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Design Feasibility of an Energy-efficient Wrist Flexion-Extension Exoskeleton using Compliant Beams and Soft Actuators*

Ali Amoozandeh Nobaveh[1,†] and Brandon Caasenbrood[2,†]

Abstract— **Passive and active exoskeletons have been used over recent decades. However, regarding many physiological systems, we see that the majority explore both active and passive elements to minimize energy consumption while retaining proper motion control. In light of this, we propose a design that combines compliant mechanisms as passive support for gravity balancing of the hand's weight and soft actuators as active support for wrist flexion-extension. Our approach offers a safe, lightweight solution that intrinsically complements and supports the wrist's degrees of freedom. We hypothesize that the proposed soft wearable device is able to increase the range of motion and reduce muscle fatigue while being energy-conservative by balancing of the passive and active subsystems. In this work, we perform a design feasibility study for such soft wrist exoskeletons, particularly focused on wrist flexion-extension rehabilitation. Through optimization, geometries for the required functionality of the compliant beam and soft actuator are obtained, and their performance as separate subsystems is evaluated by simulations and experiments. Under the appropriate inputs, we show that the system can introduce a controllable bifurcation. Through experiments, we investigate such bi-stability and explore its usefulness for rehabilitative support of wrist flexion-extension. In short, the proposed wearable can offer a viable, energy-efficient alternative to traditional rehabilitation technologies.**

I. INTRODUCTION

Exoskeletons are a particular research field of robotics aimed at developing wearable robotic systems that augment, enhance, or restore human motor functions [1]. Although the term *"exoskeleton"* was originally used in biology to describe the support and protection of soft organs, it is nowadays recognized as distinct wearable robotic devices that are actively controlled and (fully) autonomous [2]. Besides their applicability in industry [3], they also play a vital role in rehabilitation technology, especially in cases of muscular dystrophy, like post-stroke or Duchenne disease [4]. Persistent and early physical therapy has been shown to restore or improve the longevity of muscular functionality of the paretic limbs [5], [6]. In light of this paradigm, wearable exoskeletons have paved the path towards in-house rehabilitation that can prolong therapy exposure and thereby, its effectiveness. However, there is a lack of solutions to the *"one-strategy-fits-all"* problem, and consequently, proposed exoskeleton frameworks often benefit only a limited group of target patients.

A. Related works

Passive exoskeletons are being developed primarily to demonstrate that by balancing the weight of body parts, it is possible to significantly reduce the amount of muscular effort, which is required to perform a specific task [7]. This type of exoskeleton is developed to reduce the complexity, cost, and weight of assistive devices [8].

In most of the developed passive exoskeleton concepts, a rigid body mechanism has been used with a close kinematics to the human body and a set of springs or elastic bands to store energy [9], [10]. This combination, however, is inconsistent with the ultimate goal of exoskeletons being slender, and not impeding natural human motion. As a solution to these issues, a relatively new branch of passive exoskeletons has been developed in recent years, utilizing compliant mechanisms to integrate the energy storage and kinematic functionality into a single element. These exoskeletons are scarcer but slowly emerging [11]–[13].

In contrast to rigid support, soft robots continue the path of lightweight, cost-effective, and safe robotics. By exploring *soft materials*, e.g., silicone rubbers and thermoplastics, the natural compliance of the system can be significantly improved, leading to continuum-bodied motions that are difficult, if not impossible, to achieve for conventional robots. Naturally, these soft robots have major benefits for healthcare and human-robot collaboration [14], where safe and passive interactions are prioritized over the classic robotics requirements (e.g., precision and speed). Several researchers have shown the prominent role soft robots can have in rehabilitation and other assistive technologies. Polygerinos et al. developed a soft robotic glove for hand rehabilitation using pneumatic actuation [15], exploring the flexibility to partially solve the one-fits-all problem. Yap et al. designed a soft actuator with optimal joint placement to maximize the force transferal [16], and other more recent examples of soft exoskeletons include [17], [18].

A major drawback of conventional soft actuators, however, is their limited force and torque transmission. Although softness benefactors safety, it lacks the structural stiffness to efficiently transfer large loads without buckling. A possible solution here might be fiber-reinforced strategies [15], [19] or soft robots that explore semi-rigidity (similar to most physiological systems in biology). Alternatively, Yang et al. proposed a vacuum-actuated muscle-inspired pneumatic structure (VAMPs) that purposefully introduces buckling

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instability for enhanced actuation strength, whose motions resemble linear actuators in rigid robots [20].

Our contribution(s)

In retrospect to the previous literature, we propose a synergy between the distinct fields of passive muscular support using compliant mechanisms and active support using soft robotics technology. In this work, we primarily focus on a wrist exoskeleton that naturally compliments and supports the flexion-extension of the patient's hand(s). Furthermore, we perform a design feasibility study for a soft wrist exoskeleton. In our analysis, we seek a system that enhances both the range of motion and reduced muscle fatigue while attaining high energy-efficiency through the synergy between passive and active elements. Our contributions presented in this work are listed below:

- Development of a novel and safe design for wrist-support used for gravity-balancing;
- Optimization-driven design of compliant beam and soft robotic actuator;
- Experimental verification of the mechanical behavior of the designed soft wrist exoskeleton, exploring its controllable bi-stable nature for enhanced wrist mobility.

This work is organized as follows. First, we discuss the design of the proposed wrist exoskeleton – detailing the passive gravity balancing and the bi-stability mechanism that allows bi-directional mobility through one actuation input and the external stimulus from the user (or patient). Second, we show results from simulations and experiments of each individual sub-component to show the feasibility of the final integrated design. Following, we investigate the bi-stability and its role in the hand's range of motion, followed by a brief discussion and conclusion.

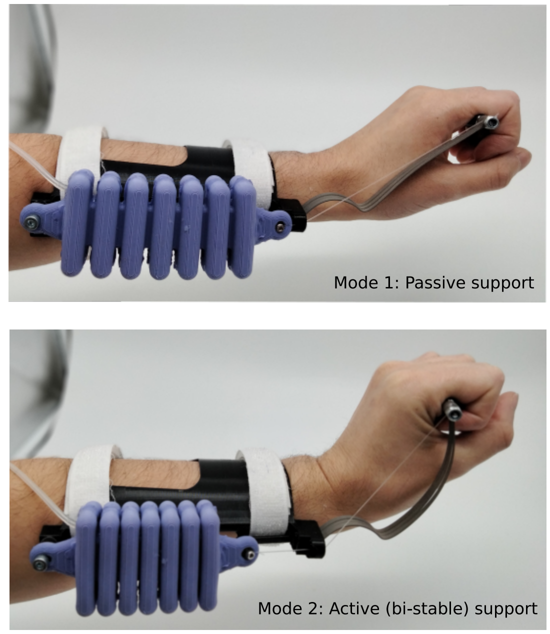

Fig. 1. An illustration of the energy-efficient wrist exoskeleton that explores the fields of passive compliant beams and active soft actuation. In the zero-input situation (Mode 1 – passive support), the compliant beam balances the weight of the hand. The integration of the soft actuator (Mode 2 – active support) extends the range of motion.

II. Design overview

A monolithic compliant mechanism can be an appropriate option for passive gravity balancing of hand without adding a considerable weight or occupying a large space [7], [8]. Among a wide range of possible compliant mechanism topologies, a cantilever beam is chosen as it has a relatively simple and slender topology, which benefits comfort and ease of design. The design objective of the beam is to keep the hand in the straight position without any muscular effort while still giving the user the flexion and extension freedom. In addition to the described passive beam, a tendon-like soft actuator is introduced as an active parallel subsystem. This tendon can help the user cover the hand's range of motion with a minimal effort. As such, the compliant beam balances the hand's dead weight while the soft actuator helps with the hand's movement in its range of motion. The tendon free-end is connected via a thin nylon rope to the endpoint of the compliant beam. This rope passes through the hole at the beam's grounding near the wrist joint. The beam is designed in such a way that the straight posture maximizes the distance between the hand interface and the hole at the wrist joint. Therefore, contraction of the active tendon induces a bi-stable flexion-extension motion around the straight posture of the hand. As a result, only by inducing a minimal flexion or extension from the user, the soft actuator can start helping the hand with its contraction to reach either side of the motion range. In other words, the active actuator for the tendon is designed in such a way that its contraction can have dual effects, similar to the flexor-extensor muscles embedded in the human forearm. The operation of the passive and active subsystems together and their interface with the patient's hand is illustrated in Fig. 1 and Fig. 2. The following subsections detail the design process for each component of this exoskeleton.

A. Compliant beam (passive support)

The ultimate goal of the monolithic passive compliant element is to mimic the motion of the hand during wrist flexion-extension while providing balancing support for the hand's weight as a pendulum around the wrist joint.

The steps for designing the required beam are as follows. First, a method for parametrizing the beam shape using B-splines is developed, whose shape is defined by a set of control points. Second, an optimization routine is used to satisfy the balancing kinetostatic requirements by tuning these control points as design parameters. In each iteration, the optimizer forms a beam shape based on the control parameters and evaluates the balancing behavior by a set of endpoint incremental loadings from zero to twice the hand's weight. The displacements upon these loadings are numerically computed using a self-developed finite element model. The reader is referred to [12] for more detail.

As for the beam shape parametrization, a degree-four B-spline is used to interpolate the spine shape and the width of

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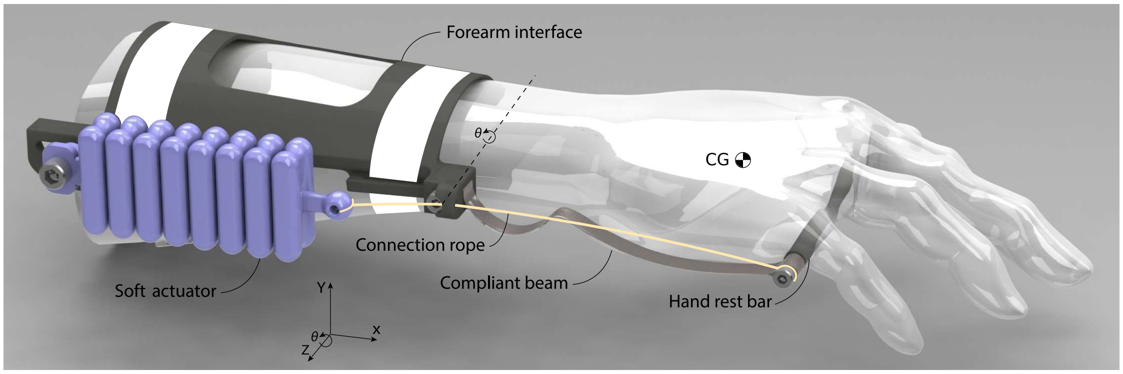

Fig. 2. An illustration of the concept of the wrist exoskeleton. The compliant beam and soft actuator are both connected to the forearm interface at their grounding. The beam's endpoint is directly connected to the hand bar, and the soft actuator is connected via a nylon rope to the hand interface. The bar is sitting under the roots of the user's fingers.

the beam based on seven optimized control points as shown in Fig. 3. The B-spline is chosen to have an open uniform knot vector to ensure equal weight for each control point and to make the first and last optimized control points coincide with the first and last parameters of the beam itself. Using B-spline interpolation not only reduced the parameters, but also caused a smooth transition between the optimized parameters of the beam and avoided any discontinuities that could have happened due to sudden dimensional changes in the finite element model.

The beam is designed to be made with a 2D fabrication method out of planar material. Therefore, a rectangular cross-section with a fixed width ($9\,\mathrm{mm}$) is selected for it. Based on this assumption, a set of parameters including two Cartesian coordinates and one thickness at each control point (21 parameters) are subjected to optimization.

For the optimizer, the *Multi Start* option from *optimization toolbox* of Matlab® is used. To increase the chance of having a better optimum solution, five random starting points are selected for the main *fmincon* function with *Interior-Point* as the algorithm. Upper and lower constraints are set for coordinates to keep the beam shape in a limited design space.

TABLE I
THE FLEXION-EXTENSION AND PRONATION ANGLES OF THE BEAM'S ENDPOINT FROM EXPERIMENTS AND SIMULATIONS.

	Extension+/Flexion- (deg)		Pronation (deg)	
load (g)	model	experiments	model	experiments
$\delta = 0$	+57	+50	0	0
$\delta = 100$	+50	+43	1.1	9.3
$\delta = 200$	+32	+26	2.5	25.7
$\delta = 300$	+10	+14	4.6	34.1
$\delta = \delta_W$	-5	+2	7.4	35.2
$\delta = 500$	-15	-7	10.6	37.7
$\delta = 600$	-22	-12	14.2	37.9
$\delta = 700$	-27	-15	18	38.3
$\delta = 800$	-32	-21	22.1	38.7

It is desired that the mechanism has a minimum effect on supination-pronation angle and keeps it as straight as possible. However, as the mechanism is on the side of the hand and the beam is designed to be planar, it is inevitable to have some pronation. The range of motion for supporting the user hand is set to reach $\theta = \pm 30°$ for flexion-extension from straight posture with 50% less muscular effort. The hand's weight assumed to be $\delta_W = 400\,\mathrm{g}$, with its center of gravity (CG) in $0.07\,\mathrm{m}$ from the wrist joint itself in X direction, and $0.045\,\mathrm{m}$ in Z direction from the outer side of hand as illustrated in Fig. 2.

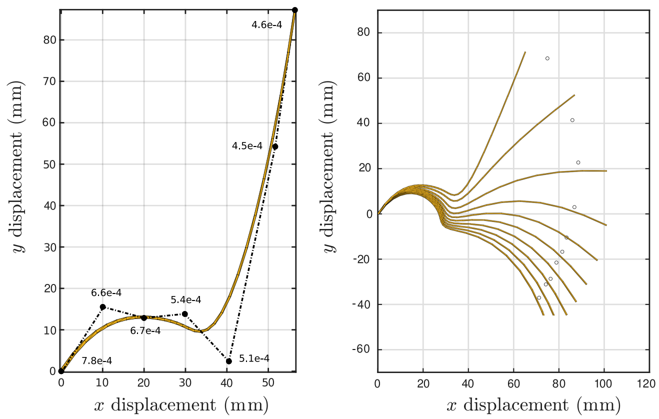

Fig. 3. (Left) The optimized control points and thicknesses in each point and the final beam shape based on the control parameters. (Right) The (nonlinear) elastic deformation of the complaint beam under incremental vertical loadings at the tip.

A self-developed solver using geometrically nonlinear co-rotational beam elements, based on the Euler-Bernoulli formulation, is used for modeling of the beam behavior. The details of this solver can be found in [21]. The final topology of the beam and the thickness at each of the control points (in millimeters) are shown in Fig. 3 (left).

To validate the beam performance, experiments are conducted to show the flexion-extension and inevitable pronation behavior of the beam. The prototype for these experiments is made from spring steel (AISI 301) with an elastic modulus of $E = 190\,\mathrm{GPa}$ and a shear modulus of $G = 72\,\mathrm{GPa}$ (identical to simulation). The experiments include clamping the

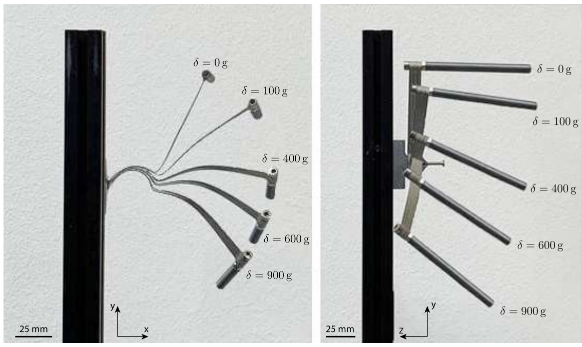

Fig. 4. An illustration of experimental results. Here the load δ is applied at the CG of the wrist, illustrating both flexion-extension and pronation mobility in the complaint beam.

beam at the grounding point and using different suspended weights $\delta \in [0, 800]$ g at the endpoint. The deflection of the beam is measured by comparing the endpoint position of the loaded and unloaded states with image processing using images from two perpendicularly located cameras, one for flexion-extension and one for pronation. The experiment is shown in Fig. 4. The endpoint positions of the beam, which result from the experiments, are shown with $\circ$ in Fig. 3. In this figure, it is possible to compare the flexion-extension angles under different loadings from experiments and modeling. The angle information is shown in Table I.

B. Vacuum-actuated soft actuator (active support)

In this section, we will discuss the development of the active soft robotic muscle that ultimately acts as an active supportive layer to the wrist exoskeleton. In particular, for patients that suffer from muscular dystrophy, where muscle forces are significantly reduced, the soft actuator could potentially substitute or increase muscular strength. Similar to human muscles, we aim to seek a soft actuation subsystem that can undergo rapid contraction and relaxation. Here, we have chosen a vacuum-actuated muscle-inspired soft actuator design loosely inspired by the designs of [20]. These soft actuators use buckling of elastomeric structures to generate muscle-like motions when negative pressure (vacuum) is applied; the external (ambient) pressure causes their co-operative, reversible collapse. The soft material chosen to develop the soft actuator is two-component platinum cured silicone *Smooth-Sil 945* with a 45A shore hardness ($E = \pm 1.52$ MPa Young's modulus). To design the geometry of these soft actuators, we again explored an optimization-based design approach. However, contrary to the previous beam optimization, a nonlinear-topology optimization scheme is used to find the optimal two-dimensional material layout that maximizes contraction-like displacement when subjected to negative differential pressure. This forms a highly-nonlinear optimization problem, and the reader is referred to [22] for more detail on the method and solver. The optimization result can be found in Fig. 5, together with the experimental results of the final soft muscle.

To validate the performance of the vacuum-driven soft actuator, we developed a uni-axial benchmark test that can measure the contraction displacement, denoted here by $s(t)$, with respect to the negative pressure input $u(t)$. The intrinsic length of the soft actuator is given by $s_0 = 160.25$ mm. The pressure input is defined as a relative pressure difference, i.e., the pressure potential with the ambient pressure $u(t) = p(t) - p_{\text{atm}}$ with $p_{\text{atm}} \approx 101.33$ kPa. To investigate the force transmission of the proposed soft actuation, we subject the system to a range of suspended weights $\delta \in [0, 1000]$ g. In Fig. 5, we show an example of such benchmark tests for $\delta = 400$ g (i.e., the average weight of the human hand). The evolution of all contraction ratios for the remaining loading scenarios are also shown in Fig. 5.

First of all, as expected, we observe that the soft actuator has a bounded contraction range, roughly $\pm 30\%$ contraction for $\delta = 0$, and $\pm 65\%$ contraction for $\delta = 1000$ g. Please note that the increase of $+35\%$ work range is caused by pre-stretching of the soft actuator. Regarding the force transmission, we observe that the soft actuator is sufficiently strong to produce a wide range of forces – whose loading conditions should not exceed $\delta > 1000$ to prevent tearing. For the no-load condition ($\delta = 0$), we observe that the necessary differential pressure required is less than -35 kPa, slowly increasing towards $-50\,\text{kPa}$ for increasing loads δ. Clearly, the trend of the contraction w.r.t. to the pressure input is nonlinear. However, the trend w.r.t. the loading δ appears to be closely linear. These observations indicate that the soft actuator possesses dual control functionality, namely as an actuator and a variable stiffness. As a result, a merit benefit of such soft actuation systems is their low energy consumption, only requiring an influx of energy when switching between the states. Although this process is nearly reversible (assuming minimal temperature exchange with the environment), it is difficult to realize it in practice.

Remark 1: Let it be clear that the proposed soft wrist exoskeleton is not fully back-driveable. Only contractile motion of the soft actuator is feasible, yet positive pressures can ensure that the rest configuration is reached faster. However, (large) compressive forces will eventually result in structural buckling. A possible solution to the system in Fig. 2 might be antagonistic soft muscle pairs, similar to a flexor-extensor muscles.

III. Exploring the bi-stability property of the soft wrist exoskeleton system

To show the feasibility of the overall design, we combine the compliant beam of Section II-A and the vacuum-actuated soft actuator in Section II-B, giving rise to the soft wrist exoskeleton as shown in Fig. 1. The goal is to investigate the feasibility of such design to support the whole range of motion of hand by exploiting the aforementioned bi-stability property. If the internal pressure in the soft actuator is $u = 0$, the exoskeleton will default to its gravity balancing stability for the load $\delta = \delta_W$, *i.e.*, $\theta = 0$ (or close to zero, see Table I). This is the global minimizer of the potential energy of the wrist exoskeleton without control. However, if the internal pressure in the soft actuator is non-zero, this induces a force

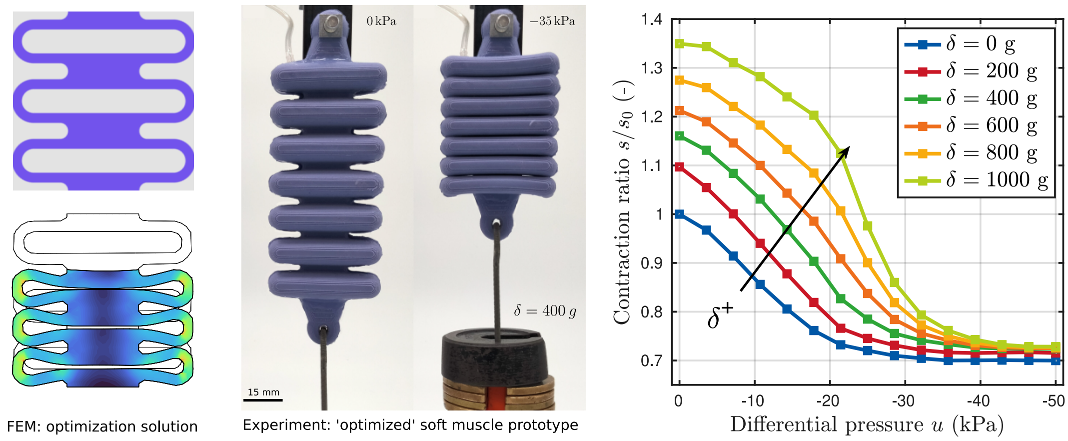

Fig. 5. (Left) Exemplary benchmark test of the vacuum-actuated soft actuator subjected to a loading weight of $\delta = 400$ g. Here we clearly see the soft subsystem being sufficiently strong to lift the suspended-weight for a differential pressure of $u = -35\,\text{kPa}$. (Right) The quasi-static evolution of the contraction displacement $s(t)$ against different input pressures $u(t)$ for different suspended weights δ. As expected, a decreasing slope for larger suspended weights can be seen. Although the actuator can lift $\delta > 1000g$, the contraction amplitudes significantly diminish beyond its practical use.

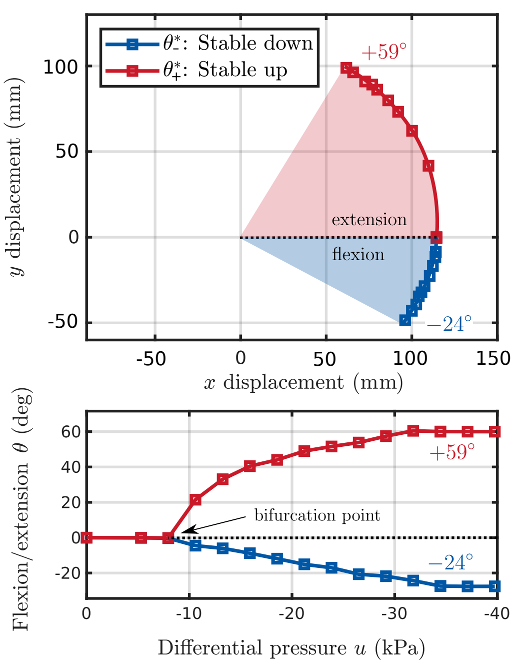

Fig. 6. Experimental results of the bi-stability analysis. Here we see the transition of one stable equilibrium ($\theta^* = 0$) for low-pressure, to an unstable equilibrium and two stable equilibrium pairs θ^*_-, θ^*_+.

inbalance between the gravitational component of the hand, the compliance of the beam, and contraction force of the soft actuator. Since the tendon of the soft actuator is in its most extended form in the straight posture of the hand ($\theta = 0$), the contraction of the tendon can lead to two unique (stable) equilibria. In this analysis, we load the wrist exoskeleton with the average human-hand weight δ_W, and investigate the bi-stability for various pressure inputs $u \in [-40, 0]\,\text{kPa}$. These experimental results are shown in Fig. 6.

As can be seen, there indeed exist pairs of stable equilibria for identical input conditions. We see that for low differential pressures (*i.e.* $u \geq -8\,\text{kPa}$), only one stable equilibrium $\theta^* = 0$ exists (passive gravity balancing). Further increasing the negative pressure input u, we observe a so-called bifurcation point, where $\theta^* = 0$ is now unstable and two new stable equilibria appear, $\theta^*_- < 0$ (down/flexion) and $\theta^*_+ > 0$ (up/extension), respectively. The change of stability in the zero position θ^* is actually beneficial, as it allows the user to switch to flexion or extension with limited effort. It is worth noting that the pair of equilibria can be tuned accordingly by changing the input u. Let it be clear that the magnitude of the flexion equilibrium is lower than the extension angle since the soft actuator has to further deform the compliant beam. Nevertheless, given the full actuation range of the soft actuator, we observe a substantial range of support ($|\theta^*_-| \leq 24°$ for flexion, $|\theta^*_+| \leq 59°$ for extension) for wrist mobility, indicating such hybrid system can be a viable, energy-efficient alternative to traditional wrist exoskeletons.

IV. Discussion

The results of the compliant beam and soft actuator experiments indicate that both subsystems satisfied the required share for balancing and actuating the user's hand, while the combination of both subsystems demonstrated the hypothesized bi-stable behavior, indicating that such design can support the hand in flexion-extension direction over a broad range of motion. This is further supported by the fact that the dead weight of the hand is balanced during actuation, the active support can greatly minimize the actuation energy. Additionally, both passive and active elements do not require energy to maintain an angle. Thus, when the device is turned off, it can be employed as a fully passive support or as a posture fixation for the hand's flexion-extension.

It is worth mentioning that this device is created to function as a semi-active gravity balancer. As a result, the orientation of the user hand must always match the direction illustrated in Fig. 2. However, this is not a significant concern because the majority of the device's intended users, *e.g.*, people with Duchenne disease, preferably have their hands on the armrest of the wheelchair.

The device demonstrates a substantial support angle (83°) for hand flexion-extension. This range is sufficient for the majority of the hand placement and range of motion tasks required to perform activities of daily living [23]. However, this range can still be enhanced in order to make it more symmetrical in terms of flexion and extension. For instance, a closed-loop control scheme can be proposed that counteracts the stiffness of the compliant beam under compression.

Nevertheless, there are still some limitations to the current design. First, additional experiments, modeling, and testing are required with patients to arrive at a final design that well integrates both subsystems – especially for user clinical tests. The proposed bi-stable property can still be improved by tweaking the individual subsystems and the connections between them. Additionally, patient tests might highlight constraint or comfort issues, e.g., limited wrist supination-pronation, that are currently not investigated. In future work, we will extend to patient trails.

V. Conclusion

A novel design for energy-efficient wrist exoskeletons is presented that utilizes a compliant mechanism and soft actuation to exploit both passive and active support elements simultaneously. The compliant beam is optimized, and the experiments indicated that it can balance the hand's dead weight over the desired range of motion. The topology-optimized soft actuator has demonstrated the ability to sustain the needed contraction force for hand flexion-extension. The combination of these two subsystems into a singular exoskeleton proved that the proposed bi-stable behavior of exoskeletons is capable of achieving both flexion and extension via actuator contraction in an extended range of motion (83°). This unique soft exoskeleton can be used in two modes: *(i)* when the passive part maintains the hand in its straight position and allows for a range of motion with lower muscular force; *(ii)* by engaging the soft actuator and active support of the body. Given these preliminary insights, such a semi-active wrist exoskeleton can be utilized to support other body parts while complementing the part's natural movement and requiring reduced energy consumption.

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