Energetic Analysis of the Ankle Musculoskeletal Complex with a Passive Spring Exoskeleton

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Abstract

The augmentation of human capabilities has been at the forefront of exoskeleton development in the past decade. Recently, passive devices have been developed and studied as an option for reducing the metabolic cost during gait. In this dissertation, a computational model of the ankle musculoskeletal complex is developed with the intention of studying the mechanics and energetics of this joint during the gait cycle, whether aided by an exoskeleton or not. The developed model is composed of the soleus and the tibialis anterior, as well as a linear stiffness spring in parallel with the joint, which provides an added moment during stance. Each muscle is modeled as a Hill-type muscle coupled to an energy expenditure muscle model. This allows for the simultaneous analysis of changes to both mechanics and energetics of the muscles when spring stiffness is varied between 0 and 200 Nm/rad. The total metabolic cost associated with the joint is also analysed and a minimum is reached for an intermediate stiffness of 150 Nm/rad. This finding supports the idea of reducing the metabolic cost of walking using passive devices, while providing evidence that altering the natural gait cycle, without careful analysis, can lead to an increase in its metabolic cost.
Keywords: Biomechanics, Musculoskeletal modelling, Gait, Metabolic cost, Passive exoskeleton

1. Introduction

Walking is the most common of human movements and, although it is one of the most complex completely integrated movements, it is one that has been perfected throughout the centuries [1]. When walking, humans keep energy expenditure to a minimum, by for example adjusting their step length and arm motion. Despite this natural tendency towards minimizing energy costs during walking, humans still spend an overwhelming amount of energy on this activity, especially in demanding conditions [2].

Developing and studying strategies to further reduce this cost could, therefore, prove useful in many settings. For instance, by effectively reducing the metabolic cost of walking, and thus allowing individuals to walk farther or carry more weight, but also by reducing fatigue, increasing mobility and reducing the risk of injury [2, 3].

The reduction of the metabolic cost of walking is the goal of many exoskeleton technologies being developed [4]. One of the most promising exoskeletons developed with this purpose is the ankle exoskeleton developed by Collins et al. [2]. While most of these solutions are active, this one is passive, that is, it relies solely on passive elements, namely a spring for actuation, making it lighter. Recently, a prototype inspired by that of Collins et al. was developed at IST by Machado [5]. This ankle exoskeleton is quasi-passive, meaning that, while relying on passive elements for actuation it makes use of electronic components for controlling said actuation system.

In order to further develop these technologies, a deeper understanding of the mechanisms underlying the mechanics and energetics of the ankle during gait, both aided and unaided, is required. This work focuses on developing a computational model for studying the ankle complex muscles' mechanics and energetics when a linear stiffness spring is added to the joint and analysing the potential decrease in metabolic cost.

2. Background

2.1. Energetics of Gait

Most of the energy expended by individuals during the day goes towards walking [2]. Metabolic energy, which comes from ingested and stored nutrients, is required for muscle contraction and relaxation. Therefore, the metabolic cost of walking "is set by muscles that act to perform work on the center of mass, swing the legs relative to the center of mass, and support body weight" [6]. Since gait is a repetitive motion, it is often analysed as a cycle. The gait cycle "is defined as the time interval between two successive occurrences of one of the repetitive events" of the motion [7], the most usual occurrence chosen being the initial contact (IC) of one of the feet. The event between the IC of a foot and the IC of the same foot is a stride. The gait cycle is divided in two phases: the stance phase, when the foot is in contact with the ground, and the swing phase, when the foot is no longer in contact with the ground. The stance phase makes up 60% of the cycle while the swing phase corresponds to the remaining 40% [7].

Since the overall gait pattern is similar, both intra and inter-subject, it is possible to analyse gait data according to normalized universal patterns [1]. For metabolic cost analysis the most relevant data to analyse is the power generated by each joint during the cycle. From the reference curve for joint power (figure 1), it is clear that the ankle is the joint that produces more power during the gait cycle, namely during push off. In addition to this, it is estimated that the plantarflexors expend approximately 27% of the metabolic energy used for walking [2]. Therefore, a reduction in the metabolic cost of gait should inherently be focused on the ankle.

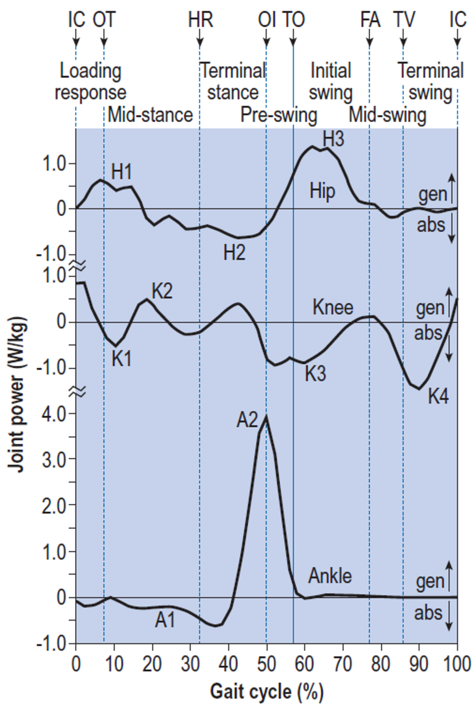

Figure 1: Variation of joint power along the gait cycle in the sagittal plane [7].

Measuring the metabolic cost of human activities provides insight into their demand on the body. The analysis and comparison of these measurements, for different activities or for the same activity performed under different conditions, can help find ways to reduce energy expenditure or quantify the benefit provided by an external device, such as an ankle exoskeleton. In a laboratory setting, metabolic costs are usually measured by indirect calorimetry.

Calorimetry methods only provide a whole body estimate of energy expenditure, when often the expenditure of single muscles or joints is of greater interest. In light of this limitation, many researchers have developed energy expenditure models which use variables acquired during movement analysis, such as muscle forces, lengths and activations, in order to estimate the energy expended by a given muscle [8]. One such model, which has been employed by many researchers, is that developed by Umberger et al. [9]. In addition to providing average muscle specific energy expenditure, this model also "allows the time profile of the metabolic rate to be computed" [10]. Therefore, providing information on metabolic consumption throughout the movement.

2.2. Locomotion Aids for Metabolic Cost Reduction

Recently, researchers have been focusing on ankle exoskeletons to reduce the total metabolic cost of walking. For example, the team lead by Collins [2] developed a passive exoskeleton and an active, tethered, exoskeleton, the latter allowing both torque and work to be provided to the ankle [2, 11]. On the other hand, Mooney et al. [4] developed an active, battery powered, exoskeleton. The passive exoskeleton makes use of a passive clutch, that controls the locking and unlocking of the mechanism, and a series spring, that stores energy during stance phase dorsiflexion and releases it at push off. According to their findings, a metabolic cost reduction of $7.2 \pm 2.6\%$ was achieved when compared to walking without the exoskeleton [2]. The active exoskeleton developed by Mooney et al. relies on brushless DC motors connected to winch actuators that act on the struts connecting the shank and the foot. During test trials a metabolic cost reduction of $8 \pm 3\%$ was achieved [4].

The main goal of the passive exoskeleton prototype developed by Collins et al. was to determine if it was possible, and if so, how, to reduce the metabolic rate of walking without providing an additional energy source, that is, by using solely passive components when developing an ankle exoskeleton. The design utilized in their experiments worked with a spring acting in parallel with the calf's muscles which off-loaded the muscle force, leading to a decrease in metabolic energy consumed during walking.

The exoskeleton produced a torque pattern similar to that of the ankle, thus reducing the moment produced by the plantarflexors and, consequently, reducing their activation. As expected, this effect was particularly noticeable in the soleus, since this is a uniarticular muscle acting solely on the ankle and so the exoskeleton closely mimics its actuation [12]. Increasing the spring stiffness above a certain threshold (180 Nm/rad) led to an increase in metabolic cost. It is important to note that, according to the authors, the effective mechanical stiffness of the exoskeleton was about 33% lower than the nominal spring stiffness [2]. This increase in metabolic cost could be due to several factors, such as an increase in dorsiflexor activity, namely of the tibialis anterior, to counteract the added torque [2, 11], or increased knee muscle activity to prevent hyperextension during stance [2].

An increase in plantarflexor activity at the end of stance was also observed, even though the joint moment decreased. This could suggest that the plantarflexors are shortening sub-optimally at the end of stance. Indeed "the plantarflexor muscle–tendon units seem tuned for near-optimal efficiency and power production during unassisted locomotion" [12]. During gait, the muscles produce near isometrical, albeit eccentric force, while the Achilles tendon lengthens, storing mechanical energy. This is an energetically efficient strategy since near isometrical force requires little energy and, at push off, the shortening of the muscles, allied to the tendon's recoil, generates the required power burst. Moreover, the architecture of the plantarflexor muscle-tendon units allows the "muscle fibers to operate at favorable lengths and velocities during positive work production" [12] at push off. Changes to these fine-tuned muscle-tendon units' operating ranges can lead to an increase in energy expenditure.

The goal of the active prototype was two-fold: analyse the effect of adding work, and of adding torque, to the biological system. The findings were that "both techniques reduced effort-related measures at the assisted ankle" [11], but while adding work reduced the metabolic cost, adding torque lead to an increase in the energy expended, most likely due to whole body effects. Another possibility for this increase stems from the disturbance of the fine-tuned plantarflexor muscle-tendon units. In order to explore this possibility, the authors conducted further analyses using musculoskeletal models.

3. Models

3.1. Hill-type Muscle-Tendon Model

Hill-type muscle models provide a fair description of the dynamic behavior of real muscles and are typically used due to their simplicity and low computational cost [13]. In this work a Hill-type muscle-tendon unit (MTU) model, as presented in Geyer and Herr [14], was adopted.

The MTU model (figure 2) consists of a contractile element (CE), representing the muscle fibers, which at rest has zero tension but when activated is able to shorten, an elastic series element (SE), representing the stiffness of the tendon, an elastic parallel element (PE), which represents the stiffness of structures parallel to the muscle fibers [15], and a buffer element (BE). The PE is engaged when the CE stretches beyond its optimal length, while the buffer element prevents the collapse of the contractile element if the series element is slack, that is, if $l_{MTU} - l_{CE} < l_{slack}$. l_{MTU} is the length of the muscle-tendon unit, l_{CE} is the length of the contractile element and l_{slack} is the length of the SE when slack [14].

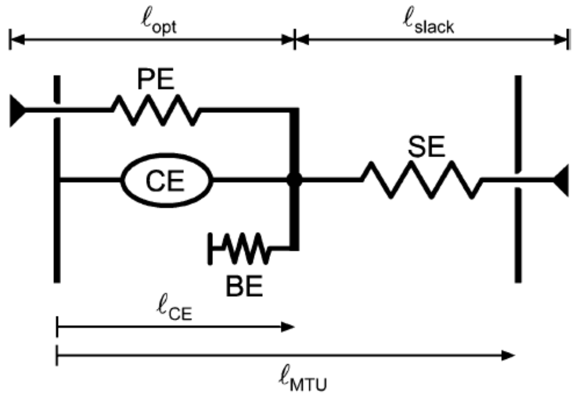

Figure 2: Representation of the Muscle Tendon Unit (MTU) [14].

3.2. Muscle Energy Expenditure Model

In this work a model for muscle energy expenditure, developed by Umberger et al. [9] and used at length by Ackermann [15] in his doctoral thesis, was adopted. This model was chosen since it is founded on mammalian and human muscle experimental data and it accurately "accounts for muscle heat production during submaximal and eccentric muscle activities" [15].

The total rate of muscle energy expenditure ($\dot{E}$) can be expressed as a sum of three terms: the activation and maintenance heat rate ($\dot{h}_{am}$), the shortening/lengthening heat rate ($\dot{h}_{sl}$) and the mechanical work rate of the contractile element ($\dot{w}_{ce} = f^{ce}v^{ce}$). The total rate of muscle energy expenditure ($\dot{E}$) is thus given, in Watt (W) by:

$$\dot{E} = -f^{ce}v^{ce} + m_{musc}\dot{h} \quad (1)$$

It is important to note that the total heat rate ($\dot{h}$), given by the sum of the two heat rates above, cannot fall below 1 W/kg, since this is the resting energy rate for human skeletal muscle *in vivo*, and while "eccentric muscle work can be performed more efficiently than an equivalent amount of concentric work and can perhaps even cause a net absorption of heat, studies have shown that active lengthening cannot result in a net synthesis of ATP" [16]. So the total instantaneous power ($\dot{E}$) is prevented from becoming negative, that is, $\dot{E} \geq 0$ for every instant. Finally, integrating this quantity over time, for each muscle, yields the total energy expenditure required to produce the movement. The total metabolic cost, in J/kg, is given by equation 2, where m represents the subject's mass and T is the motion duration.

$$E_{tot} = \frac{1}{m} \int_{t=0}^{T} \sum_{i=1}^{N_{musc}} \dot{E}_i \, dt \tag{2}$$

4. Dynamics of the Ankle Joint

4.1. Problem Formulation

This work aims to analyse the effect of an ankle exoskeleton on the energetics of the ankle joint. For that, a computational model of the ankle musculoskeletal complex is developed, where a linear stiffness spring is added to the joint. The computational model is implemented in Matlab and has two different versions. The first considers the two main plantar flexor muscles, soleus and gastrocnemius, and the main dorsiflexor, the tibialis anterior. And the second one considers solely the soleus and the tibialis anterior.

4.2. Methodology

4.2.1 Reference Gait Model and Data

Data for the ankle joint kinematics was obtained from the Neuromuscular Locomotion Model developed by Geyer [14, 17]. This model, implemented in Simulink, has seven segments driven by 14 muscles, which are modeled as Hill-type muscles [18]. The model simulates walking at a speed of 1.3 m/s, and the soleus, gastrocnemius and tibialis anterior were considered for energy cost. The data retrieved from the simulation were the ankle joint angle, the total moment produced at the joint, as well as the activation, length of the muscle tendon unit (l_{mtu}) and velocity of the contractile element (v_{ce}) of each muscle considered.

4.2.2 Energetic Ankle Model

The muscles were modeled using the Hill-type muscle model coupled to the energetic model previously described. This approach allows for the estimation of the energy expenditure of each individual muscle considered in the analysis. Since the considered model only accounts for the soleus, gastrocnemius and tibialis anterior, the muscle specific parameters required were retrieved solely for these muscles (table 1).

Table 1: Muscle specific parameters required by the Hill-type model and by the muscle energy expenditure model [14, 15].

Parameter	l_{slack} [m]	l_{opt} [m]	v_{max} [$l_{opt}s^{-1}$]	f_{max} [N]	ft [%]
Gastrocnemius	0.40	0.05	12	1500	50
Tibialis Anterior	0.24	0.06	12	800	25
Soleus	0.26	0.04	12	4000	20
Parameter	ω	c	N	K	ϵ_{ref}
	0.56	0.05	1.5	5	0.04

The muscle-tendon model was implemented in Simulink. The implementation was adapted from the Neuromuscular Locomotion Model [14, 17]. Only the Simulink blocks corresponding to the Hill-Type muscle-tendon units of the muscles of interest were kept for this work. Each block receives as input the activation and MTU length of its corresponding muscle and outputs the velocity, length and force of the contractile element, as well as the length of the tendon.

The muscle energy expenditure model computes the total energy expenditure of the muscle (equation 2), given the force, velocity and length of the contractile element and the activation. In addition to these, the model also provides the time profile of all portions of $\dot{E}$. The energy expenditure model was implemented in Matlab as a function, which receives as input the parameters and variables required and outputs the aforementioned values.

The spring added to the model stores energy during stance phase dorsiflexion and releases it at push off [5]. Since the spring has linear stiffness the moment it produces is given by equation 3, where K_s is the spring's stiffness, θ_0 is the joint angle at spring activation and θ is the instantaneous joint angle.

$$M_{spring} = K_s(\theta - \theta_0) \tag{3}$$

4.2.3 Optimization

Admitting that the movement of the ankle joint is not altered by the effect of the spring, that is, that the kinematics remain the same, the total moment of the ankle complex will remain the same. Thus, the moment generated by the muscles active during this period, the triceps surae, which are contracting eccentrically during controlled dorsiflexion and concentrically during push off, will be reduced compared to when the spring is not active. A reduction of the energy expended by the triceps surae is expected to be associated with this decrease in moment produced. This reduction in energy expended depends on a multitude of factors which will be influenced by the reduction in muscle moment production. Most importantly, since the muscles are producing less moment, and hence less force, the activation will be reduced. Once the reduced activation is obtained it can be input, along with the MTU length, in the ankle complex model in order to ascertain if there is a reduction in muscle energy expenditure due to the presence of the spring.

In order to compute the reduction in muscle activation, optimization strategies must be employed. Dynamic optimization allows considering the dynamic behavior of the muscle-tendon unit as a whole, whereas static optimization suffices when the tendon is neglected. Routines for multiobjective optimization implemented in Matlab's Optimization Toolbox allow for the optimization of parameters in Simulink models. As stated before, the Hill-Type muscle-tendon unit model used in this work was implemented in Simulink, and has the activation of the muscle as one of its inputs. Therefore, a multiobjective optimization routine allows for the optimization of the activation based on a certain objective function.

While taking into account the tendon's behavior and the dynamic aspects of the MTU, this optimization strategy fails in an important aspect. Since it is unable to make use of a biologically relevant objective function, it does not allow for the solution of the redundancy problem. The muscle where the impact of the exoskeleton is more relevant is the soleus, as it is the uniarticular muscle acting at the ankle joint during stance phase dorsiflexion [12]. Thus, restricting the energetic analysis to this muscle is a valid approximation of the physiological phenomenon.

With the increase in spring stiffness there will be periods in which the moment generated by the spring is higher than that initially generated by the soleus. Therefore, in order for the kinematics to remain unchanged, the stance phase antagonists, particularly the tibialis anterior (TA), will have to compensate for this moment which is being added to the joint. In order to emulate this behavior the multiobjective optimization approach was used to compute the activation of the TA.

Once the activations of the muscles are obtained they can be input in the ankle complex model, along with the MTU lengths, in order to obtain the energetic cost associated with the motion when the spring is present.

5. Results

The developed ankle complex model was applied to the data generated by the Neuromuscular Locomotion Model [14, 17] for the ankle joint. The workflow was the following: the ankle complex model was run without any stiffness added to the joint, so that the moment generated by each muscle could be saved as the initial muscle moment (M^I_{muscle}); a given stiffness was added to the model and the multiobjective dynamic optimization routine was used to obtain the muscles' activations; the obtained activations and MTU lengths were input in the ankle complex model to compute the changes in the muscles' mechanics and energetics. Seven different values of spring stiffness (K_s), between 50 Nm/rad and 200 Nm/rad, were tested and compared to the instance where no stiffness was added to the joint. Results were only analysed for the stance phase, since the spring is only active during this phase and thus, any changes to the muscles' mechanics and energetics occurring during swing are due to numerical instabilities and should not be considered in the analysis.

5.1. Optimization

The activation is the only parameter of the muscle-tendon model which is manipulated in this analysis, as the MTU length is equal for all K_s values, since kinematics are fixed. Therefore, the proper estimation of the activation is crucial to the validity of the results obtained. As previously mentioned, multi-objective optimization is used to compute the activation, required by each muscle, to produce the moment which offsets the one produced by the spring. As spring stiffness increases, less moment is required from the plantarflexor and more moment is required from the dorsiflexor. Total joint moment is computed according to equation 4, where the TA's moment is subtracted since it is antagonistic to the movement. As spring stiffness increases the moment required from the muscles spanning the joint decreases. The total moment at the joint is the sum of the total muscle moment and the moment produced by the spring. As expected, given that the kinematics are fixed, the variation in total joint moment across stiffness conditions is negligible, and likely due to numerical instabilities in the optimization procedure (figure 3).

$$M^T = M_{soleus} - M_{TA} + M_{Spring} \tag{4}$$

5.2. Ankle Complex Model

Once the activation values obtained through the optimization routine are validated, they can be input in the developed ankle complex model, in order to analyse the resulting changes in mechanics and energetics of the muscles considered in the model.

5.2.1 Soleus

The soleus is the muscle which the exoskeleton most closely resembles. It is, therefore, the muscle which should benefit the most from the moment being added to the joint. Indeed, the force required of the plantarflexor, and thus its activa-

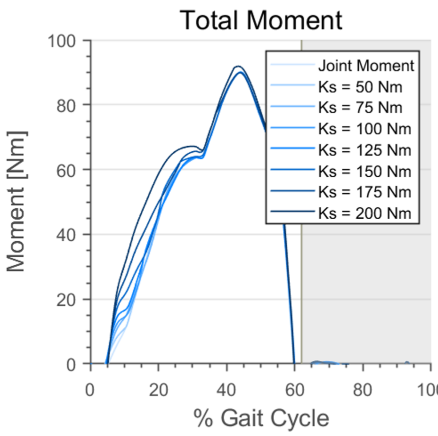

Figure 3: Total moment being generated at the joint. Darker blue indicates higher spring stiffness.

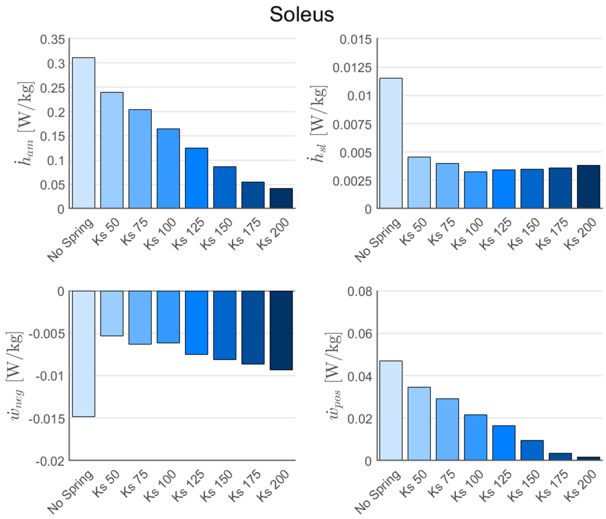

Figure 4: Heat and work rates, computed by the muscle energy expenditure model for the soleus, for different K_s values. Top left: $\dot{h}_{am}$; top right: $\dot{h}_{sl}$; bottom left: $\dot{w}_{neg}$; bottom right: $\dot{w}_{pos}$. Darker blue indicates higher spring stiffness.

tion, decrease steadily with the increase in spring stiffness (figure 7ab). Another change brought on by the exoskeleton is the increase in contractile element length (l_{ce}) during controlled dorsiflexion (figure 7c), which naturally leads to a decrease in tendon length (l_{se}) during the same period (figure 7d), since the length of the MTU remains unchanged. This increase in length causes the lengthening velocity of the CE (v_l^{ce}) to increase, since it has to lengthen more in the same time period. But it also leads to an increase in shortening velocity (v_s^{ce}), because at push off (PO) the CE is more distended, so it has to contract faster (figure 7e).

Initially, the averages of all portions of the muscle energy expenditure rate ($\dot{E}$) decrease with increasing spring stiffness (figure 4). But, while the activation and maintenance heat rate ($\dot{h}_{am}$) and the positive work rate ($\dot{w}_{pos}$) averages always decrease, the shortening and lengthening heat rate ($\dot{h}_{sl}$) and the negative work rate ($\dot{w}_{neg}$) averages start to increase after a certain threshold of K_s is reached. For $\dot{h}_{sl}$ this threshold is $K_s > 100$ Nm/rad, while for $\dot{w}_{neg}$ it is $K_s > 50$ Nm/rad.

Despite the increase in both $\dot{h}_{sl}$ and $\dot{w}_{neg}$ for stiffnesses over 100 Nm/rad, the total metabolic cost of the stance phase, in what concerns the main agonist decreases steadily with stiffness (figure 6). Indeed it seems to decrease more between successive stiffness conditions when stiffnesses are higher than 100 Nm/rad (on average 33%), than when they are lower (on average 20%). This suggests that the decrease in activation and maintenance heat rate and positive work, make up for the imposed changes in muscle mechanics which lead to the increase in shortening and lengthening heat rate and negative work.

5.2.1 Tibialis Anterior

The tibialis anterior, after initial contact occurs, acts as an antagonist for the remainder of the stance phase. So it will have to compensate for any excess moment the spring provides. Initially, an increase in spring stiffness requires an increase in force generation by the tibialis anterior at early and late stance, but afterwards, this increase extends to mid-stance (figure 7b). Naturally, in order for force production to increase, the muscle's activation must also increase (figure 7a). The behaviour of CE length and tendon length is the opposite of that observed for the soleus, that is, with increasing stiffness l_{ce} decreases while l_{se} increases (figure 7cd). For each value of K_s, lengthening velocity peaks at the instant at which the muscle stops producing force during mid-stance (figure 7e).

This increase, in both activation and force production, leads to the increase in the average of all portions of the muscle energy expenditure rate (figure 5). This increase across portions of the muscle energy expenditure rate leads to an increase in the metabolic cost associated with the tibialis anterior during stance (figure 6). Initially, for stiffnesses lower than 75 Nm/rad, the increase in metabolic cost between successive stiffness conditions is low (on average 14%), afterwards this increase is significantly higher (on average 41%). This is in line with the fact that, initially, the muscle is only required to produce more moment at early and late stance, and gradually it continues to be recruited throughout mid-stance.

5.2.1 Total Metabolic Cost

The total metabolic cost of the stance phase for

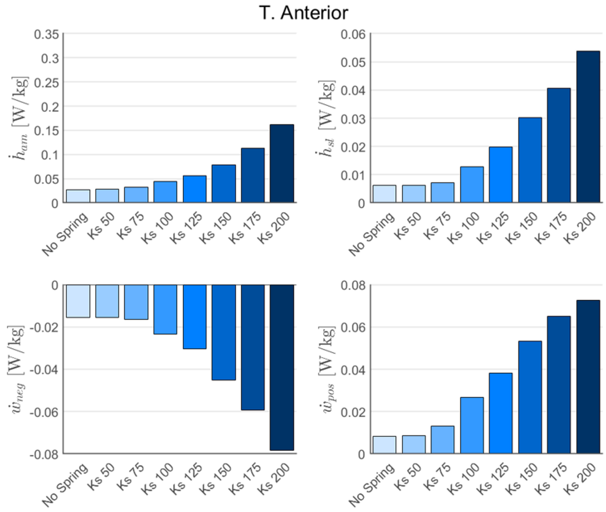

Figure 5: Heat and work rates, computed by the muscle energy expenditure model for the tibialis anterior, for different K_s values. Top left: $\dot{h}_{am}$; top right: $\dot{h}_{sl}$; bottom left: $\dot{w}_{neg}$; bottom right: $\dot{w}_{pos}$. Darker blue indicates higher spring stiffness.

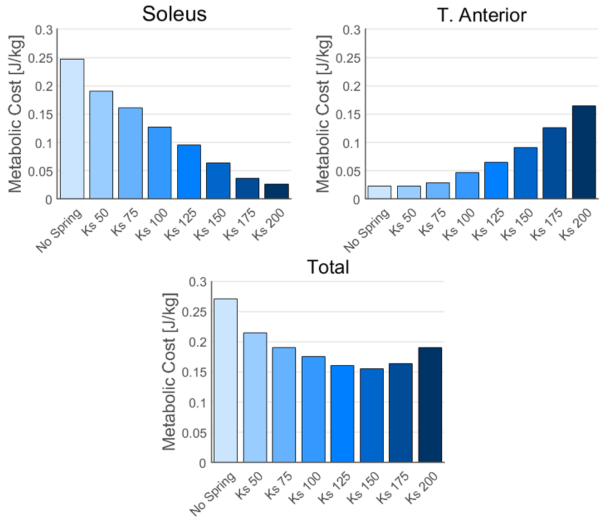

Figure 6: Metabolic cost of the stance phase, computed by the muscle energy expenditure model, for different K_s values. On top for both the soleus and tibialis anterior, below the total metabolic cost.

the agonist-antagonist pair analysed is simply the sum of the metabolic cost related to each individual muscle (figure 6). And, while increasing the stiffness of the ankle exoskeleton reduces significantly the metabolic cost associated with the soleus, it also leads to an increase of the metabolic cost associated with the tibialis anterior. Eventually, the increase in metabolic cost associated with the antagonist surpasses the benefit the exoskeleton provides to the agonist. There is, however, an optimal value of K_s for which the metabolic cost of the pair is minimal. This suggests that the soleus is a muscle whose activation and force production is expensive, and that reducing its recruitment, even if it means increasing it elsewhere, will reduce the total metabolic cost of the movement.

Indeed, for lower stiffnesses, the decrease in metabolic cost associated with the soleus is higher, on average 20% between consecutive K_s values, than the increase of the cost associated with the tibialis anterior, which is on average 14%. While, for higher stiffness values, the opposite is true, a 41% average increase of the metabolic cost associated with the tibialis anterior is observed while the reduction in soleus metabolic cost is on average 33%. The minimum in metabolic cost for the ankle complex musculature corresponds to a stiffness of 150 Nm/rad, which leads to a decrease of 0.1151 J/kg (42.57%) in metabolic cost.

5.3. Discussion

The results obtained in this computational analysis support the fact that, adding a parallel stiffness to the ankle joint, will have a beneficial impact on the metabolic cost associated with the joint during walking, at least up to a certain point. This benefit stems from changes in both the mechanics and energetics of the muscles spanning the joint.

The most important of these changes is the reduction in the moment that is required of the plantarflexor, since the parallel spring will generate part of the required moment. This, naturally, reduces the force and activation required of the muscle. As expected, this decrease in force and activation lead to a decrease in metabolic cost associated with the muscle. However, significant changes also occur to the intrinsic mechanics of the soleus muscle-tendon unit, namely the increase in l_{ce} (figure 7c) and decrease in l_{se} (figure 7d) all throughout controlled dorsiflexion. As mentioned in section 2.2, the plantarflexor MTU is tuned for optimal efficiency during unaided gait. Thus, adding a parallel spring disturbs these finely tuned mechanics. This effect is noticed particularly on the shortening and lengthening heat rate ($\dot{h}_{sl}$), which increases for higher stiffnesses. Since the muscle is no longer working in its optimal operating range it will expend more energy for shortening and lengthening its fibers.

Another consequence of the increase in l_{ce} during controlled dorsiflexion is the increase in lengthening velocity, as the muscle must stretch more in the same amount of time, and in shortening velocity, as it will have to contract more at push off. The increase in lengthening velocity is responsible for the increase in negative work rate observed with increasing stiffness. However, the increase in shortening velocity does not lead to an increase in positive

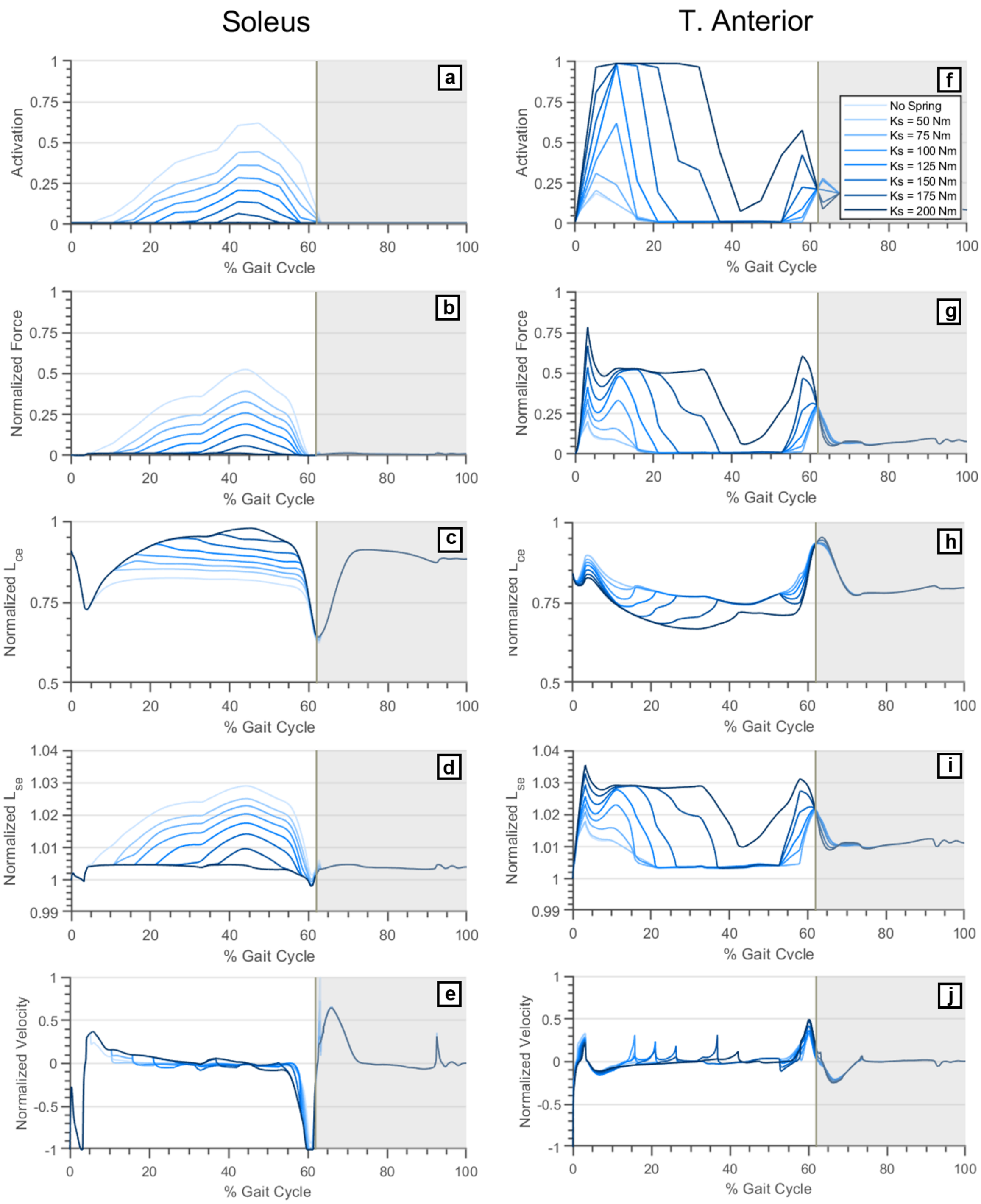

Figure 7: Time profiles for the mechanics of the soleus (left) and tibialis anterior (right). Darker blues indicate higher spring stiffness.

work rate. When the spring is added to the joint, during controlled dorsiflexion, the muscle is indeed producing more work than before and the tendon is accumulating less energy, while at the same time, the spring is stretching and accumulating the mechanical energy that the tendon would be accumulating otherwise. At push off the muscle contracts faster than it would if unaided, but producing less force, since it is not working in its optimal range and both the tendon and spring recoil, providing the remaining power boost required for PO (figure 1).

With the increasing stiffness, the reduction in the metabolic cost associated with the joint is curbed by the fact that, when the moment the spring provides surpasses the physiological moment generated by the soleus, it must be offset by the tibialis anterior. Initially, the tibialis anterior must only compensate for the increase in agonistic moment occurring at early and late stance, but eventually this increase extends to mid-stance, and the benefit experienced by the agonist no longer surpasses the added cost to the antagonist.

The findings in this work match those of Collins et al. [2] who, when measuring the metabolic cost, through indirect calorimetry, of the gait cycle in subjects wearing a passive exoskeleton, obtained a minimum for a stiffness of 180 Nm/rad, corresponding to a decrease in metabolic cost of 7.2%. They identified the increase in tibialis anterior activation as a possible contributing factor for the increase in metabolic cost for the higher stiffnesses tested. The effective stiffness of the exoskeleton they developed was around 33% lower than the nominal value. So in reality the minimum obtained by the researchers corresponds to a stiffness of around 120 Nm/rad, while the next value tested, 240 Nm/rad, actually corresponds to a nominal stiffness of about 160 Nm/rad, already higher than the stiffness for which the minimum metabolic cost was obtained in this work, which was 150 Nm/rad. The increase in knee muscle activity was another contributing factor identified, but the analysis of the knee joint was not addressed in this project. The other contributing factor identified was the increase in plantarflexor activity at the end of stance. This was not observed with the ankle model developed, because, while the work developed by Collins et al. was based on experimental acquisitions, the work developed for this dissertation was purely computational and considered the kinematics fixed, using the moments generated at the joints during unaided gait to estimate the activations. Therefore, a decrease in moment generated will, most certainly, be tied to a decrease in activation in this model.

The metabolic cost reduction obtained in this work is significantly higher than that obtained by Collins et al., a decrease of 42.57% compared to a decrease of 7.2% obtained by these researchers. This difference is due to the fact that the metabolic cost computed in this analysis only considers the contribution of the soleus and tibialis anterior. While the smaller decrease published by Collins et al. was obtained through indirect calorimetry and thus takes into account the whole body cost.

6. Conclusions

Human performance augmentation is one of the focuses of exoskeleton development [4]. One of the ways in which human performance augmentation is being pursued is in the reduction of the metabolic cost of human activities, such as gait. Despite being perfected over centuries of human evolution, locomotion still accounts for most of the energy expended during the day [2]. Reducing the cost of this activity could help individuals whose labor is walking intensive to maintain their quality of life for longer as well as to reduce their risk of injury.

A special interested is taken in assisting the ankle joint, since this is the joint which most energy expends during gait. In this dissertation, a computational model of the ankle musculoskeletal complex, comprised of the agonist-antagonist pair soleus - tibialis anterior, was developed. A parallel stiffness was added to the model, making use of a spring, in order to study the effect a passive device would have on the joint. Muscles were modeled as Hill-type muscles coupled to an energy expenditure muscle model. This allowed for the simultaneous analysis of changes to both mechanics and energetics of the muscles when spring stiffness was varied between 0 and 200 Nm/rad.

The analysis conducted using the musculoskeletal model developed, reached results which are congruent with those published by Collins et al. [2], which were obtained experimentally. The developed model estimates the metabolic cost associated with the individual muscles spanning the ankle during stance and allows the analysis of the changes that occur in the mechanics and energetics of the muscles due to the moment being added by the spring. For the data analysed, obtained for the Neuromuscular Locomotion Model [14, 17] walking at 1.3 m/s, a reduction of 42.57% in the metabolic cost associated with the ankle was obtained, when the stiffness added to the joint was 150 Nm/rad.

Provided with experimental acquisitions of the unaided gait cycle, the developed model can be used to estimate the metabolic cost associated with the ankle joint and help select the spring stiffness which would correspond to the greatest reduction in metabolic cost for the individual in question. This is an important contribute to streamlining the fabrication process of the device, since instead of having to test several spring stiffnesses in a laboratory setting, only the acquisition of the natural gait cycle is required to estimate the stiffness best suited to each individual.

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References

[1] D. A. Winter. *The Biomechanics and Motor Control of Human Gait*. University of Waterloo Press, Waterloo, 1st edition, 1987.

[2] S. H. Collins, M. Bruce Wiggin, and G. S. Sawicki. Reducing the energy cost of human walking using an unpowered exoskeleton. *Nature*, 522(7555):212–215, 2015.

[3] B. Chen, H. Ma, L. Y. Qin, F. Gao, K. M. Chan, S. W. Law, L. Qin, and W. H. Liao. Recent developments and challenges of lower extremity exoskeletons. *Journal of Orthopaedic Translation*, 5:26–37, 2016.

[4] L. M. Mooney, E. J. Rouse, and H. M. Herr. Autonomous exoskeleton reduces metabolic cost of human walking during load carriage. *Journal of NeuroEngineering and Rehabilitation*, 7:1–11, 2014.

[5] A. Machado. *Concept Development for Lower Limb Exoskeleton*. PhD thesis, Instituto Superior Técnico, 2020.

[6] T. M. Griffin, T. J. Roberts, and R. Kram. Metabolic cost of generating muscular force in human walking: Insights from load-carrying and speed experiments. *Journal of Applied Physiology*, 95(1):172–183, 2003.

[7] M. W. Whittle. *An Introduction to Gait Analysis*. Elsevier Ltd, 4th edition, 2007.

[8] A. D. Koelewijn, D. Heinrich, and A. J. van den Bogert. Metabolic cost calculations of gait using musculoskeletal energy models, a comparison study. *PLoS ONE*, 14(9):1–19, 2019.

[9] B. R. Umberger, K. G. Gerritsen, and P. E. Martin. A model of human muscle energy expenditure. *Computer methods in biomechanics and biomedical engineering*, 6(2):99–111, 2003.

[10] A. M. Gonabadi, P. Antonellis, and P. Malcolm. Differences between joint-space and musculoskeletal estimations of metabolic rate time profiles. *PLoS Computational Biology*, 16(10): 1–27, 2020.

[11] R. W. Jackson and S. H. Collins. An experimental comparison of the relative benefits of work and torque assistance in ankle exoskeletons. *Journal of Applied Physiology*, 119(5): 541–557, 2015.

[12] R. W. Jackson, C. L. Dembia, S. L. Delp, and S. H. Collins. Muscle-tendon mechanics explain unexpected effects of exoskeleton assistance on metabolic rate during walking. *Journal of Experimental Biology*, 220(11):2082–2095, 2017.

[13] P. Bujalski, J. Martins, and L. Stirling. A Monte Carlo analysis of muscle force estimation sensitivity to muscle-tendon properties using a Hill-based muscle model. *Journal of Biomechanics*, 79:67–77, 2018.

[14] H. Geyer and H. Herr. A Muscle-reflex model that encodes principles of legged mechanics produces human walking dynamics and muscle activities. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 18(3): 263–273, 2010.

[15] M. Ackermann. *Dynamics and Energetics of Walking with Prostheses*. PhD thesis, University of Stuttgart, 2007.

[16] R. H. Miller. A comparison of muscle energy models for simulating human walking in three dimensions. *Journal of Biomechanics*, 47(6): 1373–1381, 2014.

[17] H. Geyer, A. Seyfarth, and R. Blickhan. Positive force feedback in bouncing gaits? *Proceedings of the Royal Society B: Biological Sciences*, 270(1529):2173–2183, 2003.

[18] S. Song and H. Geyer. Regulating speed and generating large speed transitions in a neuromuscular human walking model. *Proceedings - IEEE International Conference on Robotics and Automation*, pages 511–516, 2012.