Development of a musculotendon model within the framework of multibody systems dynamics

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Abstract

The main aim of this study is the development of a musculotendon model and its implementation in a multibody dynamics code with natural coordinates already existent. This model is a Hill-type muscle model assembled in series with a tendon model and it intends to simulate the dynamic contraction of the musculotendon unit in order to analyze the interaction between the muscle and the tendon and its influence in the movement, providing values of forces, lengths and velocities of both.

To study the mechanics of the human movement, the musculotendon model was integrated in the code in a forward dynamics perspective that allows for the determination of the system motion for a given set of muscle activations, and also in an inverse dynamics perspective that allows the calculation of the muscle activations, and consequently the musculotendon forces, that are needed to execute a presented movement. The inclusion of musculotendon actuators in the system leads to a problem of muscle redundancy that is solved through optimization methodologies.

A biomechanical model of the whole body, where the head, arms and torso are consider as a single rigid body, and in which the muscle apparatus of the lower limb is constituted by forty-three muscle was developed to analyze the musculotendon model. Experimental data of gait, running and jumping were acquired in a biomechanics laboratory. The results showed that the tendon has a significant influence in certain muscle groups along the movements analyzed. The results are compared with the muscle model and discussed, as well as, some conclusions are taken together with possible future developments.

Keywords: Multibody dynamics, Inverse and Forward Dynamic, Musculotendon Contraction Dynamics, Musculotendon Force, Biomechanical Model

1 Introduction

Along recent years, research and development studies in human movement are quickly progressing due to the activities of scientists in different areas like biomechanics, health science, sports science, prosthetics orthotics, among others. Scientific research in this field allows for a better understanding of normal and abnormal human movement characteristics and the development of new and innovative ways to increase the quality of live of people and reduce the health care costs. The recognition and evaluation of movement abnormalities has been performed through the analysis of gait and other human movements like running and jumping (Jal ´on & Bayo, 1994). These human movements, in recent times, are considered as a routine procedure in many diagnostic and rehabilitation procedures that include applications like: the design of a rehabilitation program, the planning and evaluation of surgical outcomes and the improvement of sports techniques and performance (Jal ´on & Bayo, 1994).

The analysis of the human movement depends greatly on the use of multibody formulations as kinematic or dynamics tools. The developments occurred in multibody dynamics allowed it to become an important tool in the design, promote and simulation of articulated mechanical systems in great detail (Amirouche, 2006).

The movement of the human body is mainly of the responsibility of the muscle. The central nervous system (CNS) excites muscles causing the development of forces that are transmitted by tendons to the skeleton, causing its movement. Muscles and tendons are therefore an interface between the CNS and the articulated body segments (Zajac, 1989) and the study of such interface if of great interest for the scientific and medical community as it allows for a better understand how a specific muscle contribute to a given movement (Hoy et al, 1990), to improve the applications described above and to develop prosthetic, orthotic designed and functional neuromuscular stimulation systems to restore lost or impaired motor function (Zajac, 1989).

The function of this interface, the musculotendon unit, can be affected by the elastic properties of the tendon allowing a dynamic interaction between muscle and tendon that will influence the force transmission, energy store and the control of joint position and movement accuracy (Magnusson et al, 2008). Therefore, the development of non-invasive methods based on musculoskeletal modelling and computer simulations to study the interaction between the muscle and the tendon and their influence on the movement are very important in different fields of study.

To define the contraction properties of muscles, several mathematical models are developed, standing out the ones proposed by Hill and Huxley (Pandy,2001; Vilimek, 2007; Salinas-Aliva et al, 2009). The Huxley-type model, derived from the fundamental structure of muscle, estimates the forces in cross-bridges which makes the analyze very complex. The muscle dynamics are defined by multiple differential equations that have to be numerically integrated. Therefore, these models are computationally time-consuming when used for modelling forces in systems with multiple muscles. The Hill-type model is the one that is more often used for many researchers because, mainly, the dynamics are governed by one differential equation per muscle, making modelling computationally viable (Buchanan et al, 2006; Millard et al, 2013).

The crucial goal of this thesis is to implement a musculotendon model that takes into account the influence of tendon in muscle contraction. This model is adapted from the work developed by Zajac (Zajac, 1989) and it is able to determine the tendon and muscle force developed in a given movement, as well as, the length and velocity variation of both. The musculotendon model considers a mechanical Hill type model, where the force-length-velocity relationship, the pennation angle and the muscle activation are accounted, together with, the elastic properties of tendon.

Also, this work aims to incorporate the musculotendon model in APOLLO (Silva, 2003), a program of 3D multibody system dynamics analysis with natural coordinates, allowing the inclusion of the tendon characteristics in the biomechanical system.

A Biomechanical model was created to study the influence of the tendon in the musculotendon unit when the model is subjected to activities like walking, running and jumping.

2 Musculotendon System Modelling

The physiological musculotendon behavior that begins with a neural activation signal and ends with the muscle contraction (Silva, 2003), is studied in order to understand the dynamics of muscle tissue.

The dynamics of muscle tissue can be, therefore, divided into activation dynamics and contraction dynamics (Zajac, 1989), as represented in the next figure (Figure 3.1). The neural excitation u(t) acts through the activation dynamics to create the muscle activation a(t). This activation will give energy to the muscle cross-bridges and muscle force is developed, through musculotendon contraction dynamics.

Musculotendon contraction dynamics corresponds to the transformation of muscle activation in to musculotendon force.

The Hill-type model, presented in Figure 3.3, was used in this work to describe the dynamics of contraction since it considers the mechanical properties of the muscle and tendon

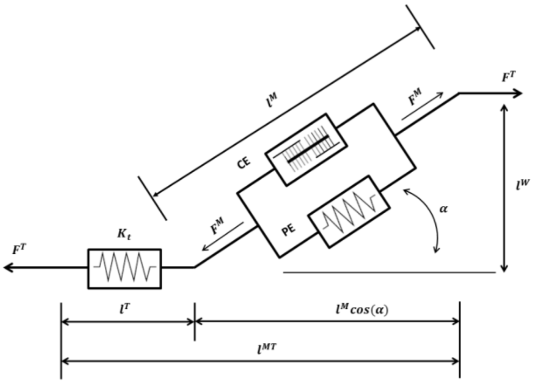

Figure 1: Mechanical Musculotendon Model that describes the musculotendon contraction dynamics.

The musculotendon length, that is represented by l^{MT}, results from the sum of tendon length l^T and muscle fibers length l^M taking into to account the pennation angle, as represented in Equation 1.

$$l^{MT} = l^T + l^M \cos(\alpha) \quad (1)$$

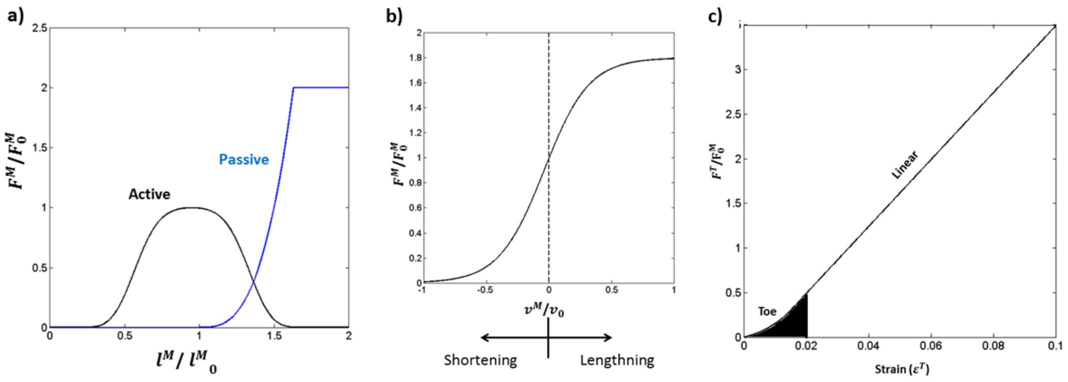

Figure 2: a)Active and Passive Muscle Force-length relation. b). Force-velocity Muscle Curve. c) Force-Strain Tendon Curve

The tendon is defined as a spring-like elastic component with a constant stiffness K_t that depends on its elastic properties, placed in series with the contractile component (Lorenz & Campello).

But its turn, the muscle is represented by a contractile element (CE) in parallel with passive one (PE). This element produces a force that depends on the force-length-velocity relation of the muscle and on the activation level. Regarding the PE, which is used to simulate the elastic properties of the muscle (i.e., the different levels of collagenous tissue) (Silva, 2003), generates a force that depends only on the muscle length. The sum of the forces generated in these two components, represents the resultant muscle force (Equation 2).

$$F^M = F^M_{CE} + F^M_{PE} = \frac{fl(l^M)fv(v^M)}{F^M_0} a^M + F^M_{PE}(l^M)($$

As stated above, the force produced by the muscle is transmitted to the skeleton by the tendon, so the force that is exerted by the tendon, or musculotendon unit, is the one responsible for the movement. The tendon force depends on the pennation angle between muscle and tendon, as expressed in Equation 3.

$$F^{MT} = F^T = F^M \cos(\alpha) \quad (3)$$

The musculotendon model was implemented as seen in Figure 3.

The peak isometric active force (F_0^M), the optimal muscle fiber length (l_0^M), the optimal fiber pennationangle (α_0), the maximum shortening velocity (v_0) and the tendon slack length (l_s^T) are characteristic parameters of each musculotendon unit that are needed and their values were obtained from Delp (Delp et al, 1990).

As represented in Equation 4 the contraction dynamics of the musculotendon is characterized by a first order differential equation (Martin and Schovanec). The derivative of the musculotendon force will be the model result that allows the determination of the force for the next time step through its numerical integration. Hence:

$$\frac{\partial \tilde{F}_a^{MT}}{\partial t} = K_t[v^{MT} - \frac{v^M}{\cos(\alpha)}] \quad (4)$$

Where K_t is the tendon stiffness calculated according to Zajac (Zajac, 1989) as $K_t = 30/\tilde{l}_s^T$, v^{MT} and v^M is the musculotendon and muscle velocity respectively and α is the pennation angle.

The musculotendon length (l^{MT}), velocity (v^{MT}) and force ($\tilde{F}_a^{MT} = \tilde{F}_a^T$) are the model input variables.

The musculotendon length (l^{MT}) is the distance between the origin and insertion of the muscle (DeWoody et al, 1998) and it is determined by the sum of the lengths of the line segments that define the muscle path (Delp & Loan, 1995). The musculotendon velocity (v^{MT}) is determined by the sum of the velocities of the line segments along the muscle (Salinas-Aliva, 2009).

Looking to Equation 4, the derivative of the musculotendon force depends on tendon stiffness and tendon velocity, which is represent by Equation 5.

$$v^T = v^{MT} - \frac{v^M}{\cos(\alpha)} \quad (5)$$

Once the muculotendon force is already known, the muscle velocity v^M and the pennation angle α must be determined to calculate the tendon velocity. The steps described below and seen in Figure 3 must be taken into account.

1st Step – Determination of tendon length

The normalization of tendon slack length $\tilde{l}_s^T$ by optimal fiber length l_0^M is denoted by l_s^T and defines the compliance of the tendon, once the tendon elasticity is proportional to l_s^T (Zajac, 1989). Thus the tendon is considered.

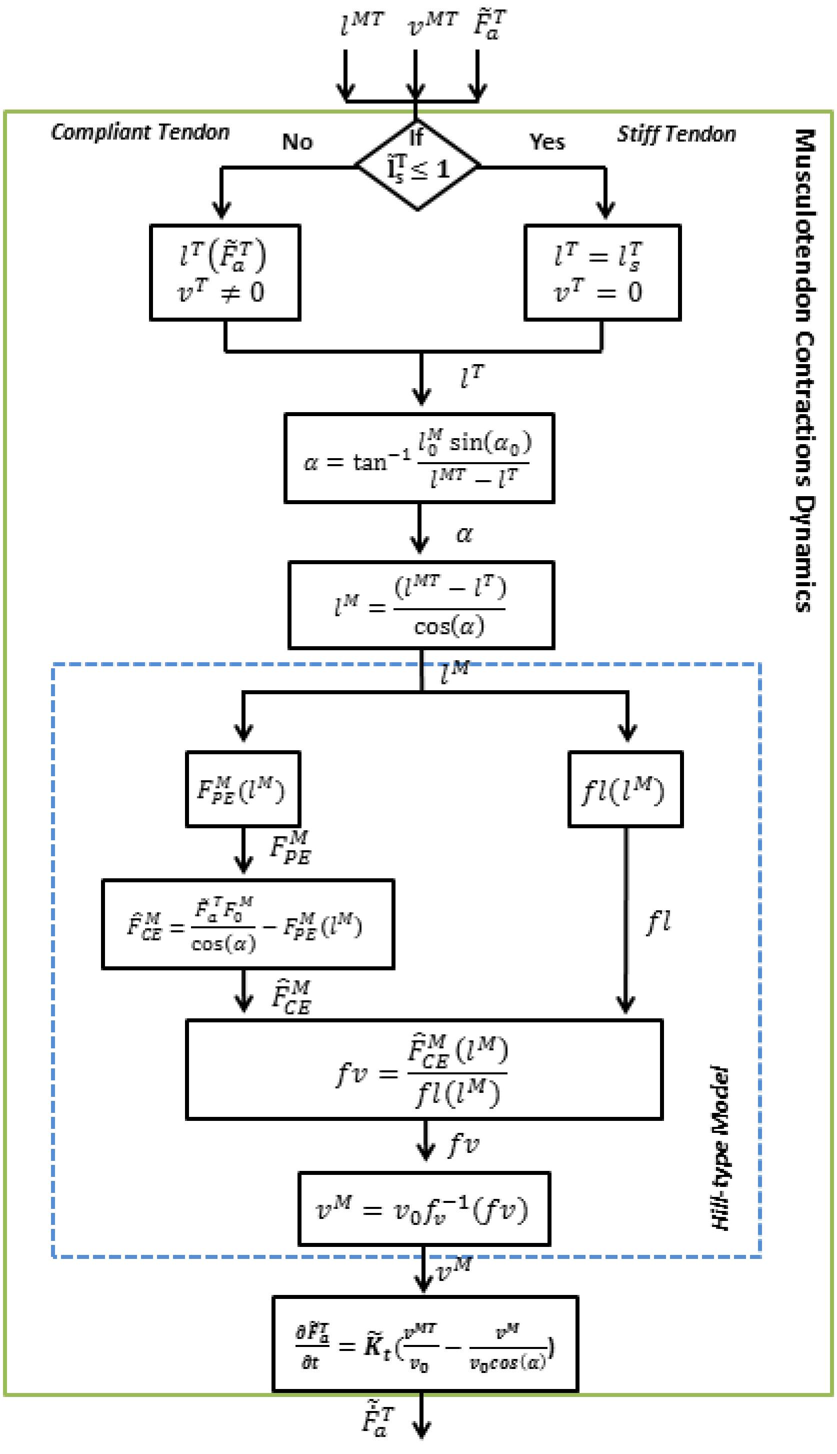

Figure 3: Musculotendon Model.

compliant when l_s^Ts is higher than 1, and stiff if otherwise

A tendon's stiff ($\tilde{l}_s^T \leq 1$) will be treated as inextensible which implies that is length does not change over time, i.e., it is equal to the slack length, and as so the tendon velocity is zero.

On the other hand, a compliant tendon will be described by the inverse force-strain curve (Figure 2 c) (Zajac, 1989) that could be represented as follow (Equation 6) (Martin & Schovanec).

$$l^T(\tilde{F}_a^T) = \begin{cases} l_s^T\left(1 + \frac{\ln\left(\frac{\tilde{F}_a^T}{0.10377} + 1\right)}{91}\right), 0 \leq \tilde{F}_a^T \leq 0.3086 \\ l_s^T\left(1 + \frac{\tilde{F}_a^T + 0.26029}{37.526}\right), 0.3086 \leq \tilde{F}_a^T \end{cases} \quad (6)$$

2^{st} Step – Determination of muscle length

Before determining the muscle length, the pennation angle must be calculated. Once the muscle thicknness (l^w) is constant, this angle can be determined through the following equality:

$$l^w = l_0^M \sin(\alpha_0) = l^M \sin(\alpha) \leftrightarrow \alpha = \sin^{-1}\left(\frac{l_0^M \sin(\alpha_0)}{l^M}\right) \qquad (7)$$

Using Equation 1 and 7, the expression that allows the calculation of the angle between the muscle fibers and the tendon is obtained:.

$$\alpha = \tan^{-1}\left(\frac{l_0^M \sin(\alpha_0)}{l^{MT} - l^T}\right) \qquad (8)$$

The muscle length can now be calculated by solving the Equation 1 as shown in Figure 3.

3^{st} Step – Determination of Passive Muscle Force and of Force-length

The passive force and active force ($fl(l^M)$) are governed by Equations 9 and 10 respectively (Silva, 2003).

$$F_{PE}(l^M) = \begin{cases} 0, & l_0^M > l^M(t) \\ 8\frac{F_0^M}{(l_0^M)^3}(l^M - l_0^M)^3, & 1.63 l_o^M \geq l^M(t) \geq l_0^M \\ 2F_0^M & l^M(t) > 1.63 l_o^M \end{cases} \qquad (9)$$

$$fl(l^M) = F_0^M \exp[-[\left[\frac{9}{4}\left(\frac{l^M(t)}{l_0^M} - \frac{19}{20}\right)\right]^4 - \frac{1}{4}\left[-\frac{9}{4}\left(\frac{l^M(t)}{l_0^M} - \frac{19}{20}\right)\right]^2] \qquad (10)$$

4^{st} Step – Determination of the Force-velocity (fv)

Through the Hill-type model equations, an expression for the muscle velocity is obtained. Considering the relation expressed in Equation 2 and Equation 3, an equation in order v^M is achieved (Equation 11).

$$v^M = v_0 f_v^{-1}\left(\frac{\frac{\tilde{F}_a^T F_0^M}{\cos(\alpha)} - F_{PE}(l^M)}{fl(l^M)}\right) \qquad (11)$$

So the force-velocity is given by:

$$fv(v^M) = \frac{\frac{\tilde{F}_a^T F_0^M}{\cos(\alpha)} - F_{PE}(l^M)}{fl(l^M)} = \frac{\hat{F}_{CE}^M(l^M)}{fl(l^M)} \qquad (12)$$

Where $\hat{F}_{CE}^M$ is the maximum available contractile force.

In the cases where a singularity is present $fl(l^M) \to 0$, the condition $fl(l^M) > 0.1F_0^M$was considered in order to maintain f_v between physiological values (Millard et al, 2013).

5^{st} Step – Determination of the Muscle-velocity (fv)

The inverse of the force-velocity muscle relation (Anderson, 2007), represented in Figure 2b), which allows the calculation of the muscle velocity according to the force calculated in Equation 12 will be expressed by as:.

$$v^M = -v_0(0.18 \log\left(\frac{\frac{fv}{F_0^M}}{-\frac{fv}{F_0^M} + 1.8}\right) + 0.04) \qquad (13)$$

3. Multibody Dynamics

The analysis of the movement in this study was carried in the framework of a multibody dynamics analysis with natural coordinates. An existing Fortran code called APOLLO was adapted in order to integrate the musculotendon model.

3.1Integration of Musculotendon Model within APOLLO

3.1.1 Inverse Dynamic Analysis

In the flowchart present in Figure 4 a) it is possible to observe the integration of the musculotendom model in an inverse dynamics analysis.

This type of analysis aims to determine the internal and external forces developed in/by the multibody system. The solution of this problem requires an optimization technique to resolve the EOM in order to the unknowns, the Lagrange multipliers and the muscle activations.

For each time step the vector that contains the position, velocity and acceleration of each segment are known. Otherwise, the fully activated musculotendon force is determined through numerical integration ensuring this value always for the next time step. This force will be essential, together with musculotendon length and velocity, to obtain the the maximum contractile force, the passive force and the pennation angle through the musculotendon model. These values will integrate the EOM.

After the muscle activation is determined, the musculotendon and muscle force of the current time step is obtained as shown in 'Update Forces' section (Figure 4 a)).

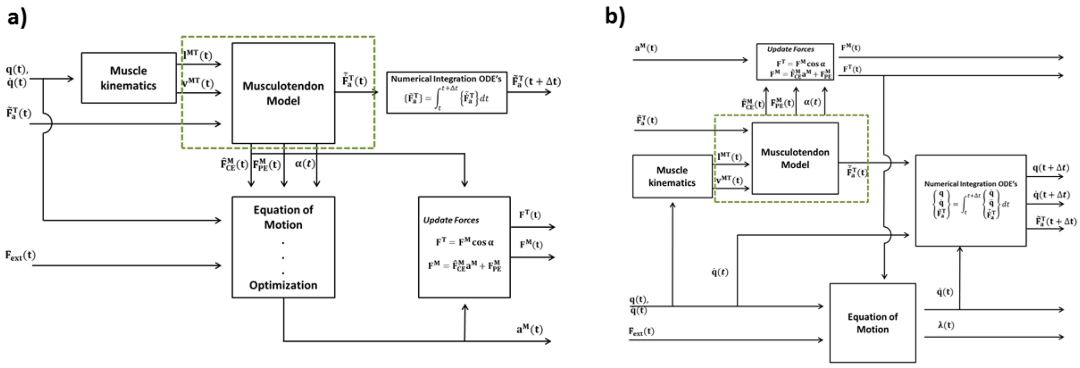

Figure 4: a) Inverse Dynamic Analysis. b) Forward Dynamic Analysis.

3.1.2 Forward Dynamic Analysis

The flowchart presented in Figure 4 b) shows how the musculotendon model was integrated in the forward dynamic analysis.

Unlike what happens with the previous analysis, a forward analysis aims to obtain the movement of a multibody system resulting from the application of external forces. From the beginning the muscle activations are known and the positions, velocities and accelerations vectors must be determined.

Through musculotendon model also the maximum contractile force, the passive force and the pennation angle are determined. These variables together with the muscle activation allow the calculation of the musculotendon forces that will be integrated in the EOM. As mentioned before, through EOM the Lagrange multipliers and the acceleration vectors are obtained.

As the muculotendon force is controlled by a differential equation, the value of the fully activated musculotendon force derivative is assembled to the vector $\dot{y}$ to be integrated with the velocities and acceleration variables. This result in a vector y that contains the positions, velocities and fully activated musculotendon forces for the next time step.

4 Biomechanical Model

The main propose of this work is to develop a model that allow the study of the main kinematic and dynamic patterns during different types of human activities like walking, running and jumping. Due to its complex kinematic structure, it can be used to simulate human movements in forward and inverse dynamics analysis.

This Biomechanical Model (Figure 5) is defined by 20 rigid bodies and 29 unit vectors and 27 degrees-of-freedom. And features 43 muscles in the lower extremity (Silva, 2003).

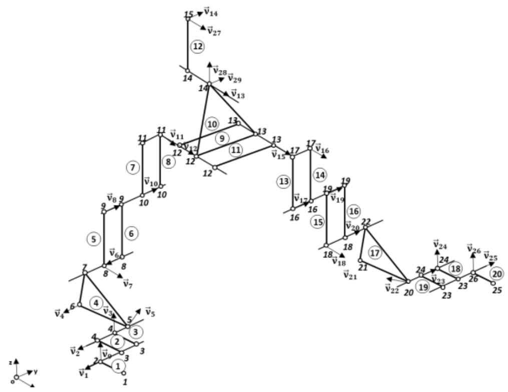

Figure 5: Biomechanical Model

5 Results and Discussion

A kinematic data acquisition was performed using a markers protocol at the Lisbon Biomechanics Laboratory (LBL) in *Instituto Superior Técnico.* In this protocol, to tracking the movement, 54 markers were placed at specific anatomical points of interest for this study. Markers placement was based on the protocol developed by Malaquias (Malaquias, 2013) and Gonçalves (Gonçalves, 2010) work.

The data acquired in the laboratory was processed in order to perform a kinematic and kinetic analysis. The muscles were integrated and the results were obtained.

In this chapter, the results will be present and analyze in order to realize the influence of each muscle in the different phases of the movement. Also, the results obtained through the muscle Hill model developed by Pereira (Pereira, 2009) will be display to compare and understand the influence of the tendon in the movement.

The results will be analyzed mainly in the sagittal plane and the muscle will be grouped according to their most important function. Quadriceps femoris composed by the vastus medialis (VM), vastus intermedius (VI), vastus lateralis (VL) and rectus femoris (RF); Hamstrings composed by the semitendinosus (ST), semimembranosus (SM), biceps femoris (long (BFLH)) and short head (BFSH)); triceps surae, composed by soleus and the two head of the gastrocnemius (medial (GM) and lateral (GL)); ilipsoas composed by iliacus (I) and psoas; ankle plantarflexors composed by the triceps surae, tibialis posterior (TP), peroneus brevis (PB) and longus (PL); and the ankle dorsiflexors composed by the tibialis anterior (TA). Some muscle will not be included in the results due to almost no activity present in the movement analyzed.

5.1. Gait Analysis

The gait cycle, movement of the lower limbs during walking, consist of one cycle of swing and stance phase by one limb, in this case the right one (limb represented in Figure 6 with red muscles). The stance phase consists in the period where the right limb is in contact with the ground, and the swing phase where is not (Figure 6).

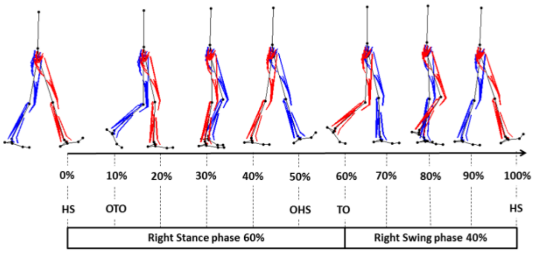

Figure 6: Gait Cycle

Taking into account the different phases of gait and analyzing the figures above, it is possible to identify the contribution of each muscle for the movement observed. The analysis starts with the HS where the dorsiflexors, mainly the Tibialis Anterior (TA) and the hamstrings are activated. This initial activity of the dorsiflexors aims to control the landing of the calcaneus on the surface due to the weight admitted in HS. The contraction present in the hamstring allows the weight transfers from a single support to a double support, giving stability to the body. Also, in this initial phase, the plantarflexors are activated. Tibialis posterior, gastrocnemius (Figure 7) present a small activation to guarantee joint stabilization.

When the OTO phase occurs, the body weight is totally transferred to the right leg. Muscle activity and consequently musculotendon force of the triceps surae (Figure 7) was detected to control the rotation of the leg around the ankle and the stabilization of the ankle joint. Even before 30%, the moment when the left foot passes a point below the left hip joint, muscle activation and consequently musculotendon force are observed in the hamstrings and in the quadriceps femoris. This muscular activity will promote the stabilization of the knee joint, once the right leg is supporting all the body weight.

After mid stance phase, the ankle plantarflexors develop significantly musculotendon force. This force is mainly realized by the triceps surae (Figure 7), and it is responsible to bring the body forward. Besides, a contraction of the quadriceps femoris (rectus femoris and vastus) occurs close to the TO. This muscle activity is responsible to the extension of the knee and to ensure this extension while the impulse given by the foot is transmitted to the hip, pelvis and trunk, when there is a forward inclination (Silva, 2003). The stance phase ends when occurs the TO, moment when the ankle plantarflexion force decrease.

A swing phase begins now and muscle activation of the tibialis anterior is observed. This activity will maintain the foot in a stable position and prepare it to the next HS. During this phase, the flexion of the hip joint must occur. Activations of the hip flexors, Psoas and iliacus, are activated to produce that movement.

At the end of the swing phase, occur an increase activity of the hamstrings, mainly, the biceps femoris. This muscle presents activity to decelerate the lower leg and foot, until the nearly full extension of the knee, to prepare the leg for the new HS.

The results obtained in both models must be compared to understand the influence of the tendon in gait. The principal difference present is the presence of passive force in the muscle model. According to the force-length muscle relationship this means that the muscle length is longer compared with the isometric muscle length.

In the musculotendon model, the variation of tendon length enables the muscle to work at a more optimal velocity v_0^M and length l_0^M . When muscle working in this zone, the activation needed to developed the force required will be lower than the present in the muscle model.

For example, in Figure 7, where are represented the results of triceps surae significant differences are detected. If the muscle force obtained with the musculotendon model were compared with the contractile muscle force obtained (dash lines with the respective color) the differences are too small, which means that the passive force is almost nonexistent.

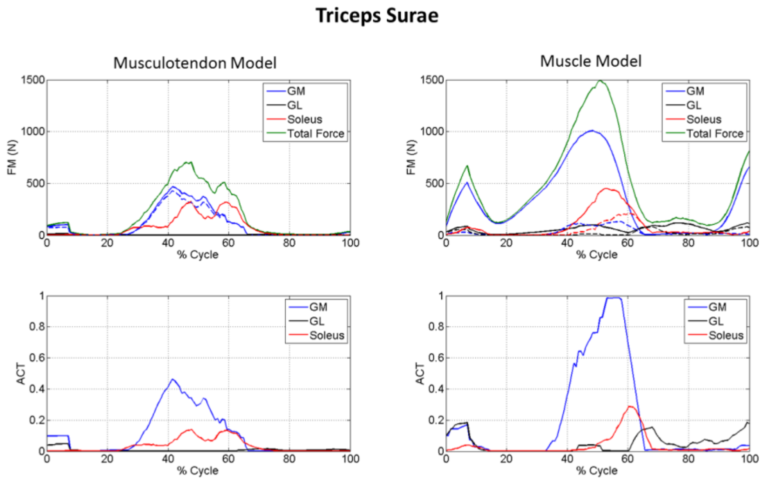

Figure 7: Muscle Force and Muscle Activation of the triceps surae obtained in the musculotendon model and in the muscle model in gait analysis

Otherwise, when analyze the results of the muscle model, although the muscle force produce by those muscle are greater, most of them correspond to passive force. The muscle is then working in a non-favorable zone, when it needs more activation to obtain the contractile force to archive to the moment desired.

5.2. Run Analysis

Running, like walking, is an activity characterized by a cycle which repeats over time. In this analyses the cycle beginning and end with HS. Unlike what happen in gait cycle, running only is divided into a support phase, when a foot is on the ground and a recovery phase in which both feet are off ground.

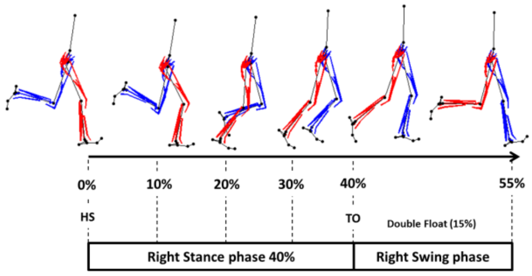

Figure 8: Running Cycle

The results present in this work are referent to the support and recovery phase of the right limb.

This only represent 55% of the running cycle (Figure 8), once is not observable opposite heel strike (OHS) and the final HS of the right limb.

Through the results obtained, high levels of muscle activation and consequently the muscle forces are present in the stance phase, as expected. Between the 40% and 55%, double float phase, there is a decrease of those levels.

After the HS and before the mid stance phase occurs a period of absorption. In this period the body's center of mass decrease and its velocity decelerates horizontally (Hamner & Delp, 2010). Strong muscle activity is present in the knee (quadriceps femoris) and hip extensors (gluteus maximus and hamstring). Quadriceps femoris are activated in this period to prepare the limb for the ground contact and to absorb the shock of the impact. The hip extensors, gluteus maximus and the hamstring, mainly the semimembranosus, also present high levels of activations to contribute to the body support. The activation of these muscles, together with the triceps surae, will provide the acceleration of the body vertically until the mid stance.

Forward propulsion of the body in this phase is provided initially by hip extensors discussed above, and after mid stance by the ankle plantarflexors (Hamner & Delp, 2010), soleus, gastrocnemius (GM, GL) (Figure 9) and peroneus longus.

In the beginning of the swing phase tibialis anterior, iliacus and psoas are activated to prepare the foot for the next HS, and to flex the hip joint, respectively.

During double float phase, occurs an increase activity of the hamstrings to decelerate the lower leg and foot and preparing to the new HS.

Comparing the two models, some differences are also found. In triceps surae (Figure 9), the presence of a tendon improves the results. The muscles work in optimal zone

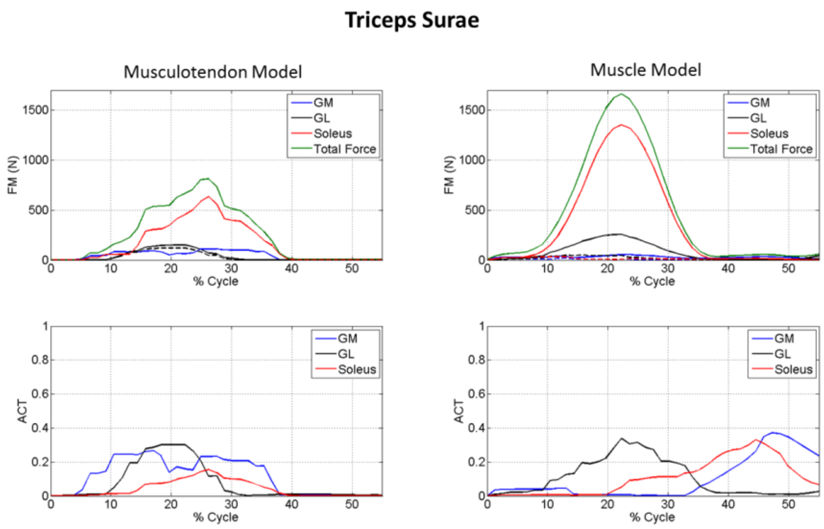

Figure 9: Muscle Force and Muscle Activation of the triceps surae obtained in the musculotendon model and in the muscle model in Running Analysis

allowing the production of more force with less activation. In the results of the muscle model, the muscle force are mainly compose by a passive component and the muscle activation are at the same levels compare to the other model, but the contractile force produce is almost zero. Also, the gastrocnemius, either the medial or the lateral, should be active until the TO and not after as happen in the Muscle model. The contraction of those muscles before the TO is very important to bring the body forward.

5.3. Jump Analysis

Jumping is divided biomechanically into three phases: preparation, action and recovery (Bartlett,2007) (Figure 10).

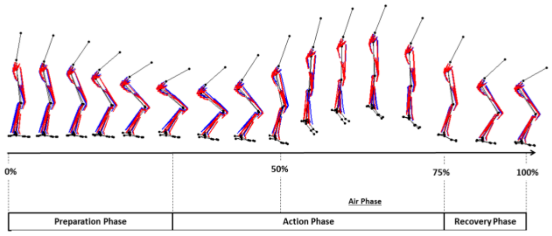

Figure 10: Running Cycle

on
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ankle. To allow this movements the hamstrings (semimembranosus, semitendinosus and biceps femoris), and the hip flexors, mainly rectus femoris (RF), ilipsoas, contract eccentrically.

The action phase involves the hip and knee extension and the ankle plantarflexion through shortening contraction of the muscle responsible for that movement and drive the body vertically upwards. Analyzing the results obtained, the hip extensors (bicep femoris, semitendinosus, semimembranosus and gluteus maximus present muscle activity and consequently muscle force that enable the movement. The quadriceps femoris are activated to promote the knee extension and the gastrocnemius, soleus (Figure 11), peroneus longus, tibialis posterior are also activated to enable the plantarflexion.

In the air phase, the muscle activity observed is must lower than the remains cycle. Finally, the opposite muscle groups, dorsiflexors (TA), hip flexors (RF, ilipsoas) and knee flexors (gastrocnemius and hamstrings), present muscle activity in recovery phase to decelerate the movement and control the landing.

The integration of the tendon in the model also influences the results in the jump but passive force is now present in both models due to the type of movement. Analyzing the results obtained by the two models, the same conclusions already taken in the previous section could be made for the triceps surae (Figure 11).

6. Conclusions

The objectives propose for this work was successfully fulfilled. A valid musculotendon model was developed within the framework of multibody systems dynamics and a biomechanical model composed by 43 muscle actuators was created to study activity, like walking, running and jumping.

Analyzing the results obtained along the cycle, the muscle activities and the muscle force obtained are consistent with the movement observed. When compared this model with the muscle model, the influence of the tendon is evident in all the analysis. The main differences are the percentage of passive component in certain moments of the cycle, and the intensity of the muscle force

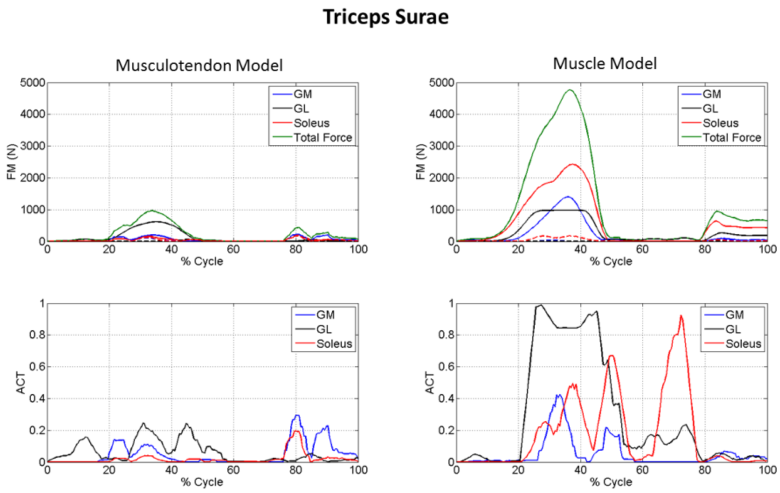

Figure 11: Muscle Force and Muscle Activation of the triceps surae obtained in the musculotendon model and in the muscle model in Jumping Analysis

realizes taking into to account the activity present. The changing on tendon length allow the muscle to contract or elongate with a length and velocity more close to the optimal one, decreasing the activation needed to produce the force to realize a certain movement.

Although, in some muscle the presence of the tendon is not so significant, it is important to considerer tendon in the models to guarantee that results for other muscles are more physiological and more close to the reality.

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