The AMP-Foot 2.0:
Mimicking Intact Ankle Behavior with a Powered Transtibial Prosthesis

Pierre Cherelle, Arnout Matthys, Victor Grosu, Bram Vanderborght and Dirk Lefeber

Abstract— **Almost all of the transtibial prostheses that are available on the market are purely passive devices. They store energy in an elastic element at the beginning of a step and release it at the end in order to move the body forward. The main problem with these prostheses is that only the energy that has been stored in the elastic element is used for the push-off, unlike for non-pathological ankles where the muscles provide extra energy. There are a few prostheses who use active components for this energy input. In this article, the authors propose a new design of an energy efficient, powered transtibial prosthesis to mimic intact ankle behaviour, the AMP-Foot 2.0. The main idea behind our research is to have the actuator work longer with a lower power rating while the produced energy is stored in elastic elements and released when needed for propulsion. The device is designed to provide 100% of push-off for a 75 kg subject walking at normal cadence on ground level.**

I. INTRODUCTION

Until the '80s the focus in the design of prosthetic feet was on trying to restore basic walking and enabling the amputee to fulfill basic tasks. The prostheses and materials used slowly evolved until the so-called conventional feet, which were still very basic but from these prostheses onward things like weight of the prosthesis and amputee comfort became more important. The most common of these conventional feet are probably the SACH-foot, or Solid Ankle Cushion Heel [1], and the uni-axial foot [2].

Driven by the desire of amputees to walk more naturally, to reduce metabolic cost and even in some cases practicing sports, prosthetic feet were significantly improved over time. In general, today's prosthetic feet can be classified into three categories: Conventional feet (CF), "Energy-storing-and-returning" (ESR) feet and bionic feet. ESR feet, compared to CF feet, are capable of storing energy in elastic elements and returning the major part of it to assist in forward propulsion [3]. Hereby the push-off is improved and thus moving forward is made easier for the amputee. Examples of the first ESR feet are the Seatle foot [3] and the Jaipur foot [1]. Thanks to better knowledge and understanding of the human gait and biomechanics, new types of ESR prostheses [4] were developed as the Flex-foot, the Springlite foot, the VariFlex, the Re-Flex, etc.

In 2010, researchers at the Vrije Universiteit Brussel, Belgium, have developed the Ankle Mimicking Prosthetic Foot (AMP-Foot 1.0) [5], an articulated ESR-type foot shown in Fig. 1. It is one of the first devices to use locking mechanisms to store harvested energy during the dorsiflexion (DF) phase of stance, and to release it at push-off, compared to conventional ESR feet which acts like torsion springs.

To improve even more the push-off properties of passive prostheses, Collins et al. [6] have developed the so-called Controlled Energy Storing and Returning foot (CESR foot). Rather than storing energy during stance, the CESR foot uses the weight of the body at initial contact to harvest energy and releases it when needed [7].

All of the prostheses described so far use only the energy provided by the amputee himself to mimic the behavior of a healthy ankle. But there are also possibilities to inject energy into the system using an external power source. Currently, most of the developed powered devices are still on a research level. To name a few, at the Massachusetts Institute of Technology, the MIT Powered Foot Prosthesis [8] has been developped using a high power electric actuator with series elasticity, better known as the Series Elastic Actuator (SEA) [9]. At the Arizona State University, the SPARKy project (Spring Ankle with Regenerative Kinetics) [10] uses a Robotic Tendon actuator [11] to power an artificial foot. At the Vrije Universiteit Brussel, a transtibial (TT) prosthesis using Pleated Pneumatic Artificial Muscles (PPAM) was developed [12] to demonstrate the importance of push-off during gait.

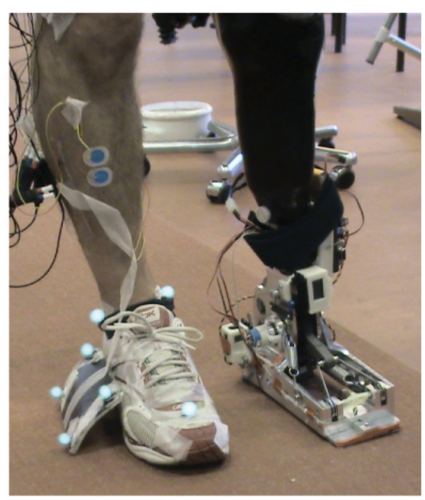

Fig. 1. AMP-Foot 1.0.

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When studying the state-of-the-art in TT prostheses, one can conclude that passive energy storing devices (ESR feet) are energetically efficient but do not provide the extra power needed for propulsion during walking. On the other hand, actuated devices are able to provide the necessary energy, but need heavy and bulky actuators capable of producing high torques in small periods of time. With the AMP-Foot 2.0, the authors propose a new concept which enables the use of low power actuators storing energy in springs and releasing it when needed. The AMP-Foot 2.0 is presented in the next section. The mechanical design of the prosthesis, which is currently under construction will be described in detail in section II-B, followed by conclusions in the last section.

II. THE AMP-FOOT 2.0

In this section, the working principle and mechanical design of the AMP-Foot 2.0 is described. From biomechanical data analysis [13], it is known that a healthy ankle joint produces energy during walking. As mentioned before, to imitate this, an external power source is needed. The main objective of this research is to retrieve as much energy as possible from the gait and to implement an electric actuator with minimized power consumption. The idea behind the AMP-Foot 2.0 is to use a spring, called the plantar flexion (PF) spring, to accumulate energy from the dorsiflexion phase of stance while the actuator is injecting energy into another spring, called the push-off (PO) spring, during the complete stance phase. By using a locking system, the energy stored in the PO spring, before HO occurs, is kept into the system and released for push-off. This way it is possible to reduce the actuator's power and thus its size while providing the full torque needed for propulsion during walking.

A. Working Principle

In Fig. 2 the essential parts of the AMP-foot 2.0 are drawn. The prosthesis consists of 3 bodies (the leg, the foot and a lever arm) pivoting around a common rotation axis A (the ankle axis). θ is the angle between the foot and the leg while ϕ represents the angle between the foot and the lever arm. The PF spring (stiffness k_1) is placed between a fixed point p on the foot and a cable that runs over a pulley a to the lever arm at point b and is attached to the lever arm at point c. The distance between the rotation axis and respectively the pulley a, point b and c, is taken to be L_1, L_2 and L_3. The PO spring (stiffness k_2) is placed between the motor-gearbox-ballscrew assembly and a fixed point d on the lever arm. The distance between the rotation axis and the fixed point d is L_4.

A critical part, which is not drawn in Fig. 2 is the mechanism which locks the lever arm to the leg when energy is injected into the system. The working principle of this locking mechanism will be discussed in more details in section II-B.

To illustrate the behavior of the AMP-Foot 2.0, one stride of a gait cycle will be discussed in detail. Fig. 3 shows the torque-angle characteristic of the AMP-Foot 2.0 and of an intact ankle [13]. The stance phase, which is illustrated in Fig. 4, is subdivided into several key points [14] which are used to describe the working principle of the AMP-Foot 2.0. A detailed description of one gait cycle is given below.

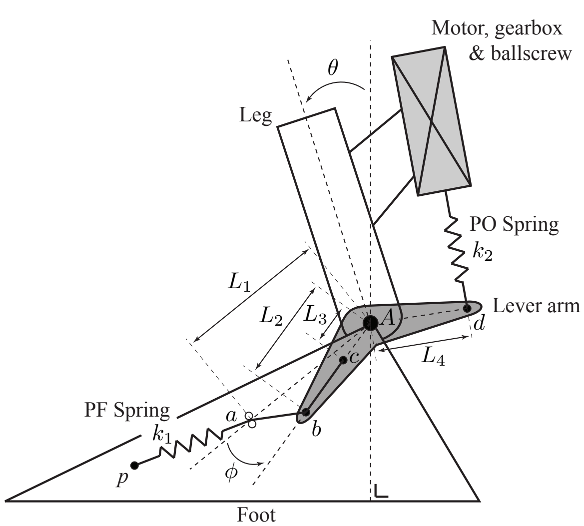

Fig. 2. Schematics of the AMP-foot 2.0.

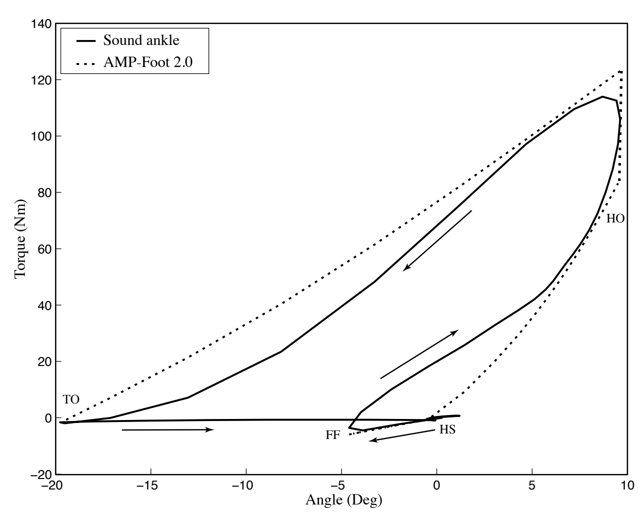

Fig. 3. Torque - Angle characteristics of a sound ankle and of the AMP-Foot 2.0.

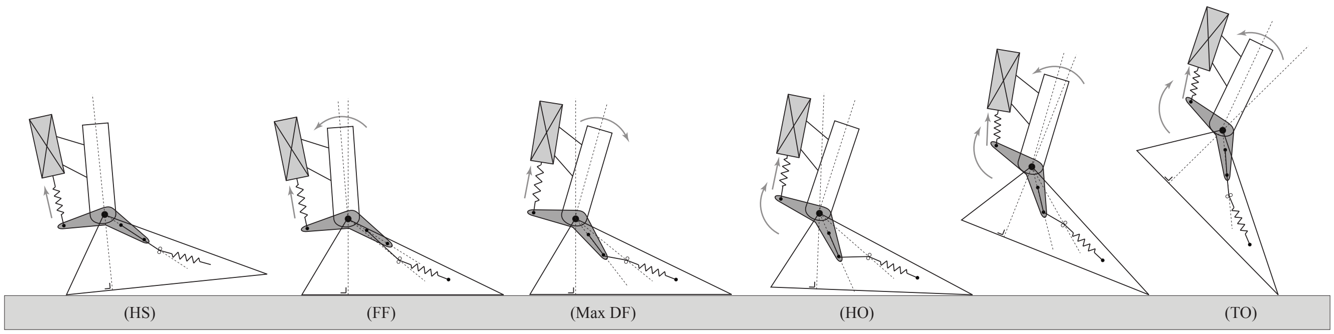

Fig. 4. Working principle of the AMP-Foot 2.0 during the stance phase.

1) From heel strike (HS) to foot flat (FF): A step is initiated by touching the ground with the heel. During this phase the leg is moving backwards until θ $(=\phi)$ reaches approximatly $-5°$. Since the lever arm is fixed to the leg, the PF spring is elongated and generates a DF torque around the ankle axis which is calculated as follows:

$$T_1 = k_1(l_1 - l_0 + V_{0,1})\frac{L_1 L_3}{l_1} sin\phi \tag{1}$$

in which
T_1 = Torque applied by the PF spring to the lever arm and thus to the ankle.
k_1 = Spring constant of the PF spring.
l_0 = Distance between the fixed points a and c when $\phi = 0$.
$V_{0,1}$ = Pretension of the PF spring.
l_1 = Distance between the fixed point a and c or

$$l_1 = \sqrt{L_1^2 + L_3^2 - 2L_1 L_3 cos(\phi)} \tag{2}$$

During this period the motor loads the PO spring. But since the motor is attached to the leg and lever arm is locked to the leg, the PO spring is loaded without delivering torque to the ankle joint. The prosthesis is not affected by the work done by the actuator.

2) From (FF) to heel off (HO): In this second phase, the leg moves from $\theta = -5°$ to $\theta = +10°$. Until the leg reaches $\theta = 0°$ the torque of the system is given by (1). From $\theta = 0°$ to $\theta = +10°$ the lever arm length is adjusted and thus the torque becomes:

$$T_1 = k_1(l_1 - l_0 + V_{0,1})\frac{L_1 L_2}{l_1} sin\phi \tag{3}$$

in which L_2 represents the distance between the ankle axis and the fixed point b. This is done by using two different connection points b and c (Fig. 2), on the lever arm, which are respectively active when $\theta > 0$ and $\theta < 0$. This way it is possible to mimic the change in stiffness of a sound ankle. During this phase the motor is still injecting energy into the system by loading the PO spring.

3) At heel off (HO): Because the angle between the PO spring and the lever arm is $\pi/2$, the torque excerted by the spring (no pretension) on the lever arm is given by

$$T_2 = k_2 l_2 L_4 \tag{4}$$

with
T_2 = Torque applied to the lever arm by the PO spring.
k_2 = Spring constant of the PO spring.
l_2 = Elongation of the PO spring.

The torque T_1 excerted by the PF spring on the lever arm is given by (3). At the moment of HO, all the energy which is stored into the PO spring is fed to the system by releasing the locking mechanism. $T_1 \leq T_2$ and as a result of this, both PF and HO springs tend to rotate the lever arm with an angle ψ to a new equilibrium position. In other words, T_1 and T_2 respectively evolves to new values T_1' and T_2' such that $T_1' = T_2' = T'$ with $T_1' \geq T_1$ and $T_2' \leq T_2$. The torque at the ankle becomes

$$T' = k_1(l_1' - l_0 + V_{0,1})\frac{L_1 L_2}{l_1'} sin(\phi + \psi) \tag{5}$$

in which

$$l_1' = \sqrt{L_1^2 + L_3^2 - 2L_1 L_3 cos(\phi + \psi)} \tag{6}$$

The effect of this is an instantaneous increase in torque and decrease in stiffness of the ankle joint as depicted in Fig. 3.

4) from HO to toe off (TO): In the last phase of stance, the torque is decreasing until toe off (TO) occurs at $\theta = -20°$. Since the two springs are now connected in series, the rest position of the system has changed according to the elongation of the PO spring. As a result of this a new equilibrium position is set to approximately $\theta = -20°$. The actuator is still working during this phase.

5) Swing phase: After TO, the leg enters into the so called swing phase in which the whole system is resetted. While the motor turns in the opposite direction to bring the ballscrew assembly back to its initial position, return springs are used to set θ back to $0°$ and to close the locking mechanism. From now on, the device is ready for a new step.

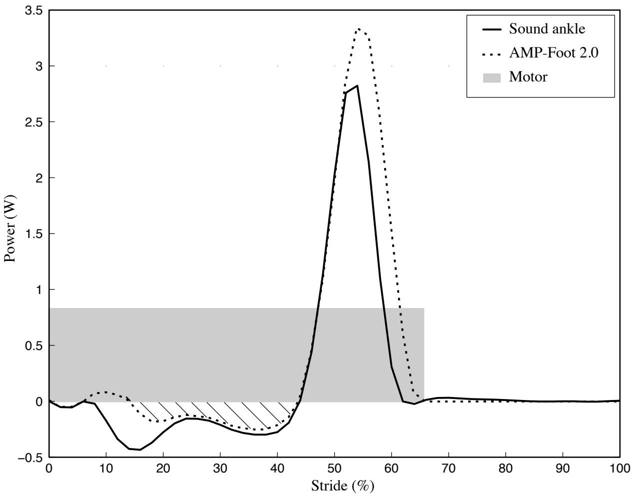

Fig. 5. Ankle power during one stride. The solid line represents the power generation of a sound ankle while the dotted line represents the resulting power of the AMP-Foot 2.0. The gray rectangle shows how the actuator power is spread over one gait cycle while the shaded area represents the energy harvested with the PF spring.

TABLE I

LEVER DIMENSIONS AND SPRING STIFFNESSES

$L_1 = 80\ mm$	$k_1 = 300\ N/mm$
$L_2 = 60\ mm$	$V_{0,1} = 5\ mm$
$L_3 = 30\ mm$	$k_2 = 120\ N/mm$
$L_4 = 60\ mm$	$V_{0,2} = 0\ mm$

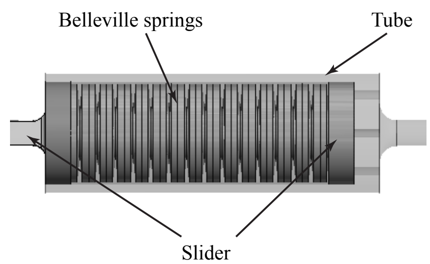

Fig. 6. Section representation of a disc spring assembly. 29 disc springs are stacked in series on a slider which moves into a tube.

Compared to other powered TT prostheses, the electric actuator of the AMP-Foot 2.0 is working the whole time. During stance the motor actuates a spring-like load while the reset requires almost no power. This way it is possible to deliver the necessary power and torque for an ankle joint during walking in normal conditions with a low power actuator. As a result of this the electric motor can be downsized considerably compared to the peak power requirements of a non-pathological ankle. Fig. 5 shows the power generation of a sound ankle and of the AMP-Foot 2.0 during one gait cycle. Clearly, the mechanical power during one stride remains the same, but instead of providing it all at HO, the mechanics of the AMP-Foot 2.0 allows an accumulation of energy during the complete stance phase.

B. Mechanical Design

According to Winter [13] a 75 *kg* subject walking at normal cadence (ground level) produces a maximum joint torque of 120 *Nm* at the ankle. This has been taken as a criterion. Moreover, an ankle articulation has a moving range from approximatly $+10°$ at maximal dorsiflexion to $-20°$ at maximal plantarflexion. Therefor a moving range of $-30°$ to $+15°$ has been chosen for the system to fulfill the requirements of the ankle anatomy. The length of the lever arms, spring stiffnesses and pretension named in Fig. 2 are given in TABLE I

For the PF spring (k_1), a belleville spring assembly, which is shown in Fig. 6, is used because of its compactness en ability to provide extremely high forces. This assembly consists of a tube in which a slider is moving to compress the disc springs. To achieve the desired, as linear as possible, spring characteristic, 29 belleville springs are stacked in series. For the PO spring (k_2), two tension springs with each a stiffness of 60 N/mm are used.

The two spring assemblies are equipped with strain gauges which allows a force measurement with a resolution of $\pm$ 1.5 N. To measure the position of the lever arm, and the leg with respect to the foot, two absolute magnetic encoders (Austria Micro Systems AS5055) are used with a resolution of $\pm$ 0.08°. While the magnets of the encoders are glued to the ankle axis (which is fixed to the foot), the two hall sensors are fixed on the lever arm, respectively on the leg. As a result of this, the resulting torque at the ankle can be calculated using the mathematical model of the mechanical system which has been discussed before. To detect the important triggers during the stance phase (IC, FF, HO, TO), two Force Sensing Resistors (FSR) are placed on the foot sole: one at the heel and one at the toes. These triggers will be used to control the motor and to lock or unlock the locking mechanism. A Maxon Brushed DC motor (60 W) has been chosen in combination with a gearbox and ballscrew assembly, which is described in

TABLE II

MOTOR AND TRANSMISSIONS

Motor	Maxon RE 30 - 60 *W* $T_{cont.} = 51.7\ mNm$ $T_{peak} = 150\ mNm$
Transmission stage 1	Maxon GP32BZ i = 5.8:1
Transmission stage 2	Maxon ballscrew GP32S ∅10x2

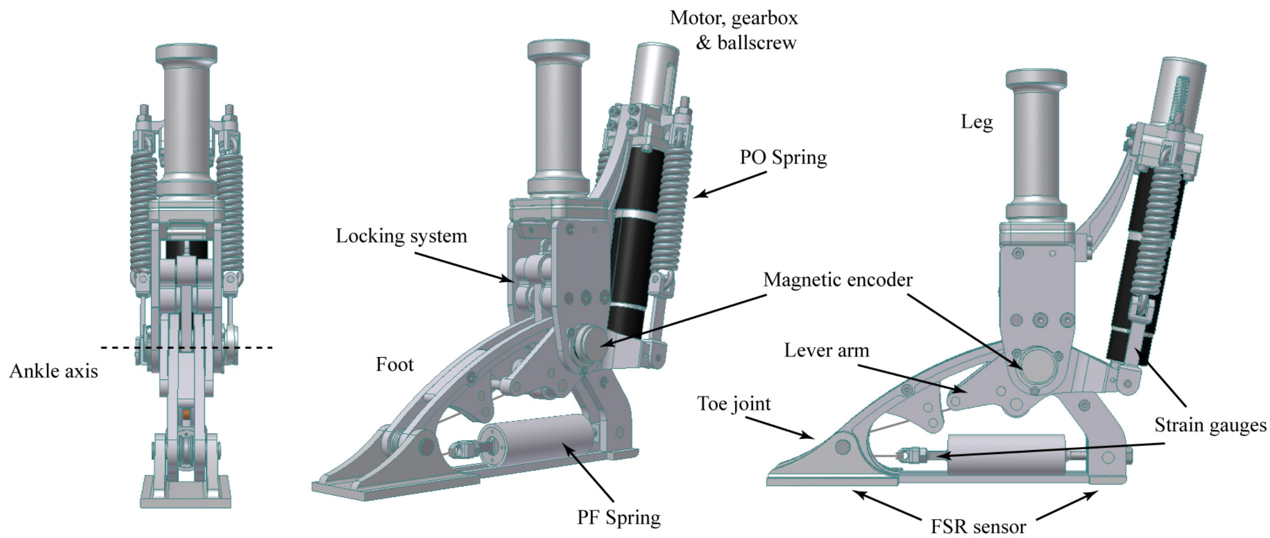

Fig. 7. CAD representation of the AMP-Foot 2.0.

TABLE II. The positioning of the motor and other hardware have been chosen in view of the range of motion and optimized for compactness of the system. Fig. 7 shows the CAD design of the AMP-Foot 2.0 which is currently under construction. Moreover, the AMP-Foot 2.0 is equipped with a wireless data transfer system for data acquisition.

As mentioned before, a critical part of this mechanical system is the locking mechanism. The requirements for this locking system are:

- Ability to withstand high forces.
- As compact and lightweight as possible.
- Locking at a fixed angle.
- Unlock under maximum load.

To achieve this, it has been chosen to work with a four bar linkage moving in and out of its singular position. This principle has already proved its effectiveness in [15], where it is used to lock the knee joint of a walking robot. Fig. 8 shows the schematics of the four bar linkage when locked (a) and opened (b). When the four bar linkage is set in its singular position, it is in unstable equilibrium.

Therefor to ensure locking, the system is allowed to move a bit further than its singular position. When the singular position is past, the load forces the mechanism to continue moving in the same direction. To keep it in equilibrium, a mechanical stop blocks the system. A solenoid is then used to push the mechanism back past its singular position when triggered. Because close to its singular position, the transmission coefficient of the four bar linkage tends to infinity, the resulting force (or torque) which has to be applied to unlock the system is greatly reduced. Fig. 9 shows the transmission coefficient and the resulting force necessary for unlocking under maximal load in function of the lever arm angle.

It can be estimated that the maximum resulting load which can be applied to the lever arm, e.g. when PF spring and PO spring are fully extended (at maximal dorsiflexion), is more or less 40 Nm. In this case, and if the four bar mechanism

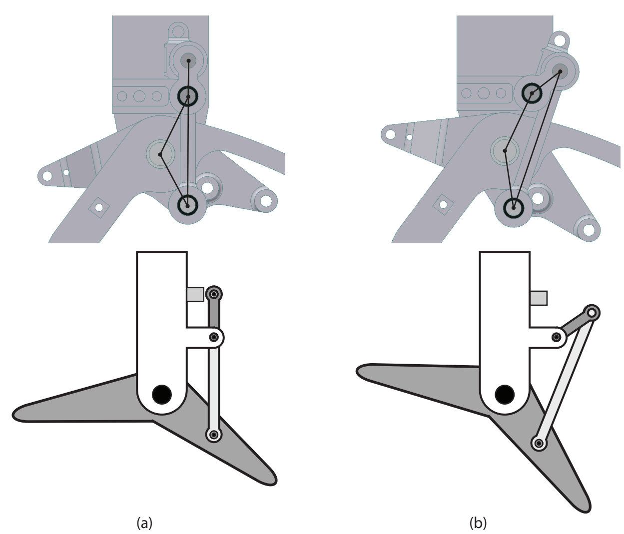

Fig. 8. CAD representation and schematics of the four bar mechanism in locked (a) and unlocked (b) position.

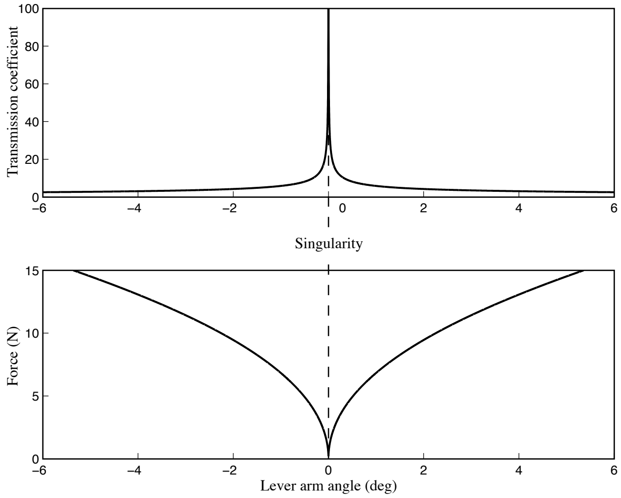

Fig. 9. Transmission Coefficient and resulting force of the four bar linkage mechanism close to its singular position (0°).

is past its singular position of a few degrees, the resulting force needed for unlocking should be less than 10 *N*. Of course, this is a worst case senario. Having the PO spring completely extended at maximal dorsiflexion is certainly not optimal. This would mean the motor has to stop moving between HO and TO. A better control strategy is to make the motor move during the complete stance phase as shown in Fig. 5. Therefor, depending on the way the motor is controlled, the resulting force needed to unlock the four bar linkage will be reduced.

III. CONCLUSIONS

In this article, the authors propose a new design of an energy efficient powered transtibial prosthesis mimicking non-pathological ankle behaviour, the AMP-Foot 2.0. The inovation of this study is to harvest energy from motion with a PF spring while storing energy produced by a low power electric motor into a PO spring. This energy is then released with a delay at the right time for push-off thanks to the use of a locking system. The prosthesis is designed to provide a peak output torque of 120 *Nm* with a range of motion of 45° to fullfill the requirements of a 75 *kg* subject walking on level ground at normal cadence. Its total weight is estimated at ± 2 *kg* which corresponds to the requirements of a healthy foot. The AMP-Foot 2.0 is currently under construction. Preliminary tests should be performed in february 2012.

IV. ACKNOWLEDGMENTS

This work has been funded by the European Commissions 7th Framework Program as part of the project VIACTORS under grant no. 231554.

REFERENCES

[1] A. P. Arya, A. Lees, H. C. Nerula, and L. Klenerman, "A biomechanical comparision of the sach, seattle and jaipur feet using ground reaction forces." *Prosthetics and Orthotics International*, vol. 19, no. 1, pp. 37–45, 1995.

[2] J. C. H. Goh, S. E. Solomonidis, W. D. Spence, and J. P. Paul, "Biomechanical evaluation of sach and uniaxial feet." *Prosthetics and Orthotics International*, vol. 8, no. 3, pp. 147–154, 1984.

[3] B. J. Hafner, J. E. Sanders, J. M. Czerniecki, and J. Fergason, "Transtibial energy-storage-and-return prosthetic devices: A review of energy concepts and a proposed nomenclature," *Journal of Rehabilitation Research and Development*, vol. 39, no. 1, pp. 1–12, 2002.

[4] R. Versluys, P. Beyl, M. V. Damme, A. Desomer, R. V. Ham, and D. Lefeber, "Prosthetic feet: State-of-the-art review and the importance of mimicking human ankle–foot biomechanics," *Disability and Rehabilitation: Assistive Technology*, vol. 4, no. 2, pp. 65–75, 2009.

[5] B. Brackx, M. V. Damme, A. Matthys, B. Vanderborght, and D. lefeber, "Passive ankle-foot prosthesis prototype with extended push-off." *Advanced Robotics (in review)*, 2011.

[6] S. H. Collins and A. D. Kuo, "Controlled energy storage and return prosthesis reduces metabolic cost of walking," *ISB XXth Congress - ASB 29th Annual Meeting*, 2005.

[7] A. Segal, K. E. Zelik, G. K. Klute, D. X. Morgenroth, M. E. Hahn, M. S. Orendurff, P. G. Adamczyk, S. H. Collins, A. D. Kuo, and J. M. Czerniecki, "The effects of a controlled energy storage and return prototype prosthetic foot on transtibial amputee ambulation." *Human Movement Science*, pp. 1–30, 2010.

[8] S. K. Au and H. M. Herr, "Powered ankle-foot prosthesis," *IEEE Robotics & Automation Magazine*, vol. 15, no. 3, pp. 52–59, 2008.

[9] G. A. Pratt and M. M. Williamson, "Series elastic actuators," *Proceedings of the IEEE/RSJ International Conference on Intelligent Robots and Systems*, pp. 399–406, 1995.

[10] J. Hitt, T. Sugar, M. Holgate, R. Bellman, and K. Hollander, "Robotic transtibial prosthesis with biomechanical energy regeneration," *Industrial Robot: An International Journal*, vol. 36, no. 5, pp. 441–447, 2009.

[11] K. W. Hollander, R. Ilg, and T. G. Sugar, "Design of the robotic tendon," *Design of Medical Devices Conference*, pp. 1–6, 2005.

[12] R. Versluys, A. Desomer, G. Lenaerts, O. Pareit, B. Vanderborght, G. V. der Perre, L. Peeraer, and D. Lefeber, "A biomechatronical transtibial prosthesis powered by pleated pneumatic artificial muscles," *International Journal of Modelling, Identification and Control*, vol. 4, no. 4, pp. 1–12, 2008.

[13] D. A. Winter, "The biomechanics and motor control of human gait: Normal, elderly and pathological." *Waterloo Biomechancs*, vol. 2, 1991.

[14] J. Perry and J. M. Burnfield, *Gait Analysis: Normal and Pathological Function*. SLACK Incorporated, 2010.

[15] G. van Oort, R. Carloni, D. J. Borgerink, and S. Stramigioli, "An energy efficient knee locking mechanism for a dynamically walking robot." *IEEE International Conference on Robotics and Automation*, pp. 9–13, 2011.