A lightweight and compact lockable parallel spring enhances the performance of a powered ankle-foot prosthesis

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***Abstract*— Lower-limb amputation has a major impact on locomotion and quality of life. The last few decades have seen the emergence of powered devices aiming at replacing the missing limb and restoring the lost mobility. These devices endeavor to reproduce as accurately as possible the biomechanics of lost biological joints. However, the engineering toolbox lacks actuators having the same force and power density as human muscles. Electromagnetic motors with gearboxes are usually chosen for their high power density but their poor force density imposes limitations in terms of backdrivability and noise due to the high reduction ratio required. This work proposes a lockable parallel spring mechanism providing a substantial amount of joint torque required for an artificial ankle-foot prosthesis, thus lowering the torque requirements on a complementary motor to a minimum. This allows a smaller gearbox to be selected, consequently improving the weight, bulkiness and backdrivability of the system. Experimental data collected on a testbench suggest the lockable parallel spring, tested in isolation, exhibits negligible hysteresis and an instantaneous engagement. It provides 24% more torque and harvests 53% more energy than a fixed parallel spring, better reproducing the biological dynamics.**

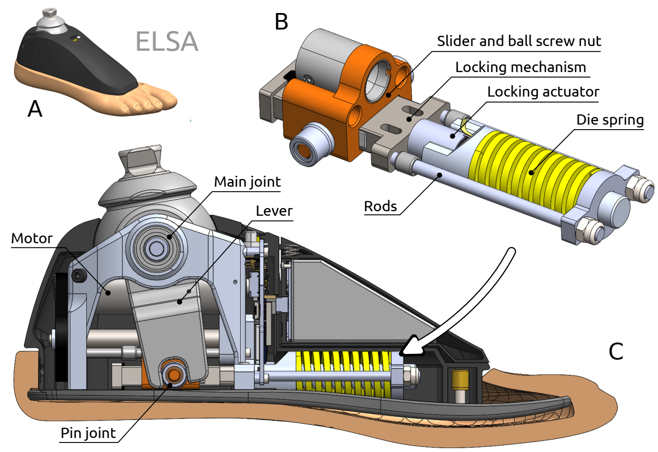

Fig. 1. System overview: (A) ELSA ankle-foot prosthesis, (B) Lockable parallel spring (LPS) assembly, (C) Cut view of the LPS inside the ELSA prosthesis.

I. INTRODUCTION

The world is currently facing a rising prevalence of lower-limb amputations, mostly due to peripheral vascular diseases, in particular diabetes mellitus [1]–[3]. Patients suffering from this condition are severely impacted in terms of locomotion capabilities and quality of life. To overcome this limitation, they are fitted with prostheses replacing the missing limbs. The vast majority of these devices are passive in the sense that they cannot provide the net positive energy required during daily life activities like overground walking. To cope with the lack of propulsion, uni-lateral amputees tend to compensate with the remaining joints, the knee and/or hip, often leading to osteoarthritis [4]. Furthermore, the weakening of muscle strength after amputation imposes an even bigger burden in terms of propulsion and balance [5], [6]. The altered biomechanics also leads to an asymmetric gait inducing back pain for a majority of individuals [7].

The decade-long development of active ankle prostheses, capable of providing mechanical work, is expected to normalize the gait of amputees, thus limiting the onset of biomechanics related comorbidities. These devices attempt to reproduce biomechanically relevant dynamics by delivering a fraction of the energy required during daily-life activities. Many ankle assistive devices have been developed targeting this goal [8]–[13]. Whereas these devices show promising results in terms of mechanical capabilities, they do not yet fulfill all the requirements needed for a complete appropriation by the amputated users. Indeed, not only is the mechanical outcome important but also, if not more, the safety, weight, encumbrance, energetic autonomy, comfort and cosmesis, including noise and appearance. Technical choices made by the aforementioned designs sometimes make it impossible to reach all these appropriation features at once.

In an effort to downsize the bulkiness of powered devices, it is common to add a compliant element in the system, either in series or in parallel. This spring can deliver energy at a high power during the gait and thereby lowers the peak power and/or torque of the actuator [14]. For series elastic actuator (SEA), as reviewed in [15], the motor is placed in series with the spring. Since the load displacement is equal to the sum of the motor and the elastic element displacements, power requirements on the actuator can, in certain circumstances, be improved. Downsizing of the actuator, increasing the efficiency and/or lowering system size and weight are the direct benefits of this [16]. Environmental disturbances are also absorbed by the elastic element, protecting the actuator from shocks. However, the series spring also reduces the control bandwidth.

In the parallel configuration, compliance is added in parallel with the actuator, such that the spring and the

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motor contributions add up to the total joint torque, thus lowering the torque required from the motor. This allows a lower gear ratio transmission to be selected which, in turns, leads to a better backdrivability and lower reflected inertia. Such a parallel compliant element has been reported in [12], [17]. In particular, we demonstrated in previous simulations [18] and early developments of the Efficient and Lightweight Spring Ankle (ELSA) prosthesis [19], [20] that combining a parallel spring that can be selectively engaged and disengaged with an actuator leads to a globally energy efficient system. Indeed, considering overground walking, adding such a parallel spring helps to lower the torque requirement of the complementary active module by 50% (Fig. 2B). We showed that this corresponds to a theoretical 24% reduction of electrical power consumption of the active side due to reduced losses in the actuator. An additional benefit of this hybrid architecture is the possibility to walk passively, the spring providing a baseline torque even with no action from the actuator. The complementary actuator can inject energy on demand, for activities that require assistance.

However, introducing a clutch/locking mechanism in the system in order to (dis)engage the parallel spring comes with drawbacks, as reviewed in [21]. Indeed, it is challenging to produce a locking mechanism with an infinite number of locking positions, large force capabilities, low power consumption, and being lightweight. The added benefit of a lockable parallel spring should be balanced with the added complexity needed to address these constraints.

In this paper, we propose the design and testbench validation of a lockable parallel spring (LPS) fulfilling these constraints. Rollers wedging between non parallel surfaces are leveraged to produce a very lightweight and compact locking mechanism to couple the joint motion to a parallel spring. Section II details the electromechanical design and Section III the related controller. The following Section IV provides the results of a testbench validation. These results are discussed in the last section.

II. MECHATRONIC DESIGN

The development of a lockable parallel spring is motivated by the singular kinematics and kinetics of the human ankle during walking. Indeed, the position and torque trajectories of the joint suggest that a substantial amount of torque could be generated passively from the motion, as shown in Fig. 2. In order to achieve the best performance, the parallel elastic element should be selectively engaged during the stance phase and disengaged during the swing phase. Its engagement position should be the minimum plantarflexion angle following heel strike (MPA), maximizing the harvested elastic energy from the motion. However, it should be disengaged during the swing phase in order not to impede the foot return motion, ensuring the complementary motor does not have to work against the spring. It should provide a silent, instantaneous engagement and exhibit negligible hysteresis to maximize the amount of energy returned to the user. The complementary actuator could then provide the difference between the target joint torque and the torque generated by the parallel element as depicted in the same figure. The stiffness of the spring should be tailored to each user to ensure a comfortable and energy efficient operation.

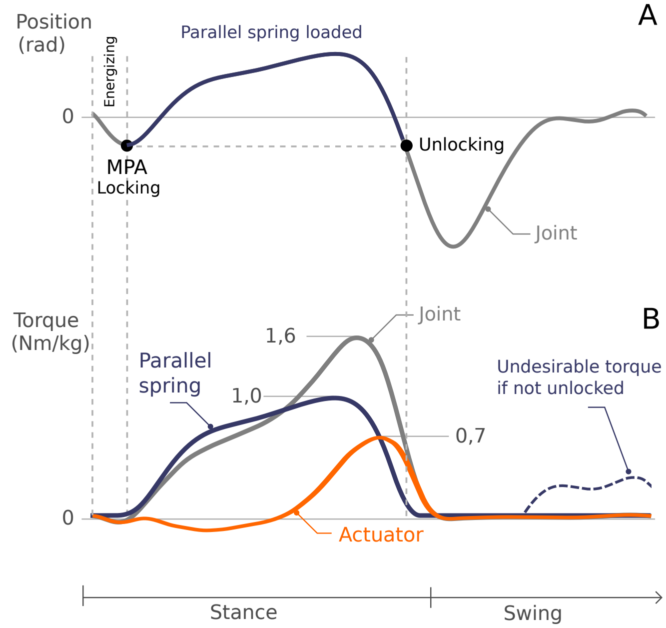

Fig. 2. Ideal parallel spring behavior: (A) Ankle joint position with locking events, (B) Ideal torque contributions from the motor and the parallel spring according to simulations [18]. Joint reference trajectories are taken from [22].

The design of the proposed lockable parallel spring is inspired by the Achilles tendon and the plantar fascia of the human ankle acting as elastic energy storage elements during walking [23]. The LPS mechanism can be unidirectionnally linked to the ball screw linear nut slider of the prosthesis actuator, i.e. the motion of the parts can be either coupled in one direction or completely free. The nut slider is itself connected to the main joint through a lever and pin joint (Fig. 1B and C). When the locking mechanism is energized, plantarflexion of the joint remains free (unidirectionnally locked) but as the speed reverses and the joint enters dorsiflexion, the LPS engages instantly.

This paper proposes an innovative lockable parallel spring (LPS) specifically tailored for seamless integration with the ELSA prosthesis [24]. This prosthesis is composed of an actuation unit affixed to a carbon fiber foot plate. The actuation unit comprises a BLDC motor, a ball screw transmission, custom electronics and a 3D printed (Nylon, PA12) shell. The rigid motor-transmission unit is complemented by the novel lockable parallel spring described below and depicted in Fig. 1. Notably, the entire device is engineered to fit compactly within the confines of the biological foot volume. The lockable parallel spring total weight amounts to 128 g.

A. Locking principle

The wedging of rollers between non-parallel surfaces is leveraged to create a unidirectional locking action with an infinite number of locking positions. This happens by bringing rollers in contact with a carriage and a slider, as

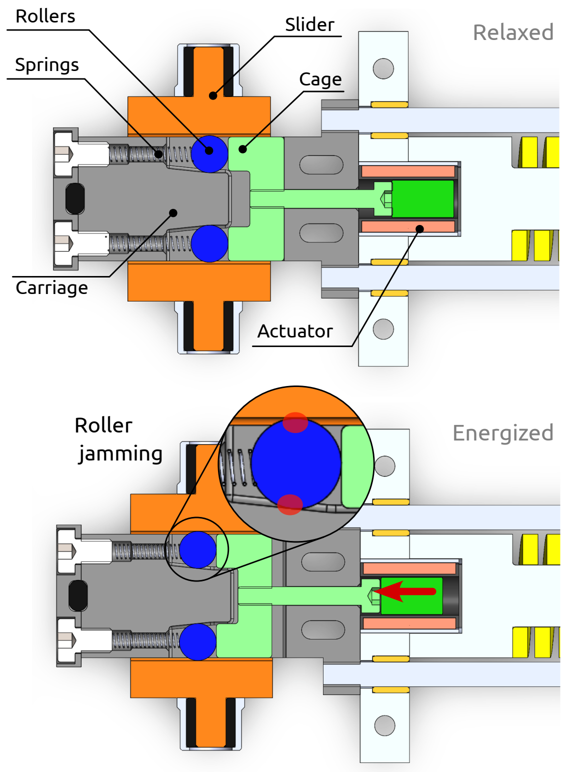

Fig. 3. Locking mechanism operation (cut view): when energized, the actuator pushes the cage that brings the rollers in contact with the slider and carriage surfaces leading to a unidirectional wedging configuration.

shown in Fig. 3. It cannot be unlocked under load and does not require power to remain locked. Disengagement happens automatically when the load disappears. In such friction based mechanisms, the normal contact forces are much larger than the tangential forces to be locked. This implies mechanically performant materials that can handle large fatigue contact stresses. SAE5200 bearing steel hardened by quenching was selected to achieve this. The locking system weights $51\,\mathrm{g}$, i.e. a very low weight given the high forces transmitted.

B. Locking actuator

The locking mechanism is monostable: a custom short stroke linear actuator is energized to move the roller cage against internal springs. This brings the rollers in contact with the carriage and the nut slider. Wedging of the rollers then happens automatically as soon as the joint transitions to dorsiflexion. As depicted in Fig. 4, the linear actuator is composed of a PCB winding stator with back iron and a moving magnet. The coil is optimized to generate the maximal thrust under the nominal working conditions (10% duty cycle), leading to an almost constant 4 N force along the 3 mm stroke. The actuator power consumption is $10\,\mathrm{W}$, but only active for a short amount of time at heel strike. Over a typical gait cycle, the actuator energy consumption is around $1\,\mathrm{J}$.

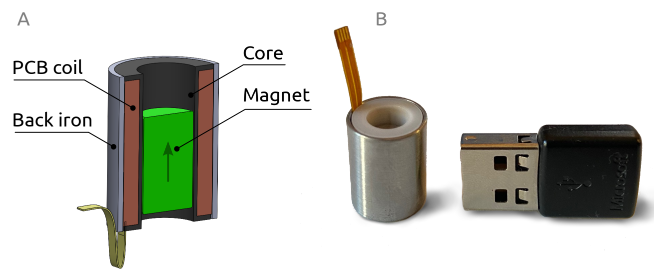

Fig. 4. Custom linear actuator: (A) Cut-view, (B) Actuator next to a USB dongle for size comparison.

C. Spring embodiment

The elastic characteristic is rendered by a spring and a transmission system. A pulling motion on the carriage is transmitted by rods to a spring that consequently deforms, producing an elastic response. Regarding the spring embodiment, following a first attempt with polymers [25] that exhibited too much hysteresis, a compression die spring (5410 from Lesjofors, Sweden) was selected with a stiffness of $177\,\mathrm{N/mm}$, resting length of $38\,\mathrm{mm}$ and weight of $77\,\mathrm{g}$ including its transmission. In combination with the joint lever, it produces an apparent stiffness at the joint level of $442\,\mathrm{Nm/rad}$. The spring was selected based on simulation [18] in order to best reproduce literature data [22] for a $80\,\mathrm{kg}$ user.

D. Graceful degradation

As a safety feature, the parallel spring always engages above a certain joint angle (mechanical stop at $0\,\mathrm{rad}$, pylon vertical), no matter the locking state, providing elastic support in a degraded mode, i.e. the spring engages latter in the stance phase than when actively controlled. In combination with passive motor damping with appropriately chosen dissipation resistors, this creates a critically damped passive spring-damper system, allowing the user to walk comfortably even with the device being turned off.

III. Engagement control

As shown in Fig. 5, in its active walking operation, the prosthesis engages a state machine encompassing three distinct states: stance, push-off, and swing [24]. During the stance state, the coil is energized in response to the detection of heel strike, i.e. when a plantarflexion motion is sensed. This motion is identified through the confluence of a plantarflexing speed and the attainment of a specific plantarflexion position. The coil remains energized throughout early stance until the moment speed reverses. At this juncture, the mechanism transitions into a self-locking configuration, steadfastly maintaining this state until the push-off phase, where the spring undergoes complete unloading. The mechanism is monostable, causing it to automatically reset during the swing phase waiting for a new heel strike to be detected. The unidirectional and infinitely adjustable locking mechanism presents an added advantage by eliminating the need for precise timing of the gait phase. This innovative

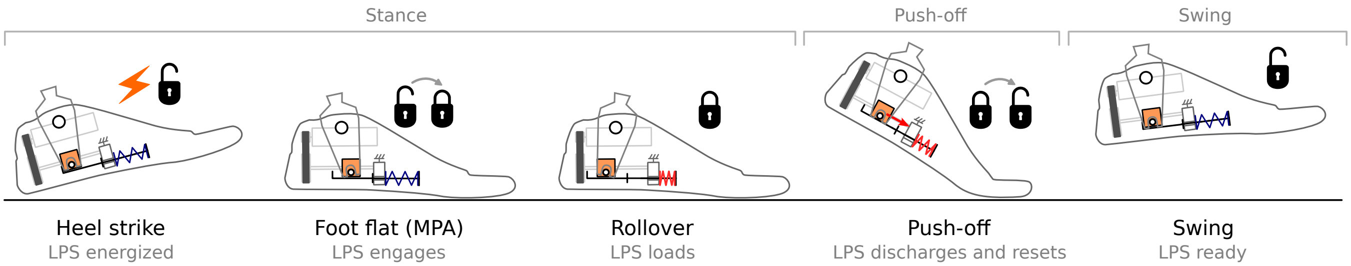

Fig. 5. Snapshots of the complete walking gait cycle from heel strike to swing with the main LPS events displayed.

design streamlines control requirements to the simple detection of heel strike and sufficient push-off plantarflexion to reset the device. Also, by design, the engagement position can change at each step depending on the kinematics of early stance.

IV. Benchtop validation

The suggested locking mechanism underwent testing on a specialized testbench to evaluate its performance across key metrics, including engagement speed, hysteresis, and peak torque.

A. Setup

A dedicated testbench capable of actuating the ankle joint while recording the reaction torque was used to evaluate the LPS. This device includes a single 220 W motor (Nanotec DB87S01), 40:1 gearbox (Apex Dynamics PEII090-040), and 2:1 belt drive. Joint torque is measured by a custom torque sensor and joint position is measured by a hall effect sensor AS5048 (accuracy 0.02°). A custom controller tracks reference joint trajectories. The LPS itself was mounted in a partially assembled ELSA prosthesis composed of only strictly required components such as the structural frame, joint lever, and axle. This way, only the LPS contribution is measured with no impact from the actuator or its gearbox working in parallel. The device was clamped on a fixed mounting plate and the pylon was actuated by the bench actuator. The prosthesis custom electronic board powered by the testbench was used to control the LPS actuator. The complete setup is depicted in Fig. 6.

B. Protocol

The testbench was instructed to replicate a typical ankle position profile, derived from prior experiments involving amputees wearing the prosthesis and illustrated in Fig. 7B. This trajectory is based on [22] but with a reduced amplitude push-off to accommodate the prosthesis limited range of motion. In the initial test condition, corresponding to normal operation, the LPS actuator was manually triggered by the operator slightly before the maximal plantarflexion angle (MPA) and energized for a fixed $200\,\mathrm{ms}$ duration. In this condition, the spring then physically engaged as soon as the speed reversed, corresponding to the initiation of dorsiflexion. To facilitate manual operation of the locking actuator, the gait cycle frequency was lowered to $0.2\,\mathrm{Hz}$ compared to a normal walking frequency of $1\,\mathrm{Hz}$. This first experiment was compared to a control condition where the LPS remained inactive and the parallel spring was engaged at its mechanical stop, i.e. when the joint position passes the vertical $(0\,\mathrm{rad})$, in order to emulate a fixed spring. Torque and position data were recorded at $100\,\mathrm{Hz}$.

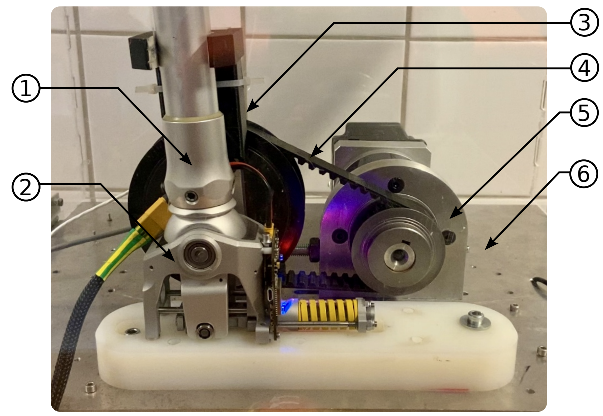

Fig. 6. Testbench setup: (1) pylon, (2) LPS mounted inside a partially assembled prosthesis, (3) custom torque sensor, (4) belt transmission, (5) motor and gearbox, (6) mounting plate.

C. Data processing

The torque and position signals were filtered using an 8^{th} order low-pass Butterworth filter with a cutting frequency of $10\,\mathrm{Hz}$. Cycles were identified from the recording and the mean trajectory over 10 cycles was computed. Since data was collected on a testbench repeating the same motion, variability over the 10 cycles was very low and is not reported in the result figures. Energy storage is computed by integrating the torque-position profile during the loading phase.

D. Results

The resulting mean trajectories for both test conditions are illustrated in Fig. 7. A movie showing the experiment is also provided as supplementary material. Under the control test condition, where the spring engages at the mechanical stop, a linear torque-position relationship is observed, exhibiting the anticipated spring stiffness of $442\,\mathrm{Nm/rad}$ and a peak torque of $75\,\mathrm{Nm}$. The overlapping loading and unloading trajectories of the spring indicate a very low hysteresis, suggesting an excellent energy return, as could be expected from a die spring.

In the experiment involving the LPS, the torque starts to rise earlier, specifically just after the minimum plantarflexion angle following heel strike (MPA). The apparent stiffness then gradually increases before reaching the nominal spring value. This is attributed to the rollers progressively wedging between the contact surfaces. This corresponds to a small internal motion within the locking mechanism, i.e. $\delta_e = 2\,\mathrm{mm}$ (LPS) or $0.045\,\mathrm{rad}$ (joint equivalent). The peak torque achieved is $93\,\mathrm{Nm}$, 24% higher than that in the control condition, implying enhanced energy storage (53% higher). Additionally, hysteresis remains negligible, affirming an energy return comparable to the control condition, albeit with a higher magnitude given the higher amount of energy stored. The locking actuator was activated for a fixed $200\,\mathrm{ms}$ per cycle, leading to a $2.8\,\mathrm{J}$ energy consumption per cycle. This duration is longer than what should be required during a normal walking cycle, thus overestimating the typical consumption of the system. Indeed, the testbench cycle was slower than normal walking and the actuator was not shutdown upon detection of dorsiflexion even though the mechanism is self-locking under load.

A reference torque-position profile for a $80\,\mathrm{kg}$ user, from [22], is also depicted as reference in Fig. 7. The LPS condition contributes a more substantial portion of the reference torque, implying that a complementary motor would need to supply less torque to track the trajectory. The LPS also provides a loading response closer to the biological target than the control condition. In particular, the quasi-zero torque portion of the curve beyond MPA illustrates that the parallel spring is disengaged as expected during the simulated swing phase (foot return).

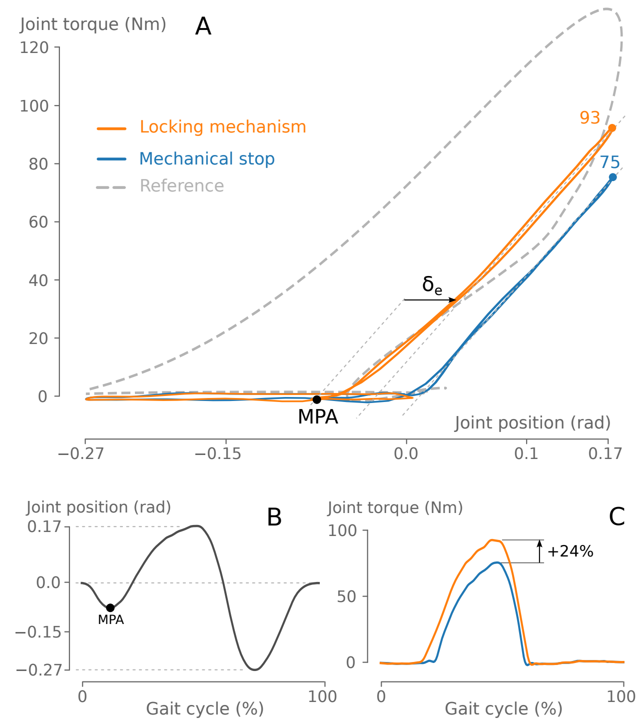

Fig. 7. Benchtop validation: (A) Torque-position trajectories averages over several cycles with active locking and without (mechanical stop). The LPS engages instantly after the MPA and exhibits a progressively increasing stiffness over δ_e. Reference from [22]. (B) Position profile tracked by the bench. (C) Torque profiles for both configurations showcasing a 24% increase in peak torque with the LPS.

V. Discussion & Perspectives

Using a parallel spring holds the promise of significantly reducing the requirement of the motor in a powered ankle prosthesis, pending that this spring could be disengaged during the swing phase. The preceding section highlighted a significant increase in peak torque and harvested elastic energy when employing a locking mechanism compared to a fixed parallel spring. Designing a lightweight locking mechanism with features such as infinite locking positions, low power consumption, high force capability, and self-locking under load poses a technical challenge, as emphasized in [21]. We showed that at the cost of a small added weight, an actuator working in parallel to this lockable parallel spring would need to generate much less torque. This creates a virtuous circle, enabling the selection of a lower gear ratio, a more compact and lighter transmission, thereby improving backdrivability by minimizing the reflected inertia at the joint. This offsets the small added weight of the passive spring. Additionally, the presence of a parallel spring allows for graceful degradation in case of empty battery, motor failure or during low activity tasks, by providing a baseline torque, enabling users to continue walking, albeit with no active assistance.

To the best of our knowledge this is the first time such a compact, high torque and energy efficient lockable parallel spring for a prosthesis is presented and validated. This design improves on preceding work from [12], [26] by proposing an even lower power, more compact and lighter solution. Furthermore, it only requires heel strike detection for control since physical engagement relies on a self-energizing mechanics driven by the joint motion. Even though the infinite number of locking positions feature and low noise level were not formally validated in this paper, it comes as a direct consequence of the mechanical design. This could be validated during experiments involving more variability in the kinematics.

It is also important to note that the minimum plantarflexion angle (MPA) at which the mechanism engages is not determined in advance but will vary with every step and user. Indeed, it results from the dynamics at heel strike, including the prosthesis resistance and ground reaction forces. Moreover, when ascending a ramp or during stairs ambulation, there is usually no plantarflexion after heel strike. In that case, the spring is simply engaged by the mechanical stop and the locking mechanism is not used. On the contrary, when walking down a ramp, depending on the prosthesis damping properties, a significant amount of plantarflexion can be experienced. In such a case, the LPS can be triggered later than the MPA so that it does not engage too early and does not generate too much resistance against dorsiflexion. This is especially needed when walking downstairs as the heel strike related plantarflexion is more pronounced and locking early might prevent a complete rollover. Finally, the mechanism would work equally well for any walking speed, the actuator having a bandwidth substantially faster than the walking dynamics.

To further validate performance, additional tests should be conducted at nominal gait speed. The mechanism should also be validated with a higher stiffness spring targeting heavier subjects. Moreover, given the high stress levels on the components, longer-term studies are crucial to evaluate fatigue performance of materials. Finally, integrating the mechanism into a complete motorized prosthesis and testing with amputee users are essential steps to assess its efficacy in daily-life activities and its proper synergy with the complementary active motor.

VI. Conclusion

This paper described the design and validation of a novel lockable parallel spring to be fitted within an ankle prosthesis. The aim was to reduce the torque requirements of a complementary motor, thus lowering the gearbox ratio, bulkiness and weight of the total device. A lockable solution, engaging at the minimum plantarflexion angle following heel strike, optimally harvested elastic energy from the gait cycle without impeding the foot return during the swing phase. The solution relied on wedging rollers to create a unidirectional locking action. A small and energy efficient monostable actuator was used to trigger the locking action. The system is lightweight and capable of sustaining significant torque typically experienced during walking. Compared to a fixed spring, 24% more torque could be provided by the passive spring, hence a reduction on the requirements on a complementary motor. Moreover, this passive spring provided graceful degradation rendering the behavior of a passive elastic device if ever the motor was inoperative. Future work includes testing the mechanism inside a prosthesis with an amputee subject performing daily-life locomotion activities.

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