Development of a human walking model comprising springs and positive force feedback to generate stable gait

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Abstract

Biomechanics is a very important field of study when one considers the importance that motion, such as walking, has in our everyday life. Although we perform tasks such as running and climbing stairs in an automatic way, without even thinking of it, these tasks are extremely complex from a mechanical and biological point of view. Nowadays, the progression of science towards the understanding of the human body and locomotion has been exciting, with huge advancements being held in fields such as prosthetics and humanoid robots. In the latter case, humanoid robots such as ASIMO by Honda and Petman by Boston Dynamics are now able to perform much of the tasks required for human locomotion. However, much of the state-of-the-art humanoids are not as robust as desirable, generating a gait pattern that is frequently dependent on pre-defined joint trajectories. Inspired in neuro-musculo-skeletal models, principles of spring mechanics, feedback control theory and a careful study of the human gait cycle, we present in this work a new reflex-type controller to generate a human walking pattern in a 2D simulation. Developed in a Matlab, Simulink and SimMechanics environment, this control strategy comprises the motion of the three main joints of walking (ankle, knee and hip) in the sagittal plane.

Keywords: Human Gait, Reflex Control, Humanoid Robots, Force Feedback Control, Biomechanics

1 Introduction

The word “robot” is actually not very old. It was first presented by the Czech playwright Karel Capek in 1921 [1], derived from the Czech word “robota” that means labor. The first modern programmable and industrial manipulator robot in the world, Unimate, was only introduced in 1961, having worked on a General Motors assembly line in New Jersey [2]. Unimate was definitely a revolution as it was able to lift and stack hot pieces of die-cast metal and spot welding, a task that was difficult and dangerous if done by humans.

According to Bill Gates [3], robots will have an extremely important role in the future society, assisting humans in dangerous jobs, providing physical assistance, helping health care workers to diagnose, treating patients that live in the other side of the world, perform search and rescue operations and even providing companionship for the elderly (as Paro, a seal therapeutic robot [4]).

Amazing advances have taken place in various fields such as in prosthetics; artificial devices that can replace missing body parts, such that in some cases these devices can even provide certain advantages when compared with biological parts. Such discussion was held in early 2008 when Oscar Pistorius, a runner that uses transstibial prosthesis limbs, was in the process of competing in the 2008 summer Olympics, and the advantage he had compared to runners who had biological ankles was on the table [5].

Many attempts have been made in order to develop a robot walking mechanism that can be compared to the human gait pattern. Although every one of them has its advantages and disadvantages, most of them require pre-defined joint trajectories in order to develop a stable walking pattern. For the development of such robots that can adapt to different unknown ground environments and learn from its errors, it is necessary to find research groups that unite engineers, neuroscientist, psychologists and even philosophers [6].

Biomechanics in general and the study of human movement in particular is fundamental when one considers the importance that walking and running have in human life. As David A. Winter wrote, “Walking is very important for meeting the world, for growing up, for retreating to solitude, for returning to join again, for carrying the day’s tasks, for belonging” [7].

Walking is an amazing and complex interaction between the Central Nervous System (CNS), muscles and gravity, being remarkable that we do not have to think about such task in our day to day life (fortunately, for that would be a great time-consuming task). Nowadays, neuroscience is attempting to fully uncover the neuronal dynamics that enable us to walk and precisely trying to discover the generation of the signals that guide our muscle. For instance, some neuroscientists have hypothesized that the...
signals from the brain to the muscles are predictions of sensory input and not commands like was prior belief [8].

A usual method to generate human like walking in humanoids is the ZMP technique, used in very known robots such as ASIMO [9] and HRP-4 [10]. The ZMP is the point where the vertical inertia and gravity add to zero [11]. At each step, the robot calculates where its foot should be placed in order to have zero moment in the horizontal plane. This method generates a joint trajectory that satisfies the ZMP constraints with an online tracking mechanism that guides the joints to the desired trajectory in a more robust way, overall assuring a dynamical stable walking pattern. This technique is very good when one considers stable and controlled environments. There is however some drawbacks such as the resulting “bent-knee” kind of motion that looks unnatural and limits the type of walking styles that can be applied to the robot.

Methods such as reinforcement learning [12], learning by imitation [13] and passive-walking [14] have also been explored with great results. The latter approach tries to take advantage of the non-linear properties of passive spring-damper components and the interaction between the robot and the environment, like gravity forces. Using this method, robots such as Petman (in Figure 1.3) by Boston Dynamics [15] achieve a more human-like walking pattern compared to those that use the ZMP approach.

In the context of locomotion, a humanoid robot should know when to change between walking and running and adapt to surface changes such as slippery ground, slopes and obstacles along the way. There is a strong need in uniting robotics, biomechanics and cognition research in order to develop a humanoid robot that is not only capable of mimicking the human walking pattern but also able to make decisions and solve problems as it moves. A lot can be also learned from robotics into cognition as the researchers have tools in which they can apply their theories and measure the outcome [16].

The basis of this master thesis work is Geyer and Herr’s article entitled “A Muscle-Reflex Model That Encodes Principles of Legged Mechanics Produces Human Walking Dynamics and Muscle Activities” [17]. As explained later, in this paper, Geyer and Herr described how they developed a stable human-like walking pattern for a planar biped, using seven Hill-type muscles at each limb. Although this method is not currently applied to humanoid robots, it is surely a good alternative to state of the art solutions, as it does not depend on pre-defined references from human motion and it has a strong background in human biomechanics and locomotion.

The purpose of this new model is to be applied to the control of small humanoid robots such as the Bioloid by Robotis [18]. This robot is a very cheap humanoid robot, which is perfect for educational purposes. With this platform, is possible to develop good applications in engineering, inverse kinematics and kinetics. In previous years, some works such as [19] were done in the Mechanics Department of Instituto Superior Técnico in exploring this humanoid robot platform.

In section 2 I will briefly describe the gait cycle and then move to the presentation of [17] section 3. In sector 4 I describe how the joints can be approximated as springs. In section 5 I describe the model developed in the context of this master thesis, divided into three sub-sections: ankle, knee and hip control. In section 6 it’s discussed the optimization procedure and in section 7 the resulting walking pattern obtained with this model. In the final section, I conclude and explore a little bit the possibilities of improving this model, in order to apply it to a humanoid robot.

2 Human Gait Cycle Analysis

The human gait cycle consists of one step, it begins when one foot initially touches the ground and ends when the same foot touches the ground again after completing its two main phases, stance phase and swing phase [20].

![Figure 1 - The gait cycle divided into stance and swing phases](image)

In the context of this work and the development of the model presented in section 5, it is important to explain first the reference angular positions in the ankle, knee, hip and trunk. In the ankle, we defined the zero angle when the internal angle between the foot and the leg is 90 degrees. In this case we defined dorsiflexion as positive angular displacement and plantar-flexion as negative angular displacement. We defined the zero angle at the knee as when the leg is in line with the thigh and so, as the knee flexes, its angular position increases. The angle of the hip joint is defined as the angle between a line that moves from
the trunk downwards and the thigh. The angle is considered to be positive when the hip flexes and negative when it hyperextends. The trunk’s angular position is defined as the angle between the trunk and a vertical line. This angle is considered to be positive as the trunk tilts forward and negative otherwise.

The first sub-phase of stance (approx. from 0% to 10% of the gait cycle) is the loading response and begins when the foot first touches the ground, which is called initial contact. Normally, this event is also called as heel strike. This phase ends with toe-off by the opposite leg. This phase also initiates the first double-support phase of the gait cycle as the two feet are touching the ground. The functions of this phase are shock absorption, weight-bearing stability and preservation of progression [21]. Mid-Stance is the second sub-phase of stance (approx. from 10% to 30% of the gait cycle), starting with opposite toe off and ending with heel off by the supporting limb. It starts the first single-support phase of the gait cycle which ends with initial contact by the opposite foot. The goal of mid-stance is to stabilize the limb and trunk and to achieve progression over the stationary foot.

The goal of the third sub-phase of stance (approx. from 30% to 50% of the gait cycle) is the progression of the body beyond the supporting foot. Terminal Stance starts with heel off and ends with initial contact or heel strike of the opposite foot. It is also the second and last part of the first single limb support phase of the gait cycle. During this phase the center of pressure moves to the forefoot. Pre-Swing (approx. from 50% to 60% of the gait cycle) begins a task that can be named as limb advancement. That task begins at the end of the stance to prepare the posture needed to complete the advancement of the limb. This sub-phase begins with initial contact by the opposite foot and ends with heel off by the current foot, as it starts swing phase. Also, this phase starts the second period of double-support of the gait cycle, as the weight begins to be transferred from the trailing leg to the opposite leading leg. The freedom created by this transfer of body weight to the leading leg is used to prepare the ipsilateral leg for leaving the ground.

Initial Swing (approx. from 60% to 73% of the gait cycle) is the first sub-phase of swing. It begins with toe off by the epsilateral limb (when the foot leaves the ground) and ends when the swinging foot is adjacent to the stance foot. The main goal of this sub-phase is the advancement of the limb. Mid-Swing (approx. from 73% to 87% of the gait cycle) begins when foot is adjacent to the stance foot and ends when the tibia of the swinging limb is vertical. The goal of this sub-phase is also limb advancement. Terminal-Swing (approx. from 87% to 100% of the gait cycle) is the last sub-phase of swing and the end of the gait cycle. This sub-phase begins when the tibia of the swinging leg is vertical and ends at initial contact or heel strike, when the foot touches the ground, beginning a new stance phase. The goals of this sub-phase are to complete limb advancement and to prepare the limb for stance.

3 A Neuro-Muscular-Skeletal Model of Human Gait

In [17], Geyer and Herr present “A Muscle-Reflex Gait Model That Encodes Principles of Legged Mechanics Produces Human Walking Dynamics and Muscle Activities”. This model is based on muscle reflexes and on the principle that locomotion requires little control if principles of legged mechanics and its dynamics are exploited in the model itself. The human model is represented with a trunk and two three-segmented legs, in which they applied 14 Hill-type muscles (7 on each leg) that can be compared directly to important muscles in locomotion. The basis for this model is the bipedal spring-mass model, based on the fact that, if the legs behave like springs in stance, walking and running emerge naturally as a consequence of the forward speed of the human model. Although this model is very simple to reproduce a human-like walking pattern, the displacement of the center of mass is similar to that in walking and running.

To achieve human-like locomotion, three main changes were done on top of the bipedal spring-mass model [17]. To begin with, each spring of the model was replaced with a three-segmented leg (composed by thigh, shank and foot) and muscles were added spanning the ankle and knee in order to achieve compliant stance behavior. Secondly, the point mass was replaced with a trunk and hip muscles for its balance control. In the end, swing leg control was added, allowing this model to enter in cyclic locomotion.

In the first step of their model development, in order to achieve compliant stance behavior, they added 4 muscles to each limb. To begin with, a Soleus muscle (SOL) and a Vasti muscle group (VAS) were added, represented in picture A of Figure 2. Both these muscles activities depend on positive force feedback $F^+$, a spinal reflex observed in cats and suggested in humans that can generate compliant behavior in neuromuscular legs. With the force feedback mechanism, the stimulation of a muscle is the sum of a pre-stimulation and a time-delayed (activation to contraction delay) gained force. For example, the function of the Soleus in stance is designed as:

$$ S_{SOL, stance} = S_{0,SOL} + G_{SOL} F_{SOL}(t - \Delta_{1TD}) $$

(3.1)

The $F^+$ of SOL and VAS, along with the segmentation of the limb, introduces a problem in which a large extension torque at the ankle can lead to knee hyperextension and
vice-versa. To counter this effect, the Gastrocnemius (GAS) and Tibialis Anterior (TA) muscles were added (picture C of Figure 2). The Gastrocnemius also uses F+ during stance, preventing knee hyperextension and generating an overall compliant leg behavior. The Tibialis Anterior, contrary to the other 3 muscles, uses positive length feedback L+, preventing ankle overextension when large torques are applied to the knee. However, it is necessary to add a negative force feedback F- in TA’s control, to avoid this muscle to act against the SOL if the active ankle extensors preserve the torque equilibrium between the knee and ankle. An inhibition at the VAS was also added if the knee extends beyond a 10° threshold.

Figure 2 - Sequence of modifications applied to the spring-mass model by Geyer and Herr in [17].

The second part of the model development was to replace the point mass with a trunk and add its balance control. Herr and Geyer hypothesized that one could accomplish such task activating the hip muscles proportionally to the trunk’s velocity and forward lean in the inertial system. They added to each leg a Gluteus muscle group (GLU) and a HFL (Hip Flexor Muscle) muscle group, both activated proportional to the trunk’s forward lean and angular velocity. Finally, they added a Hamstring muscle group (HAM) with stimulation similar to GLU (picture D of Figure 2). This muscle counters knee hyperextension resulting when a large hip torque is generated as the GLU pulls back the heavy trunk. Also, these 3 muscles stimulations are a function of the amount of body weight the leg bears, as the hip torques can only balance the trunk if the leg bears sufficient weight on the ground (picture E of Figure 2).

The third last step of the model, was to add swing leg control (picture F of Figure 2), allowing the model to enter in cyclic locomotion. In this step, it is assumed that the importance of each leg in stance is proportional to the amount of body weight that same leg bears. For that reason, they inhibit the F+ of the ipsilateral leg’s VAS in proportion to the weight the contralateral leg bears. This action is important as it enables the knee to flex while the ankle extends, thus pushing the leg forward and off the ground. There is also another swing stimulation mechanism, as the stimulation of the HFL is increased at the same time that the GLU’s is decreased by a fixed amount in double-support.

![Figure 3](image-url) - Control switching between stance and swing reflexes. The model’s feet has two sensors at the Heel and Ball that detect the ground and enable the transition between the two controllers. The stimulations $S_m$ to the different muscles depend on the mechanical input $M$.[17].

During swing, the leg has a ballistic motion with SOL, GAS and VAS muscles silent and only TA’s L+ active in order to provide foot clearance with the ground. The HFL is stimulated during swing with its stretch reflex $L+$, facilitating leg protraction during this phase. This muscle is also dependent on the trunk’s forward lean as this affects the required protraction speed. Another important feature for gait stability is to enforce leg retraction before landing. This latter mechanism is achieved with three muscle reflexes. First, they inhibit HFL’s L+ proportionally to the stretch that the HAM receives in swing. This L+ is necessary to compensate for the hip rotation that results when the passive knee rotates into full extension during leg protraction. The second and third reflexes are the $F+$ of GLU and $F+$ of HAM. Both reflexes function is to transfer part of the protraction momentum into leg lowering and retraction at the same time as the leg halts. Lastly, it is important to say that the switch between stance and swing control is achieved with sensors in the model’s feet that detect the ground (Figure 3).

4 Joints as springs

This work is based on the idea that joints can be approximately reduced to torsional springs working with feedback as was previously described. In the studies [22] and [23] the authors exploit the idea of a torsional spring working as an ankle with some defined stiffness. The way they do this is by analyzing the moment-angle graph of the ankle (Figure 4) during the gait cycle and the relationship between these two measurements.

Starting from initial contact, the ankle undergoes a controlled plantarflexion with the intention of lowering the
foot to the ground in a gentle way. After toe strike the ankle begins to dorsiflex (controlled dorsiflexion) and later in the stance phase starts to plantarflex (powered plantarflexion). This last plantarflexion mode of the ankle, that generates high torque and power, is most probably essential for the forward propulsion of the body. These studies analyze the behavior of the ankle in controlled plantarflexion and powered plantarflexion.

Figure 4 - Ankle’s angle-moment graph. The figure denotes the regression lines draw according to controlled-dorsi-flexion and powered plantar-flexion phases and the mean regression line ($\bar{k}$ defines the characteristic stiffness of the ankle as calculated in [22]) : a: Heel Contact; b: Foot Flat; c: Maximum ankle moment; d: Toe-Rocker; e: Terminal Stance; f: Swing.

One can conclude from observation of the ankle angle-moment graph that during the controlled dorsi-flexion and powered plantarflexion phases of the gait, the ankle seems to exhibit a linear behavior. In [22] the authors calculated the characteristic stiffness of the ankle as the mean between the slopes of the regression lines that passes through the two phases being analyzed in the angle-moment graph. In the development of this master thesis model, the ankle was modeled as a spring and so was necessary to define its initial stiffness, being these studies as an inspiration to such task. The knee and hip mechanism were designed in a similar way.

5 Model Development

Prior control work by Geyer’s and Herr’s [17] was developed in Matlab/Simulink environment and so, as this is the basis of the model developed in the context of this master thesis, we also used this tool. In [17] was also developed in SimMechanics a graphic representation of a human model in 2D (Figure 5), composed by the feet, shanks, thighs and a trunk. Such environment was also used in this master thesis work.

The ankle control strategy was mainly inspired in the function of the TA and SOL. In [17] the stimulation (S) computed to the TA’s and SOL’s MTUs depends if the leg which the muscles are attached is currently in stance or swing phase. The stance control is active providing at least one of two cases; the heel sensor detects that the foot is touching the ground or the ball sensor detects that the foot is touching the ground and its being produced a plantarflexion moment to the ankle joint. This last condition ensures that in the end of stance, when the ankle is at power plantarflexion stage, the control system switches to swing when its being developed a dorsiflexion moment. The ball and heel sensors are then a very important mechanism to make the transition between stance and swing controls systems.

Figure 5 - SimMechanics human model created in [17] and used in this work.

The stance ankle control is mainly composed of a force feedback mechanism inspired in TA and SOL function. This mechanism works essentially as a spring with some spring constant and reference position. After analyzing the moment-angle curve for the ankle, it was decided to divide the stance control system of the ankle in two main phases; controlled plantarflexion in the beginning of stance, from heel strike or initial contact to toe strike and from that point forward comprising both controlled dorsiflexion and powered plantarflexion, ending with toe off. Each of these two phases has its own spring constant and reference spring position.

The reference position from heel strike to toe strike is modeled as the angular position of the ankle in initial contact. In this way, after heel strike, the ankle passively plantarflexes as was previously described and is developed a counter dorsiflexion moment. The force feedback mechanism is not required in this first stage of stance. The spring stiffness is chosen as to appropriately reproduce the
moment-angle curve of the ankle during controlled plantarflexion.

During the second part of stance, divided as was previously described, the reference position of the virtual ankle's spring is chosen as the angular position of the ankle in the moment of the transition, i.e., toe strike. After this moment, the spring starts to accumulate energy as the force feedback mechanism is now active. This mechanism increases the ankle initial spring stiffness in this stage, and so the ankle’s slope of the moment-angle curve increases. During this stage, energy is generated in the ankle, as can be visualized by the area between the ankle’s moment-angle curve, being one of the main reasons for the suggestion the propulsive function of the ankle during walking, as was previously described. To guarantee the realistic function of the force feedback mechanism, is applied to force sensory information a delay called “Activation to Contraction Delay” that has to do with the time between the activation of a muscle and its contraction. This time-delay is extremely important in the control system. It would lead to an algebraic loop if one used the force feedback at the same time-step, because the system wouldn’t know what to calculate first. Also, the stiffness of the ankle during this stage only changes if the force feedback is positive, guaranteeing that the stiffness of the ankle can only increase and not decrease.

This main stance control present at the ankle is complemented with a control system based on the Gastrocnemius (GAS) muscle (Figure 6). As a muscle spanning two joints (the knee and the ankle), it has as main function to synchronize them. The contracting activity of GAS produces plantarflexion at the ankle and flexion of the knee.

Knowing that the knee has two phases of flexion and another two of extension throughout the gait cycle and after analyzing the knee’s moment-angle curve, it was decided as the reference position for this virtual spring the angular position of the knee at initial contact (around 5° in normal walking). In this way, one can visualize the knee stance flexion just after initial contact, followed by knee extension, which provides stability. The calculation of moment is also done using a force feedback mechanism that, like happened in the ankle’s control system (and for the reasons explained before), increase the stiffness of the spring as more force is applied into the joint. This mechanism requires force sensory information which is time-delayed.

As this is a two joint spring, it requires two reference positions, one for the ankle rotation and another for the knee rotation. These reference positions are taken as the angular positions of the joints in the moment of feet adjacent. As the muscles only produce compressive forces and not extensive ones, this control mechanism is only allowed to produce plantarflexion moment.

The chosen control for the ankle in swing is a simple torsional spring with both proportional and derivative part. We know from literature that the ankle is almost at a neutral position at initial contact, as explained in the previous chapters, and so we took the reference position of this virtual spring as zero degrees, with the ankle neither dorsiflexed nor plantarflexed. As the ankle is plantarflexed after toe off, it starts to dorsiflex reaching its neutral position soon after entering the swing phase. Added to this proportional behavior also present during the ankle’s stance control system, is the derivative behavior, in order to stabilize the movement that the ankle undergoes. The velocity of the ankle is sensed and multiplied by a derivative constant, which is subtracted to the proportional part – proportional derivative control.

Like the ankle’s control strategy developed in this work and the one developed in [17], the knee control strategy is also separated into two parts: stance and swing. The transition between these two control systems is made using the ball and heel sensors in both feet that are able to detect the ground. So, if at least one of these sensors detects that the foot is touching the ground, the knee control system jumps into stance control, otherwise goes to swing. The knee’s stance control system is divided in two parts, one related with the main function of the knee is stance, mainly related to the mono-articular muscle group VAS and another with a synchronizing mechanism between the ankle and knee, inspired in the function of the bi-articular GAS.

Figure 6 - Part of the ankle’s stance control inspired in the function of the Gastrocnemius (GAS) muscle.
mainly in the leg's ballistic motion. In this phase of the gait cycle, the knee takes as reference position its angular position at the moment of toe off. The moment force that is sent to the knee joint is then the multiplication of the displacement of the spring from its reference angular position by a constant (Kp of spring) that represents the spring's stiffness.

Figure 7 - Part of the knee's swing control system inspired in the function of the VAS muscle.

This part of the knee swing control system is supported by a virtual spring that mimics the function of the bi-articular HAM. This muscle group spans both the knee and hip joints, and is particularly important in the latter part of the gait cycle, from feet adjacent until initial contact or heel strike. Just as was modeled in [17], it is applied to this control system a force feedback mechanism that increases the initial stiffness of this virtual spring. There is however the need to allow only positive values of the moment force to be fed back in order to insure that the stiffness of the spring increases and not decreases.

As this is a bi-articular spring, there is the need to have two reference positions, one used in the calculation of the knee angular displacement and another in the calculation of the hip angular displacement. The reference angular position of the knee is taken as the angular position of this joint in the moment of feet adjacent and the reference angular position for the hip is a constant value of zero degrees. So, the function of this muscle is activated by extension of the knee and flexion of the hip, both lengthening this virtual spring that produces a compressive force in order to counter that tendency, producing knee flexion in this case.

As the previous control systems strategies, we mainly divided the hip control systems into two phases, stance and swing. The hip stance control system is composed by two sub-control systems, one that is related to the maintenance of the trunk's equilibrium (Figure 8), and another that comprises a spring with a force feedback mechanism.

The trunk’s equilibrium mechanism is very similar to the one implemented in [17], except for the fact that here we don’t use any muscles to create the desired moment force output. In this case we developed a virtual spring in the trunk with a reference angular position of 10° (as seen in [17], a little bit of the trunk’s forward lean, helps the human model to walk better). In the development of this sub-control systems, we must first recognize if the model is currently in stance because we assume that each leg is only capable of controlling the trunk’s equilibrium if the leg bears some body weight in the ground, i.e., is in stance phase.

For means of simplification, we didn’t model this mechanism proportional to the amount of body weight the leg bears. As can be visualized in the upper position of Figure 8, the model is considered to be in stance phase if at least one of the two ground sensors (ball and heel) detects the ground. In the case where none of these sensors return 1, the output of the “OR” system is 0 and the output of all this sub-control system is also 0. Just like we’ve done in the ankle’s Swing control system, here we present a derivative response related to the trunk’s velocity, as Geyer and Herr present in their muscles GLU and HFL control strategy.

Figure 8 - Hip Stance control part that has as main function the maintenance of trunk’s equilibrium.

The second part of the hip’s stance control system comprises a spring with a force feedback mechanism, resembling the control strategies used in the GLU and HFL muscles. This spring has as reference position 0° and so, for example, if the hip flexes, it's generated a counter extension moment.

The first part of the hip's swing control system is very simple, being composed of a simple spring without any feedback mechanism. As previously mentioned, this system is activated once there is no sensor in the foot which can detect the ground. This simple spring modeling is due to the fact that in swing, we take advantage on gravity and the ballistic motion of the leg. In swing, the hip’s control system is supported by a bi-articular spring that resembles the action of the bi-articular muscle Hamstrings (HAM) which spans both the knee and the hip. As was said
in the previous section (knee control strategy) the activity of this muscle group is particularly important in latter swing.

Just like in the knee, there is a force feedback mechanism (which must be always positive) that increases the initial spring stiffness (Kp0 bi-articular). As this is a two-joint spring, there must be two reference positions for the spring, one for the knee and another for the hip. These reference angular positions are the same as the ones utilized before in the knee control strategy.

5 Optimization

Throughout the development of this work’s biomechanical model, it was necessary to add some unknown variables which represent important aspects of the model. These variables represent the joint’s initial stiffness, the bi-articular spring stiffness, the force feedback gains and derivative constants. In total, there is 21 parameters that have to be tuned and optimized in order to achieve a stable human walking pattern.

The parameter tuning procedure was partly made “by hand” (analyzing the moment-angle curves for normal walking and correcting the parameters by observation of the resulting gait pattern) and partly achieved throughout an optimization procedure using Simulink Response Optimization (Figure 9). This procedure requires a set with the parameters initial values and a cost function that analyses the resulting gait pattern at each iteration and computes a value that is used to compute the upgraded values in the next iteration. Typically, the optimization procedure tries to maximize the cost function (for example, the number of steps given by the human model or the distance travelled) or minimize it (for example, the energy expended).

Inspired in some works such as [17], [24] and [25] we used a number of different approaches to try to optimize this set of parameters and acquire the best possible human gait pattern.

The first, and more basic quantity used in the cost function was the simulation duration. What we thought was that a good gait pattern simulation should take more time to simulate before the system stops (due to the stopping criteria mentioned above). This quantity was then transformed in a point giving system, in which the cost function output value is increased as the simulation duration is greater.

We then added a systems that increase the output value of the cost function as the number of steps given by the human model increases, as well as the distance travelled. That work was mainly inspired in the optimization procedure of [17]. On top of that, we also rewarded step length consistency. In this way, the model has the tendency to not only increase its simulation duration but also to increase the distance travelled by giving steps.

We wanted the human model to walk in a more natural, homogenous and smooth way, and so we used the quantities of the waist vertical position and forward velocity to measure the deviation from ideality (beyond the normal fluctuations in these measurements). Also, it was developed a system mainly based in [24] and [25], which measures the moments involved in the ankle, knee and hip and minimizes that quantity.

We also added to this cost function a term that as to do with knee hyperextension which decreases the output value if the knees angle goes below one degree. Also very important to note was the term added to impose a certain angle to the trunk’s angular position (in this case was chosen zero degrees). This term decreases the cost function’s output value by the amount of the quadratic deviation from this equilibrium position over time.

Last but not least, it was developed more complex system of point giving that as to do with the precisely way in which humans walk. In this system, for example, the cost function’s output value is penalized if the front foot starts to leave the ground, for that is not normal in the context of the human walking pattern.

6 Simulated walking pattern

Comparing [17] with the resulted walking pattern shown in Figure 10, the approach of the right leg to the floor is very similar in both situations. However, one can see that in our work, the knee is much more flexed at the time of initial contact when compared with [17]. The gait determinant related to the knee flexion in early stance is clearly represented. One can also visualize the controlled plantar-flexion motion in early stance, in which the foot is lowered “gently to the ground. One can also visualize the heel off by the back foot and beginning of the controlled dorsiflexion phase by the front foot. The action of the Gastrocnemius in late stance can also be observed in Figure 10. Also, after the leg enters swing phase, the hip passes from extension into flexion and the ankle dorsiflexes to its neutral position. From that point forward, the Hamstrings muscle in [17], and
the bi-articular spring that spans the hip and knee joints in our model starts to work and the flexion motion of the hip is decelerated. The leg is pushed against the ground and the heel of the leg that is currently in swing touches the ground, initiating another double-support phase of the gait cycle.

In Figure 11, it’s represented the ankle’s angle-moment curve obtained with our model. It is only illustrated the portion of the curve until the model falls. The curve is similar in function to the one obtain by [17] and the one derived from literature. However, there is a great discrepancy from literature when one analyses the angle-moment curves of the knee and hip from our model. The model is able to take two steps before falling, but because the optimization procedure is not completed, there is still further work to do in improving this model to achieve a stable gait. First, one should review the control strategy applied to it and study the importance of the bi-articular springs for an overall stable gait. Then, a great work as to be performed in optimizing the control parameters discussed earlier. It has also to be explored the Matlab/Simulink configurations in the optimization procedure, to apply the best algorithm to perform such complex task.

The resulting simulated human gait pattern

![Figure 10 - The resulting simulated human gait pattern](image)

Unfortunately, the knee and hip angle-moment graphs are not particularly well in the context of the human gait characteristics. In the next chapter is discussed the possible further developments that can increase the potential of this model.

7 Conclusion and Future Work

Giving the fact that current state-of-the-art approaches to generate human-like walking have great disadvantages such as the requirement in pre-defined joint trajectories, in this work a mechanical reflex model to generate human-like walking in a 2D simulation was developed. Mainly inspired in [17], “A muscle-reflex model that encodes principles of legged mechanics produces human walking dynamics and muscle activities”, in this work the gait cycle and its characteristics, force feedback control and the functionality of joints as springs were analysed.

Geyer and Herr developed a neuro-muscular model that comprises seven Hill-type muscles and generates an overall compliant leg behavior, according to certain muscle reflexes. In this work, we took all the muscles of that model and substituted by individual angular springs applied to each of the principal joints (ankle, knee and hip) in sagittal-plane walking. On top of that control strategy a force feedback mechanism was added that enables the springs to not only accumulate energy as they compress but also to generate energy in the process. Following the literature [11,25], in our model two springs spanning two joints that resemble the function of the Gastrocnemius and Hamstrings and help to synchronize the joint’s activities, were also added.

In the overall model, there are 21 parameters which have to be optimized in order for the human model to achieve a stable walking pattern. That work was one of the greatest challenges of this work. In this way, we tried different approaches to the optimization procedure were taken such as maximizing the number of steps given by the model and the minimizing the energy consumption (by minimizing the sum of the moments in the joints).

The resulting walking pattern resembles much of the characteristics of the gait cycle such as controlled plantar-flexion in early stance and the ankle’s angle-moment curve. The model was able to give two steps before falling down. The complexity of the model and the number of parameters to optimize, opens doors to further work optimizing spring stiffnesses and force feedback gains.

One of the main further works should be in understanding the contribution of the different terms involved with the cost-function in the optimization procedure. That work would bring many insights about the important features that enable humans to walk efficiently. Another important future work would be to fully understand the contribution of the force feedback control mechanism for human walking, as well as the bi-articular springs.

After the fully development of that model and the achievement of a stable walking pattern in 2D simulation, the next step would be to improve the model into 3D environment, adding joint motion in the frontal and transverse planes. One has to have in mind that a lot of
energy efficiency in human walking comes from the motion from these two planes, which is represented in the gait determinants. The optimization cost-function has to be reformulated to account for this new environment.

Providing that the human-model is now able to walk in a 3D environment, it is necessary to increase its robustness, providing different conditions and environments to the model’s simulation and optimize the results once again. This work can be performed in resemblance to works such as [24] and [25].

In the end, after going through all the above steps, one could implement this controller to a humanoid robot to achieve a reflex and human-like type of walking. This controller would be able to perform calculations as the robot or the person is walking, changing its function according to perturbations and the environment.

Bibliography


