Conclusion

This work has introduced a mechanical gait model which is able to mimic pathological patterns, specifically a gait crouch which is characteristic of the cerebral palsy. This kind of models is a potential useful tool for the physician understands the nature of a particular pattern during the gait. This kind of adapted designs in addition can be a remarkable aide to simulate the evolution of a specific treatment so that treatment planning is possible.

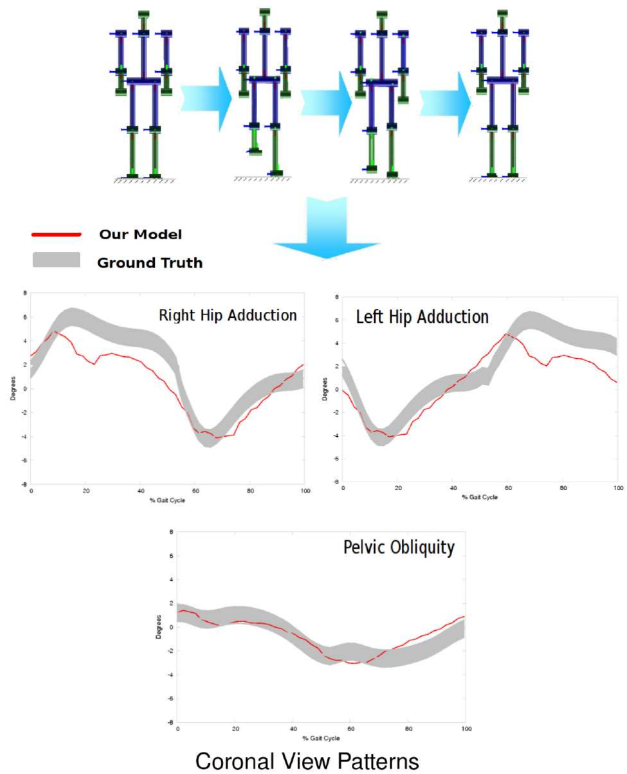

Figura 4-9.: Simulation (top) and the patterns obtained (bottom) for coronal view.

Acknowledgments

This work was partially funded by the project number 14510 of 2011 accepted by the DIB (Dirección de Investigación sede Bogotó) of the National University of Colombia.

Tabla **4-4**.: This table shows the correlation factor calculated between both actual and simulated kinematic patterns.

Sagittal View	R. Hip Flex.	0.974 ± 0.016
	R. Knee Flex.	0.71 ± 0.021
	L. Hip Flex.	0.968 ± 0.016
	L. Knee Flex.	0.71 ± 0.021
Coronal View	R. Hip Add.	0.918 ± 0.013
	L. Hip Add.	0.891 ± 0.013
	Pelvic Obliq.	0.876 ± 0.02

Tabla **4-5**.: This table shows the correlation factor calculated between both actual and simulated kinematic patterns.

Sagittal View	R. Hip Flex.	0.9514 ± 0.012
	R. Knee Flex.	0.8638 ± 0.1101
	L. Hip Flex.	0.9529 ± 0.00098
	L. Knee Flex.	0.8810 ± 0.0425
Coronal View	R. Hip Add.	0.9188 ± 0.023
	L. Hip Add.	0.891 ± 0.023

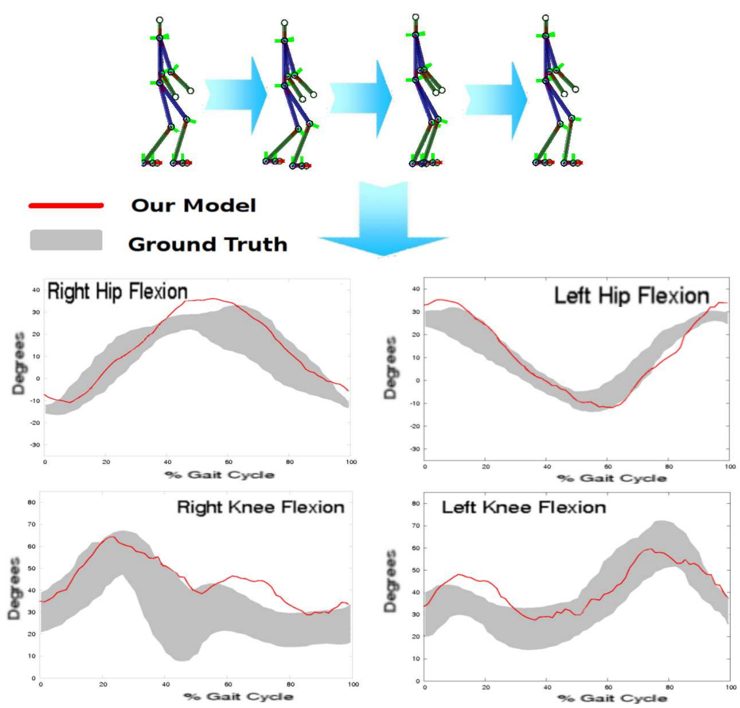

Figura 4-10.: Simulation (top) and generated pathological patterns (bottom) in the sagittal view.

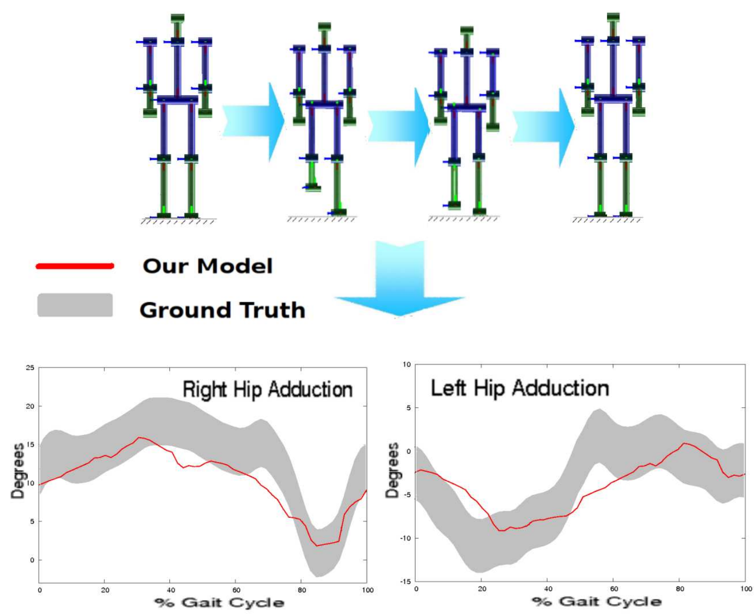

Figura 4-11.: Simulation (top) and generated pathological patterns (bottom) in the coronal view.

5. Conclusions and Future Work

The gait is a complex combination of several motion patterns, generated by the interaction of different systems, under the restriction that the energy consumption is minimum. In the clinical routine, the gait examination is one of the most important tools for identitying motion disorders, it is used as a biomarker of some neuro-muscular illnesses like the cerebral palsy or Parkinson disease, supporting diagnosis and facilitating follow up. Interestingly, during this analysis it is possible to evaluate the effectiveness of a specific treatment and the particular response of the multiple gait subsytems. However, these analyses are not actually carried out in the clinical practice, among others because this is very time consuming and examiner dependent. Moreover, measures are contaminated by noise during the capture or, by invasive devices like markers, which inevitably alter the natural motion gestures so that a proper comprehension of the gait phenomenon is impossible. Overall, most clinical examinations are devoted to capture a general gait picture and hence treatments are globally addressed. The actual utility of devising gait models is that they improve the understanding of some sub-system patterns from the captured data, even if they are noisy, and to plan more specific treatments.
The gait models reported in the literature are very limited when describing particular variations of a pathological motion. Gait models can be coarsely divided in two large groups: physical based models and musculoskeletal representations. Physical models use an inverted pendulum or a spring mass system that allow an energy consumption analysis and a global dynamic description. These models nevertheless fail to mimic pathological patterns because of the strong simplifications and restrictions, i.e., pendular models represent only the single stance while the double stance is completely omitted. Musculoskeletal representations, typically more complex than physical-based models, are able to simulate muscle patterns at each phase of the gait by adding either a Hill's model or prior information coming from data obtained from actual gait laboratories. Nowadays, musculoskeletal models associate each leg segment to a Hill's model. However, a main drawback of this approach is that there is no interaction between the different segments and hence simulations are quite far from experimental data [15]. Best performances are obtained by combining the prior model with observations coming from actual data [15], but these models completely neglect some kinetic relationships and must be tuned by an experienced observer. In contrast, physics based models are simple and usually tuned with a small number of variables. This simplicity, nonetheless, leads to most physics based models to miss important phases of the gait cycle. The fusion model proposed in this thesis simulates the single stance phase by an articulated

double pendulum system, while the double stance is properly simulated by a spring mass component that stands for the important knee motion contribution. The proposed model in addition emulates a whole skeleton structure by animating this architecture from the CoG trajectory, simulating a large set of gait patterns, including pathological conditions and isolating the role of certain subsystems. This scheme uses two important information sources, i.e., kinetic and kinematic gait components, maintaining the possibility of an energy consumption analysis. The fusion method assumes that gait kinematic patterns must follow basic physical principles. The underlying trajectories are generated by an adapted version of the compass pendulum representation which has been extended with elements that store energy, thereby describing a larger number of abnormalities than it has been possible so far. These trajectories are then regularized by actual learned paths, completing thereby the rule of fusion by means of an inverse kinematic approximation.

The whole strategy allows a natural simulation of non-linear gait patterns, representing many kinds of movements. Evaluation demonstrated the models advantages with respect to the classical approaches, like Garcia's model, and the great similarity of the obtained trajectories with the gold standard (patterns broadly described in the biomechanical literature). Accuracy and efficiency were herein assessed under several metrics. Validation was carried out by animating the same human leg structure with two different representations for the CoG: the proposed and Garcia's models. From each simulation we estimated several kinematic patterns like the CoG trajectory and the hip and knee flexion-extention movements, considered as the most representative gait patterns for determining whether a motion is pathological or not. The first metric was a simple correlation factor, and simulations showed in average that our method obtained a 0,96 correlation coefficient while Garcia's model only achieved a 0,35 for a normal pattern. When simulations were performed for pathological patterns, our method remained in a 0,90 correlation coefficient (Garcia's model do not simulate pathological patterns). Likewise, the Fréchet distance was used as a second metric since it allows to establish how close two trajectories are. Again, the proposed model outperformed the classical strategy: while the proposed strategy obtained a 0,1 Fréchet distance, Garcia's model yielded a 0,35 Fréchet distance for the normal gait. Overall, the largest error was obtained during the non linear parts of the gait cycle like the heelstrike and the hip moments. During the crouch gait simulation herein presented, the Fréchet distance maintained a comparable figure when tracking the ground truth and distortion was minimal.

On the other hand, an appropiate extraction of the CoG is essential since this biomarker is an efficient indicator of the normal/abnormal gait pattern, indeed it constitutes a major indicator in pathologies such as hemiplegia, paraplegia or distonia. Also, an optimal walking, in terms of the energy, can be defined as the CoG movement with minimum energy expenditure. Therefore, a pathological gait can be analyzed in terms of energy using the CoG, which tracks the transfer of potential to kinetic energy (recovery), i.e. normal gait patterns lose 40 % of the energy in this transfer, a higher loss is pathological [18]. We compare the CoG described by the physical based model and a clasical Garcia's model, obtaining 0,84 correlation factor while Garcia's model only achieved a 0,52. This evaluation confirmed what was already shown before, the proposed method follows better non linear parts of the gait cycle like the heel strike phase while the classical representation fails to consider the automatic change of the balancing legs, as illustrated in the Figure 3.2.

Additionally, it is worth to strengthen out the simplicity of the presented model and its ability to represent pathological movements by only tuning two parameters in the physical approach and with a relatively simple fusion strategy. The model may be used as a complementary tool of the classical gait analysis to determine an illness degree of any subsystem, either by correlating the compromise of any leg segment with the clinical data, or by perturbing the model and therefore the resultant trajectory. The proposed approach can be easily extended to simulate other pathologies or even to find more dynamic gait relationships that describe a particular movement. Finally the proposed model opens up an actual possibility of understanding more complex gait phenomenon, that also could be used for example in the prothesis alignment.

5.0.5. Acknowledgments

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A. Appendix A

A.1. Products

This master thesis allowed to obtain some publications, as research products. These products are listed below.

- Article **Title:Análisis Teórico y computacional de la marcha normal y patológica: Una revisión. Authors:** Christian Cifuentes, Fabio Martinez, Eduardo Romero - vol.18 no.2 Bogotá July/Dec. 2010. "Med" Magazine. Medicine School of Universidad Militar Nueva Granada (Clasif:A). Print version ISSN 0121-5256.

- Article accepted in the 7th International Seminar on Medical Information Processing and Analysis- SIPAIM 2011 as an oral presentation in Bucaramanga, Colombia on December 2011.**Title:** Physics-based model to simulate the Parkinsonian gait. **Authors:** Christian Cifuentes, Fabio Martinez, Eduardo Romero. Published in the Proceedings of the conference.

- Article accepted for poster presentation in the 7th International Joint Conference on Computer Vision, Imagin and Computer Graphics Theory and Applications - VISIGRAPP 2012, in Rome-Italy. February 2012.**Title:** A 3D Physics-based model to simulate normal and pathological gait patterns. **Authors:**Christian Cifuentes, Fabio Martinez, Eduardo Romero. Published in the Proceedings of the conference.

- Article submitted in the Journal of Neuroengineering and Rehabilitation. May 2012. **Title:** Simulations of normal and pathological gaits from a fusion knowledge strategy. **Authors:** Fabio Martinez, Christian Cifuentes, Eduardo Romero

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