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Conceptual Design of an Energy Efficient Transfemoral Prosthesis

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Abstract— **In this study, we present the conceptual design of a fully-passive transfemoral prosthesis. The design is inspired by the power flow in human gait in order to have an energy efficient device. The working principle of the conceptual mechanism is based on three storage elements, which are responsible of the energetic coupling between the knee and the ankle joints. Design parameters of the prosthesis have been determined according to the energy absorption intervals of the human gait. Simulation results shows that the power flow of the system is comparable with human data. Finally, an initial prototype is presented as proof of concept.**

I. INTRODUCTION

The research interest on transfemoral prostheses is clearly motivated by their crucial impact to human life. A still open challenge is the design of prostheses that can adapt to various walking conditions and can be energy efficient with respect to metabolic energy consumption and external actuation.

With this respect, passive transfemoral prostheses can be considered mechanically efficient since they are not actuated. However, amputees are required to consume a large amount of metabolic energy (about 60% extra) to compensate the lack of the lost muscles [1]. Moreover, these prostheses are designed with constant mechanical characteristics and, therefore, are not able to adapt to different conditions.

On the other side, micro-processor controlled (intelligent) transfemoral prostheses can adapt by means of internal, intrinsically passive, actuators. In [2], the dynamical behavior of the prosthesis relies on the control of a magneto-rheological damper, which produces the required breaking knee torque during walking. In [3], a micro-processor controlled transfemoral prosthesis with a magneto-rheological damper is compared with a conventional passive knee and a healthy human for different walking speeds in order to show the effect of adaptation of the knee to the gait pattern.

Another group of prostheses, called active (powered) transfemoral prostheses, are capable to inject power in order to provide ankle push-off generation and reduce the extra metabolic energy consumption [4], [5], [6], [7], [8]. Some of the design studies have been focused on the transfemoral prosthesis with energy storage capabilities in order to reduce the power consumption [9], [10], [11]. In particular, adjustable springs have been employed to provide energy storage and release. Electrically powered knee and ankle prosthesis presented in [5], also includes a spring in parallel to the ankle motor unit and its initial tests have been reported in [12]. Additionally, the studies of soft actuators for the transtibial prostheses in [13], [14], [15] have shown that the energy efficiency of the system can be improved by storing the energy during stance phase and releasing it to provide active ankle push-off generation.

Commercial transfemoral prostheses are also available, among which the passive Mauch GM [16], 3R80 [17], the micro-processor controlled RheoKnee [16], Smart Adaptive [18] and the active C-Leg [17], PowerKnee [16].

In this study, we propose a conceptual design of a fully-passive transfemoral prosthesis. The concept is mainly based on mimicking the energetic behavior of human gait to improve the energy efficiency in terms of metabolic energy consumption. To derive such kind of mechanism, power analysis of human gait and port-based representation are exploited. By analyzing the relations between the energy absorption intervals occurred during the gait, a working principle of the conceptual mechanism with three storage elements has been identified. Following the working principle, design parameters have been obtained with respect to the possible energy absorption intervals. The power flow of the mechanism is evaluated by simulation and an initial prototype is introduced to validate the energetic behavior of the overall system.

II. THE HUMAN GAIT

In this Section, we analyze the healthy human gait from an energetic point of view so to highlight the main features that should be considered in the design of prostheses. With this respect, we exploit the potential functionality of port-based analysis and modeling tools [19].

A. Power flow

In order to grasp the nature of walking, we analyze the bio-mechanical data of the human gait, as been presented by Winter in [20]. In particular, Fig. 1 depicts the power flow at the knee (upper) and ankle (lower) joints during one complete stride of a healthy human, normalized in body weight. The figure highlights three instants, i.e. heel strike, push-off and toe-off, and three main phases:

- Stance: the knee absorbs a certain amount of energy during flexion and generates as much as the same amount of energy for its extension. In the meantime, the ankle joint absorbs energy, represented by A_3 in the figure, due to the weight bearing.
- Pre-swing: the knee starts absorbing energy, represented by A_1 in the figure, while the ankle generates the main

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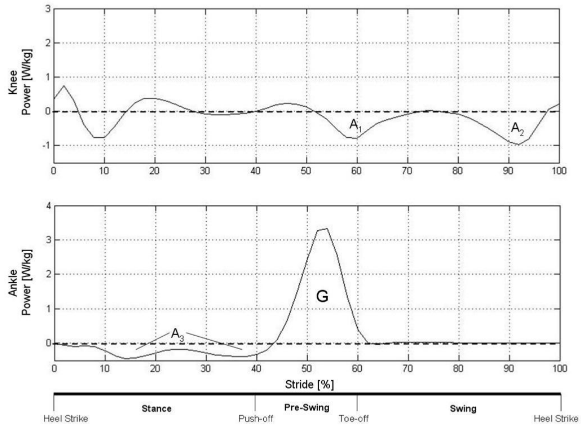

Fig. 1. The power flow of the healthy human gait normalized in body weight in the knee (upper) and the ankle (lower) joints during one stride [20]. The areas $A_{1,2,3}$ indicate the energy absorption, whereas G indicates the energy generation. The cycle is divided into three phases (stance, pre-swing and swing) with three main instants (heel-strike, push-off and toe-off).

part of the energy for the push-off, represented by G, which is about the 80% of the overall generation.

- Swing: the knee absorbs energy, represented by A_2 in the figure, during the late swing phase, while the energy in the ankle joint is negligible.

The analysis of the values of energy absorption (corresponding to the areas $A_{1,2,3}$) and generation (G) gives insightful information. It can be noted that, in the healthy human gait, the knee joint is mainly an energy absorber whereas the ankle joint is an energy generator. In particular, the knee absorbs about 0.09 J/kg during pre-swing phase (A_1) and 0.11 J/kg during late swing phase (A_2). On the other hand, the ankle absorbs approximately 0.13 J/kg during stance phase (A_3) and generates about 0.35 J/kg for push-off (G). These values show that there is almost a complete balance between the generated and the absorbed energy, since the energy for push-off generation (G) is almost the same as the total energy absorbed in the three intervals $A_{1,2,3}$.

B. Modeling

In order to deeply understand how power flows during one complete stride, we present the model of a healthy human in a port-based graphical representation. In particular, we intend to use the bond graph modeling language to visualize the human dynamical behavior so to have an intuitive approach for the design of the transfemoral prosthesis.

The basic idea behind this modeling language is that every physical system can be modeled by interconnecting simple elements, which are characterized by a particular port behavior [19]. The interconnections are realized through bonds, which represent the power flow between the different elements. The generalized representation of the power flow is made through a Dirac structure $\mathscr{D}$, as depicted in Fig. 2. The multi-bonds allow any number of storage elements $\mathbb{C}$ or $\mathbb{I}$, dissipative elements $\mathbb{R}$ and of external inputs $\mathbb{S}_{f,e}$, namely the source of flow or effort (respectively velocities and forces in the mechanical domain). Note that the Dirac structure is power continuous and not necessarily constant.

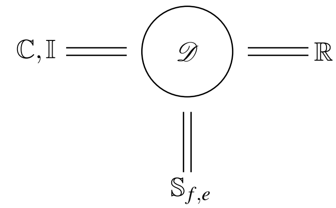

Fig. 2. Generalized representation of power flow in a dynamical system - The Dirac structure $\mathscr{D}$ defines the power-conserving interconnections between the different bonds and, therefore, how power is distributed among the ports. The multi-bonds allow any number of storage elements $\mathbb{C}$ or $\mathbb{I}$, of dissipative elements $\mathbb{R}$ and of external power source $\mathbb{S}_f$ or $\mathbb{S}_e$.

We now detail the generic representation of the Dirac structure of Fig. 2 for the specific case of the human dynamics in order to visualize how the power flows between the joints of the leg during a healthy gait, according to the data presented in [20] and plotted in Fig. 1.

The scenario is depicted in Fig. 3. Let $\mathbb{I}_t$, $\mathbb{I}_{ul}$, $\mathbb{I}_{ll}$ and $\mathbb{I}_f$ be the inertia of torso, upper leg, lower leg and foot, respectively. Let g be gravity and $\mathbb{S}_e$ a source of effort, namely the torque exerted by the hip. The 1-junctions are flow junctions and are characterized by the property that all the connected bonds are constrained to assume the same flow value, i.e. the same velocity. In particular, the figure presents three 1-junctions that indicate the angular velocities of the hip, knee and ankle joints. The three joints are interconnected such that the torso and the upper leg have the same angular velocity ω_{hip}, the upper leg and the lower leg have the same velocity ω_{knee}, and the lower leg and the foot have the same velocity ω_{ankle}. The Dirac structures $\mathscr{D}_i$, with $i = 1, \cdots, 4$ realize the interconnections between the different parts of the system and regulate how the energy is flowing. Finally, after analyzing the data in Fig. 1, we introduce three storage elements $\mathbb{C}_1$, $\mathbb{C}_2$ and $\mathbb{C}_3$, which are representing the muscle activity. In particular, they are in charge of storing the absorbed energies A_1, A_2 of the knee and A_3 of the ankle, respectively. No dissipation element $\mathbb{R}$ is considered.

C. Analysis

During human gait, the power flows from one joint of the leg to another. The core of the analysis of the gait consists in the consideration that human muscles are in charge of efficiently transferring the energy between the leg joints. The energy generated by the ankle for push-off (G in Fig. 1) is balanced by the total energy absorbed by the knee during pre-swing and late swing (A_1 and A_2) and by the ankle (A_3) during stance. This means that the ankle, in order to generate a quantity of energy in an efficient way, should exploit the energy absorbed by the knee. Therefore, a coupling between the knee and the ankle joints in terms of energy is at the basis of the efficient human gait: the energy absorbed by the knee should be transferred to the ankle.

This evaluation is crucial in the design of transfemoral prostheses and, therefore, a passive and efficient prosthesis should rely on energy transfers between the ankle, i.e. the

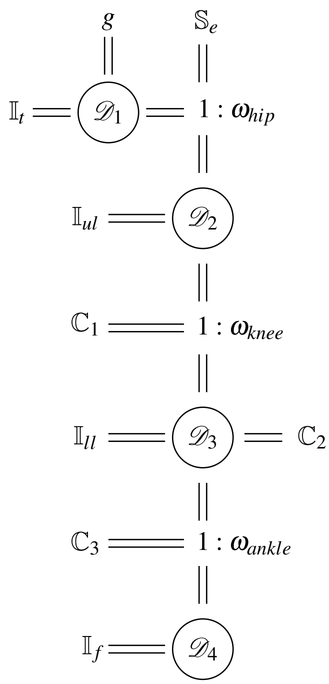

Fig. 3. Power flow in the human gait - The Dirac structures $\mathscr{D}_i$, with $i = 1, \cdots, 4$ define the power-conserving interconnections between the different bonds, i.e. the different inertias of the human leg ($\mathbb{I}_t$, $\mathbb{I}_{ul}$, $\mathbb{I}_{ll}$ and $\mathbb{I}_f$ of torso, upper leg, lower leg and foot), the joints characterized by angular velocities ω_{hip}, ω_{knee}, ω_{ankle}, the hip torque $\mathbb{S}_e$, the gravity g, the storage elements $\mathbb{C}_1$, $\mathbb{C}_2$, $\mathbb{C}_3$.

main generator, and the knee, i.e. the main absorber. This can be realized through properly designed storage elements. The three elements $\mathbb{C}_1$, $\mathbb{C}_2$, $\mathbb{C}_3$ in Fig. 3 energetically couple the knee and the ankle joints so to obtain an exchange of energy from the absorption intervals $A_{1,2,3}$ to the generation G.

III. CONCEPTUAL DESIGN OF THE PROSTHESIS

In this Section, we present the principle design for a prosthetic mechanism based on the evaluations of Sec. II. Following that discussion, the proposed concept relies on the energy transfer between the knee and the ankle joints and, therefore, on three storage elements. As summarized in Fig. 4, we introduce:

- one torsional elastic element $\mathbb{C}_1$ at the knee joint, responsible for the absorption A_1 during the pre-swing phase and for the transfer of this energy to the elastic element $\mathbb{C}_2$, by releasing it during the swing phase.
- one linear elastic element $\mathbb{C}_2$, which physically connects the upper leg (via a lever arm) and the foot and, therefore, couples the knee and ankle joints. This element is responsible for the absorption A_1 (received from $\mathbb{C}_1$) and A_2 during the swing phase and for a part of absorption $A3$ during the stance phase.
- one linear elastic element $\mathbb{C}_3$ connected between heel and a lever arm fixed on the ankle joint and responsible for the main part of the absorption A_3 during the stance phase.

It is assumed that the knee joint absorbs and generates the same amount of energy during stance phase, therefore for this phase, the knee joint is not considered as a contributor to the ankle push-off generation. For this reason, elastic elements to mimic this behavior are not included in the conceptual design. Note that the potential energy is also playing important role during the gait.

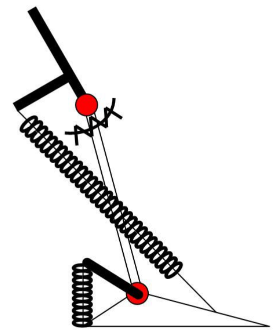

Fig. 4. Conceptual design of the proposed mechanism - The design presents three storage elements, one torsional spring at the knee joint, one linear spring between upper leg and foot via a lever arm and one linear spring at the ankle joint via a second lever arm.

A. Energy storage during swing phase

The swing phase of the human gait is an energy absorption phase for the knee joint and, therefore, the energy absorbed at the knee joint has to be transferred to the ankle joint. For the storage purpose, two elastic elements are employed, namely $\mathbb{C}_1$ and $\mathbb{C}_2$ during this phase. Part of the kinetic energy due to ankle push-off (A_1) is stored in the torsional spring $\mathbb{C}_1$. This energy is transferred to the the second storage element $\mathbb{C}_2$ during late swing phase. Specifically, $\mathbb{C}_2$ stores part of the energy transferred to the knee joint due to push-off (A_1) and the energy of the lower leg (A_2) during the swing motion.

Now the total absorbed energy (A_1 and A_2) has to be transferred from the knee to the ankle joint so to support the ankle push-off. To have an energy efficient system, this transfer should be realized without any dissipation. With this aim, we propose a design that allows the change of the attachment point of the elastic element $\mathbb{C}_2$ without changing its length. The energy transfer from the knee to the ankle joint is (ideally) without any use energy if the length is kept constant by defining a proper trajectory. Consequently, just after the full-flexion of the knee joint, the position of the attachment point moves freely from the heel (P_{fh}) to the upper part of the foot (P_{fu}) and, at the end of the swing phase, the attachment point moves freely back from the upper part of the foot to the heel. Hence, the total absorbed energy (A_1 and A_2) is transferred to the ankle joint in an energy-efficient manner. Fig. 5 shows the swing phase of the prosthesis conceptual design with particular attention to the changes in $\mathbb{C}_1$ and $\mathbb{C}_2$ ($\mathbb{C}_3$ is not depicted for simplicity). Note that the point P_u identifies the attachment point of the spring in the upper leg.

B. Energy storage during stance phase

The stance phase of the human gait is an energy absorption phase mainly for the ankle joint. The energy is stored at

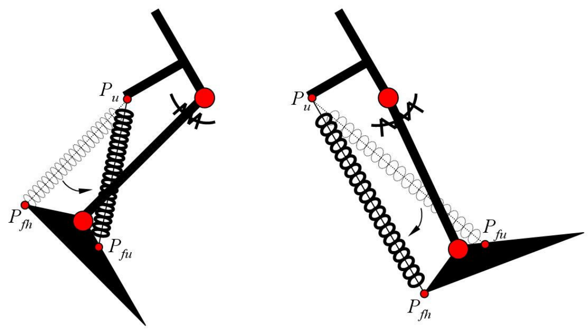

Fig. 5. The working principle at swing phase - After pre-swing phase, the attachment point of the spring $\mathbb{C}_2$ is changed from the heel (P_{fh}) to the upper part of the foot (P_{fu}) (left). At the end of the swing, the spring is loaded and its position changes back to the P_{fh} (right). The point P_u is the attachment point of the spring on the lever arm of the upper leg.

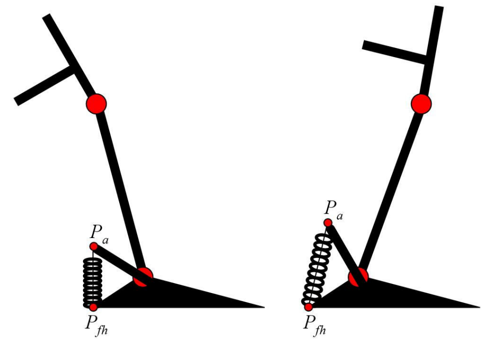

Fig. 6. The working principle at stance phase - In the beginning of the stance phase, the spring $\mathbb{C}_3$ is unloaded (left). At the end of the stance, spring is loaded (right). The point P_a is the attachment point of the spring on the lever arm and P_{fh} is the attachment point of the spring on the heel.

the ankle joint by means of the elastic element $\mathbb{C}_3$. During the stance phase, i.e. while ankle is in dorsiflexion motion, a braking torque is applied to the ankle in order to bear the weight of the body. Instead of dissipating the energy by using a brake system, we propose a design in which the storage elements provide the brake torque and, therefore, store the energy (A_3) during stance phase for delivering to the ankle push-off. In order to achieve this, a spring is connected between the heel (P_{fh}) and a lever arm (P_a), which is fixed to the ankle joint. Fig. 6 shows the stance phase of the prosthesis conceptual design with particular attention to the changes in $\mathbb{C}_3$ ($\mathbb{C}_1$ and $\mathbb{C}_2$ are not depicted for simplicity). Due to the kinematic relations between the lower and upper leg, element $\mathbb{C}_2$ is also employed to store some part of A_3.

At the end of the stance phase, the storage elements $\mathbb{C}_2$ and $\mathbb{C}_3$ are loaded and, therefore, are ready to release the total energy of all absorption phases (A_1, A_2, A_3) for the ankle push-off. Note that the swing storage element $\mathbb{C}_1$ is only active during the swing phase and the stance storage element $\mathbb{C}_3$ is only active during the stance phase. Therefore, there is no undesirable interference of the storage parts during the motion. Since the activation and deactivation of the storage elements can be realized when the velocities of the related joints (during walking) are zero, ideally no dissipation is present.

IV. DESIGN PARAMETERS WITH HUMAN KINEMATICS

In this Section, we derive the design parameters for the conceptual mechanism by using the energy absorption values of the healthy human gait. In particular, for both swing and stance phases, we identify the storage elements by using the bio-mechanical data for a human of 1.8 m height and 80 kg weight [21].

A. Swing Phase

The elastic constants of the springs employed for swing phase are derived from the energy values of the absorption intervals A_1 and A_2.

In particular, the elastic constant k_1 of the torsional spring $\mathbb{C}_1$ is determined from the energy A_1, i.e.: $A_1 = \frac{1}{2}k_1\delta s_1^2$, where δs_1 is the radial deflection of the torsional spring $\mathbb{C}_1$ and is equal to the variation of the knee angle $\delta\theta_k$, which is about 0.84 rad during this interval (between 52% and 72% of the stride). It follows that $k_1 = 0.261$ Nm/kg/rad.

The elastic constant k_2 of the linear spring $\mathbb{C}_2$ is determined from the sum of A_1 and A_2, i.e.: $A_1 + A_2 = \frac{1}{2}k_2\delta s_2^2$, where δs_2 is the deflection of the spring $\mathbb{C}_2$ and is given by $\delta s_2 = |\,\mathbf{P^{P_uP_{fu}}}\,| - s_{2_0}$, in which the magnitude of $\mathbf{P^{P_uP_{fu}}}$ is the length of the $\mathbb{C}_2$ element, when attached between P_u, and P_{fu}, and s_{2_0} is its initial length, which is 0.43 m in the beginning of late swing (see Fig. 5). It follows that $k_2 = 33.06$ N/m/kg.

B. Stance Phase

As stated in Section III, during stance phase, the energy is stored in both $\mathbb{C}_2$ and $\mathbb{C}_3$. It should be noted that, this parallel structure leads to smaller elastic constant for the element $\mathbb{C}_3$, which can be considered as an advantage for the design.

The elastic constants k_3 of the linear spring $\mathbb{C}_3$ is derived from the energy value of the absorption interval A_3.

During the stance phase, the deflection δs_2 of the storage element $\mathbb{C}_2$ is given by $\delta s_2 = |\,\mathbf{P^{P_uP_{fh}}}\,| - s_{2_0}$, in which the magnitude of $\mathbf{P^{P_uP_{fh}}}$ is the length of the element $\mathbb{C}_2$, when attached between P_u and P_{fh}, and s_{2_0} is its initial length, which is 0.52 m at the end of swing (see Fig. 5).

The deflection δs_3 of the stance storage element is given by $\delta s_3 = |\,\mathbf{P^{P_aP_{fh}}}\,| - s_{3_0}$, in which the magnitude of $\mathbf{P^{P_aP_{fh}}}$ is the length of the element $\mathbb{C}_3$, when attached between P_a and P_{fh}, and s_{3_0} is its initial length, which is 0.16 m at the beginning of roll-over (see Fig. 6).

The elastic constant k_3 of the stance storage element $\mathbb{C}_3$ can be found from the energy value of the absorption interval A_3, i.e. $A_3 = \frac{1}{2}k_2\delta s_2^2 + \frac{1}{2}k_3\delta s_3^2$, where k_2 is the elastic constant of the storage element $\mathbb{C}_2$. It follows that $k_3 = 694.45$ N/m/kg.

V. SIMULATION & RESULTS

In this Section, we simulate the conceptual mechanism in MATLAB Simulink environment. The model has been derived by using Kane's method [22]. To demonstrate the power absorption performance of the mechanism, simulation has been done for swing and stance phases separately.

The model of the prosthesis mechanism during the swing phase is considered in a sagittal plane with torso fixed in the Newtonian reference frame. Since the elastic element $\mathbb{C}_3$ for the stance phase is not active in this phase, it has not been considered in the model. For the simulation of the swing phase, the hip torque from healthy human data has been applied to the system as an external input.

The model of the prosthesis mechanism during the stance phase is considered in a sagittal plane with foot fixed in the Newtonian reference frame. Since the torsional spring on the knee $\mathbb{C}_1$ is not active in this phase, it has not been considered in the model. For the simulation of the stance phase, in addition to the hip torque, forces from the other leg which are assumed to be acting on the torso have been applied to the system as an external input.

Note that, since the model has been built to see the feasibility of the conceptual mechanism, all the mechanical losses and mass of elastic elements are neglected in the simulation. As previously stated in Sec. III, the action of the knee joint during stance phase is not considered as a contributor to the ankle push-off generation. For this reason, power flow for the elastic element to mimic this behavior is not included to the simulation results.

In the first simulation, the system has been constrained to the human kinematics in order to see the power flow in the system for the most ideal case. Fig. 7 illustrates the behaviors of the conceptual mechanisms (dashed lined) compared to the healthy human gait [20] (continuous line). The figure shows that the energy stored during the stance phase is almost the same in the two cases. However, during swing phase, the profile of power flow of the conceptual mechanism is different from the biological data: this is due to the power generation of element $\mathbb{C}_1$. In particular, since the element $\mathbb{C}_1$ generates all the energy stored during pre-swing phase (A_1), extra torque is exerted on the knee joint. Hence, this extra torque should be compensated by the hip joint, which, therefore, requires an extra metabolic energy. Even though the system appears to be mechanically efficient, the working principle of the storage element $\mathbb{C}_1$ should be changed so to avoid this extra generation. On the other side, during stance phase extra force from the other leg is required to satisfy the natural walking pattern which is expected from the non-linear elastic behavior of the natural ankle joint during dorsi-flexion in normal walking.

In the second simulation, we simplified the system by eliminating the element $\mathbb{C}_1$ in order to decrease the complexity of the system and examine only the energetic performance of the coupling element $\mathbb{C}_2$. The model has been simulated under the external forces and torques from the data of healthy human gait [20]. Since, the element $\mathbb{C}_1$ has been excluded from the system, elastic constant of the element $\mathbb{C}_2$ is changed to store only the energy of the interval A_2, i.e. $A_2 = \frac{1}{2}k_2\delta s_2^2$. It follows that, for this simplified model, $k_2 = 18.03$ N/m/kg and consequently $k_3 = 1194.45$ N/m/kg.

Fig. 8 illustrate the behaviors of the conceptual mechanism, with only $\mathbb{C}_2$ and $\mathbb{C}_3$ (dashed lined) compared to the healthy human gait [20] (continuous line). It can be observed that the energetic behavior of the mechanism during stance phase is comparable with the healthy human gait (approximately 85% of the possible amount of energy) whereas, during swing phase, the lack of the element $\mathbb{C}_1$ decreased the performance of the energy storage. Only 55% of the possible amount of energy can be stored during swing phase, which results in a storage of approximately 64% of the possible amount of energy that can be stored in the gait. Even though the performance has been decreased, the amount of energy that is stored in the system is still considerable for a passive system. On top of this energy, extra energy should be injected to the system in order to realize the ankle push-off generation. Since the system is fully passive, this energy will be generated with extra torque from the hip and the extra forces from the sound leg. The application of the forces and torques to compensate this energy is dependent on the human adaptation and the concept is evaluated with simulation under the ideal conditions.

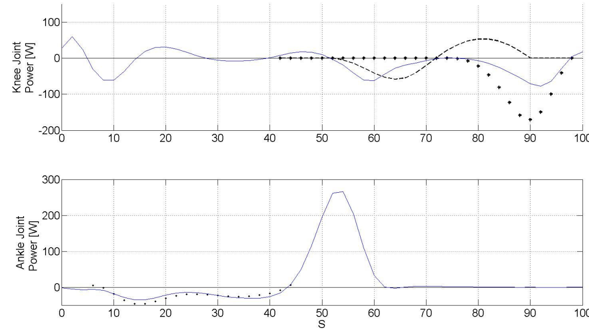

Fig. 7. The power flow of the healthy human gait [20] (continuous line) and the power flow for the conceptual mechanism (dashed line) during one stride.

In order to see the adaptation and the performance of the system in real conditions, we have built an initial prototype with the two storage elements $\mathbb{C}_2$ and $\mathbb{C}_3$ in a scale of $1:2$ according to the average human dimensions [23]. The animation of the working principle in 3D CAD is reported in Fig. 9 and a picture of the realized prototype is shown in Fig. 10 in a side-view to illustrate the elastic elements. The attached video to this paper shows the prototype during normal walking on a treadmill.

VI. CONCLUSIONS AND FUTURE WORKS

In this study, we propose a conceptual mechanism inspired by the power flow in the human gait for a transfemoral prosthesis. The conceptual mechanism consists of three elastic storage elements for the three absorption intervals in the healthy human gait. The working principle of the concept with these three storage elements has been described and the simulation has been implemented to see the power flow of the mechanism during the gait. Simulation results showed the possibility to store almost all the energy that can be stored during the gait in the expense of extra metabolic energy. Therefore, one of the elastic element that has a dominant effect on this metabolic energy requirement has

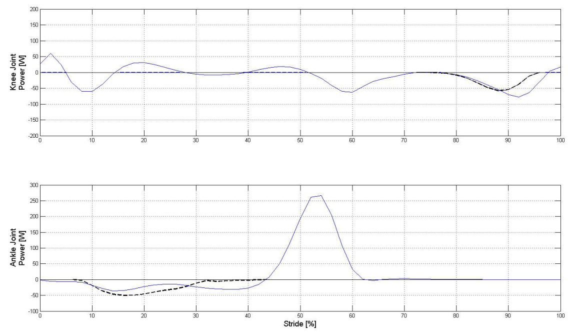

Fig. 8. The power absorption of the elements $\mathbb{C}_2$ (upper) and $\mathbb{C}_3$ (lower) during one stride in natural cadence.

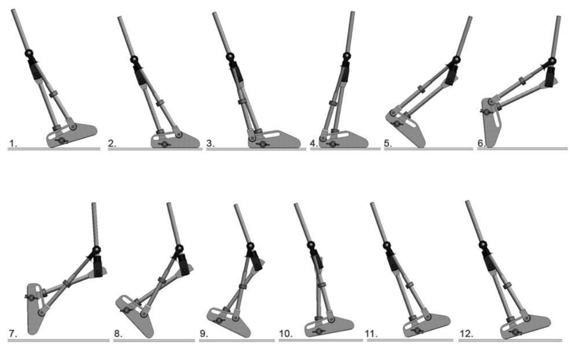

Fig. 9. Animation of one stride (from heel-strike to heel-strike) of the 3D CAD representation of the initial prototype, in scale $1:2$ with respect to the average human dimension.

been eliminated from the system and the simplified system has been simulated in order to see the contribution of the remaining storage elements to the energetic performance. This simulation showed that considerable amount of energy (64% of total absorption) can be stored in the system for ankle push-off generation. Since the system is fully passive, the rest of the energy should be provided from the human as a metabolic energy. Therefore, initial prototype in half scale of human dimensions has been built in order to check the feasibility of the concept in real conditions.

Evaluation of the concept will be done in a test setup with the initial prototype to improve the mechanism. The third elastic element will be implemented to the system by different working principle to improve the energetic performance of the mechanism.

VII. ACKNOWLEDGEMENTS

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Fig. 10. Side-view of the initial prototype in scale $1:2$ with respect to the average human dimension.

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