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Cherelle, Pierre; Grosu, Victor; Vanderborght, Bram; Lefeber, Dirk; Flynn, Louis; Junius, Karen; Moltedo, Marta

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- The Ankle Mimicking Prosthetic Foot 3 -
Locking Mechanisms, Actuator Design, Control and Experiments with an Amputee

Pierre Cherelle, Victor Grosu, Louis Flynn, Karen Junius, Marta Molledo, Bram Vanderborght and Dirk Lefeber
Robotics & Multibody Mechanics research group
Dept. of Mechanical engineering
Vrije Universiteit Brussel, Belgium.

Abstract — The Ankle Mimicking Prosthetic Foot 3 is an energy efficient bionic foot using the principle of optimal power distribution. The main challenge behind this research is focussed on retrieving as much energy as possible from the gait and to incorporate an electric actuator with minimized power consumption. The Ankle Mimicking Prosthetic (AMP-) Foot 2, as a proof-of-concept, showed the advantage of using the Explosive Elastic Actuator capable of delivering the full ankle torques ($\pm 120 \, \text{Nm}$) and power ($\pm 250 \, \text{W}$) with only a 60 W motor. The novelty of this work is focussed on the addition of an extra locking mechanism to the AMP-Foot 3 which bring additional assets such as natural adaptability to different walking speeds and slopes and an improved energy storage during early stance implying a greater reduction of the power requirements of the motor to only 50 W while still being able to produce the peak output torque and power necessary for walking.

In this article, the authors present the AMP-Foot 3 focussing on its critical mechanical parts, its actuation design and its control. Experiments with an amputee are also presented to show its effectiveness.

Index Terms — Bionic feet, transtibial, transfemoral, prosthetics, compliant actuation.

I. INTRODUCTION

Today’s prosthetic feet can be divided into conventional feet (CF), energy-storing-and-returning feet (ESR) and bionic feet, as presented in [1]. The aforementioned new generation prosthetic devices are part of the bionic feet family and have been named ‘propulsive bionic feet’ according to an updated categorization presented in [2]. The state-of-the-art in propulsive bionic feet currently consists of 26 devices, from which 19 have been developed in the USA [3], [4], [5], [6], 6 in Belgium [7], [8], [9], [10], [11] and 1 in China [12]. The leading entities in this field are undoubtedly the research teams of Herr et al. (MIT - USA) [13], [14], [15], Sugar et al. (ASU - USA) [16], [17], [18] and Goldfarb et al. (Vanderbilt) [19], [20]. Most of the developed devices are still on a research level, but represent a preview of tomorrow’s commercial prosthetic devices.

The Ankle Mimicking Prosthetic (AMP-) Foot 2 and 3, developed at the Vrije Universiteit Brussel, are part of these propulsive bionic devices but use a unique type of actuation, namely the Explosive Elastic Actuator (EEA). This new type of actuation technology consists of a spring in series with a locking mechanism placed itself in series with a Series Elastic Actuator (SEA). This mechanism has the advantage of coupling or decoupling the electric drive from the output of the devices. This has major implications on the power requirements of such system and improves significantly safety issues in human-robot interactions. The torque requirements of the EEA are similar to the SEA in a bionic foot. But thanks to the ability of decoupling the drive from the output, the motor can work for a longer time period (typically 2 to 3 times for a prosthetic ankle) and therefore reducing by the same amount its power requirements as shown with the Ankle Mimicking Prosthetic (AMP-) Foot 2 concept[7], [21]. The novelty in the AMP-Foot 3 prototype
(shown in Fig. 1) is the design and incorporation of an extra locking mechanism, namely a resettable overrunning clutch. The addition of this locking system to the device has the advantage of improving energy storage during the early stance phase and naturally adapting the prosthesis to different walking speeds and slopes.

In this article, the authors give a detailed description of a novel locking mechanisms and the EEA design used in the AMP-Foot 3 prototype. In section II, the main requirements for bionic feet are stated. These represent the requirements used as general criteria for the development of the AMP-Foot 3. Section III then gives a detailed description of the most critical mechanical parts of the prosthesis, namely its novel locking mechanisms and their implications on its actuator design. In section IV, the authors give a brief description on how the device is controlled and in section V, preliminary experiments with an amputee are presented. The article will then be closed with conclusions and future perspectives in section VI.

II. BIONIC FEET REQUIREMENTS

According to systematic gait analysis from Winter [22], a subject walking at normal cadence produces a peak torque at the ankle joint of approximately $1.6 \text{Nm/kg}$ of bodyweight in a very small amount of time ($\pm 0.2 \text{s}$ for a walking rate of $1 \text{step/s}$), consuming hereby on average $0.35 \text{J/kg}$ of mechanical energy per step. In accordance the generated power at push-off reaches peak values of approximately $3.5$ to $4.5 \text{W/kg}$. Considering a $75 \text{kg}$ subject a maximum torque output of approximately $120 \text{Nm}$ is required with a power output of about $250 \text{W}$ as can be seen in Fig. 2 (a) and (b). These approximate values are generally taken as a criterion for the development of the so-called propulsive devices. Therefore, throughout the following sections these data will be used as a reference.

III. THE AMP-FOOT 3 MECHANICAL DESIGN

In this section, the authors present the mechanical design of the most critical parts to the functioning of the AMP-Foot 3. At first the general assembly of the prosthesis is highlighted, followed by a description of the resettable overrunning clutch and the design and dimensioning of the EEA.

A. The AMP-Foot 3 Main Parts

In Fig. 1 the main parts of the AMP-Foot 3 are depicted. The device consists of a foot part, a leg part and 2 lever arms pivoting around the ankle axis. The foot is connected to lever arm 1 through a compliant crank slider mechanism. Lever arm 1 and lever arm 2 are held together by means of a resettable overrunning clutch. Lever arm 2 is fixed to the spring placed in series to the electric drive which itself is fixed to the leg and additionally, lever arm 2 and the leg are connected by means of the four bar linkage locking mechanism. When locked, the leg and lever arm 2 are rigidly connected while when unlocked, it is the Series Elastic Actuator (SEA) that determines the position of lever arm 2 with regard to the leg. On the foot part, and more precisely on the slider of the crank-slider mechanism are the planterflexion (PF) springs which allows the prosthesis to function as an efficient ESR foot in passive mode (when the motor is not activated and the four bar locking mechanism closed). In the ballscrew assembly, the PO spring is placed in series with the electric drive. This SEA together with the four bar locking mechanism form the so-called Explosive Elastic Actuator (EEA). The main difference with the AMP-Foot 2 assembly is the additional resettable overrunning clutch that permits greater energy storage during the dorsiflexion phase of stance and automatic adaptivity to different walking speeds and slopes. Other design differences are noticeable as the use of a crank-slider mechanism instead of cables and pulleys to elongate the PF spring assembly and the use of a fully integrated compression PO spring compared to externally placed tension springs used in the AMP-Foot 2 prototype.

B. The Resettable Overrunning Clutch

The novelty behind the AMP-Foot 3 concept is not only its actuation principle but also its improved ankle mimicking characteristics and energy storage during early stance together with its natural ability to adapt to different walking speeds and slopes. From the reference data [22] shown in Fig. 2 it can be seen that a sound ankle torque-angle characteristics has 3 different rest positions, meaning 3 different angle positions for which the resulting torque is zero: one at initial contact (IC) with ankle angle $0^\circ$, a second one shortly after the foot is stabilized on the floor (FF) at about $-5^\circ$ and the third and last one at the end of stance when the toes are lifted from the ground (TO) at approximately $-20^\circ$. As explained in [7], the
AMP-Foot 2 has only the ability of mimicking 2 of these rest positions, namely at IC (0°) and TO (−20°). Because of this, the AMP-Foot 2 is unable to gather all the motion energy of its wearer during the early stance phase. Therefore adding the extra rest position at FF has the potential of increasing the stored energy in the PF springs. It is also known that walking faster requires an increased step length and joint stiffness [22]. As a matter of fact, when the stride length is increased, the ankle angle at which the foot stabilizes also increases and thus the rest position of the ankle evolves to a lower value. A similar reasoning can be held for walking on slopes. When walking uphill, the plantarflexion phase between IC and FF is reduced, therefore decreasing also the second rest position angle somewhere between 0° and −5° depending on the slope. On the other hand the maximum dorsiflexion (at about heel off - HO) will increase slightly compared to level ground walking. In contrast, walking downhill will increase the plantarflexion phase between IC and FF, hereby increasing the rest position angle. To realise these adaptations naturally, the lever arm used in the AMP-Foot 2 had to be decomposed in two separate parts (lever arm 1 and lever arm 2) interconnected by means of a mechanism with the following characteristics:

- When engaged, the locking mechanism be-
Fig. 3. The resettable overrunning clutch. (a) Lever arm 2 with the integrated freewheel mechanism consisting of an inner and outer ring, pretension springs and 7 locking cylinders. (b) Lever accommodated with 7 bars to disengage the overrunning clutch. This lever is placed in between 2 symmetrically placed freewheels. (c) General outlook of the resettable overrunning clutch assembled with lever arms 1 and 2. The unlocking is actuated by a servomotor compressing a spring on a slider mechanism.

- The resettable overrunning clutch is designed to withstand the total forces and torques of the ankle-foot system.
- Lever arm 1 and 2 have to be interconnected at the ankle joint with a rotative degree of freedom.
- Both levers must be able to move freely in one direction and to be locked instantaneously in the other direction.
- The locking must be continuous and without any backlash (unlike a ratchet and pawl mechanism).
- The position of the 2 lever arms must be reset when unloaded, before or during the swing phase.
- The mechanism must be as small and as lightweight as possible.

As a matter of fact, locking devices are widely used in robotic applications for, among others, better energy management. Their principles are sometimes very old but were improved over the years to match the requirements of specific tasks. In general one can classify locking devices into three main categories: mechanical locking, friction-based locking and singularity locking. Mechanical lockings are mechanisms in which the position of its components (e.g. wedge, pawl, etc) determines the locking or unlocking state. Examples are latches [23], ratchets [24], dog clutches [25], cam-based lockers [26] or hydraulic locks [27]. On the other hand, friction-based locking devices depend on friction in order to prevent motion between two parts. Typical examples are the electromagnetic brake [28], the overrunning clutch [29], non-backdriveable gearing [30], self amplifying brakes [31] and capstan mechanisms [32], to name a few. Singularity locks finally are characterized by a position dependent transmission ratio which becomes infinitely high when brought in a singularity, such as four bar linkages [33] and non-linear transfer ratio mechanical systems [34].

Taking the aforementioned criteria into consideration, a new type of friction-based locking mechanism has been developed, called a ‘resettable overrunning clutch’ which is shown in Fig. 3.

It consists of 2 resettable continuous one way clutches placed between lever arm 1 and 2. This clutch system is based on the so-called freewheel principle consisting of spring-loaded steel rollers inside of a driven cylinder. Rotating in one particular direction wedges the cylinders against the outer ring of the system making it rotate in unison. Rotating in the opposite direction, the steel rollers just slip inside the mechanism allowing separate movements of both inner and outer rings. As such, a freewheel is not yet resettable. To realise this, a lever (called the reset lever as can be seen in Fig. 3 (b)) accommodated with bars is placed between both overrunning clutches as such that the bars fills the empty space between the steel rollers and the inner ring. When the clutches are unloaded, rotating the lever with respect to the inner ring pushes the rollers against the pretension springs, and thus disengages the clutch allowing it to rotate freely in both directions. It is important to note that an energy efficient disengagement of the overrunning is only possible when the system is unloaded. To ensure proper unlocking, a servomotor in series with a compression spring on a slider is attached to the reset lever of the clutch as can be seen in Fig. 3 (c). During the gait (when the locking mechanism is loaded) the spring is
compressed until the servomotor reaches a singular position. The principle is nothing less than a small scale Explosive Elastic Actuator (EEA). The servo motor being in a singular position, it has no torque to provide to keep the spring compressed. Once the load is removed from the clutch, and because the spring is compressed, the locking is automatically and instantaneously disengaged. This overrunning clutch is designed to keep up to 160 Nm of torque which is about 30% more than the maximum designed ankle torque.

During the first phase of gait (from IC to FF), the PF springs are not elongated since lever arm 2 moves in its free direction with respect the lever arm 1. In contrast, 2 small tension springs (also called the reset springs) are elongated to provide the small dorsiflexion torque during this phase. However mechanical stops have been placed such that the maximum possible angle between the 2 levers would not exceed approximately 12°. This has been added to the system to avoid an excessive plantarflexion angle at the beginning of a stride which could deteriorate the comfort of the wearer. If for any reason the plantarflexion angle would exceed this limit, the PF springs would simply be elongated creating a strong dorsiflexing torque helping the amputee to stabilize the foot. As soon as the movement of the leg changes direction, the overrunning clutch is engaged and the PF springs starts gathering motion energy from the wearer’s gait.

As such, the presented resettable clutch mechanism is a rotative, continuous, one way locking without backlash with the ability to bear the total ankle forces and torques and the possibility to be disengaged (and reset) when unloaded (at the very beginning of the swing phase). All these features fits completely the previously presented requirements of the AMP-Foot 3 prototype. The weight of the locking mechanism is approximately 0.3 kg.

C. The Explosive Elastic Actuator

As a matter of fact, the previously presented ‘resettable overrunning clutch’ also has its impact on the dimensioning and design of the EEA of the AMP-Foot 3. Unlike the AMP-Foot 2 prototype, in which the motion energy during the early stance phase could only be gathered between ankle angles of approximately 0° and +10°, as soon as the overrunning clutch is engaged at FF (according to Winter [22] about −5° for normal walking on level ground) the PF springs are elongated. This represents an extra 4 to 5 J that can be stored in the PF springs of the 9 J storeable energy during this phase of gait. This gain in energy is therefore non neglectable. Thus, of the total energy necessary for walking, evaluated at 26 J by integrating the power curve shown in Fig. 2(a), the AMP-Foot 3 actuator theoretically has to provide 26 J − 9 J = 17 J. When considering a walking speed of approximately 1 stride/sec (which is slightly faster than normal cadence) the stance phase takes about 0.6 sec (66% of a stride). Consequently the overall power rating of the electrical drive is calculated as follows:

$$P = \frac{E_{generated}}{\Delta t} = \frac{17 J}{0.6 s} = 28.3 W$$

(1)

In which P represents the power, $E_{generated}$ the generated energy and $\Delta t$ the lapse of time during which the energy $E_{generated}$ is produced. According to this estimation the overall power rating of the electric motor is reduced to only 28 W (without consideration of the motor’s and possible gearbox’ efficiencies). An estimation of the efficiencies of the actuation system of 60% leads us to an indication of a possible candidate around 50 W. It was also decided to work with a linear actuator (e.g. a ballscrew). A ballscrew was selected with a lead of 2 mm. To ensure a correct plantarflexion angle at toe-off (TO) the PO spring (with stiffness $k_{pO} = 180 N/m/mm$) has to be elongated by ±15 mm at the end of stance. Thus the maximum force acting on the ballscrew is

$$F_{max} = 15 mm \cdot 180 N/mm = 2700 N$$

(2)

The following design constraints have to be taken into account for the motor-transmission-ballscrew
choice:

- Maximum axial load (spring) : $F_{\text{max}} = 2700\, \text{N}$
- Maximum deflexion of spring : $S_{\text{max}} = 15\, \text{mm}$
- Ball screw lead : $L = 2\, \text{mm}$
- Screw speed during the stance phase:
  Loading between initial contact (IC) (0% of stride) and TO (66% of stride) with walking speed of 1 stride/sec.

\[
\dot{\theta}_{\text{screw}} = \frac{\dot{x}}{L} = \frac{15\, \text{mm}}{2\, \text{mm}} \frac{60\, \text{s}}{\text{min}} = 682\, \text{RPM}
\]

- Screw speed during the swing phase:
  Between TO (66% of stride) and IC (100% of stride)

\[
\dot{\theta}_{\text{screw}} = \frac{\dot{x}}{L} = \frac{15\, \text{mm}}{2\, \text{mm}} \frac{60\, \text{s}}{\text{min}} = 1324\, \text{RPM}
\]

- Peak torque applied on screw:
  ($\eta_{\text{BS}} = \pm 75\%$)

\[
T_{\text{screw}} = \frac{F \cdot L}{2\pi \eta_{\text{BS}}} = \frac{2700\, \text{N} \cdot 2\, \text{mm}}{2\pi \cdot 0.75} = 1146\, \text{Nmm}
\]

These calculations have led to a configuration consisting of a 24V Maxon motor ECi-40 (50W), a 5.8 : 1 one stage planetary gearbox and a 2mm lead ballscrew. To ensure a successful reset of the actuation system during the swing phase (approximately 0.3s), it is decided to provide the motor with a 15V power supply. During the swing phase, almost no torque has to be exerted by the motor. Taking this into account, calculations of the maximum motor torque and motor speed are shown in Eq. (6) and Eq. (7).

\[
T_{\text{motor,peak}} = \frac{T_{\text{screw}} R}{\eta_{\text{gearbox}}} = \frac{1146\, \text{Nmm}}{0.87} \frac{1}{5.8} = 0.227\, \text{Nm}
\]

\[
\dot{\theta}_{\text{motor,max}} = \frac{\dot{\theta}_{\text{screw}}}{R} = 7680\, \text{RPM}
\]

As shown in Fig. 4, the PO compression spring is mounted between the motor and the ballscrew nut. This assembly is completely different than the one used in the AMP-Foot 2 [21]. Its advantage is its compactness and safety in case of mechanical failure. As such the EEA is not yet complete. An additional locking mechanism is needed to decouple the actuator from the device. This locking mechanism is placed in the leg and fixes the lever arm to the leg when desired. When walking during the early stance phase, the locking mechanism is closed. As a result of this, lever arm 2 is fixed to the leg and only the PF spring is influencing the ankle joint. During that time, the AMP-Foot 3 is acting like an efficient passive ESR foot. In parallel to this, the linear actuator compresses the PO spring. But since the lever is fixed to the leg, this has no impact on the ankle kinematics or dynamics. At the moment of heel-off, the locking mechanism is disengaged, releasing hereby the energy stored in the PO spring and transmitting it to the ankle joint. The effect of this release is a sudden clockwise rotation (according to the schematics in Fig. 1) of the lever arm elongating the PF spring further and hereby providing push-off to the amputee. Between heel-off and toe-off, the motor continues pulling the PO spring, acting this time as a SEA between the leg and the foot. As soon as the swing phase starts, the complete prosthesis undergoes a hardware reset to get ready for a new step.

Similar to the AMP-Foot 2, the requirements for this locking mechanism are the following:

1) The locking position of the lever with respect to the leg must be the same for every step.
2) The locking mechanism must be able to withstand very high forces and torques.
3) Unlocking of the mechanism must be disengaged passively or by an external actuator (i.e. a servo motor).
4) Unlocking of the mechanism must be done when bearing its maximal load.
5) Unlocking of the mechanism should happen with a low energy consumption.

Since the four bar linkage locking used in the AMP-Foot 2 [7], [21] has shown good results, it was chosen to use the same mechanism. However, to demonstrate that unlocking could happen with a minimum of energy, it was chosen to do it passively, by hitting a mechanical stop when walking. Of course this unlocking position is then dependent of the angle at which the mechanical stop is placed. For the concept, this mechanical stop is only manually adaptable. In the case of a more elaborated product, it might be interesting to include a low power (non-backdriveable) actuator to change the position of the mechanical stop in between steps in function of the walking speed and slope.

In Fig. 5 the four bar linkage locking mechanism is shown in 2 different positions. In the AMP-foot 3 prototype, this locking mechanism was placed in the foot (while still attached to the leg). This design was elaborated in order to reduce the adaptor height of the prosthesis and to make it possible to unlock the four bar linkage in singular position by hitting point C (Fig. 5) on a mechanical stop by moving the leg forward. More details on the working principle of such a locking mechanism can be found in [21].

In the next section, an overview is given on how the AMP-Foot 3 and the EEA were controlled during the preliminary experiments.

IV. CONTROL OF THE AMP-FOOT 3

Similar to the AMP-Foot 2, the control of the AMP-Foot 3 is divided into two layers. The high and the low level control. The low level control consists of regulators sending commands to the power section of the brushless DC motor, while the high level controller detects the state of the prosthesis and the intention of the user. For the low level control, a maxon ESCON drive has been used providing a PID regulation on current. The high level controller of the AMP-Foot 3 is a state machine determining the phases of walking of a subject based on heel and toe contacts. These phases are Initial Contact (IC), Foot Flat (FF), Heel-off (HO) and Toe-Off (TO) as shown in Fig. 6. Each of these phases correspond to a particular combination of heel and toe contacts detected by means of Force Sensing Resistors (FSR). Table I shows the 4 phases based on this trigger information. As soon as the heel touches the ground (IC), until the moment the toes are lifted from the floor (TO), the state machine detects that the prosthesis is in the stance phase and the motor starts compressing the PO spring hereby loading the EEA (red arrow). This continues until the four bar locking mechanism is unlocked by pushing it out of its singular position against a mechanical stop. During this same period the movement of the leg compresses the PF springs (green arrow) providing the wearer with the necessary torque output during the early stance phase. From IC to FF the overrunning clutch does not constraint the movement of the leg with regard to the lever arm and fixes both at the moment the dorsiflexion phase is engaged (from FF to approx. HO). When push-off (PO) occurs, the energy stored in the PO spring is fed to the ankle prosthesis (blue and red arrow) providing a peak torque and power output to the amputee. At toe off the torque is returned to zero magnitude allowing the overrunning clutch to reset. Between TO and a new IC, the swing phase is detected in which the AMP-Foot 3 is reset and brought back to initial position to undergo a new step.

The output of the high level controller is then fed into the low level controller. As such the control is not adaptive to the user’s walking speed or to the slope of the terrain. Future perspectives in adaptive control will be given in section VI. For these experiments, a proper walking detection and a start/stop function has been implemented to allow natural walking, opening doors, put a step back if needed, and many other daily life scenarios. Default, the electric drive is disabled. If the user starts to walk, the controller detects the walking pattern. After 2 completed steps the motor provides push-off to the amputee. If the walking pattern changes or stops, the controller immediately detects that the user has stopped walking and disables the motor. Since the AMP-Foot is an efficient energy storing and returning foot when the motor is disabled, it still allows natural mobility when no push-off is needed.
V. EXPERIMENTS WITH AN AMPUTEE

In this section, the authors present the data captured while testing the AMP-Foot 3 with an amputee. A male amputee subject with a right tranfemoral amputation participated in the experiments and provided written and informed consent. A short video is available on https://www.youtube.com/watch?v=JrSQFM7n1wU.

A. Methods

The AMP-Foot 3, as shown in Fig. 1, was tested with Mr. A., a transfemoral amputee. To complete the prosthetic foot, the single subject who has been subjected to clinical test has been using his own knee prosthesis, an Ossur Mach Knee. In this experiment, Mr. A. was asked to walk at self selected speed on a treadmill (level ground and 4° uphill slope) while data was recorded on an SD card. The used sensors to record the data are listed in Table II.

B. Results & Discussion

In Fig. 7, the data of level ground (a, c and d) and 4° uphill walking (b, d and f) at the same self-selected speed of approximately 4.5 km/h with the AMP-Foot 3 is presented. These graphs show the ankle angle and lever arm angle in function of time, the torque in function of time and the ankle torque-angle characteristic of the corresponding time-data. The torque output is calculated based on the force acting on the PF spring and the encoder information as detailed in [35]. This data set was recorded when the subject started to walk. Therefore, one can see that during the two first steps, the prosthesis does not provide any propulsive forces but still acts as an efficient ESR foot. This is due to the high level controller having to detect 2 complete walking sequences before enabling the actuation of the prosthesis. During these 2 first steps no push-off is provided to the amputee but the PF spring is still acting in the system, and thus providing a high torque output between 60 and 70 Nm. This is an interesting asset for a prosthetic foot. In case of power failure (or any other type of actuation failure) the prosthesis can still be used in its passive mode and because of its articulated joint feels more comfortable than most rigid prosthetic feet. In addition, the EEA is intrinsically safe because the electric drive is behind a locking mechanism. At the end of both
Fig. 6. Phases of walking with the AMP-Foot 3. At initial contact (IC), the motor starts compressing the PO spring hereby loading the EEA. This continues until the four bar locking mechanism is unlocked by pushing it out of its singular position against a mechanical stop. During this same period the movement of the leg compresses the PF springs providing the wearer with the necessary torque output during the early stance phase. From IC to FF the overrunning clutch does not constraint the movement of the leg with regard to the lever arm and fixes both at the moment the dorsiflexion phase is engaged (from FF to approx. HO). When push-off (PO) occurs, the energy stored in the PO spring is fed to the ankle prosthesis providing a peak torque and power output to the amputee. At toe off the torque is returned to zero magnitude and when the foot is lifted from the floor, the prosthesis is reset to get ready for a new step.

TABLE II
Sensory system of the AMP-Foot 2.1.

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Measurement</th>
</tr>
</thead>
<tbody>
<tr>
<td>Load cell</td>
<td>Force acting on the PF spring</td>
</tr>
<tr>
<td>Force Sensing Resistors (FSR)</td>
<td>Heel and toe contact</td>
</tr>
<tr>
<td>Magnetic encoder 1</td>
<td>Angle between leg and foot</td>
</tr>
<tr>
<td>Magnetic encoder 2</td>
<td>Angle between lever arm and foot</td>
</tr>
</tbody>
</table>

As expected, some differences occur between walking on level ground and 4° uphill slope. On the data set for level ground walking in Fig. 7 (a) it is noticeable that the ankle rest angle is approximately $-6^\circ$. The reason for this is because the used shoes have a heel of about 1 cm height. Due to the resettable overrunning clutch, the prosthesis naturally adapts to the walking slope. The result of this is a shift to the right of a few degrees as can be seen in Fig. 7 (f) compared to Fig. 7 (e). As a matter of fact, when the heel strikes the floor during uphill walking, the plantarflexion angle before reaching foot flat is, in this case 3° smaller compared to level ground walking - on average $-8^\circ$ for uphill walking compared to $-11^\circ$. experiments it is also noticeable that the prosthesis immediately stops providing push-off when the amputee stops walking. One can also see that the angles of the lever arm and ankle are following each other. Because of play in the system some differences can occur, especially when the ankle torque output is high. On the contrary, the following steps do present a PO spring deflexion and after unlocking of the four bar mechanism, a difference for the lever arm and ankle angle. In general it can be seen that the ankle torque-angle characteristics for both walking experiments present a loop similar to the reference data, indicating an energy generation during walking.
for level ground walking. One can also see that the maximum dorsiflexion angle before push-off occurs reaches approximately $15^\circ$ (uphill walking) instead of $11^\circ$ (level ground walking) and that the maximum plantarflexion angle before toe-off reaches approximately $-16^\circ$ (uphill walking) instead of $-18^\circ$ (level ground walking). Because the high level controller is not yet programmed for slope walking, the torque output of approximately $125 \text{ Nm}$ at push-off remains the same for both experiments.

With these experiments the ability of the AMP-Foot 3 to provide push-off to its user and to naturally adapt to different slopes is shown.

**VI. CONCLUSIONS & FUTURE WORK**

In this article, the authors have presented a detailed description of the locking mechanisms and its implications on the EEA design used in the AMP-Foot 3 prosthesis. The main requirements for the design of bionic feet were highlighted and used as general criteria for the development of the prototype. Furthermore, a brief description on how the device is controlled and preliminary experiments with an amputee were presented. Experiments were conducted on a treadmill to show the effect of the locking mechanism and EEA while walking on level ground and uphill. Future work will consist of upgrading the high level control of the prosthesis adding actuation adaptivity to different walking speeds and walking slopes. As a matter of fact the actual design is naturally adaptable but it would be much more comfortable for the user if more power and torque was generated when walking faster or uphill and less power and torque when walking slower or downhill. After the treadmill experiments, the subject was free to walk outside with the AMP-Foot 3 as can be seen in the video on the following link: [https://www.youtube.com/watch?v=JrSQFM7n1wU](https://www.youtube.com/watch?v=JrSQFM7n1wU).

**VII. ACKNOWLEDGEMENTS**

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The concepts of the AMP-Foot Prostheses fall under patent nr. WO2011033341 A1.
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