Use of Compliant Actuators in Prosthetic Feet and the Design of the AMP-Foot 2.0

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Abstract. From robotic prostheses, to automated gait trainers, rehabilitation robots have one thing in common: they need actuation. The use of compliant actuators is currently growing in importance and has applications in a variety of robotic technologies where accurate trajectory tracking is not required like assistive technology or rehabilitation training. In this chapter, the authors presents the current state-of-the-art in trans-tibial (TT) prosthetic devices using compliant actuation. After that, a detailed description is given of a new energy efficient below-knee prosthesis, the AMP-Foot 2.0.

1 Introduction

Experience in clinical and laboratory environments indicates that many trans-tibial (TT) amputees using a completely passive prosthesis suffer from non-symmetrical gait, a high measure of perceived effort and a lack of endurance while walking at a self-selected speed [28, 20, 3]. Using a passive prosthesis means that the patient's remaining musculature has to compensate for the absence of propulsive ankle torques. Therefore, adding an actuator to an ankle-foot prosthesis has the potential to enhance a subjects mobility by providing the missing propulsive forces of locomotion. In the growing field of rehabilitation robotics, prosthetics and wearable robotics, the use of compliant actuators is becoming a standard where accurate trajectory tracking is not required. Their ability to safely interact with the user and to absorb large forces due to shocks makes them particularly attractive in applications based on physical human-robot interactions. The approach based on compliance on a mechanical level (i.e. passive compliance), compared to introduced compliance on the control level (i.e. active compliance), ensures intrinsic compliance of the device

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at all time, enhancing hereby system safety. Therefore, this type of actuator is preferred in novel rehabilitation robots where safe human-robot interaction is required. In the particular case of trans-tibial (TT) prostheses, compliance of the actuation provides even more advantages. Besides shock absorption in case of collision with objects during walking, energy provided by the actuator (e.g. electric motor) can be stored into its elastic element (e.g. spring in series). This energy can be kept for a moment and released when needed to provide propulsion of the subject [7]. As a result of this, the electric drive can be downsized so as the overall weight and inertia of the prosthetic device to improve the so-called 3C-level, i.e. comfort, control and cosmetics.

Compliant actuators can be divided into actuators with fixed or variable compliance. Examples of fixed compliance actuators are the various types of series elastic actuators (SEA) [19], the bowden cable SEA [22] and the Robotic Tendon Actuator [14] to name a few. On the other hand the PPAM (Pleated Pneumatic Artificial Muscles) [25], the MACCEPA (Mechanically Adjustable Compliance and Controllable Equilibrium Position Actuator) [6, 8] and the Robotic Tendon with Jack Spring actuator [15, 16] are examples of variable stiffness actuators. For a complete state-of-the-art in compliant actuation, the authors refer to [9].

In this chapter, the authors present the current state-of-the-art in powered trans-tibial prostheses using compliant actuation and a brief analysis of their working principles. A description of the author's latest actuated prosthetic foot design will then be given, i.e. the AMP-Foot 2.0. Conlusions and future work will be outlined at the end of the chapter.

2 Powered Prosthetic Feet

In this section, the authors present the current state-of-the-art in powered ankle-foot prostheses, better known as ”bionic feet”, in which the generated power and torques serve for propulsion of the amputee. The focus is placed on devices using compliant actuators. For a complete state-of-the-art review of passive TT prosthesis comprising ”Conventional Feet” and ”Energy Storing and Returning” (ESR) feet, the authors refer to [24].

2.1 Pneumatically Actuated Devices

Pneumatic actuators are also known as ”antagonistically controlled stiffness” actuators [9] since two actuators with non-adaptable compliance and non-linear force displacement characteristics are coupled antagonistically. By controlling both actuators, the compliance and equilibrium position can be set.

Klute et al. [17] have designed an artificial musclo-tendon actuator to power a below-knee prosthesis. To meet the performance requirements of an artificial *triceps surae* and *Achilles* tendon, an artificial muscle, consisting of two flexible pneumatic actuators in parallel with a hydraulic damper, and placed in series with a bi-linear, two-spring implementation of an artificial tendon, was build into the ankle-foot prosthesis.

Goldfarb et al. [21] at Vanderbilt University have developed a powered transfemoral prosthesis using knee and ankle pneumatic actuation.

Developed within the Robotics & Multibody Mechanics Research Group at Vrije Universiteit Brussel, Belgium, the Pleated Pneumatic Artificial Muscle (PPAM) [23] was originally intended to be used in bipedal walking robots. It is a lightweight, air-powered, muscle-like actuator consisting of a pleated airtight membrane. Its advantage compared to other artificial muscle comes from the unfolding of the pleated membrane. Because of this there is virtually no threshold pressure, hysteresis is reduced when compared to other types of muscles, and contractions of over 40% of the initial length are possible. Whithin the IPAM (Intelligent Prosthesis using Artificial Muscles) Project [25], a TT prosthesis using Pleated Pneumatic Artificial Muscles was developed to demonstrate the importance of push-off during gait [25].

In general, drawbacks of pneumatic systems are the high cost of pressurized air production and supply requirements for autonomy. Therefore, electric actuators are preferred in novel prosthetic designs.

2.2 Electrically Actuated Devices

At the Massachusetts Institute of Technology (MIT), the Powered Foot Prosthesis [1] has been developped using a combination of a spring and a high power series elastic actuator. Its working principle consists of loading a spring during the controlled dorsiflexion phase and to activate a torque source (SEA) in parallel when peak power is needed. As a result of this, energy is added to the system to provide push-off. A peak output torque of 140 Nm and power output of 350W is applied with a torque bandwidth up to 3.5Hz. This prosthetic device has shown its effectiveness by improving metabolic economy of walking individuals with TT amputation [2], on average by 14% compared to evaluated conventional prostheses. Further research at the MIT led to the developement of the Powerfoot BiOM sold by iWalk [10]. The BiOM is a Bionic lower-leg system to replace lost Muscle function that approximates the action of the ankle, Achilles tendon and calf muscles by propelling the amputee upwards and forwards during walking.

At the Arizona State University, the SPARKy project (Spring Ankle with Regenerative Kinetics) [12] uses a Robotic Tendon actuator (including a 150W DC motor) [14] to provide 100% of the push-off power required for walking while maintaining intact gait kinematics. The first prototype (SPARKy 1) [11] was shown to store and release approximately 16J of energy per step while an intact ankle of a 80Kg subject at 0.8Hz walking rate needs approximately 36J [13]. A second prototype was built (SPARKy 2) with a lighter and more powerfull roller screw transmission and brushless DC motor. Both design's working principle rely on a SEA attached between the heel and the leg. This robotic tendon is controlled to provide the ankle torque and power necessary for propulsion during gait. The third prototype (SPARKy 3) [4] was designed to actively control both inversion and eversion as well as plantarflexion and dorsiflexion while providing high power for running and jumping.

At the Vrije Universiteit Brussel, a compact, low-weight and energy efficient trans-tibial prosthesis powered by electric drives was proposed to improve the amputee's gait [26]. The challenge was to design a device respecting the ankle-foot requirements that mimics a natural ankle behavior during walking. It was shown that by incorporating a modified MACCEPA [8] into the design, an acceptable approximation of the ankle characteristic is obtained. The prosthesis contains two unidirectional springs in parallel, connected to two lever arms. By connecting one of the lever arms to a locking mechanism it is shown that the energy efficiency is greatly improved. The actuation comprises a 150W motor with gearhead transmission connected to a ball screw mechanism through a timing belt. The Powered Below-knee Prosthesis's behavior is adjustable depending on amputee's gait speed by regulating the pretension of the springs. It is capable of providing 100% of the required push-off power, consuming only 22.19J per step for a 75Kg subject walking at normal cadence on level ground.

Further research at the Robotics & Multibody Mechanics Research Group [5] led to the design and development of the Ankle Mimicking Prosthetic Foot (AMP-Foot) 2.0. Fig. 1 shows some of the named prosthetic devices.

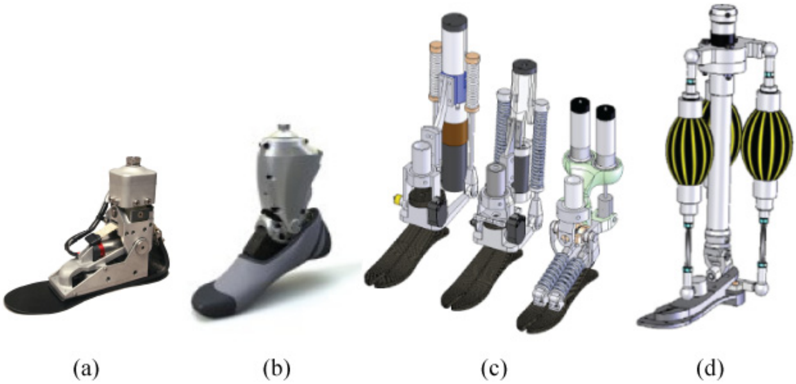

Fig. 1 (a) MIT Power Foot Prosthesis. (b) The BiOM from iWalk. (c) SPARKy 1, 2 and 3 (from left to right). (d) Trans-tibial Prosthesis using Pleated Pneumatic Artificial Muscles.

3 The Amp-Foot 2.0: A New Energy Efficient Concept

The main objective of this research is to harvest as much energy as possible from the gait and to implement an electric actuator with minimized power consumption. The concept of the AMP-Foot 2.0 relies on the use of a "plantar flexion (PF)" spring, to store energy from the controlled dorsiflexion phase of stance while an electric actuator is loading a "push-off (PO)" spring during the complete stance phase. Due to the use of a locking mechanism, the energy injected into the PO spring can be

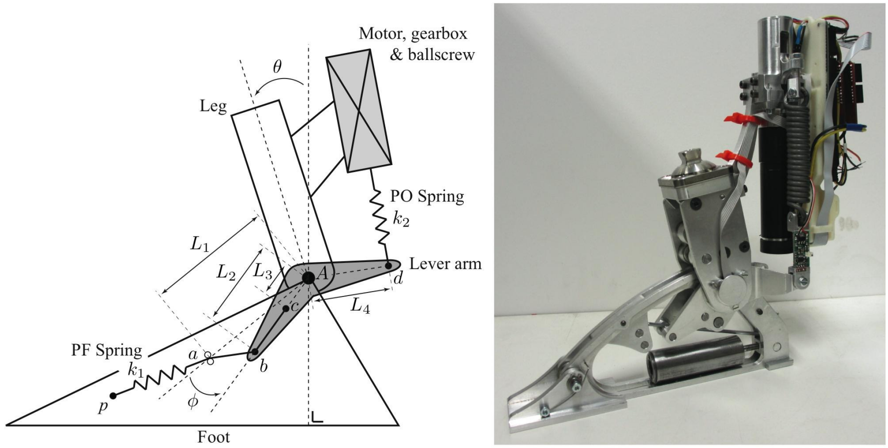

Fig. 2 Schematics and picture of the AMP-Foot 2.0

delayed and released at push-off. This way, the actuator's power is significantly reduced and so is its size and weight while still providing the full torque and power needed for locomotion.

In Fig. 2, the essential parts of the AMP-Foot 2.0 are represented. The device consists of 3 bodies pivoting around a common axis (the ankle), i.e. the leg, the foot and a lever arm. As mentioned before, the system comprises 2 spring sets: a PF and a PO spring set. The PF spring 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*, while the PO spring is placed between the motor-ballscrew assembly and a fixed point *d* on the lever arm. Not drawn in Fig. 2 is the locking mechanism which provides a rigid connection between the leg and the lever arm when energy is injected into the system. Its working principle is discussed further in the text.

To maintain a consistent notation through the chapter, symbols used in the schematics are described:

L_1 = distance between ankle axis (A) and point a.

L_2 = distance between ankle axis (A) and point b.

L_3 = distance between ankle axis (A) and point c.

L_4 = distance between ankle axis (A) and point d.

θ = angle between foot and leg.

ϕ = angle between foot and lever arm.

k_1 = Plantar Flexion spring stiffness.

k_2 = Push-Off spring stiffness.

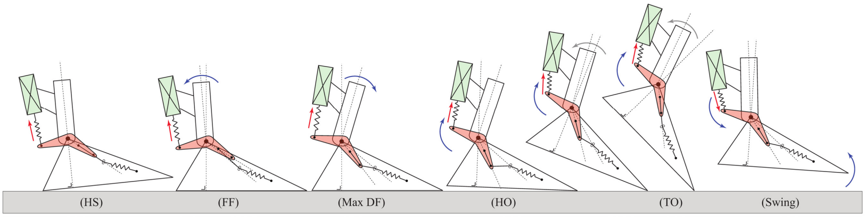

Fig. 3 Behavior of the AMP-Foot 2.0 during a complete stride

To illustrate the behavior of the AMP-Foot 2.0, one complete gait cycle is divided into several phases, shown in Fig. 3, and the working principle of the prosthetic device during each phase is explained.

3.1 *Principle of Optimal Power Distribution*

As mentioned before, the gait cycle is divided in 5 phases starting with a *controlled plantarflexion* from heel strike (HS) to foot flat (FF) produced by muscles as the *Tibialis Anterior*. This is followed by a *controlled dorsiflexion* phase ending in *push-off* at heel off (HO) during which propulsive forces are generated mainly by the *Soleus* and *Gastrocnemius* muscle groups. In the *late stance* phase, the torque produced by the ankle decreases until the leg enters the so-called *swing* phase at toe off (TO). Once the leg is engaged in the swing phase, the foot resets and prepares for the next step.

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

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

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 $\phi = 0$ i.e. $l_0 = L_1 - L_3$.
$V_{0,1}$ = Pretension of the PF spring.
l_1 = Distance between the fixed point a and c i.e.

$$l_1 = \sqrt{L_1^2 + L_3^2 - 2L_1L_3cos(\phi)} \tag{2}$$

During this period the electrical drive pulls the PO spring. 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. Therefore the prosthesis is not affected by the forces generated by the actuator.

From (FF) to heel off (HO):
When the foot stabilizes at FF, the leg moves from $\theta = -5°$ to $\theta = +10°$. Until the leg reaches $\theta = 0°$ the torque of the system is given by Equation (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_0^* = Distance between the fixed points a and b when $\phi = 0$ i.e. $l_0^* = L_1 - L_2$.
l_1^* = Distance between the fixed point a and b i.e.

$$l_1^* = \sqrt{L_1^2 + L_2^2 - 2L_1 L_2 cos(\phi)} \tag{4}$$

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.

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

$$T_2 = k_2 l_2 L_4 \tag{5}$$

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 Equation (3). At the moment of HO, the locking mechanism is unlocked and as a result of this, all the energy which is stored into the PO spring is fed to the system. Since $T_1 \leq T_2$ 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 $T' \geq T_1$ and $T' \leq T_2$. The torque at the ankle becomes

$$T' = k_1(l_1' - l_0' + V_{0,1})\frac{L_1L_2}{l_1'}sin(\phi + \psi) \quad (6)$$

in which

$$l_0' = l_0^* = L_1 - L_2$$
$$l_1' = \sqrt{L_1^2 + L_2^2 - 2L_1L_2cos(\phi + \psi)}$$

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

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 mechanism back to its initial position, return springs are used to set θ back to $0°$ and to close the locking mechanism. At this moment, the device is ready to undertake new step.

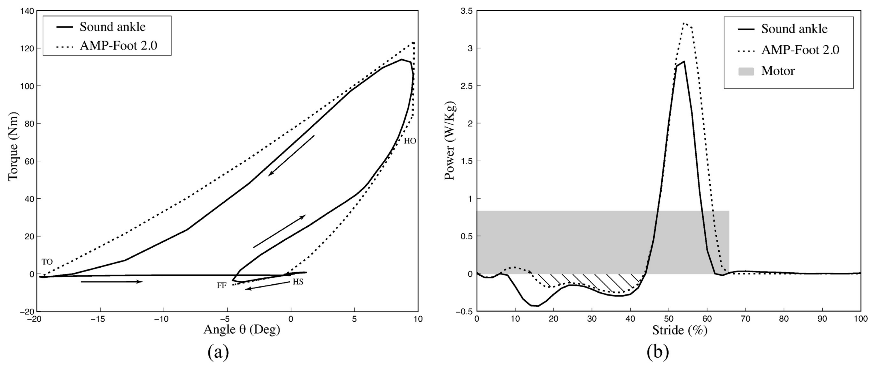

Fig. 4 (a) Torque-Angle characteristic of the AMP-Foot 2.0 compared to abled-bodied ankle-feet according to gait analysis conducted by D. Winter [27]. (b) Ankle power during one stride. The solid line represents the power generation of a sound ankle while the dotted line reprensents the resulting power of the AMP-Foot 2.0. The gray rectangle shows how the actuator power is spread over one gait cycle while the shade area represents the energy gathered from the controlled dorsiflexion with the PF spring.

Table 1 Lever arm dimensions

L_1 = 80 mm	L_2 = 60 mm
L_3 = 30 mm	L_4 = 60 mm

3.2 Mechanics and Design

According to Winter [27] 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. Therefore a moving range of $-30°$ to $+15°$ has been chosen for the joint to fulfill the requirements of the ankle anatomy. The length of the lever arms named in Fig. 2 are given in TABLE 1. The foot is made to match a European size 43 with a ankle height of approximately 8 cm. The largest part of the prosthesis has a width of 5 cm and is located at the toes to enhance stability. This way the prosthesis fits in a shoe which is significantly more comfortable for the amputee. A description of the elements used in the prosthesis is given.

Spring Sets:
As described in the previous section, the AMP-Foot 2.0 uses two spring sets. For the PF spring (k_1), a belleville spring assembly, which is shown in Fig. 5, 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. The PF spring has a stiffness of approximately 300 N/mm. For the PO spring (k_2), two tension springs with each a stiffness of 60 N/mm are used.

Actuation:
To achieve the requirements of a able-bodied ankle-foot complex, an actuator with a good ”power and strength to weight” ratio, high mechanical efficiency is needed. A Maxon Brushed DC motor (60 W) has been chosen in combination with a gearbox and ballscrew assembly, which is described in TABLE 2. 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.

Locking Mechanism:
As mentioned before, a critical part of this mechanical system is the locking mechanism. This locking must be able to withstand high forces while being as compact and lightweight as possible. The crucial and challenging part is that the system must be unlocked when bearing its maximum load and last but not least,

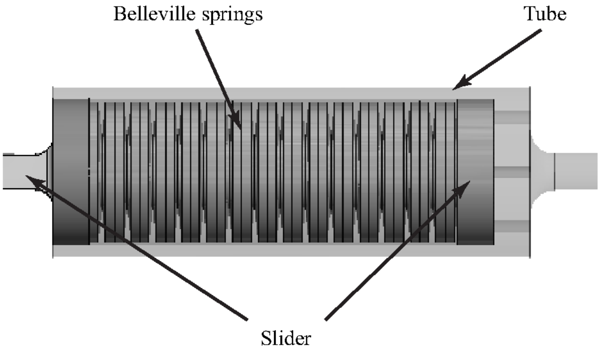

Fig. 5 Section representation of a disc spring assembly. 29 disc springs are stacked in series on a slider which moves into a tube.

Table 2 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 ϕ10x2 $\eta_{transmission 1\&2} = +/-75\%$

this unlocking must require a minimum of energy. Fortunately, the lever arm has to be locked to the leg at a fixed angle. These requirements have been taken as criteria and 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 [18], where it is used to lock the knee joint of a walking robot. Fig. 6 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. Therefore 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 (Mecalectro, 12VDC, 5W) 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. 7 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 is past its singular position of a few degrees, the resulting force needed for unlocking is estimated to 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. 4. Therefore, depending on the way the motor is controlled, the resulting force needed to unlock the four bar linkage will be reduced.

Sensors:
The two spring assemblies are equipped with custom made loadcells 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.

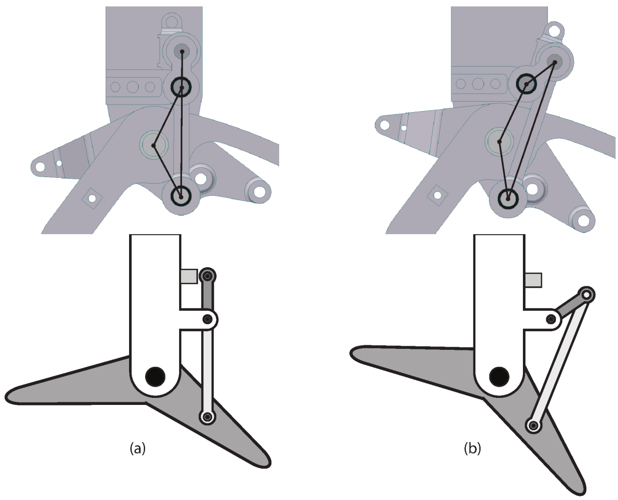

Fig. 6 CAD representation and schematics of the four bar mechanism in locked (a) and unlocked (b) position

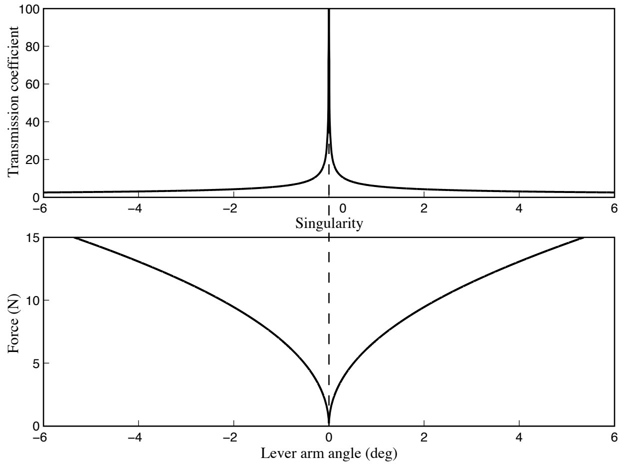

Fig. 7 Transmission Coefficient and resulting force of the four bar linkage mechanism close to its singular position (0°)

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.

4 Conclusions

In this chapter, the authors propose a new design of an energy efficient powered transtibial prosthesis mimicking able-bodied ankle behavior, the AMP-Foot 2.0. The inovation of this study is to gather energy from motion during the controlled dorsiflexion 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 a favourable 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 approximately 45° to fullfill the requirements of a 75 kg subject walking on level ground at normal cadence. Its total weight is $\pm$ 2.5 kg which corresponds to the requirements of an intact foot. The prototype is completely built and hardware and control are currently being tested. Experiments with amputees will follow.

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