A Variable Stiffness Elbow Joint for Upper Limb Prosthesis

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Abstract—One of the main research trends toward next-generation prostheses and bionic aids is to better replicate human motor behaviours and to improve the interconnection with the human sensory-motor architecture. One of the natural characteristics of the human arm of utmost importance in our interaction with the environment is our ability to vary the mechanical impedance of our joints by commanding the co-contraction of antagonist muscles. Integration in prostheses of such features is currently under studies. The introduction of physical variable impedance in the mechatronic structure of the devices could at the same time improve interaction and robustness and allow for more sophisticated controls with the goal of naturalness of motion.

The system proposed in this paper is a variable stiffness elbow joint for upper limb prostheses that reproduces mechanical abilities of the human joint, in terms of performance, inherent compliance and natural behaviour. This variable stiffness mechanism can be actively controlled by the user, and by using an agonist-antagonistic configuration of proper elastic elements, its output functions are similar to the models of the human muscle. The design and mechanical implementation of the device are detailed in this document together with its experimental validation and characterisation.

I. INTRODUCTION

Losing an upper limb alters significantly the quality of life of a person, mainly by decreasing its autonomy and its capabilities to perform activities of daily living (ADL). In the development of prostheses, the main engineering challenges are to design mechanical systems with human-like characteristics and abilities and to implement an intuitive and simple way to control the system [1].

Functional and structural anthropomorphism in prosthetic devices is not only an issue of shape or kinematic performance but it should also include the limb impedance to have robust, safe and natural interactions with unknown and dynamic environments, i.e the real world [2]. Based on the study of neurophysiological basis of natural motion control such as the co-activation of antagonist muscles, Hogan proposed that prostheses should have user-controllable impedance properties [3]. Investigating the performances of a compliant artificial upper limb with an impedance controller, Sensinger et al. had found that users do not deliberately adjust the impedance level in pointing and trajectory tracking tasks if it is around a suitable baseline [4]. Yet, more recently, Blank et al. found out that preferred impedance levels could be task-dependent (in contact force minimization and trajectory tracking tasks), implying some benefits for user-modulated impedance in prosthetic limbs [5]. A natural way to control variable impedance prostheses could be to correlate their impedance to the co-contraction of antagonist muscles and their motion to the difference of the muscle signals, as suggested in [3], [4]. In addition, based on the equilibrium-point hypothesis and the notion of synergies, a motor control theory associated to physiological variables used by the central nervous system could be developed, leading then to a natural and effective way to control the impedance and kinematics of multi degree-of-freedom (DoF) prostheses [6]. In parallel, current research efforts are made to improve the extraction of neural information from the muscle signals to overcome their current limitations [7]. Upper limb prostheses with user-controllable impedance and tests on amputees (previous studies were on healthy subjects) are thus required to evaluate potential improvements in the control and interactions of such devices.

To have a controllable impedance, variable impedance actuators (VIA) have been widely investigated recently (see [8] for an extended review). Two main designs are used: active VIA when the impedance behaviour is modulated through software control using rigid or compliant mechanisms with fixed impedance properties - passive VIA when the impedance is modified by mechanical reconfiguration of adaptable compliant mechanisms [8]. Compliant mechanisms have the benefits to protect the devices by decoupling the output link from the rest of the system in case of external impacts [9]. In addition, the elastic energy stored in the compliant elements could be used to reduce the energy consumption of the system [10] or to outperform the motor abilities, as in explosive movements [9], [11]. This approach seems promising for prostheses to have safe, fast and robust interactions. Regarding the control, active VIA allow the experimental validation and characterisation.

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design of lightweight robots [9] which could be preferable for prostheses. However, integration of compliant elements already increases the complexity of the system. On the other side, passive VIA can remain compliant even when the actuators are disabled or faulty [8], [9]. Passive VIA with locking mechanisms would then reduce the power consumption of the system as its mechanical configuration could be locked for any positions of the actuators [12]. For instance, holding a position with a specific impedance to carry an object will not require any power if the locking systems are passive. However, mechanical complexity of such systems is increased as another actuator is required [13], and therefore is a critical challenge to fit the design requirements for upper limb prostheses. To the best knowledge of the authors, there is for now no passive VIA for upper limb prostheses. Commercially available upper limb prostheses are mechanisms with impedance properties that are discrete (either rigid or free), like in the “free-swing” mode, or not user-controllable. Prostheses developed in research [4], [14], [15] are active VIA or do not have user-controllable impedance. Moreover, existing passive VIA in robotics usually do not fit the design requirements for a human upper limb joint, mainly in terms of shape, dimensions and mass [16]–[18].

A specific work is thus needed to design passive VIA for upper limb prostheses to evaluate benefits of such soft architectures for users. Main expected benefits are better user performances in doing different tasks and lower power consumption while integrating a user-controllable impedance. In addition, if the device has a similar mass distribution of a human arm and mimics the output functions of human muscles [19] (achievable in robotics as suggested by [20]), it may be able to react as a human arm in case of external load and facilitate its integration with the amputee (e.g. in the osseointegration process [21] by altering the forces and vibrations transmitted to the residual limb). As a first step, this work focuses on the design of an elbow joint with a passive variable stiffness actuator (VSA). The system is based on an explicit stiffness variation mechanism with electromechanical actuation (Fig. 1). It has an anthropomorphic shape (human-like dimensions and mass) and still keeps suitable performances for the use case of prostheses. To better fit the requirements of prostheses, a novel design approach is introduced, based on the distribution of the actuators along the device. Moreover, the non-linear springs are positioned in an antagonistic configuration as the human muscles configuration to get a similar functional behaviour. Despite an implementation and analytical functions which are not anthropomorphic, final approximated functions of the system are similar to the human ones.

The concept of the system is presented in Section II. Its design is detailed in Section III, its mechanical implementation in Section IV and the experimental validation is presented in Section V. Conclusions are drawn in Section VI.

![Fig. 2. Principle schemes and possible prosthetic architectures of various VSA configurations. Scheme 2a and architecture 2b for antagonistic configuration. Scheme 2c and possible architecture 2d for an ESV configuration. Scheme 2e and architecture 2f of the proposed distributed ESV configuration.](image)

**II. SYSTEM CONCEPT**

Design priorities for prostheses are the overall shape, dimensions and weight of the system, its output performances, power consumption and functional behaviours (refer to section III-A for the specific targeted design requirements).

Controlling the stiffness through an adaptable compliant mechanism requires two motors. The first step in the design of passive VSA is thus to select the motor layout. The two main layouts are the antagonistic setup and the independent setup [13]. The first one consists in the use of two opposed motors (Fig. 2a) and is inspired by the human muscles configuration [16]. It is therefore possible to move the joint when the two motors act in the same direction and to co-contract the two elastic elements to stiffen the joint when the motors move in opposite directions. The two motors have then to be on the same side of the joint (Fig. 2b). This constitutes a large constraint in the design of a prosthesis as its shape and dimension should be human-like, but it has the benefit to locate all the elements in the distal part of the joint so that majority of transhumeral amputees could use it. In the second configuration, one motor (M1) is used to drive the output position and the other one (M2) to adjust the stiffness [17], [18]. A particular case of this setup is the explicit stiffness variation (ESV) configuration (see Fig. 2c for its most common scheme), when the control of the motion and the stiffness can be decoupled, in absence of external torque. It is thus possible to have more freedom in the overall design of the system, and also to size each motor for its specific function, at the expenses usually of the human likeness of the system. However, the goal of this work is to design a passive VIA for upper limb prostheses to evaluate, in future works, its benefits for amputees. Therefore to develop
a first prototype as a proof of concept, primary focuses are on the anthropomorphic shape, dimensions, behaviour and consumption. The second configuration seems then more suitable and the system could be at least sufficient for full arm and transhumeral amputees with small residual limb. A mechanical optimisation will be performed in a next step.

The second design choice concerns the spring configuration. In the antagonistic motor setup, two elastic elements are needed and the elastic transmission should be non linear to be able to adjust the stiffness. In the second setup, it is possible to use only one elastic element. However, to keep a human-like configuration, two elastic elements are placed in an agonist-antagonist (A-A) configuration, ensuring also a bi-directionality of the stiffness of the joint.

With the common design of an ESV configuration, M1 and the elastic elements should be on the same side of the output joint and, regarding the physical implementation, it is more convenient to place M2 and the elastic elements on the same side. Therefore, the classic approach of an ESV configuration still constraints the design of the system to have all the elements on the same side of the joint (Fig. 2d). Thus, toward the goal of this work, a novel architecture is introduced based on an ESV configuration with A-A configuration of the springs and a distribution of the actuators along the system. Its principle scheme (Fig. 2e) is equivalent to the scheme Fig. 2c considering the motor M1 in the inertia of the output link. The simplified architecture of the proposed device is shown in Fig. 2f. M2 and the elastic elements are located on one side of the joint and constitute the upper segment (U-segment), analogous to the upper arm while M1 is fixed to the lower segment (F-segment), analogous to the forearm.

A third important aspect is to select the method to adjust the stiffness. Three main approaches can be found in the literature: the adjustment of the spring preload, the variation of the transmission ratio between the load and the spring, and the influence on the physical properties of the elastic element [8]. Although many research groups investigated the latter, influencing the physical properties of the elastic element is currently the most difficult approach to control the stiffness as it is generally ill-defined and highly variable. Adjustment of the spring preload is commonly the simplest way to implement variable stiffness, with simpler mechanical implementation than using the variation of the transmission ratio. However, preloading the spring requires an amount of energy which can be high as it depends on the generated deflection of the spring, whereas using the variation of the transmission ratio enables the implementation of mechanisms to reduce the energy expenditure to adjust the stiffness [17]. Yet, the mechanical complexity increases and the efficiency could then be reduced as more parts are moving. Hence, the first method is used in the proposed device.

To reduce the power consumption, locking devices are implemented within the electromechanical actuation of both functions. Ideally, locking devices should have low power consumption and be lockable at any positions. Non-backdrivable gearing mechanisms, such as worm drives, are able to protect actuators during the interactions with the environment and to reduce the energy consumption with static load cancellation [12], which are critical factors in the use of upper limb prostheses. Moreover, these friction-based locking devices can be passive if they are implemented in the chain of transmission. Yet, the efficiency of the system could then be reduced. The safety of the system (turned off or on) is provided by the physical elastic elements, allowing thus the use of non-reversible mechanisms.

III. DESIGN OF THE SYSTEM

A. Design requirements

The proposed device is composed of two segments: the U-segment (upper arm side) and the F-segment (forearm side). The rough specifications are extracted from the man-systems integration standards from NASA considering that the specifications for the U-segment and F-segment are half of the ones of respectively a human upper arm and a human forearm. They are set in term of length (len.) x diameter (diam.) x mass to 180 mm x 100 mm x 1.25 kg (for the U-segment) and 150 mm x 100 mm x 0.75 kg (for the F-segment). The overall target weight is thus set to 2 kg.

Regarding the expected output performances of the system, the needed range of motion of the elbow for upper limb daily activities is around 0°- 130° [22]. As a primary criteria the torque and the maximum speed should be in the range of the existing devices in research [14], [15] or commercially available. The torque should thus be around 10-15 Nm and the maximum speed should be between 80°/s to 250°/s.

The required range of elbow stiffness to perform ADL is still under discussion. Yet some insights can be found trough cyclic motions [23] or simulations [24]. The target range of stiffness is thus set to 10 Nm/rad to 300 Nm/rad.

Concerning the functional behaviour of the elbow joint driven by a single pair of A-A muscles, its output torque function can be approximated as

\[
\tau = R(p_1(e^{\gamma A_1}) - 1) - p_2(e^{\gamma A_2} - 1),
\]

where \( R \) represent the instantaneous lever arm, \( (p_1, p_2) \) magnitude parameters specific to each muscle, \( \gamma \) a form parameter common to all muscles and \( (A_1, A_2) \) the specific muscle activations [19]. They can be defined as

\[
A_1 = Rq(t - d) - \lambda_1 + \mu Rq(t - d) \\
A_2 = -Rq(t - d) + \lambda_2 - \mu Rq(t - d),
\]

where \( q \) is the angular position of the forearm, \( (\lambda_1, \lambda_2) \) are the specific central neural commands, \( d \) a reflex delay and \( \mu \) a parameter providing the dependency between the activation and the velocity. The joint stiffness can be thus derived as

\[
\sigma = \frac{\partial \tau}{\partial q} = R^2 \gamma(p_1 e^{\gamma A_1} + p_2 e^{\gamma A_2}).
\]

Output functions of the device should approach this model.

B. Working Principle

The detailed architecture of the system is shown Fig. 3. Each segment includes a motor unit (electric rotary motor

4https://msis.jsc.nasa.gov/volume1.htm
Fig. 3. Detailed architecture of the proposed system. M1 stands for the position motor and M2 for the stiffness motor. Gb stands for their associated planetary gearbox and W for the worm.

and a planetary gearbox) and its related transmission mechanism. The transmission of the motions is based on worm drives with low efficiency (less than 0.5) to ensure the non back-drivability of the system. Considering that the efficiency of a planetary gearbox with two stages is around 0.8, the maximum expected efficiency of the system is around 0.4.

The F-segment is composed of the position motor unit and can rotate around the shaft of the elbow joint. To rotate the F-segment, M1 generates an action on the shaft of the elbow and when this action is compensated by the total action of the elastic elements on the shaft (in absence of external torque), the shaft is fixed and so the F-segment can move around it.

The U-segment is composed of the stiffness motor unit and the elastic transmission. An A-A configuration of the elastic mechanisms is set to mimic the human muscle configuration and to ensure the bi-directionality of the joint stiffness. The elastic transmission is placed on opposite sides of the system (referred as a-side and b-side). A scheme of each side of the elastic mechanism is shown in Fig. 4. Each elastic mechanism is composed of an open timing belt (in blue on the scheme), stretched by a linear traction spring (in green) through a lever arm mechanism (in orange) which can rotate with respect to the upper arm frame and pressing on the belt via an idle pulley (in red). To adjust the stiffness, M2 drives the rotation of the pulleys to wrap more or less each belt around them. The active length of each linear springs is thus modified, changing the tension in the belt and thus the stiffness of the joint. Notice that the stiffness motor unit is acting at once on both pulleys on the upper arm side and their variation of position are the same, stretching (or releasing) both belts. In the absence of external torque, the total torque created by the two belts on the elbow shaft is null.

C. Modelling

1) Control of the position. To compute the output position of the forearm $\theta_{arm}$, two position measurements are used: the relative position $\theta_r$ provides the position of the elbow shaft with respect to the forearm and the deflection $\delta$ of the joint provides the position of the elbow shaft with respect to the upper arm, i.e the ground. $\theta_{arm}$ is therefore computed as the difference of these two values:

$$\theta_{arm} = \theta_r - \delta. \quad (4)$$

The relative position is directly proportional to the position of M1, $q_{M1}$. In absence of external torque, $\delta$ is null, $\theta_{arm}$ is then only driven by $q_{M1}$.

2) Regulation of the joint stiffness. To estimate the stiffness, measurements of the deflection $\delta$ of the joint and the angular position $\theta_r$ of the pulley $P_2$ are used. M2 is linked to $P_2$ (upper arm side) and so $\theta_r$ is proportional to the position of M2, $q_{M2}$. The stiffness is also function of the design parameters of the system, shown Fig. 4. The elbow shaft is fixed to the pulley $P_1$ (forearm side). In the following, subscript $w$ stands for the lateral a-side or lateral b-side.

Assuming that the friction is negligible, the torque $\tau_w$ generated by the belt on the output link can be computed after some calculation as

$$\tau_w = \varepsilon_w R_1 F_w \cos \alpha_w = \varepsilon_w R_1 \frac{K_s}{4} L_w \beta^2 \left( \frac{X_0}{h_w} - 1 \right), \quad (5)$$

where $\alpha_w$ is defined Fig. 4 and $\beta$ represents the ratio $L_a^{w} / L_w$ between the two lengths of the lever arm and $K_s$ the spring constant. $X_0$ is assumed to be the same for the two springs and is reached when the belt is straight. $L_w$ corresponds to the usable length of the belt between points A and B:

$$L_w = L_{0w} + \varepsilon_w R_1 \delta + R_2 \theta_2 = L_{0w} - R_1 \left( - \varepsilon_b \delta - \kappa \theta_r \right). \quad (6)$$

$L_{0w}$ is the length of the belt when $\delta$ and $q_{M2}$ are null and $\kappa$ is the ratio $R_2 / R_1$. The height of the belt $h_w$ is expressed as

$$h_w = \frac{1}{2} \sqrt{L_w^2 - D^2}. \quad (7)$$

The parameter $\varepsilon_w$ expresses the fact that the elastic transmissions are mounted in an A-A configuration to ensure the bi-directionality of the joint stiffness. It is defined as

$$\varepsilon_a = -1 \quad \text{and} \quad \varepsilon_b = 1. \quad (8)$$

The torque $\tau_{out}$ created by the full variable stiffness unit is then the sum of the torques generated by the two belts:

$$\tau_{out} = \tau_b + \tau_a = R_1 \frac{K_s}{4} \beta^2 \left( L_b \left( \frac{X_0}{h_b} - 1 \right) - L_a \left( \frac{X_0}{h_a} - 1 \right) \right). \quad (9)$$

Its related stiffness can be derived as

$$\sigma = \frac{\partial \tau_{out}}{\partial \delta} = \frac{K_s R_1^2}{4} \beta^2 \left( 2 + \frac{D^2 X_0}{4 \beta} \left( \frac{1}{h_a^2} + \frac{1}{h_b^2} \right) \right). \quad (10)$$

Therefore, in absence of external torque (i.e $\delta = 0$), the stiffness can be adjusted only with M2.
IV. MECHATRONIC DESIGN

The lateral views and sections of the system are shown Fig. [5] All annotations are detailed in the following sections.

A. Position mechanism (PM)

The PM is composed of the position motor unit and the elbow shaft. M1 and its planetary gearbox (1) are respectively a Maxon motor DCX22L GB KL 18V and a GPX22HP 35:1. The worm drive (3, 4) used for the transmission is the model A25U20 from FramoMorat. The total gear ratio of this actuation unit is thus 700:1. Its frame is fixed to the box of the position unit (5) made of aluminium alloy. The box integrates a main part, its cover and all the flanges used to fix the mechanism. The worm (3) linked to the output shaft (6) of M1/gearbox (1) can rotate freely with respect to the box (5) through the ball bearing (7). All of these components are part of the F-segment and can rotate around the shaft of the elbow (8) linked to the gear (4) and the pair of ball bearings (40, 44). Spacers (39, 43) are used to fix the bearings along the axis. It is also free to rotate with respect to the frame of the U-segment through the pair of ball bearings (41, 42).

B. Variable stiffness (VS) unit

The VS unit is divided into the stiffness motor unit and the elastic transmission. On each side, the open timing belt rotates freely through the ball bearing (15). The transmission to the axis of each pulley (27, 30) is based on the worm drive A25U20 from FramoMorat with a worm (11) and double gears (12, 13) configuration. The frame of the actuation unit is fixed to the box of the VS unit (10), which integrates the main part, two covers and all the flanges used to fix the mechanism. The upper arm frames (20, 36) are unified to this box. The worm (11) linked to the output shaft (14) of M2/gearbox (9) can rotate freely through the ball bearing (15). Each shaft (49, 50) of the pulleys (27, 30) rotates with respect to the box via pairs of ball bearings (48, 52 and 46, 51). Spacers (47, 53) are used to fix the bearings along the axis.

C. Electronics and control

All the electronic boards, softwares and librairies used in the proposed system are extracted from Natural Machine Motion Initiative (NMMI) platform[25]. Electronic boards (16) and (25) are respectively used to control M1 and M2. (16) is fixed to the frame of the forearm (2) fixed to the box of the position unit (5), and (25) on a support part (26) fixed to the box of the stiffness unit (10). The microcontroller

https://www.naturalmachinemotioninitiative.com/

V. EXPERIMENTAL VALIDATION

Table I provides the complete datasheet of the device with its specifications (Spec.). Data (experimentally validated) are extracted following the methodology suggested in [26].

Dimensions of the U-segment (#1), the F-segment (#2) and the maximum width of the joint (#3) globally matched the design requirements. The maximum width of the system is larger (110 mm) with respect to the 100 mm targeted, mainly due to the positioning of the encoders that could be modified in a future version. Besides the distribution of the actuators leads to a distributed mass similar to the human-like target distribution. Masses of the F-segment (#4) and U-segment (#5) include the electronics but not the battery. F-segment is heavier (820 g) with respect to the 750 g targeted, yet the overall mass of the prototype (#6) is less (by 225g) than the targeted value of 2 kg. The device is thus within the design requirements. Yet it is more bulky than existing devices. Therefore, if the benefits of user-modulated impedance are proven, a mechanical optimisation on the design and the materials should be done to reduce the weight and size of the system. The active rotation angle of the system (#7) can vary from -25° to 155° (in total a range of work of 180°). The
angular position of the forearm is null when the F-segment and U-segment are aligned. This range of work is included in the required range to perform ADL [22].

Besides the classic features of actuators (e.g. range of work, nominal/peak values of the torque and velocity, the efficiency), VSA are also characterized by their range of stiffness, their stiffness variation time, their maximum deflections in minimum and maximum stiffness configurations and their maximum torque hysteresis [26]. Moreover, to control the system, their output torque and stiffness should be derived as functions of the available measurements. Quasi static load-unload cycles with fixed stiffness preset experiments are presented in Section V-A to establish the deflection-torque ($\delta - \tau$) and stiffness-torque ($\sigma - \tau$) charts. Output functions based on the data are then derived and the maximum deflections, the range of stiffness, the maximum hysteresis and the efficiency are extracted. Experiments in Section V-B provides the stiffness variation time and the maximum speed.

The experimental setup is the same for all the experiments. The system is rigidly fixed on a table and powered using a power supply at 18V. The configuration has been set up such that the F-segment is vertical in the middle of the active rotation angle of the system. If needed, additional load can be fixed to the F-segment. The no load configuration is when no additional load is used. Data are acquired via Simulink schemes on an external computer.

A. Quasi static load-unload cycles with fixed stiffness preset

In this experiment, a known torque profile is applied to the joint by exploiting the gravity and adding a load of 2 kg. The range of torque is within [-5.5; 5.5] (Nm) as it is the feasible continuous range of torque with the electronics used in the prototype.[7] Seven levels of stiffness (Lv1St) are considered (0 stands for the minimum and 6 for the maximum stiffness configuration). To have a fixed stiffness preset, constant position inputs are used to control $q_{M2}$, namely $[6 \ 5 \ 4 \ 3 \ 2 \ 1 \ 0]$ (10^3 ticks) for each level. To be in a quasi static configuration, a slow periodic triangle position profile is used to control $q_{M1}$ (amplitude of 3.10^4 ticks - period of 140s).

The deflection $\delta$, the relative position $\theta_r$ and the position $\theta_s$ are recorded, and $\theta_{arm}$ is computed as in (4). The external torque $\tau_{ext}$ is thus derived by exploiting the gravity. As this experiment is set to be in a quasi static configuration, the output torque of the system $\tau_{out}$ is considered as

$$\tau_{out} = -\tau_{ext}. \quad (11)$$

The $\delta - \tau$ chart of the measured data is shown Fig. 6a. To remove a bias possibly due to the experimental setup, the data of $\delta$ are translated so that for each Lv1St $\tau_{out}$ is null when $\delta$ is null. From this graph it is possible to extract the maximum deflections in a maximum (#12) and minimum (#13) stiffness configuration and the maximum hysteresis (#14) equal to 2.5° which is around 1/10 of the range of deflection.

To get the efficiency of the system (#15) to move while holding a fixed stiffness, the current consumed by M1 is recorded, then the theoretical torque generated by M1 is derived using its datasheet from maxon motor and the global ratio of the system. This value is then compared to the actual output torque generated by the mechanism. An efficiency of 0.3 is obtained. It is then possible to get an estimation of the peak torque of the system (#8) by using the peak current of the electronics, equal to 5 A, the specifications of M1, the global ratio and the efficiency. The peak torque is estimated

6New electronics is under development to improve their performances
The stiffness associated to the model can then be derived as

\[ \sigma = \lambda_1 e^{-\mu_1 \kappa \theta} e^{\mu_1 \delta} + \lambda_2 e^{-\mu_2 \kappa \theta} e^{-\mu_1 \delta} \]  

from which the maximum (#9) and minimum stiffness (#10) are extracted. Their values fit the design requirements. The form of the model (13) is similar to the one expressed in (5). Therefore, main features of the system are within the targeted values and its behaviour can be approximated by similar functions as the model of human muscles which is very satisfying and promising for a prosthetic device.

### B. Other features

1) **Stiffness variation time.** Estimation of the variation time of the stiffness is made in two different configurations: no load configuration (#16) and 5 Nm load configuration (#17). In each configuration, the position is preset at the minimum stiffness, to have either no load or 5 Nm load torque at the joint. M1 is then turned off. Ten trials are performed, in which M2 is used at its maximum power to vary the stiffness from its minimum to its maximum. Then the mean value of the ten trials is computed. The results seem to be enough to be able to adjust the stiffness while performing low frequency movement. Such aspects will be investigated in future work.

2) **Maximum speed with fixed stiffness preset.** The maximum speed of the system (#11) is estimated in the no load configuration. For five different levels of stiffness, M1 is used at its full power over the entire range of the active rotation angle. Five trials are performed for each level. For each trial the average speed is computed. The maximum speed of the system is computed as the average value of all the speeds of the different trials. Its standard deviation is 1.14 s, which legitimates the estimation of the maximum speed by its total average value. This maximum speed is in the targeted range and therefore is suitable to perform ADL.

### C. Soft behaviour

The soft behaviour of the system is highlighted in a static configuration Fig. [7](image) At two different levels of stiffness, load of 2 kg was added to the system. Its behaviour in a dynamic situation is shown Fig. [8](image) with the response of the system after an impact at two different levels of stiffness. Photo sequences are extracted from the video attached as additional material to this paper. In a soft configuration, it is possible to see how the system is oscillating after an impact compared to the high stiffness configuration. Behaviour of the VS elbow
In this paper, a novel variable stiffness elbow joint for upper limb prosthesis has been presented. The main new feature of this system is the implementation of a passive VSA, which enables a control of the stiffness of the joint. An extensive characterisation of the system has been performed to extract its main features and a modelling of its behaviour. Regarding its anthropomorphic shape, dimensions, mass and output performances, this system is close to be human-like and should be able to be used in ADL. Finally, models of the output functions of the system are similar to the models of human muscles, which suggests a natural behaviour of the joint. Therefore, this system should be compatible as an upper limb prosthesis with natural and controllable compliance.

Future works will go toward the evaluation of the benefits of passive VSA in upper limb prostheses, regarding among others performance and power consumption of such systems, during ADL. A natural control should be implemented, firstly based on surface EMG signals of the user. Integration of this system to the amputee and investigation on the effect of soft architectures on the connection to the patient will be performed. A better understanding of the effect of these new features will lead to a mechanical optimisation on the design in view of enlarging the variety of patients that could use such systems.

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