Musculoskeletal system simulations to analyse muscle forces and movement pattern

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Abstract: The musculoskeletal system of the human is a complex system that still has a lot of unsolved mysteries. There is plenty of research being performed right at this moment which is trying to better understand the movement apparatus of the human. In spite of all research being conducted, there are no clear description of the important issues of which facts determine how much and which muscle activates in different movements. This also raises the question of the amount of force each muscle contributes with over the different joints for different movements. The aim of the thesis was to develop a musculotendon unit model for use in optimal control simulations. The model was targeting to be specialized to handle optimization problems in stretch shortening sport movements. The work was concentred on the development of a musculotendon (MT) -unit model, consisting of the muscle and its belonging tendon structure. The model included features for force-velocity and force-length relationship, elasticity of cross-bridges and the passive structures in muscles. The model was made dimensionless which opened the possibility to use it for all skeletal muscles in the body together with the muscle specific parameters. Excluded in the model was the possibility of variable muscle activity and pennation angle. The purpose of the MT-unit model was to incorporate it into a musculoskeletal (MS) model. The MS model developed and used consisted of one degree of freedom, two segments and one muscle. This model was then used in a drop jump simulation where the ground contact phase was evaluated. The muscle was assumed to be fully activated during the whole ground contact. This simulation generated realistic results.

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1- Introduction

The domain of interest of biomechanics is huge: it ranges across physiology (with a special focus on the musculoskeletal, cardiovascular, respiratory, and digestive apparatuses), pathology (orthopaedics and traumatology, maxillofacial surgery, dentistry and orthodontistry, cardiovascular and respiratory surgery), forensics (accident reconstructions, crime scene investigation), vehicle safety (car safety, helmets), ergonomics and workplace safety, defence and social security (combat and law enforcement protection, effectiveness of projectile weapons), and sport (performance optimization, protection devices). (Viceconti, Testi et al. 2006)

Its main scientific journal, the Journal of Biomechanics, was founded only in 1968; even later, the International Society of Biomechanics was founded in 1973, the European Society of Biomechanics in 1976, and the American Society of Biomechanics in 1977. While the research activity over these 30 years has been intense, the impact of biomechanics today is not yet as great as one may have expected. One of the main factors limiting the application of biomechanics results is that, to answer most practical questions, a global model is required (Viceconti, Testi et al. 2006).

The aim of the thesis was to develop a musculotendon unit model for use in optimal control simulations. The model was targeting to be specialized to handle optimization problems in stretchshortening sport movements.

• Develop a two-dimensional musculoskeletal model

• Improve the developed model according to the latest research on SSC

• Validate the model

2- Biomechanics of the human

The knowledge of the human physiology and especially the human biomechanics is of great importance when trying to model the human musculoskeletal system. The general descriptions in this chapter are relatively sparse and more focus has been placed on specific properties important for this thesis. This chapter is divided into three parts; bone and joints, skeletal musculotendon units, and stretch-shortening cycle.

The skeleton in the body consists of bone connected with joints for ability to move. Bone is a hard material which gives the skeleton very good mechanical properties (Marieb 2004).

The anatomical structure of a given joint, as well as the direction of which the attached body segments are permitted to move at the joint, have very small variations from person to person. However, differences in relative tightness or laxity of the surrounding soft tissues result in different range of movement (ROM). (Hall 2003).

The structure of the human skeletal musculotendon unit is well examined and consists of the muscle and tendon. The tendons consist of collagen and elastin and connect the muscle with the bone. The muscle is divided into smaller and smaller portions starting with the whole muscle, fascicles, fibres and finally fibrils. The fibrils are built up by sarcomeres, shown in Figure 1, which consists of actin, myosin and elastic filaments. The movement of the muscle is due to the active movement between actin and myosin filaments and the elastic filament contributes to the elasticity of the muscle. (Marieb, 2004).

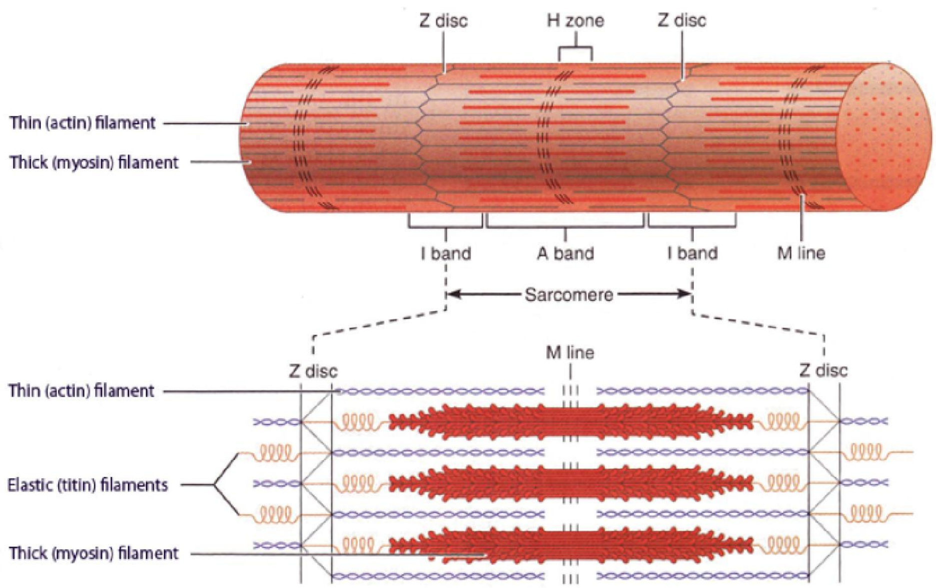

Figure 1: The myofibril (upper) and a sarcomere (lower) of a skeletal muscle. Reproduced from Marieb (2004)

An eccentric muscle action is defined as a muscle action performed during lengthening (or stretching) of the muscle and a concentric muscle action is defined as a muscle action during its shortening. The combination of an eccentric directly followed by a concentric action forms a natural type of muscle function called stretch-shortening cycle (Komi 2000). This function, or phenomenon, is present in daily life activities as walking, running and jumping. Many studies have proved that the enhancement gained from SSC mainly is due to stored elastic energy (Komi 2000).

The literature has described many factors that influence SSC in different ways. The amount of activation of the muscle before impact, called *pre-activation*, has been stated to be important (Komi 2000). Further, the change of length of the muscle fascicle compared to the tendon structure during the functional phase and the stretch reflex affects stretch-shortening cycle (Komi 2000). In vivo experiments on cats reviled that the force increase with higher speeds (Gregor, Roy et al. 1988).

Even though this has been known for a long time the ability to directly apply that on the natural movement including SSC is not straightforward.

In a specific study it has been shown that up to a speed of 14·km/h the positive external work duration is greater than the negative external work duration, suggesting a contribution of muscle fibres to the length change of the muscle–tendon units. Above this speed, the two durations (<0.1·s) are similar, suggesting that the length change is almost totally due to stretch–recoil of the tendons with nearly isometrically contracting fibres (Cavagna 2006).

In human triceps surae, a muscle with short fibres and a long tendon, the time courses of the total (muscle and tendon) length and of the length of the contractile component (CC) alone in running are completely different. The muscle tendon complex shows first an eccentric phase with negative work, followed by a concentric phase. The CC, on the other hand, is concentric all the time. Moreover, the work that is performed is done at a speed that guarantees a high energetic efficiency. It is argued that this high efficiency is an in-built property of the muscle mechanics for muscles with a compliant tendon and a low maximal velocity (Hof 2003).

3- Review: Modelling of the musculoskeletal system

It seems as it is an exponential increase in research performed in the area of modelling the musculoskeletal system and there are also good and recent review articles in the field (Pandy 2001; Fernandez and Pandy 2006; Viceconti, Testi et al. 2006; Erdemir, McLean et al. 2007). One of the largest areas of human movement is the human gait and also in this area a few recent review articles have been published (Zajac, Neptune et al. 2002; 2003). Because of this already good mapping of the research in modelling the musculoskeletal system the gain of knowledge has been great. A special interest has been put on the ability to better simulate movements involving high speed and large forces such as high jump, sprint running and many other sport activities. A property that increases in importance when the movement involves high speed is the stretch-shortening cycle phenomenon (Komi 2000), see 2.3 for more details.

The first valid question to ask is why there is a need of developing models of the musculoskeletal system. The alternative is to directly or in laboratory environments carry out measurements on the human body. Both these methods are needed because direct measurements are in many cases the only possible method to use but direct measurement of for example muscle forces is generally not feasible in a clinical setting, and non-invasive methods based on musculoskeletal modelling are therefore mostly considered (Erdemir, McLean et al. 2007). Dynamic simulations of movement, using a musculoskeletal model, allow one to study neuromuscular coordination, analyse athletic performance, and estimate internal loading of the musculoskeletal system. Simulations can also be used to identify the sources of pathological movement and establish a scientific basis for treatment planning (Delp, Anderson et al. 2007).

Phenomenological and numerical models of the musculoskeletal system are built up in many different ways depending on how the models are supposed to be used. The most accurate and sophisticated muscle models described in the literature yield infeasible computation times, even on modern supercomputers, if they are combined with optimal control techniques (Eberhard, Spägele et al. 1999). Therefore, plenty of different models are developed and a difficult part is to evaluate which model and method is best and in which cases. When using an inverse dynamic solution with static optimization a much more advanced model can be used compared to an optimal control solution but instead this optimization is only considering static conditions. A number of anatomical measurements, extremely important for biomechanical modelling, such as muscle-fibre length, tendon rest length and muscle-fibre pennation angles, can currently be obtained only via dissection (Viceconti, Testi et al. 2006). Because of this, advanced scaling tools have to be used to get the right dimensions.

There are plenty of models presented in the literature but the ability to replicate them is more or less impossible if the author of the model is not willing to supervise. As a part of the review one article was chosen to make a more thorough analysis. It turned out that 26 reference articles were needed to be able to reconstruct the model. Of these 26 references around half of them were easy to find but a few articles were considered impossible to find by the author. One of the articles was referring to unpublished material that would be sent by request which is getting quite difficult when the article is almost three decades old.

A query that could be of great interest for a researcher is the knowledge of how complicated model that has to be used. Depending on the research question to answer different levels of complexity has to be put into the model. While simple models can be helpful in identifying basic features of muscle function, more complex models are needed to discern the functional roles of specific muscles in movement (Pandy 2003). The one of the most simple models to use is a spring-mass model which were used by Bullimore & Burn (2007) to analyse running (3-5 m/s). It showed good predictions of stance tme, vertical impulse, contact length, relative stride length and relative peak force but systematically overestimated horizontal impulse, change in mechanical energy, aerial time and peak vertical displacement (Bullimore and Burn 2007). The spring-mass model is usually used to predict the external kinetic and kinematic variables of interest (Cheng and Hubbard 2004; Robilliard and Wilson 2005), or the joint torque (Cheng and Hubbard 2005).

In the study by Pandy (2003) a comparison were made between one simple and one complex model in walking. The variables of interest were how muscle forces, gravitational forces and centrifugal forces (i.e. forces arising from motion of the joints) combine to produce the pattern of force exerted on the ground. It showed that the simple model gave reasonable results for the larger questions of understanding but the far more complex model (3D) gave plenty of important detailed information (Pandy 2003).

The interest of what is going on inside the body drives the development of more complex models which includes muscles. Many models are built in two dimensions (see Table 1) making the model less complex. This introduces difficulties since the human body musculature has three-dimensional characteristics that are hard to be reduced into two dimensions. Especially, when looking at the location of the origin, insertion and via-points of most muscles, it is observed that three-dimensional vectors instead of two-dimensional vectors better represent the line of action of many muscles (Nagano, Umberger et al. 2005). Due to that reason and the improving computer capacity many of the newer models are developed in three dimensions.

The latest in model development is to create three-dimensional (3D) finite-element models that are able to represent complex muscle geometry and the variation in moment arms across fibres within a muscle. This new framework for representing muscle will enhance the accuracy of computer models of the musculoskeletal system (Blemker and Delp 2005).

3.1 Inverse dynamics-based static optimizations

Muscle force estimation using gait data combined with inverse dynamics and static optimization has been practiced for almost three decades and has become a routine tool in clinical gait analysis (Erdemir, McLean et al. 2007). The muscular load sharing problem is solved for each instant in time, by minimizing an objective function (e.g. total muscle force) subject to constraints representing the equality of the sum of individual muscular moments to the joint torques calculated from the inverse dynamics analysis (Erdemir, McLean et al. 2007). Inadequate kinematic models to represent the motion of interest and inaccuracies of experimental data have been identified as weaknesses of the methodology (Erdemir, McLean et l. 2007).

3. 2 Forward dynamics assisted tracking

Forward dynamic optimization can be performed such that solutions are less dependent on measured kinematics and ground reaction forces, and are consistent with additional knowledge, such as the force–length–velocity–activation relationships of the muscles, and with observed electromyography (EMG) signals during movement (Erdemir, McLean et al. 2007).

When muscle excitations or joint torques are available or assumed, a forward dynamics approach can be utilized that integrates the system equations to calculate the movement patterns. An initial set of muscle activations are fed into a forward dynamics model of the musculoskeletal system. The solution is compared against experimental data and the process is iterated by updating the muscle activations that best reproduce the experimental kinematics and in some cases kinetics (Erdemir, McLean et al. 2007).

The technique has been used in a variety of activities and particularly found its applications for high pace movements of sports biomechanics. A common use has been to find a set of muscle activations that can reliably reproduce the movement pattern, and subsequently perturb parameters of the optimal solution to explore injury mechanisms. This strategy is advantageous due to the more straight forward inclusion of muscle dynamics within the solution when compared to inverse dynamics-based static optimization (Erdemir, McLean et al. 2007). Although the dynamics of the muscle (activation and force generation properties) might not be influential for low pace movements, muscle force estimation for activities of high performance might benefit from this property of forward dynamics assisted data tracking (Erdemir, McLean et al. 2007).

It is possible that multiple solutions exist to track the same experimental data. Multi-objective criterion probably increased the tracking errors in favour of estimating muscular forces based on task objectives (Erdemir, McLean et al. 2007).

The approach is advantageous in that the movement is predicted. Yet, accurate knowledge of muscle excitations (forces) or joint torques is rare, eliminating the stand-alone application of this technique (Erdemir, McLean et al. 2007).

3.3 Optimal control strategies

Occasionally the experimental data might be incomplete or the movement related investigations require predictive simulations of the musculoskeletal system in novel situations for which no movement data are available. Under these circumstances, optimal control strategies that use forward dynamics are alternatives to solve for muscle excitations and forces during movements. Given an initial set of muscle excitations, system equations are first solved in a forward dynamics fashion. Then, the objective of the movement and task related constraints, e.g. static equilibrium at final time, are calculated. The objective can be a function of muscle force and kinematics. It can be related to task performance, e.g. maximum height jumping, and is usually represented in an integral form to introduce dependence on time history (Erdemir, McLean et al. 2007). The process is iterated until an optimal set of muscle excitation patterns is found that minimizes the objective and satisfies the constraints (Erdemir, McLean et al. 2007).

The technique allows for changes in motion and adaptations at the muscular control level following alterations in the system. This major advantage can lead to predictive simulations to assess changes in control of muscles and muscle forces as a result of therapeutic interventions, surgery and rehabilitation (Erdemir, McLean et al. 2007). However, the selection of an objective function can still be controversial; the criterion is clear for movements that aim for optimal performance (e.g. maximal height jumping) but for other activities (that rely on physiological function) such as walking at different speeds and non-ballistic movements, this selection relies on the investigators' preference (Erdemir, McLean et al. 2007). Computational complexity and implementation difficulties also prohibit the routine use of this technique in clinical settings and limit its use to research environments (Erdemir, McLean et al. 2007). Similar movement patterns can be obtained using optimal control simulations with different objective functions while investigating non-ballistic activities, but the muscle activation patterns might be different (Erdemir, McLean et al. 2007).

When using optimization techniques to predict muscle forces, it must be recognized that the solution is sensitive to many assumptions and variables such as PCSA. On the other hand, the joint force solutions are less sensitive to such variations, and the absolute values are more reliable (Brand, Pedersen et al. 1986).

4. Validation

All results generated from a computer model have to be validated to show that they gave reasonable results. Due to many reasons there are difficult to successfully validate the musculoskeletal computations of the muscle force estimates (Erdemir, McLean et al. 2007).

Studies of muscle force predictions usually compare muscle loading or activation patterns against EMG data as an estimate of validity. Although evaluating the temporal characteristics and intensity of muscle firing during a movement is useful, such comparisons cannot verify the magnitude of the calculated muscle force. Fortunately, alternative and more advanced analyses exist, which incorporate the quantification of muscle force sensitivity on modelling parameters and comparisons of muscle forces against direct measurements of tendon loading (Erdemir, McLean et al. 2007).

Direct validations are limited to simple musculoskeletal models, e.g. with one or two degrees of freedom, and tendon force measurements are performed on animals by surgical implantation of tendon force measurement devices. Nonetheless, the results of these studies can be used to assess the validity of objective functions used in inverse dynamics-based static optimization and the load sharing between synergistic muscles (Erdemir, McLean et al. 2007).

It is possible to predict similar muscle forces and joint reaction forces for walking using the inverse dynamics-based static optimization approach and the optimal control simulation approach. The consistency observed in these muscle force predictions suggests that if experimental accuracy can be improved, then resultant muscle forces might not depend on the simulation characteristics (Erdemir, McLean et al. 2007).

Induced acceleration analysis (IAA) provides a platform to establish the link between an isolated change in a muscle force and the corresponding changes in the movement. This ‘‘coupled dynamics’’ representation can explain some of the counterintuitive functions of biarticular muscles, such as the gastrocnemius functioning as knee extensor for specific conditions (Erdemir, McLean et al. 2007).

3.4 Musculoskeletal models in literature

The number of musculoskeletal models in literature is very large due to the large amount of different implementations. First of all there are a very simple models only using simple spring-mass models (Bullimore and Burn 2007) to very advanced models with plenty of DoF and muscle groups included (Anderson and Pandy 1999). Depending on the choice of algorithms for estimating muscle forces (3.2) different complexity of the model is allowed where inverse dynamics methods can have very complex models compared to optimal control strategies.

A special interest within this thesis was explosive movements including high speed and large forces. Consequently, the inverse dynamics that uses a static optimization is not considered as an alternative due to poor results in faster movements. A forward dynamic configuration is more appropriated to use. Further, often the aim is to find optimal movement patterns and therefore no available measured data exist to use a tracking configuration. The commonly used method is therefore an optimal control strategy. The models presented below are, by the reasons described above, models used in optimal control strategies for estimating muscle forces in movements with high speed and large forces (vertical jumps).

3.4.1 Studies investigating vertical jumps (optimal control strategy)

In a study conducted 1993 (Anderson and Pandy 1993) subjects jumped on average 5% higher during the counter-movement jump (CMJ) than they did during the squat jump (SJ), although some subjects performed equally well during both jumps. The model, on the other hand, jumped 2% higher during the SJ than it did during the CMJ. In that study the total energy delivered to the skeleton was almost the same for the CMJ and the SJ. It was also noticed that there was almost as much elastic strain energy stored during the SJ as it was stored during the CMJ. Calculations indicate that much more energy was lost as heat during the CMJ than the SJ. With this analytical result in mind, together with their own analytical and experimental findings, the authors propose that humans perform counter movements not so much to store and re-utilize elastic strain energy during jumping, but rather to increase ground contact time during the propulsion phase of the jump.

Several years later, the same researchers made new more advanced model and analyse (Anderson and Pandy 1999). The mode1 was characterized by several key features: first, it was a mode1 of the whole body; second, full three-dimensional motion was permitted by virtue of a 6 dof pelvis, 3 dof joints for the back and the hips, and 2 dof joints for the ankles; third, the feet were free to make and break contact with the ground; and fourth, the number of muscles was much greater than that considered in previous dynamic optimization studies. This increase in complexity has improved the fidelity of the mode1 in a number of ways: (1) the vertical ground-reaction force demonstrated a more gradual decrease near lift-off compared with the results obtained in previous simulations (Pandy and Zajac 1991); (2) the fore-aft ground-reaction force was reproduced more accurately than before (Pandy and Zajac 1991); and (3) the mode1 was capable of predicting not only the major movements of the body segments in the sagittal plane, but also those which occur in the frontal and transverse planes. The major limitation of the mode1 was its failure to reproduce the kinematics of the jump near lift-off. This result may be explained by the relatively fast rise time for muscle activation used in the model. gait (Anderson and Pandy 2001).

4- The musculotendon-unit model

As been stated earlier, the development of a new musculoskeletal model is not trivial and therefore not realistic to fit into a master thesis. The work of this thesis was concentred on the development of a musculotendon (MT) -unit model. As the name describe this unit consists of the muscle and its belonging tendon structure. An important criterion for the model was to make it dimensionless so the same model could be used to describe different muscles in the body even though they have different properties and dimensions. The scaling parameters used were the maximum isometric muscle force (FISO,Max) and the muscle length corresponding to the maximum isometric force (lM,opt).

During the meticulous review of literature many different MT-unit models were found as can be reed earlier (3.1.3). A decision was made to use the model reported by Pandy, Zajac et al. (1990). Even though this article is almost two decades old, the model is still used and the subject of the article were vertical jumping which includes both high speed and large forces. The MT-unit model in the chosen article was first presented at the RESNA conference 1986 (Zajac, Topp et al. 1986) and has been cited many times after.

Another question for discussion was the type of programming environment that should be used. The literature are showing a wide spread of environments. Finally decisions were made to develop the MT-unit model in *Mathematica* (Wolfram Research Inc. 2007). The decision was made mostly due to earlier familiarity with the program and its ability to treat both symbolic and numerical mathematics.

When talking about the original article or original graph it refers to Zajac et al. (1986).

4. THE MUSCULOTENDON-UNIT MODEL

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4.1 MT-unit

The MT-unit is based as been said on Zajac et al. (1986) which was using a Hill-type model. The model consists of a tendon and a muscle (Figure 3). These two components will be described in detail later.

The idea of the model was to build it up using springs and actuators. The representation is basically done in a schematic way of the representation of a real human musculotendon unit. The tendon model (T) is placed in series with the muscle model (M) and represents by a spring. The muscle model consists of a parallel elastic element (PE), a serial elastic element (SE) and a contractile element (CE) (see Figure 3). Both the elastic elements are represented by simple springs and CE is represented as an actuator.

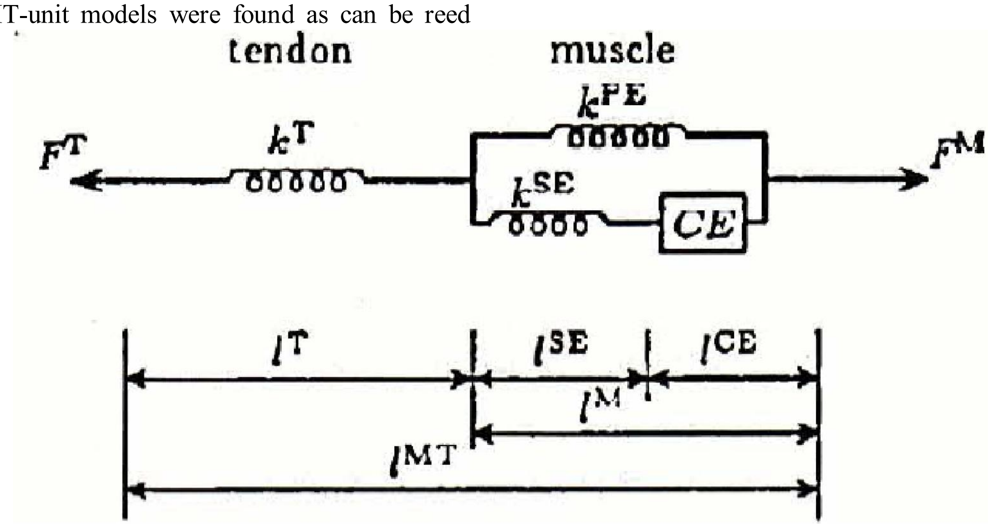

Figure 3: The musculotendon-unit. The stiffness symbol ki is not used in this report but it stands for the tangent stiffness. Reproduced from Zajac et al. (1986)

4.1.1 Muscle

The components of the muscle are as earlier stated the parallel elastic element (PE), the serial elastic element (SE) and the contractile element (CE). The individual properties of the components are described separately in own sections below.

$$\lambda_{PE} = \lambda_M$$
$$\lambda_M = \lambda_{SE} + \lambda_{CE}$$

The force of M is the sum of the force in SE and PE and the force in CE is equal to the one in SE and that is shown below.

$$\phi_{CE} = \phi_{SE}$$
$$\phi_M = \phi_{SE} + \phi_{PE}$$

This gives that at maximal isometric contraction the length of M is exactly one because they are made dimensionless.

Parallel elastic element

The purpose of the parallel elastic element was to simulate the force arising from either inter-fibre connections or elastic structures internal to the muscle fibre (Zajac, Topp et al. 1986). This was simulated by a simple spring and the force was assumed to be zero when the muscle was shorter than the optimal length. Further, it was assumed that this relationship was the same among all the muscles (Zajac, Topp et al. 1986).

$$F_{PE} = k_{PE}\, l_{PE}$$

The constitutive equation of the passive muscle was viscoelastic, as it was for most soft tissues. For slow movements, the viscous contribution was neglected and hyperelastic models were used for the constitutive equation. For more rapid movements, the combination of viscous forces and large deformations makes the models extremely nonlinear, and it was necessary to adopt explicit integration schemes to solve them (Viceconti, Testi et al. 2006). In this case the equation for PE was derived using a graph in the original article. The force equation (Eq. 6) were constructed by localization of the points (1.0, 0.0), (1.3, 1.0), (1.2, 0.5) from the original graph (Figure 5) and then make a curve fit for a second order polynomial.

4.1.2 Tendon

The tendon element (T) of the model represents the physiological tendon both internal and external to the muscle. The strain in the tendon is assumed to be the same everywhere in the tendon.

Further, it is assumed that the force-strain relationship is the same among all musculotendon units.

The tendon slack length is varying a lot depending of the muscle (Pandy, Zajac et al. 1990) and is therefore change for the specific musculotendon unit.

$$\epsilon_T = \frac{\lambda_T - \lambda_{T0}}{\lambda_{T0}}$$

The original article presented a graph (Figure 13) over the relationship between force and strain and within this graph three points were located; (0.0, 0.0), (0.03, 1.0) and (0.02, 0.5). A second order polynomial was used as fitting function and gave the following function:

$$\phi_T = \frac{25}{3}\epsilon_T + \frac{2500}{3}\epsilon_T^2 \quad when\ \epsilon_T > 0$$

$$\phi_T = 10^{-5} \quad when\ \epsilon_T < 0$$

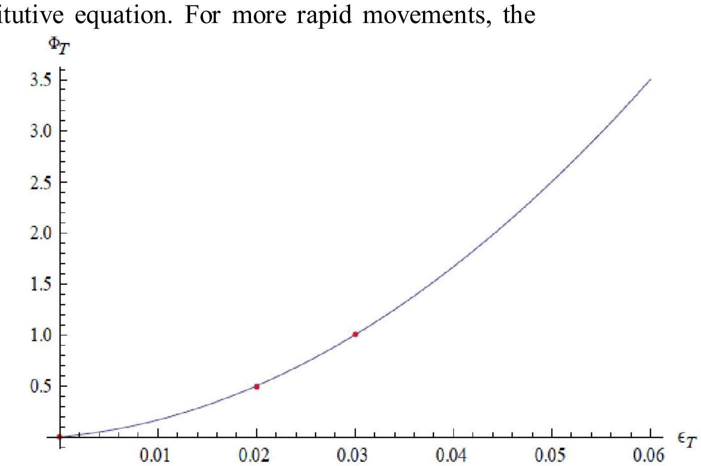

Figure 12: Force-strain relationship of T (Eq. 16).

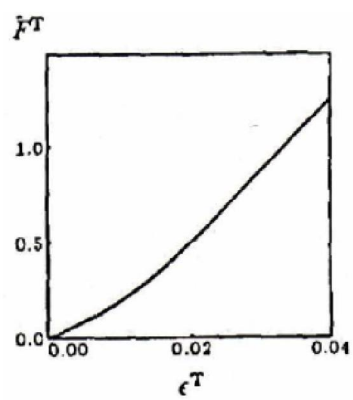

Figure 13: Force-strain graph of T.

Tendons are usually represented as an elastic element. Even though force varies nonlinearly with a change in length as tendon is stretched from its rest length, a linear force-length curve is sometimes used. This simplification will overestimate the amount of strain energy stored in tendon (Pandy 2001).

4.2 Numerical calculations

Last part described the components one by one but they have to be put up to a system representing the musculotendon unit. Due to the complexity of the system two configurations of numerical calculations were used; starting configuration and dynamic configuration. As the name reveals the starting configuration was used before the simulation to find a stable starting equilibrium for the MTunit.

The equilibrium values from the starting configuration were then used as starting values in the dynamic configuration.

4.2.1 Starting configuration

The basic thought of the starting configuration was to run a simulation in time where no input values were change and a final stable equilibrium was found.

This configuration was developed only because of the complexity of the numerics and therefore many input values did not needed a physiological explanation. Because the aim was to find a stable equilibrium, the velocity of CE was set to zero (μCE=0). The simulation time was set to 4.5 seconds with a relatively large time step (=0.1) and was chosen because it gave good values for all tried cases.

The only input value that was changing dependent on the simulation performed was the length of the MT-unit (λMT). This length was of course changing depending of the starting position.

To be able to solve for the whole MT-unit the use of a parameter solving algorithm needed to be used. This was because the system is highly nonlinear and includes many if-statements. At each time step a special algorithm was needed to solve equilibrium between the forces ?M = ?T and ?CE = ?SE.

The algorithm was based on a while statement.

The length of the tendon was used in the while statement, with an old (λTG) and a present (λT) value of the tendon length. These two were compared in the while statement and when the difference between them was less than the tolerance limit (αs=10-12) the solution were satisfying.

Inside the while statement the mean of λTG and λT were calculated and became the new λT.

The velocity of CE was thereafter calculated by subtracting the present λCE with the one from last time step, this was then divided by the time step length. This formula is used for all time steps except the first when the velocity was considered zero. The force of CE can then be calculated and because the force of SE should be the same this force was used in the force equation for SE (Eq. 9) and the new λSE was solved. This was looped until it satisfied the tolerance.

Thus, all the lengths were updated and consequently the force of M could be updated. Because the force of M should equal the force in T the force in M was used to solve λT using the equation for calculating the tendon force (Eq. 16). This was looped until it satisfied the tolerance.

The last thing done at every time step was to update λCE for use to calculate the velocity in CE at the next time step.

4.2.2 Dynamic configuration

A starting value of the length of CE has to be known to be able to run the dynamic configuration.

This value was the most important outcome from the starting configuration. As the starting configuration was in a static equilibrium the velocity of CE was set to zero. The time step was set to a constant with the length 10-4 s.

The input value was the length of the MT-unit which could come from a simple function or from a larger musculoskeletal system simulation. The driving variable in the simulation was the length of CE which was updated without any ability for correction and was updated according to this equation:

$$\lambda_{CE}^{n} = \lambda_{CE}^{n-1} + v_{CE}\, dt$$

This equation was valid for all cases except if the minimum value of λCE were reached. In that case the length was kept constant at the minimum length. A while statement was used to solve the length distribution between T and SE. It had the same basic thoughts as in the starting configuration algorithm. Here the while statement was driven by an old and a new λSE.

5 Simulation

The model developed in the last chapter is of no or very little use alone. The purpose of the MT-unit model is that it should be incorporated in a musculoskeletal (MS) model. The MS model could include up to 54 muscle groups (MG) (Anderson and Pandy 1999), each represented by one MT-unit and their specific muscle properties. It is when these kinds of MS models are included in simulations that a good MT-unit is of great use.

5.1 Musculoskeletal model

As been stated above the musculoskeletal (MS) model was aimed to be very simple and was therefore introducing many assumptions. The first assumption was that the model was developed in two dimensions (2D) instead of the three dimensions (3D) a real human have. The movement in the transversal plane were assumed to be small and neglected. Further, only two segments were included in the model; the foot and the shank. In Figure 14 the two segments can be seen, where the shank starts at the knee and ends at the ankle and the foot segment starts at the ankle and ends at ground contact. The other two thicker lines in the foot are just for visual clarity. The thinner line starting from the shank and attaching to the foot is representing the only muscle included in the model, *m. soleus*.

Figure 14: The MS model

5 Simulation

The model developed in the last chapter is of no or very little use alone. The purpose of the MT-unit model is that it should be incorporated in a musculoskeletal (MS) model. The MS model could include up to 54 muscle groups (MG) (Anderson and Pandy 1999), each represented by one MT-unit and their specific muscle properties. It is when these kinds of MS models are included in simulations that a good MT-unit is of great use.

The choice of simulation was greatly dependent of the amount of working hours that reasonably could be placed on the simulation.

5.1 Musculoskeletal model

As been stated above the musculoskeletal (MS) model was aimed to be very simple and was therefore introducing many assumptions. The first assumption was that the model was developed in two dimensions (2D) instead of the three dimensions (3D) a real human have. The movement in the transversal plane were assumed to be small and neglected.

Table 2: Muscle specific parameters of m. soleus.

Muscle specific parameters	OpenSim	Article 1	Article 2
Maximal isometric force	4000N	3150N	4235
Optimal fibre length	0.08m	0.030m	0.034
Tendon slack length	0.22m	0.264m	0.360
% - tendon	73.3%	89.8%	91.4

OpenSim – generic values of a model, Article 1- (Thelen 2003), Article 2 - (Pandy, Zajac et al. 1990)

The MT-unit model requires three muscle specific parameters; maximal isometric force, optimal fibre length at maximal isometric force and the tendon slack length (see Table 2). Table 2 shows values from three different sources with quite different values. One reason could be that different sizes of humans have been used and another that the pennation angles are different. The one used in this model is the values from OpenSim.

$$\theta_{foot} = \frac{180\,\theta_1}{\pi} - 34$$

$$\theta_{ankle} = \frac{180\,\theta_1}{\pi} - 34 + 90 - \frac{180\,\theta_2}{\pi}$$

The foot angle, θfoot , is the angle between the ground and the sole of the foot. The ankle angle, θankle , is the angle between the shank and sole of the foot and gives zero when they are perpendicular. When the angle is less than 90 degrees it gives negative values and lager than 90 degrees positive values.

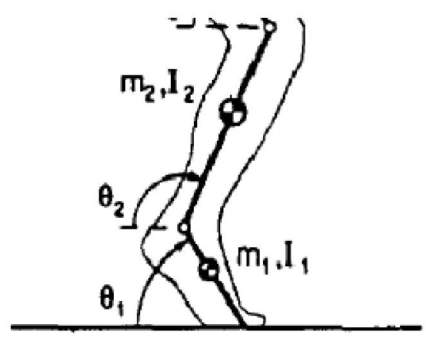

Figure 15: The MS-model. Reproduced from Pandy el al., (1990)

The human of which all anthropometric data was based on had the total mass of 76 kg. When studying Table 3 four segments are recognized but in this model only the shank and foot were modelled. The thigh and HAT were only integrated by placing a point mass in the knee so the length and inertia for those were not implemented in this model.

5.2 Drop jump simulation

The simulation itself is aiming to replicate the contact phase during a drop jump. This is a fast movement with large forces which is included in many sports (Stålbom, Holm et al. 2007). The muscle was assumed to be fully activated during the whole ground contact which was believed to be reasonable.

The maximum isometric force for *m. soleus* was set to 7000N instead of the earlier stated 4000N, for the reason that more plantar flexor muscles are usually active during this kind of activity and because *m. soleus* was the only implemented muscle in this simulation more force was given to it.

The starting angular velocity of the foot (θfoot) and ankle angle (θankle were -353 °/s and 471 °/s, respectively. This corresponds to a drop jump from 50 cm. The reason of presenting it in degrees is due to convenience of comparing it to the literature.

5.2.1 MS-model

The MS-model was developed to be simple and it had the major aim to serve as a tool for evaluating the developed MT-unit. Even though the model was supposed to be simple it still needed to serve as a sufficient base for evaluating the MT-unit. The result from the MS-model had two important roles; describing the conducted simulation and give a brief idea of the ability of generate accurate values for the evaluation of the MT-unit. The contact time during this analysis was 0.22 s. Figure 16 shows the two angles used to describe the position of the shank and foot in degrees. The upper curve is representing the foot angle and the lower the ankle angle. As can be seen the downward movement was going faster than the following upward movement. The largest moment arm for the MT-unit was shortly before the lowest part in the jump (Figure 17).

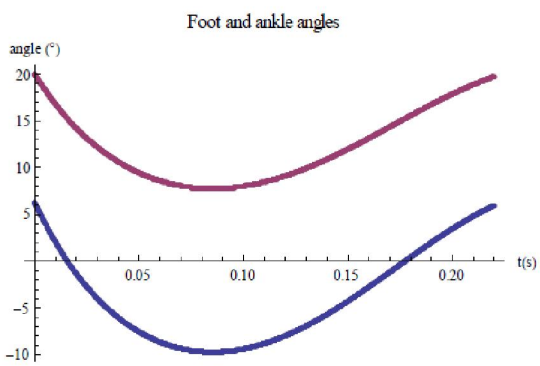

Figure 16: The angles of the foot

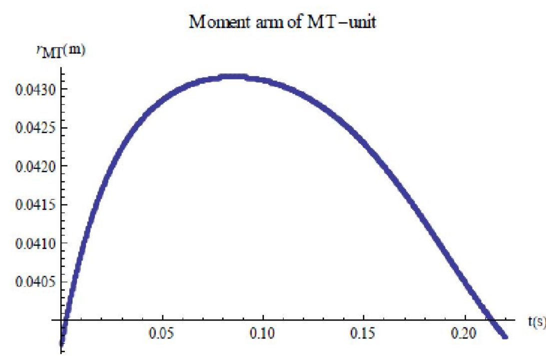

Figure 17: The MT-unit moment arm

Figure 18 is showing five pictures, from left to right, were the first shows the start position and the last the end position. The range of motion was approximately within the same region as the one found in a very fundamental empirical test.

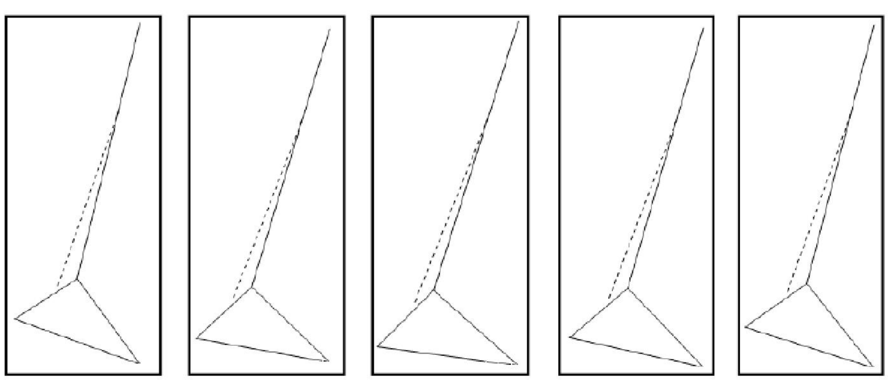

Figure 18: Schematic picture of the jump: Starting position at θfoot=20.0° and θankle **=6.3°**, l**owest position at θfoot=7.8° and** θankle **=--9.7°** and **final position at θfoot=19.7° and** θankle **=6.0°**

5.2.2 MT-unit

The most interesting results were the one for the MT-unit because it was primary this model that had been in focus during the model development. One of the aims of the MT-unit model was to make it working well with high velocity and large forces. It has been stated that under these conditions the passive structure gets a more important role and should be carefully modelled.

The figures below are showing the length of the whole MT-unit and the CE-unit respectively. The xaxis shows the time and the y-axis the actual length in metres.

Conclusions

This thesis has presented a new mathematical model of a musculotendon unit based on an old model. The model includes features for force-velocity and force-length relationship, elasticity of crossbridges and the passive structures in muscles. The model is dimensionless which makes it possible to use for all skeletal muscles in the body together with the muscle specific parameters. Excluded in the model is the possibility of variable muscle activity and pennation angle. Running the musculotendon unit model within a drop jump simulation generated realistic results. The introduction part introduced many interesting questions about muscle forces, optimal movement pattern and changes in movement pattern due to injury. Within this thesis no answers to these questions have been revealed and due to the complexity of the questions it was not expected.

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