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Bio-inspired design and validation of the Efficient Lockable Spring Ankle (ELSA) prosthesis

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***Abstract*— Over the last decade, active lower-limb prostheses demonstrated their ability to restore a physiological gait for transfemoral amputees by supplying the required positive energy balance during daily life locomotion activities. However, the added-value of such devices is significantly impacted by their limited energetic autonomy, excessive weight and cost preventing their full appropriation by the users. There is thus a strong incentive to produce active yet affordable, lightweight and energy efficient devices. To address these issues, we developed the ELSA (Efficient Lockable Spring Ankle) prosthesis embedding both a lockable parallel spring and a series elastic actuator, tailored to the walking dynamics of a sound ankle. The first contribution of this paper concerns the developement of a bio-inspired, lightweight and stiffness adjustable parallel spring, comprising an energy efficient ratchet and pawl mechanism with servo actuation. The second contribution is the addition of a complementary rope-driven series elastic actuator to generate the active push-off. The system produces a sound ankle torque pattern during flat ground walking. Up to 50% of the peak torque is generated passively at a negligible energetic cost ($0.1\,\mathrm{J/stride}$). By design, the total system is lightweight ($1.2\,\mathrm{kg}$) and low cost.**

I. INTRODUCTION

The past decade has seen a lot of research seeking to improve locomotion capabilities of lower-limb amputees, by providing safe, energy-efficient, and user-friendly prostheses replacing their missing limb [1]–[5]. A related expected impact of these research efforts is to increase the use of lower-limb prostheses by dysvascular amputees, representing 70% of all lower-limb amputees [6]. These strongly disabled patients face tremendous difficulties to use classical prostheses owing to the challenges associated to (i) providing more energy with the their other joints [7], (ii) ensuring the overall body stability, and (iii) managing the cognitive effort which is required to walk with a prosthesis, mainly if it is passive [8]. In addition to these functional objectives, the design of a prosthesis must be guided by several other important criteria: safety, weight, encumbrance, energetic autonomy, comfort and cosmesis, including noise and appearance. Prosthesis design is also highly constrained by the stump/prosthesis connexion. Although progress has been made recently towards osseo-integration [9], the vacuum-fixed socket still remains the most usual solution for achieving the human-prosthesis physical anchoring. As a consequence, stump soft tissues have to cope with large pressures associated with weight bearing and dynamical transfer of propulsive forces. Furthermore, the swing motion of the leg can lead to uncomfortable inertial efforts due to the non-negligible mass of the prosthesis. This problem is even more critical for transfemoral amputees with a short stump, and critically limits the device usability for an extended period of time. The comfortable weight limit highly depends on the stump length, location and the activity level of the user. In sum, there are strong incentives for minimizing the weight of lower-limb prostheses.

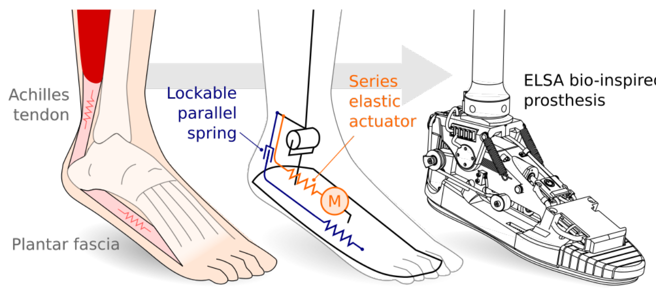

Fig. 1. ELSA active ankle prosthesis comprising a series elastic actuator (SEA) and a lockable parallel spring (LPS) inspired from the configuration of the plantar fascia and achilles tendon of the sound ankle.

These prostheses can be divided into passive and active devices. Only active ankle prostheses can provide the net positive energy being required during flat ground walking, and more complex tasks such as slope and stair ascend. This was recently demonstrated in [1], where the authors corrected the gait pattern of transtibial amputees using an active device, both regarding kinematics and metabolic cost. However, actuated systems tend to be bulkier and heavier than their passive counterparts. In an effort to reduce weight and encumbrance, existing devices embed series elastic actuators (SEA), this principle being reviewed in [10]. If correctly tuned, SEA might have a direct effect in decreasing the motor speed and thus further decrease the required peak electrical power. In sum, this offers to equip the prosthesis with smaller motors. Moreover, some mechanisms can be added to make the rendered stiffness variable, further improving this power tuning [11].

Yet, active prostheses are facing another big challenge, namely energetic autonomy. In order to maximize the efficiency, the actuator torque profile should also be minimized. Indeed, the motor torque is proportional to its current, and

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the motor Joule losses are proportional to the square of this current. Targeting this torque reduction, a common solution within lower-limb prostheses is to embed a parallel spring, passively generating torque on top of the SEA. Consequently, the actuator produces only the remaining fraction of the whole requested joint torque. As detailed in our previous work [12], ideally, the parallel element should be unidirectional and should engage only above a certain joint angle, so that no torque is produced below that angle and torque ramps up above this threshold. In existing ankle prostheses, this parallel compliant element is implemented in two different ways, depending on the joint angle where torque production is triggered. The first type engages at a fixed joint angle, that is carefully chosen to be outside the range of motion of the swing phase, see e.g. [1]. This type can only produce a torque boost in late stance. The second type can dynamically change the angle of engagement. This requires a clutch mechanism to anchor the spring when needed. Engaging early in the stance phase offers to store more elastic energy but requires the parallel spring to be disengaged during the swing phase. Such adaptive parallel springs heavily rely on appropriate locking mechanisms, e.g. in [13], most having been reviewed in [14].

As mentioned above, global weight is another critical design issue for a prosthesis. To the best of our knowledge, there is currently no untethered propulsive bionic feet as light as their biological counterpart (foot and calf respectively 1.35% and 4.20% of body weight, i.e. $1.0\,\mathrm{kg}$ and $3.2\,\mathrm{kg}$ for a $75\,\mathrm{kg}$ individual [15]).

Building upon previous simulation results [12] and early developments [16], the present work focuses on the development of a lightweight and affordable active ankle prosthesis. More precisely, this paper validates a complete ankle-foot prosthetic device embedding a parallel spring with an ultra low power locking mechanism. The proposed design takes advantage of a bio-inspired elastic nylon rope transmission leading to an implementation that is lightweight, compact and achieves high mechanical performances (Fig. 1).

The paper is structured as follows. Section II briefly summarizes our simulation results and design guidelines. Based on these objectives, Section III describes the proposed new electromechanical design and Section IV the associated controller. Experimental validation results are reported in Section V, and the paper ends with a discussion and conclusion.

II. Approach

With the objective to minimize the power consumption, weight and price of active ankle prostheses, we demonstrated — based on previous simulations [12] and early developments [16] — that combining a lockable parallel spring with a series elastic actuator (see Fig. 1) leads to a globally energy efficient system. Indeed, considering the flat ground walking task, adding a lockable passive elastic element helps to lower the torque requirements of the complementary active module by 50%, as depicted in Fig. 2. 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. The active module is needed to provide the extra energy used by the biological joint during propulsion (16J/stride for a 75Kg individual [17]).

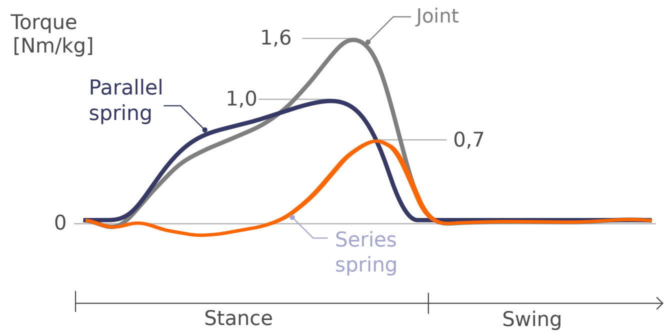

Fig. 2. Normalized torque profile of the ankle during normal speed walking (grey) [17], split between the contributions from a passive elastic element (blue) and a series elastic element (orange), according to our previous simulations [12]

However, in order to achieve such performance, the passive elastic element is selectively engaged during the stance phase and disengaged during the swing phase. Its neutral position should be the minimum plantarflexion angle following heel strike. This maximizes the harvested elastic energy but does not impeded the foot return during the swing phase. The series elastic actuator is designed to provide the difference between the target joint torque and the torque generated by the passive element. The following section details the mechanical implementation which achieves these objectives.

III. Design

This section describes the mechanical implementation of the lockable parallel spring and the series elastic actuator. The design is strongly driven by bio-inspiration. At the same time, cost, weight and external appearance are critical design constraints.

A. Lockable parallel spring

The design of the 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.

As depicted in Fig. 3, the elastic characteristic is rendered by a nylon rope (3). One end of the rope is wound around the spring loaded axle of a locking mechanism (5). The other end, after circulating around a shank lever pulley (2) and a foot heel pulley (6), can be clamped anywhere along the plantar region of the prosthesis (9). As such, the desired stiffness of the mechanism can be finely adjusted. Indeed, for a given load, the rope will deform as a function of its length. Changing the length then changes the rendered stiffness of the rope. This principle is used to adjust the desired stiffness to the user's weight and preference.

The rope drum locking system is composed of a ratchet and pawl mechanism (4,7). The small diameter of the drum in combination with the hoist configuration generates a large reduction ratio. With a $6\,\mathrm{mm}$ drum diameter (5) and a $50\,\mathrm{mm}$

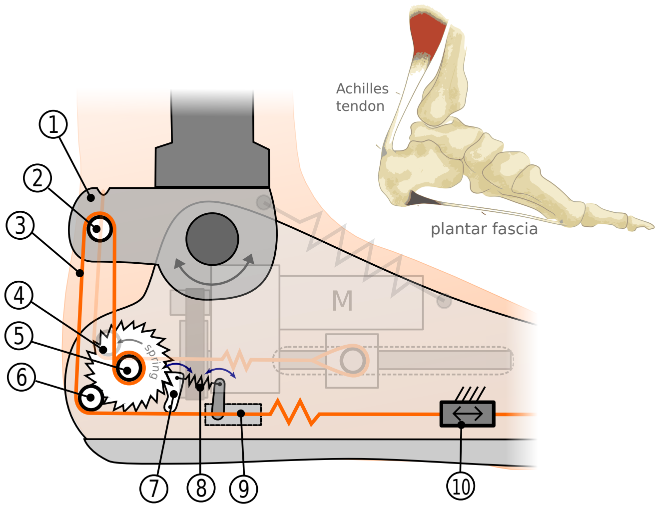

Fig. 3. Bio-inspired lockable parallel spring: (1) shank lever, (2) shank pulley, (3) nylon elastic rope, (4) ratchet wheel, (5) ratchet drum, (6) heel pulley, (7) pawl, (8) elastic link, (9) micro servo, (10) adjustable clamp.

shank lever (1), the reduction ratio is 33:1 such that a $80\,\mathrm{Nm}$ joint torque becomes only $2.4\,\mathrm{Nm}$ at the locking axle. Therefore the locking mechanism can be designed small and lightweight. The large reduction ratio also minimizes the impact of the discretized locking positions due to the teeth of the ratchet and pawl mechanism such that with 36 teeth the discretization seen at the joint is only 0.3 degrees (see [16] for details).

The pawl is actuated via an elastic link (8) coupled to a micro servo motor (9) which switches between a locked and unlocked position. The elastic link decouples the timing between actuation and engagement of the device. Indeed, as the servo closes, due to the elastic link and the asymmetric geometry of the teeth, the pawl is still able to retract as the spring (not shown) rewinds the rope during the plantarflexion motion following heel strike. As the joint speed reverses (dorsiflexion) the pawl engages and the system is loaded. The closing actuation time can thus precede the exact moment of maximum plantarflexion. Similarly, when the servo opens, due to the friction of the teeth when the system is loaded and thanks to the elastic link, the pawl keeps engaged until the load disappears. This design again imposes little constraint regarding the accuracy of the opening actuation time.

This mechanism always engages at the maximum plantarflexion angle following heel strike and this angle varies depending on the terrain, slope and gait style. Consequently, the maximum amount of elastic energy can be passively harvested, thus reducing the load on the complementary active module. From now, this system will be referred to as the lockable parallel spring (LPS).

B. *Series elastic actuator*

In order to properly render the push-off motion, an active module must provide the required mechanical energy. As depicted in Fig. 4, the proposed design uses a similar elastic transmission rope (2) coupled to a ball screw actuator. The motor (5) transfers power to the ball screw (6) via a belt-pulleys transmission (4) providing a first 3:1 gear ratio. Then the $2\,\mathrm{mm}$ pitch ball screw converts the rotative motion to a linear motion. The nut is guided by a slot on either side and pulls on the elastic rope to generate tension. The tension is transfered via a pulley (3) to the shaft lever (1) which converts the rope tension into a joint torque. By nature, this system is unidirectional, i.e. the motor can only pull, but this is compatible with the biological torque target (see Fig. 2). A soft tension spring (7) provides the little force required to bring the joint back to its neutral position during the swing phase. By controlling the position of the nut with respect to the position of the joint and given a prior characterization of the rope, one can control the applied torque. This allows the system to be seen as a torque source for higher level controllers such as, for instance, a muskuloskeletal model or an impedance controller, as reviewed in [18].

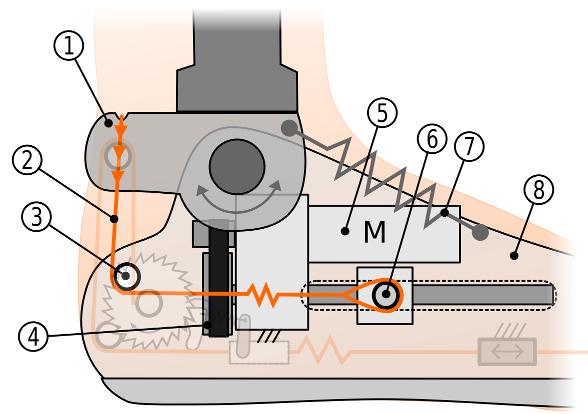

Fig. 4. Series Elastic Actuator: (1) shank lever, (2) elastic rope transmission, (3) pulley, (4) belt-pulleys transmission, (5) Motor, (6) nut on ball screw, (7) foot return tension spring, (8) foot body

C. *Prototype*

Based on the design proposed in the previous section, a prototype was made using rapid prototyping techniques. The prosthesis is a single degree of freedom device and is made as a two parts plastic assembly (PA 2200, SLS printing). It is thus both lightweight and cheap to manufacture. All other parts except the ratchet and pawl are off-the-shelf components requiring little to no machining. The complete system weights $1.28\,\mathrm{kg}$ (not including batteries) and the prototype cost is around \$1000. The parallel spring tendon is composed of two strands of Ø $2.5\,\mathrm{mm}$ nylon rope (max $250\,\mathrm{Nm}$ joint torque) while the SEA tendon is composed of one strand of Ø $2\,\mathrm{mm}$ aramid (max $250\,\mathrm{Nm}$ joint torque), both chosen to safely sustain the torque required by a $75\,\mathrm{kg}$ individual (around $120\,\mathrm{Nm}$) due to the reduced tensile strength associated with rope bending and fatigue. It is also suitable for users with a very low amputation level since this connection comes just above the joint ($110\,\mathrm{mm}$ pyramid anchor height). The device's shape fits within the envelope of a EU40 biological foot size. Fig. 5 shows a CAD model of the system and the real prototype.

The electronics is composed of the VESC open source motor controller that controls a Faulhaber BP4 brushless Serie 2264 motor and runs a ChibiOS real-time operating system. A RaspberryPi zero runs higher-level code implemented as ROS nodes and provides wireless communication. The aluminum standard pylon and pyramid adapter were kindly provided by TruLife Prosthetics (Sheffield, UK). The system can be physically scaled according to the size, torque and power requirements of the user (child, adult, ederly).

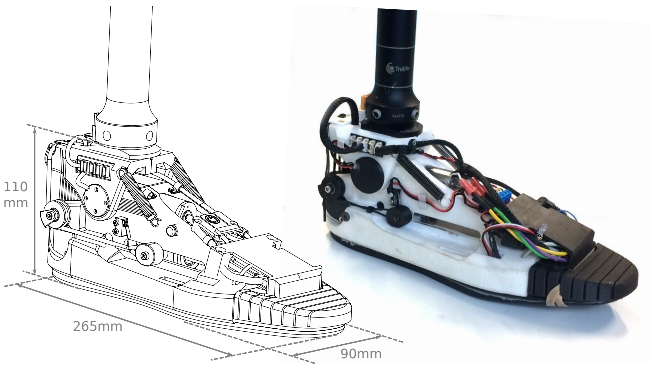

Fig. 5. CAD model of the ankle device (left) and assembled prototype to be tested (right)

IV. Control

The control of the locking mechanism and the series elastic actuator are detailed in the following sections.

A. Lockable parallel spring

The controller of the lockable parallel spring is a state machine that switches between a locked and an unlocked state depending on the walking gait phase. However, engagement and disengagement of the locking mechanism happen later than the actuation and are monitored by the controller as depicted in Fig. 6. The controller keeps track of the locking angle and has a feedback about the physical engagement state. The system is most of the time locked (only in dorsiflexion due to the geometry of the ratchet's teeth). The system unlocks only when push-off is detected, i.e. a plantarflexion event is detected after loading of the prosthesis. The actuator then switches to the unlocked position but the ratchet only opens when the load returns to zero. The system remains opened until the foot return motion of the swing phase is completed. The actuator then locks again to be ready for the next heel strike. This architecture offers graceful degradation in case the system runs out of energy. The locker remains closed and the prosthesis behaves like a passive system.

B. Series elastic actuator

The controller is based on a nested loops configuration from the motor current up to the generated torque as shown in Fig. 7. The torque controller estimates the actual generated torque $\hat{\tau}$ from an experimentally identified stiffness function

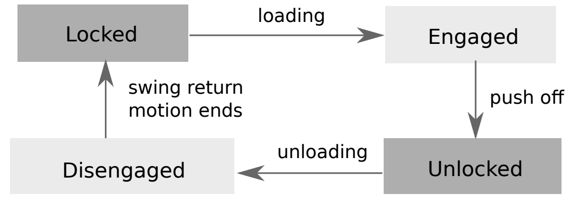

Fig. 6. Locking state machine: locking states split between the actuation and engagement states

$k(x, x_n)$ depending on the the ball screw nut position x and the nut neutral position x_n (See Fig. 4 and 8). Here, a linear model was used (Hooke's law). The nut neutral position is the joint angle dependent nut position at which no torque is generated and is also experimentally calibrated, then estimated as a function of the ankle joint angle, $x_n = \hat{g}(\theta)$. The torque controller generates a nut position reference based on the actual neutral position and the SEA inverse stiffness function $x_{ref} = k^{-1}(x_n, \tau_{ref})$

The inner current, speed and position loops are are classical nested loops provided within the motor driver. Note that the controlled variables x and $\dot{x}$ denote the nut linear position and speed along the ball screw. Those controllers are implemented as PID controllers.

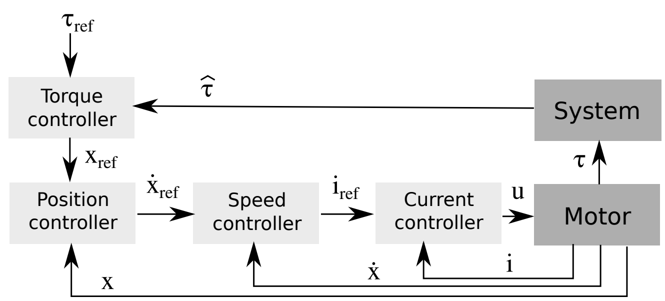

Fig. 7. SEA nested control loops with τ_{ref} the reference torque, x_{ref} the reference nut position, $\dot{x}_{ref}$ the reference nut speed, i_{ref} the reference motor current, u the motor voltage, τ the motor torque, $\hat{\tau}$ the estimated SEA torque, x the nut position, $\dot{x}$ the nut speed, i the motor current. The device is seen as a torque source for higher level controllers

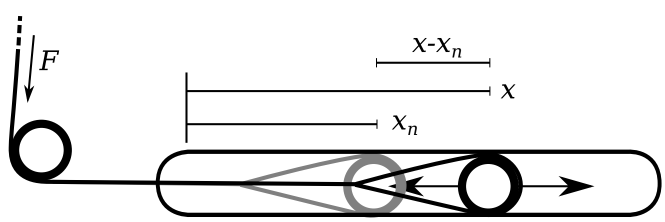

Fig. 8. A joint torque is generated by pulling on the elastic rope with force F. Using the rope elongation, i.e. the difference between the nut position x and the nut neutral position x_n, the generated torque can be estimated.

V. Validation

The system was validated on a test bench to assert whether the required torque trajectories can be generated by the device. This test bench depicted in Fig. 9 is position-controlled and imposes a walking gait trajectory to the prosthesis. At the same time, the prosthesis generates torque both via its lockable parallel spring and via the series elastic actuator. The active element is controlled to generate the missing torque between a torque pattern reference taken from [17] and the torque already generated by the passive parallel elastic mechanism. The test bench then records the total torque generated by the device. The lockable parallel spring is first tested alone to measure its true contribution, then the same tests are performed with both systems running.

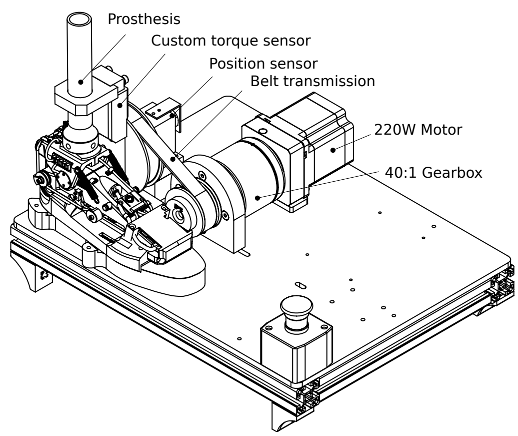

Fig. 9. Testbench setup: the prosthesis is clamped on the bench. The test bench imposes a walking gait kinematic trajectory while recording the reaction torque.

A. Lockable parallel spring testing

The contribution of the parallel spring only is depicted in Fig. 10. A good linearity can be observed with some hysteresis due to the visco-elastic properties of the nylon rope. While this leads to a loss of energy, it also prevents oscillations from appearing in the system. As shown, when clamping the rope at different locations under the foot, different stiffness characteristics can be generated to adapt to the user's preference. The servo actuation requires less than $0.1\,\mathrm{J/stride}$ which represents less than 1% of the delivered mechanical energy.

B. Full system testing

Fig. 11 shows two typical loading cycles of the prosthesis when both the SEA and LPS are active. The total torque generated by the prosthesis (blue) closely matches the reference (grey). The prosthesis internal estimation of the series elastic torque is also depicted (orange) and shows that the peak torque from the active module is only half of the total torque due to the contribution of the parallel spring.

Fig. 12 depicts the torque-position relationship of the device during the gait cycle. The total torque closely matches the reference profile with the lockable parallel spring contributing up to half of the peak torque.

VI. Discussion & Perspectives

The results presented in the previous section validate the possibility to efficiently combine an active series elastic actuator and a compact lockable parallel spring to generate a healthy torque pattern.

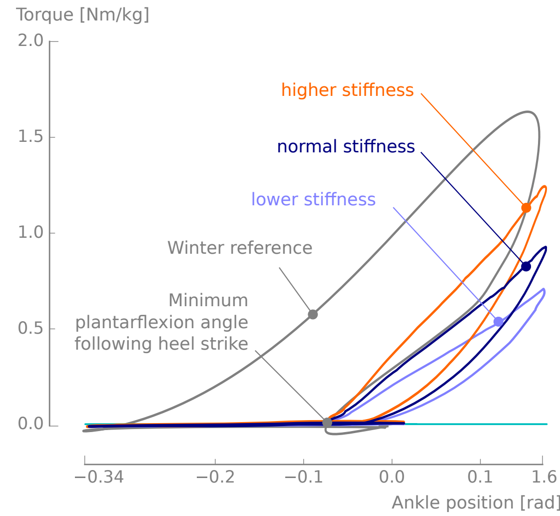

Fig. 10. Normalized torque contribution of the parallel spring for different clamping positions rendering different joint stiffnesses. The position vs. normalized torque profile taken from the reference [17] is shown for comparison (grey).

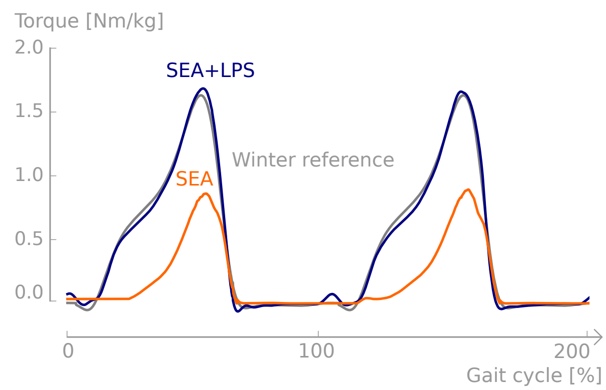

Fig. 11. Normalized prosthesis torques measured by the bench compared to a reference torque pattern from Winter [17]. Total contribution (blue) superimposed with the torque estimation from the onboard series elastic actuator (orange). The device is able to generate a complete torque pattern while the SEA has to generate only a fraction of the total torque.

A. Limitations

Due to a power limitation on the test bench, the gait frequency is currently limited to $12\,\mathrm{steps/min}$ which prevents any comparison of the electrical power consumption to our former simulation results. There is, however, no such limitation on the prosthesis itself. During these experiments, the torque measurement from the test bench was available to the prosthesis torque controller to discard any modeling error in the torque estimation. For treadmill experiments, such measurement will not be available and the internal controller will use its torque estimation instead. The torque target of the device was defined from the healthy pattern provided in [17]. This choice was made to ease comparison with other existing devices. There is no evidence that this pattern is optimal for amputees. A muskulo-skeletal model or other impedance models could be used to generate the torque reference trajectory during the gait cycle. Finally, the

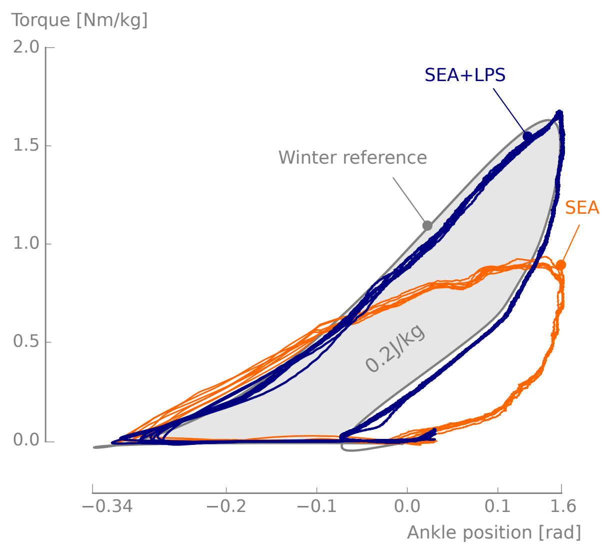

Fig. 12. Normalized torque-position trajectories of the complete system (blue) over several cycles with a sound ankle reference [17] (grey) and SEA contribution (orange). The device is able to generate the full reference trajectory as a combination of the passive elastic torque and the active series elastic torque.

parallel spring controller is currently targeting flat ground walking but could be extended to other daily life activities such as stair ascent/descent.

B. Future developments

Future developments include the synthesis and test of a closed-loop controller, which uses a foot pressure sensor insole to generate adequate motor commands. The prosthesis, implementing this embedded controller, will be tested with healthy subjects wearing an adapter and with amputees walking on a treadmill. Also, the lifetime of the ropes will be characterized.

VII. Conclusion

This works proposed the development of the ELSA (Efficient Lockable Spring Ankle) prosthetic device, which is both lightweight ($1.2\,\mathrm{kg}$, state-of-the-art $>2\,\mathrm{kg}$), affordable ($\sim$ \$1000) and with a low profile ($110\,\mathrm{mm}$ height, state-of-the-art $>200\,\mathrm{mm}$) allowing more patients to access it. It comprises a combination of an ultra low energy lockable parallel spring ($0.1\,\mathrm{J/stride}$) generating up to 50% of the peak torque during flat ground walking. This system uses a nylon rope to store and release mechanical energy during the gait. The parallel stiffness can be tuned by both choosing a string type (diameter, material) for coarse tuning and by adjusting the clamp position for fine tuning. It automatically locks at the maximum plantarflexion angle following heel strike which adapts to the subject's gait variations and the terrain. This passive system is coupled to a small series elastic actuator providing the net positive energy required during walking. The system was characterized and validated on a test bench. Future work involves embedding the control software in the device and testing its behavior with subjects wearing the device on a treadmill.

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