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Variable Stiffness Actuator Applied to an Active Ankle Prosthesis: Principle, Energy-Efficiency, and Control

Christophe Everarts, Bruno Dehez and Renaud Ronsse

***Abstract*— Series elastic actuators are very popular in rehabilitation robotics. Among other advantages, elastic elements between the actuator and the load permit to store and release energy during the task completion, such that the energy balance is improved and the motor power peak is decreased. In rhythmic tasks like walking, this reduces to design the spring stiffness such that it works at resonance. To comply with different gaits and cadences, it is therefore necessary to design Variable Stiffness Actuators (VSA). This paper proposes three contributions: (i) we apply a particular concept of VSA to an active ankle prosthesis; (ii) we discuss the relevance of using VSA to change the stiffness also within the gait cycle; and (iii) we elaborate some control strategies for this device. Our guideline is to track a mechanical design and a controller maximizing energy efficiency. We establish that a promising approach is simply to control the amount of energy stored in the elastic element.**

I. INTRODUCTION

Walking is a complex task, involving many energy exchanges across the body joints and segments, and with the environment. Human walking splits up into two phases :

- The stance phase when the foot is in contact with the ground to propel the body forward.
- The swing phase when the foot does not touch the ground, corresponding to a forward leg movement to take another step.

Figure 1 shows the ankle kinematics and dynamics (in the sagittal plane) of a healthy subject during self-paced walking [1]. In the lower panel, it is visible that a non-trivial energy flow occurs during the stance phase. During the load acceptance phase (about 200 ms after heel strike), the ankle dissipates energy (negative power, green area). Later, some energy is produced (positive power, red area) during the second part of the stance phase, corresponding to the body propulsion phase. The red area being larger than the green one, the net work of the ankle during self-paced walking is positive.

A purely passive ankle prosthesis, as most of the transtibial amputees are using nowadays, fails in reproducing this complex pattern. Even if the most sophisticated of them comprise an elastic element to store and release energy (like e.g. the flex-foot by Össur [2]), an actuator is necessary to retrieve the positive energy balance. Consequently, transtibial amputees walking with a passive prosthesis have to readapt the whole walking pattern, giving rise to an increase in the overall metabolic consumption of about 60% and compensatory movements by the intact joints (both ipsi- and contra-lateral) [3].

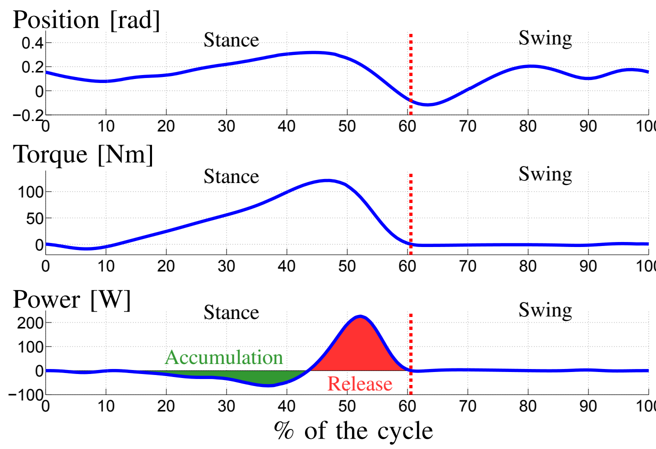

Fig. 1. Ankle position, produced torque and power during a representative cycle of walking for a healthy subject [1].

To optimally exploit the necessity to first dissipate then produce energy, the most advanced active ankle prostheses rely on the concept of Series Elastic Actuators (SEA), i.e. the serial connection of a standard actuator with an elastic element (e.g. a spring). By reducing the interface stiffness, this concept offers a lot of advantages, including greater shock tolerance, lower reflected inertia, more accurate and stable force control, less inadvertent damage to the environment, and - last but not least - the capacity to store energy [4]. Several ankle prostheses exploit this principle like e.g. the devices of Herr [5]–[7], Sugar [8]–[10] and Goldfarb [3], [11].

Putting a spring in series between the actuator and the effector in an ankle prosthesis decreases the power peak that has to be provided by the actuator. Indeed the spring stores energy at heel strike (the one that would have to be dissipated by the actuator otherwise) and releases it when the highest power peak is needed. Consequently a smaller motor can be used (the peak power is one of the key factors influencing the size of an electric motor), reducing the device cost, encumbrance and weight. For a given gait and cadence, it can be shown that it exists a particular stiffness that will bring the motor power peak to a minimum value [12], corresponding

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to a situation where the spring works at resonance.

An intrinsic limitation of the SEA is that the mechanical stiffness of the compliant element can be optimized only for a single task and/or user, since it is fixed. Consequently, people developed in the last few years the concept of Variable Stiffness Actuators (VSA). Relevant contributions in this field include the work of Bicchi et al. with *VSA*, *VSA-II* and *VSA-HD* [13]–[16], Lefeber et al. with *MACCEPA* [17], [18], Tsagarakis et al. with *AWAS* [19], [20] and Stramigioli et al. with the bond graph study of VSA [21]–[24]. Applying the concept of VSA to an ankle prosthesis would therefore allow to adapt the actuator stiffness not only to the user, but also to the detected gait and cadence using an appropriate adaptive controller.

This paper's objectives are threefold: (i) to discuss the principle of SEA and VSA in the the design of an active ankle prosthesis; (ii) to explore the relevance of changing the actuator stiffness not only across different tasks (i.e. across the gait cycles), but also within the gait cycle, to further improve the global energy balance; and (iii) to elaborate some concepts of controller to maximize energy-efficiency in the proposed design. A conceptual method using Infinitely Variable Transmission is presented to implement a VSA which can change its stiffness without changing the amount of stored energy in the spring; i.e. achieving high energy efficiency. The paper is organized as follows. In Section II, we briefly illustrate how a SEA with an optimal stiffness can decrease the power peak of the actuator of an active ankle prosthesis. In Section III, we discuss the implementation of a VSA using the concept of Infinitely Variable Transmission proposed by Stramigioli et al. [21]. We further augment the model to account for realistic actuator dynamics. Section IV discusses two control strategies of the conceptual device, again accounting for realistic actuator features (e.g. regarding efficiency). Finally, Section V concludes the paper.

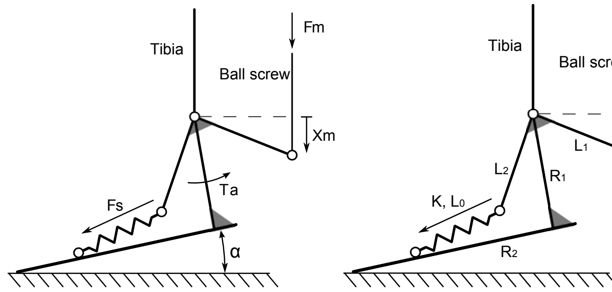
II. Series Elastic Actuators applied to an ankle prosthesis

This Section shows the usefulness of SEA in an active ankle prosthesis design. For this aim, we propose a geometric model of a conceptual prosthesis in the sagittal plane. This model was directly inspired by the series of the SPARKy devices developed by Sugar et al. [8].

The active movement is transmitted to the ankle via a ball screw through one spring. The model is shown in Figure 2 and introduces the following dynamical parameters:

$$\begin{cases} F_m, X_m & \text{Ball screw force and position} \\ F_s & \text{Spring force} \\ \alpha & \text{Ankle joint angular position} \\ T_a & \text{Ankle joint torque} \end{cases}$$

The constitutive geometric parameters (i.e. R_1, R_2, etc...), such as the springs stiffness were optimized to reduce the motor power peak (with data taken from Figure 1) with the following objective function :



(a) Dynamic (b) Geometric

Fig. 2. Ankle prosthesis model

$$\begin{aligned} f(R_1, R_2, L_1, L_2, L_0, K) &= \max(|P_m|) \\ &= \max(|F_m| \cdot |\dot{X}_m|) \end{aligned} \tag{1}$$

TABLE I
OPTIMISATION RESULTS

Variable	Value	Variable	Value
R_1 [mm]	130	L_1 [mm]	91
R_2 [mm]	80	L_2 [mm]	57
K [$\frac{N}{m}$]	140160	L_0 [mm]	114

The outcome of this optimization corresponds to the parameters given in I, giving rise to the situation illustrated in Figure 3. During the early stance phase, the motor loads the spring, which is further loaded during the load acceptance phase. During the late stance phase, the spring releases its energy (giving rise to a power peak close to the one of the healthy ankle), while power provided by the motor is very small. Without the spring, the motor would have to reproduce the power profile of a healthy ankle (peak of 170 W). With the spring, the peak power is only of about 60 W. In sum, the maximum power peak in the motor is decreased by a factor 3.

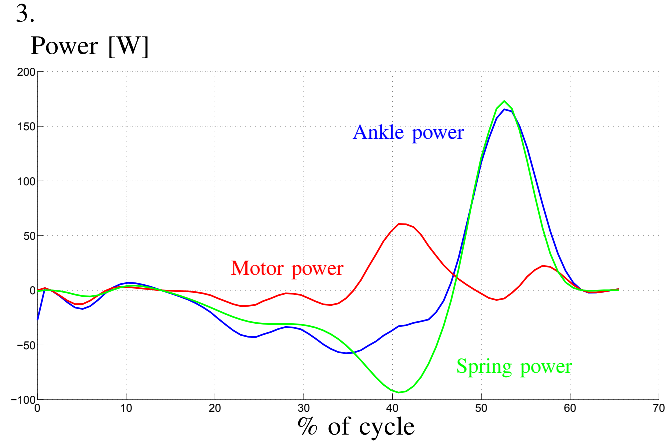


Fig. 3. Optimal power flow in the SEA ankle prosthesis, during the stance phase of walking

III. Variable Stiffness Actuators with IVT

As already mentioned, the main drawback of the SEA design is that its stiffness is optimized only for a particular gait and cadence. A natural extension of the previous approach would be to use a VSA, i.e. an elastic actuator whose intrinsic stiffness is not fixed but can be changed by means of a second motor. Such a principle has been explored in [25]. Taking this idea a little further, the actuator's stiffness could also be changed within the cycle. As such, the main motor would just have to provide the mean positive power required at the ankle - i.e. the net positive energy provided during a walking cycle, divided by the cycle duration - while the stiffness would be constantly adapted to render the desired torque. Using data from Figure 1, this mean power corresponds to approximately 10 W. This approach is analogous to a decoupling between a booster motor injecting energy into a spring with constant power, and a valve releasing just the energy necessary to the ankle for walking.

To make this approach viable in terms of energy-efficiency, it is mandatory to use an efficient mechanism to change the actuator stiffness. Said differently, this would require that changing the actuator stiffness requires little to no energy, a feature that is not encountered with most of the VSA to date. In theory, changing the actuator stiffness would require no energy if its equilibrium position is changed in parallel, such that the energy stored in the elastic element is kept constant. Recently, Stramigioli et al. proposed a concept to implement this idea [21]. It consists of a motor connected in series with a torsional spring. This spring is connected to the effector via an Infinitely Variable Transmission (IVT), i.e. a transmission that can continuously change its ratio from positive to negative values, consuming virtually no energy. An IVT allows to produce the desired output torque with an arbitrary level of energy in the spring. This mechanism is therefore not a VSA *stricto sensu*, since the spring stiffness is kept constant, but renders a variable stiffness by means of the IVT. Importantly, modifying the rendered stiffness does not modify the amount of energy stored in the spring. This concept is illustrated in Figure 4.

The model variables are listed as following:

θ_m, ω_m	Motor position and velocity.
$\Delta\theta_s, \Delta\dot{\theta}_s$	Spring relative position (i.e. overall deformation) and velocity.
θ_t, ω_t	Spring position and velocity at the load side (before the IVT). As such, $\theta_t = \theta_m - \Delta\theta_s$
$\theta_{ank}, \omega_{ank}$	Load (i.e. in this case the ankle) position and velocity
T_{in}	Torque in the motor, spring, and at the input of the IVT
T_{ank}	Ankle torque (i.e. at the output of the IVT)
n	Instantaneous transmission ratio
k	Intrinsic stiffness of the spring

Torque and velocity equations based on the power conservation in the transmission are given as follows:

$$T_{in} = -k\,\Delta\theta_s = \frac{1}{n}\,T_{ank} \tag{2}$$

$$\omega_t = \omega_m + \Delta\dot{\theta}_s = n\,\omega_{ank} \tag{3}$$

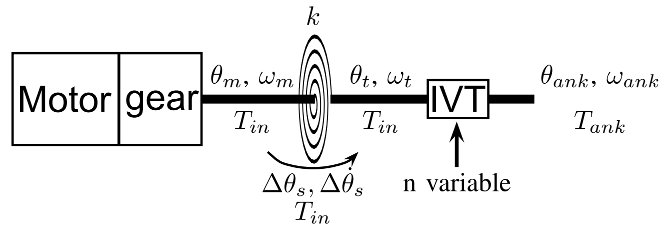


Fig. 4. VSA model with IVT

Spring torque $-k\Delta\theta_s$ appears negative because when the motor applies a torque in one direction, the spring deformation goes in the opposite direction: when the spring stores energy, its velocity is negative.

These equations further reduce to :

$$\omega_m + \Delta\dot{\theta}_s = -\frac{\omega_{ank}\,T_{ank}}{k\,\Delta\theta_s} \tag{4}$$

$$n = -\frac{T_{ank}}{k\,\Delta\theta_s} \tag{5}$$

Equation (5) captures the evolution of the transmission ratio that would have to be provided to exactly render the desired torque, i.e. T_{ank}. Assuming that this can be perfectly achieved, the only remaining differential equation is

$$\Delta\dot{\theta}_s = -\frac{\omega_{ank}\,T_{ank}}{k\,\Delta\theta_s} - \omega_m \tag{6}$$

IV. Control Strategies

The goal of this Section is to propose different control laws for the variable ω_m, and to discuss their respective energy-efficiency, taking realistic actuator dynamics into account. Two control laws are developed: one with no feedback and another with feedback on the spring deformation.

A. Actuator Model

The classical differential equations governing the dynamics of a DC-motor system are given as following:

$$U = R\,I + L\,\frac{dI}{dt} + K\,\omega \tag{7}$$

$$T = K\,I - J\,\dot{\omega} - T_f \tag{8}$$

where R represents the electric resistance, L the inductance, K the torque constant of the motor, J the inertia of the rotor and T_f the friction torque. The variables U, I, T and ω represent respectively voltage, current, torque and velocity. Assuming the electrical time constant to be negligible with respect to the mechanical time constant, and eliminating I from (7) and (8), gives:

$$\dot{\omega} = \frac{K(U - K\omega)}{J\,R} - \frac{T + T_f}{J} \tag{9}$$

By combining Equations (6) and (9) and defining the motor gear ratio to be equal to χ and the gear efficiency η_{gear}, i.e. $\omega = \chi\omega_m$ and $T = \frac{-1}{\chi\eta_{gear}}k\Delta\theta_s$, the following non-linear state-space system is obtained:

$$
\begin{pmatrix} \Delta\dot{\theta}_s \\ \dot{\omega}_m \end{pmatrix} = \begin{pmatrix} 0 & -1 \\ \frac{k}{J\chi^2\eta_{gear}} & -\frac{K^2}{\chi RJ} \end{pmatrix} \begin{pmatrix} \Delta\theta_s \\ \omega_m \end{pmatrix} + \begin{pmatrix} \frac{1}{k\Delta\theta_s} & 0 & 0 \\ 0 & -\frac{1}{\chi J} & \frac{K}{\chi JR} \end{pmatrix} \begin{pmatrix} \omega_{ank}T_{ank} \\ T_f \\ U \end{pmatrix} \tag{10}
$$

This system isolates the three inputs, namely the desired power ω_{ank} T_{ank}, the friction torque T_f and the motor voltage U; and the corresponding state variables $\Delta\theta_s$ and ω_m. Before discussing two different control laws to set the voltage U, we set the parameters to the following numerical values to be used in simulations:

- Spring stiffness : k = 3 Nm/rad
- Motor features (copying those of a Maxon EC 45 flat motor of 30 W):
  - K = 70.6 mNm/A
  - R = 6.70 Ω
  - T_{max} = 380 mNm
  - T_n = 70.6 mNm
  - ω_n = 3210 rpm
- and a compatible gear box:
  - χ = 186 with efficiency being equal to η_{gear} = 64 %

Simulations were carried out using Matlab (the Mathworks, Natick, MA), with above numerical values and the ankle data presented in Figure 1.

B. Constant voltage

The first approach is to supply the motor with a constant voltage U. In a certain range, this open loop control[1] is stable.

Figure 5 shows results of this simulation with a voltage of 21 V. The corresponding desired transmission ratio (given by (5)) varies between -0.7 and 7.6 and the spring deformation oscillates between -4 and -5.5 radians. The motor power stays close to the mean power which is 10 W, it oscillates between 9.5 and 10.2 W, representing therefore a peak power reduction by a factor 16 with respect to the healthy ankle.

This approach therefore provides an elegant way to decouple the power feeding (provided by an open-loop 10 W motor) and the control (provided by the motor controlling the transmission ratio n, theoretically spending no energy).

Figure 6 shows the results of the simulation for different input voltages. It shows that the system is stable up to 21 V and that the spring deformation and the motor speed are adapted to provide the needed power.

This is due to the fact that the power-torque curve of the DC motor (neglecting the dynamic effects and the gear efficiency, i.e. assuming $J = 0$) follows an inverted U-shape (see an example in Figure 7). The above simulations work in the descending part of the curve, therefore achieving stable behaviour: when power is required at the ankle, the spring elongation decreases, so the motor torque; due to the negative slope, this increases the motor power, tending to stabilize the spring elongation. However, this region corresponds to a small efficiency of the DC motor (below 40 %).

[1] open loop means here that there is no feedback on the voltage U. There is of course always feedback to implement the transmission ratio, through Equation (5)

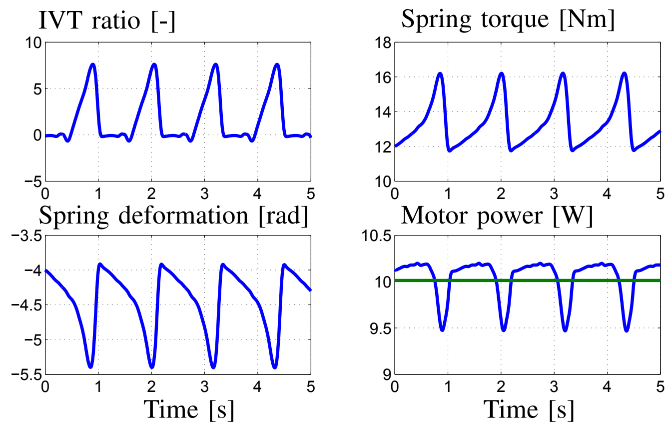


Fig. 5. Simulation results for constant voltage control

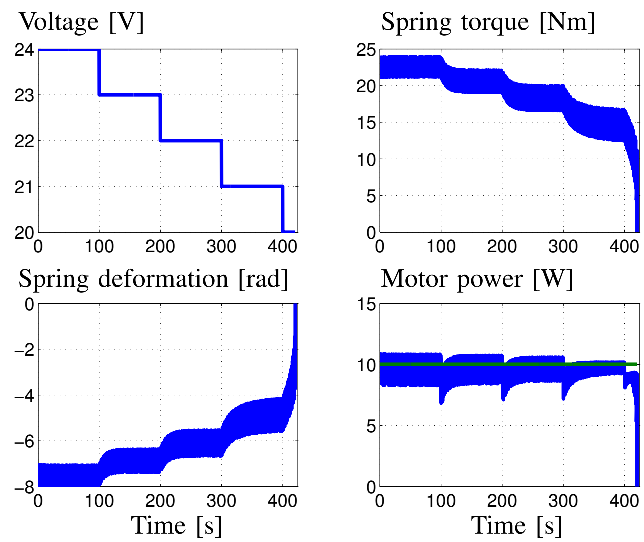


Fig. 6. Simulation results for different constant voltage control

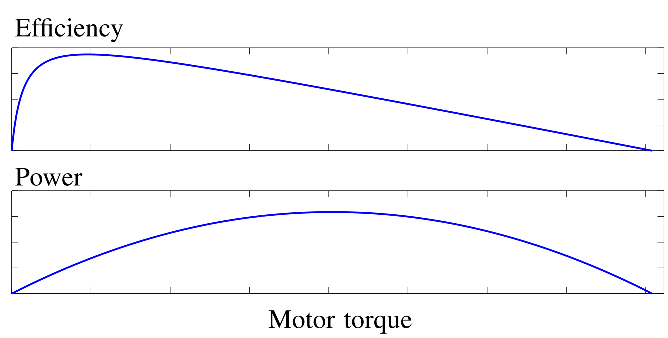


Fig. 7. Power/torque and efficiency/torque relationship of a DC-motor

Figure 8 shows the same relationships for the data from Figure 6 (i.e. with the dynamic effect taken into account). Importantly, the figure shows that the motor efficiency is indeed very bad, with this control, mainly with high input voltage.

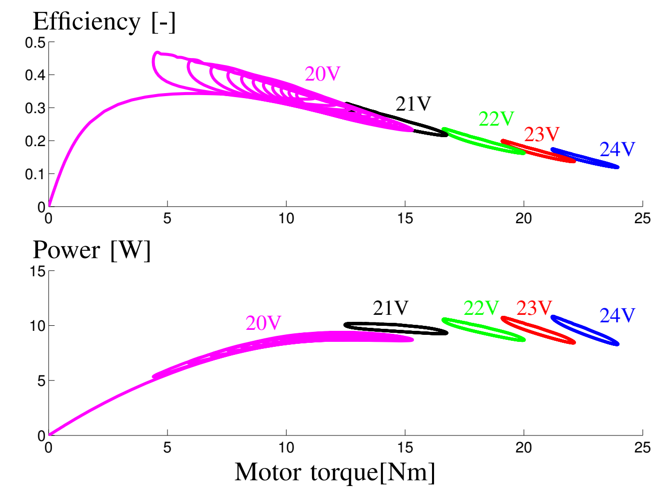


Fig. 8. Power and efficiency of the motor with different constant voltage

Figure 9 shows the electrical power, UI, and the mechanical power after the gear, $T\omega$. The mechanical power stays around 10 W but the electrical power is about 4 times higher with an input voltage of 21 V. This tendency get further worse as the input voltage increases.

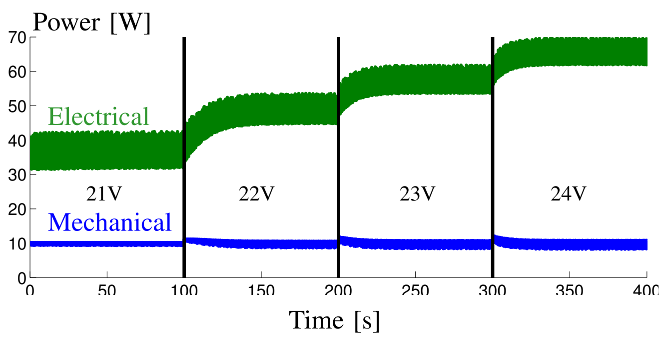


Fig. 9. Input (electrical) and output (mechanical) power of the motor with different constant voltages.

When a constant voltage is applied, the motor is not operating in its optimal area. To solve this issue, a second method is proposed, in which some feedback on the spring deformation is provided to control the motor.

C. Voltage versus spring deformation control

The second approach implements a feedback on the spring deformation to control the motor supply voltage.

The proportional law is written as follows :

$$U = U_c + k_p(\Delta\theta_s - \overline{\Delta\theta}_s) \tag{11}$$

This law combines a constant voltage similar to the feedforward term that we used in the previous section and a proportional term stabilizing the spring deformation around $\overline{\Delta\theta}_s$. Here, we use U_c = 23 V, k_p = 4 $\frac{V}{rad}$, and $\overline{\Delta\theta}_s$ = -3 rad.

Using this law in Equation (10), the following behaviour is obtained (Figure 10). Spring deformation oscillates around -3 rad, this decrease the torque applied by the motor compared to the previous section (between 6 and 13 Nm instead of 12 and 16 Nm).

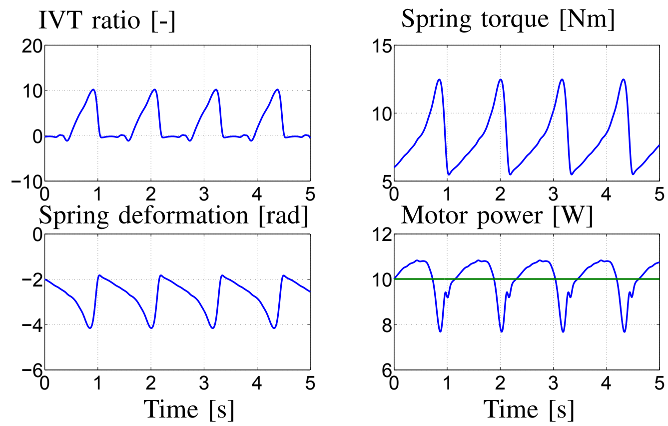


Fig. 10. Simulation results with varying voltage.

Figure 11 shows the efficiency of the motor versus the torque. With this second method, the motor operates in a better area: the efficiency oscillates between 0.25 and 0.55.

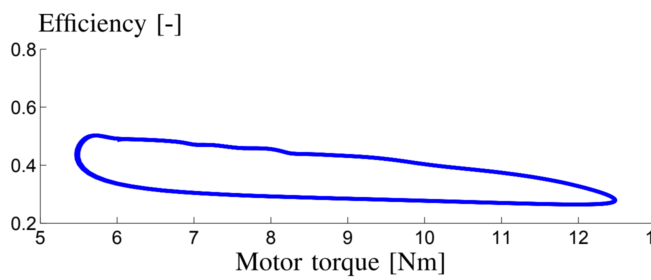


Fig. 11. Power and efficiency of the motor with controlled voltage

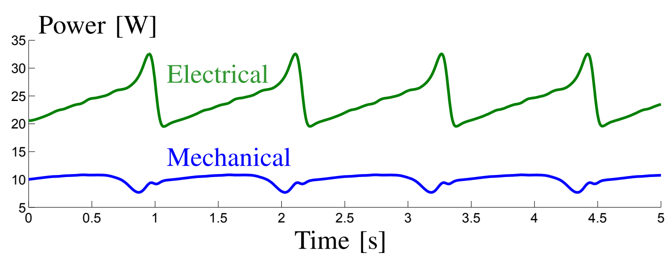


Fig. 12. Input and output power of the motor with controlled voltage

Figure 12 shows the electrical power, and the mechanical power. The mechanical power stays around 10 W (as before) and this time the electrical power varies between 20 and 32 W. This fluctuation is due to the fact that the motor efficiency depends on the torque which varies within the cycle.

V. CONCLUSION

This paper investigated the use of Series Elastic Actuators and Variable Stiffness Actuators for the design of an active ankle prosthesis, with three main objectives.

First, we quantified the peak power reduction that can be obtained through a sound design of the elastic element. With a "simple" SEA (i.e. with a fixed spring), a peak power reduction by a factor 3 was observed, pending an appropriate tuning of the spring stiffness, which worked close to resonance. However, the spring stiffness can only be optimized for a given gait and cadence. To circumvent this limitation, we introduced the concept of VSA, i.e. an elastic actuator whose intrinsic stiffness can be changed by means of a second motor.

Second, we explored potential benefits of changing the actuator stiffness not only across the cycles (i.e. to achieve resonance tuning for a given gait and cadence), but also within the cycle. This permitted to decrease the motor power by a factor 17 relative to the peak power, nearly corresponding to the minimum theoretical value, i.e. the total net energy produced by the joint during a walking cycle, divided by the cycle duration. This was achieved by means of a VSA rendering a continuously variable stiffness through a continuously variable transmission. This concept is analogous to using a flywheel in applications requiring high power peaks with a constant input power. A critical difference is that the flywheel stores kinetic energy, while the VSA stores potential energy.

Third, we proposed two control strategies for the motor feeding the spring energy. The first one was very simple, since it required the motor to be fed with a constant voltage. We showed that this strategy is stable over a wide range of input voltages, owing to an intrinsically stable torque-to-power relationship typical in DC motors. A major drawback of this approach is that it stabilized in a region where the motor efficiency is very poor, giving rise to an inefficient electromechanical conversion. This was improved by our second controller, which included a dynamical term stabilizing the spring elongation around a desired value. This permitted to improve the electromechanical conversion efficiency by about 25%.

Our results tend to validate the use of a VSA based on a continuously variable transmission for the design of an active ankle prosthesis. So far, we assumed this IVT to consume no energy for changing its transmission ratio. Future work will focus on a realistic modelling of this IVT (including its energy efficiency, actuation bandwidth, or additional weight and inertia, etc.) in order to propose guidelines for the design of light and energy efficient robotic devices. In particular, a question to be solved is whether the energy balance of the VSA using this realistic IVT (i.e. using two motors) is better than the one of a classical SEA (i.e. using a single motor) working at resonance during steady-state walking, and close to resonance nearby. We further expect to validate these results with a real platform.

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