Soft, Lightweight Wearable Robots to Support the Upper Limb in Activities of Daily Living: A Feasibility Study on Chronic Stroke Patients

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***Abstract*—Stroke can be a devastating condition that impairs the upper limb and reduces mobility. Wearable robots can aid impaired users by supporting performance of Activities of Daily Living (ADLs). In the past decade, soft devices have become popular due to their inherent malleable and low-weight properties that makes them generally safer and more ergonomic. In this study, we present an improved version of our previously developed gravity-compensating upper limb exosuit and introduce a novel hand exoskeleton. The latter uses 3D-printed structures that are attached to the back of the fingers which prevent undesired hyperextension of joints. We explored the feasibility of using this integrated system in a sample of 10 chronic stroke patients who performed 10 ADLs. We observed a significant reduction of 30.3 $\pm$ 3.5% (mean $\pm$ standard error), 31.2 $\pm$ 3.2% and 14.0 $\pm$ 5.1% in the mean muscular activity of the Biceps Brachii (BB), Anterior Deltoid (AD) and Extensor Digitorum Communis muscles, respectively. Additionally, we observed a reduction of 14.0 $\pm$ 11.5%, 14.7 $\pm$ 6.9% and 12.8 $\pm$ 4.4% in the coactivation of the pairs of muscles BB and Triceps Brachii (TB), BB and AD, and TB and Pectoralis Major (PM), respectively, typically associated to pathological muscular synergies, without significant degradation of healthy muscular coactivation. There was also a significant increase of elbow flexion angle (12.1$\pm$1.5°). These results further cement the potential of using lightweight wearable devices to assist impaired users.**

***Index Terms*—Wearable robotics, assistive robots, muscle synergies, stroke, rehabilitation.**

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I. Introduction

MORE than 795,000 people suffer a stroke each year globally. This makes stroke a leading cause of disability, with more than half of stroke survivors aged 65 and above having reduced mobility [1] and paresis of the arms [2]. Wearable assistive robots have been lauded as a potentially beneficial tool for aiding in the performance of Activities of Daily Living (ADLs), but also in rehabilitation efforts, as they allow many repetitions and high-intensity training, all features that show evidence of being beneficial in stroke rehabilitation [3]–[5].

Several advances have been made in the field of upper limb wearable robotics, and a multitude of devices have been developed [6]–[8]. These devices are traditionally created using rigid structures, which can lack compliance to human natural motions and create misalignment between robotic and human joints [9]. In the last decade, research endeavours have shifted the focus towards softer devices. Typically created from deformable materials that improve their biomimetic properties, these wearable robots are generally more ergonomic and safer [10]–[14]. The soft properties of these devices are intrinsically related to the type of actuation, being based mostly either in fluidic or cable transmissions. For example, O'Neill et al. [12] placed a textile pneumatic actuator under the arm to support shoulder abduction by inflating and pushing the arm

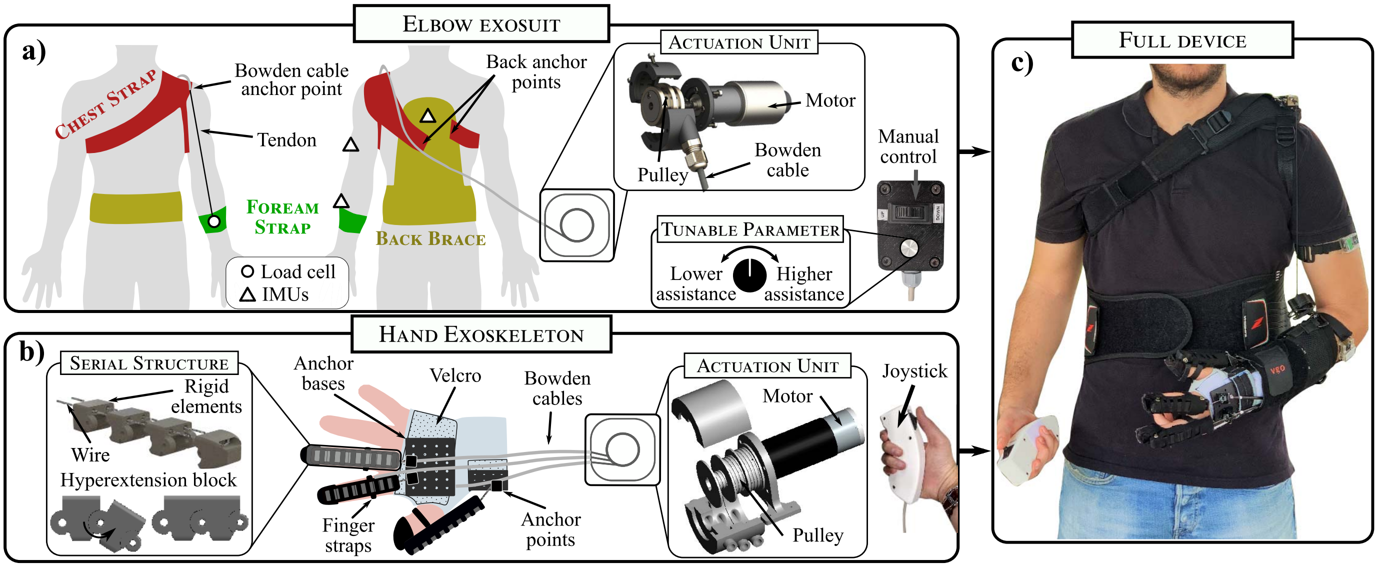

Fig. 1. a) Elbow exosuit. The exosuit is composed of three main components: chest strap, to which the Bowden cable anchor point is attached; forearm strap, where the tendon (in series with a load cell) is connected to; and the back brace, where the chest strap can be fixed. Three IMUs placed on the forearm, upper arm and back brace, respectively, allow for the exosuit's assistive torque to counteract gravity to be determined. The actuation unit and motor driver are stored remotely in a box. The tunable parameter can be adjusted by turning the knob in a joystick. b) Hand exoskeleton. A structure composed of serially linked rigid elements that prevent finger hyperextension is attached to the back of the index, middle fingers and thumb, by using a thimble-like cap and finger velcro straps. Tendons are routed through the structures and connected to an actuation unit via Bowden sheaths. The anchor points of the Bowden cables can be moved along rigid platforms (anchor bases), so their positions can be adapted to the user's preference. The Beaglebone Black (BBB), motor driver and actuation unit are stored in a box placed remotely. In series with the motor is a pulley divided in three sections (one for each tendon) with different diameters, adjusted to actuate the hand according to its first postural synergy. A joystick is used to control the current of the motor, adjusting the fingers' extension. c) The upper limb wearable robot, as a combination of the hand and elbow devices.

upwards, while 2 smaller actuators pivot the arm horizontally. Using cable-based actuation, Alicea et al. [11] employ tendons connected to a multi-diameter spool that actuate the fingers according to the first postural synergy of the hand.

An important feature common to all these devices is their light weight due to the nature of the soft components, giving way to increased portability. Such characteristics are fundamental for using robots as suitable devices for assistance with ADLs. However, they can be difficult to model and control [15] and attaching them to the body is not trivial [16]. For these reasons, hybrid wearable robots that employ features characteristic of both soft and rigid systems in a complementary manner have become increasingly popular in recent years [17]. Such exoskeletons leverage the soft components for increased compliance and comfort with the hard ones for embedding safety features and improving power transmission.

The potential of wearable robots for use in daily life has been shown before in terms of their favourable physiological impact on healthy participants, such as in reducing the muscular effort required to perform movements [10], [12], [18] or in delaying the onset of muscular fatigue [10]. Hull *et al.* [19] developed an upper limb exoskeleton that uses a pantograph mechanism to provide gravity-compensating support to the arm. A study on 12 healthy participants revealed a reduction of 37-57% in the mid deltoid muscle activity and a reduction of 55% in the Biceps Brachii (BB) when performing dynamic movements on a horizontal plane. O'Neill *et al.* [12] developed a soft wearable device for the shoulder that uses textile pneumatic actuators. They reported an average reduction of 20.1% in the activity of the Pectoralis Major (PM) in a trial with 3 healthy participants. Dudley *et al.* [20] report a reduction of 50% in the activity of the extensor digitorum muscles of a stroke patient when using a hand exoskeleton. Some research groups have also opted for using passive devices instead. Yin *et al.* [21] developed a passive exoskeleton for overhead tasks and report an average electromyographic (EMG) reduction of 39.3% on Anterior Deltoid (AD), 32.4% on medial deltoid and 32.2% on Triceps Brachii (TB) in a study with 15 healthy participants. Qu *et al.* [22] developed a passive exoskeleton for lifting tasks, and reported a reduction of 38% in the labrum-biceps muscle in a study with 8 participants.

Preliminary clinical studies have also led to encouraging results. Dinh and colleagues showed an increase in range of motion of the elbow joint when assisting a patient with bilateral brachial plexus injury, using a textile-based cable-driven exosuit [23]. A similar design, proposed by Li *et al.* [24] and extended to the shoulder, elbow and wrist joints, allowed for a 174% increase in the range of motion in hemiparetic stroke patients. Finally, Simpson *et al.* [25] showed that a soft, inflatable exosuit can improve the reachable workspace on a sample of six chronic stroke patients by alleviating the pathological flexor synergy, typical after stroke. These studies provide evidence that portable and lightweight devices can be used to aid therapy and improve range of motion in controlled, clinical environments. Their technical characteristics, however, make exosuits ideal for use in daily life, to improve independence by supporting functional tasks. According to Brose *et al.* [26], trials in natural environments should follow testing in laboratory conditions, and Gassert and Dietz [27] highlight that assistive robots should focus on supporting

ADLs. Furthermore, a recent study showed that adults view assistive robots as potentially very useful tools for helping with ADLs [28]. Nonetheless, to our knowledge, no study so far has investigated the feasibility of using wearable devices to help the upper limbs on neurological patients while performing ADLs.

In this study, we use our previously introduced soft exosuit meant for elbow assistance [13] and a soft-rigid hybrid exoskeleton supporting hand extension. We explore the feasibility of using this integrated wearable system as an assistive tool for performing ADLs on a sample of chronic stroke patients. We focused primarily on investigating the effects of the wearable robots on muscular activity while performing functional tasks. We observed that with the assistance of the devices, the participants reduced their muscular effort. Furthermore, we also investigated the effects of the robots in muscular coactivation patterns, without directly attempting to decouple abnormal muscular coactivation, as recommended by Santello and Lang [29]. Pathological synergies are typically divided in flexor synergy, characterised by elbow flexion and supination, shoulder abduction and extension, and wrist and finger flexion; and extension synergy, characterised by elbow extension and pronation, shoulder adduction and flexion, and potentially wrist extension and finger flexion [30]–[32]. We observed a slight reduction of pathological muscular coactivations of the upper limb, while healthy ones were kept intact. This study provides encouraging evidence to the benefits of using these technologies as a means to support post-stroke patients in everyday life.

II. Methods

A. Elbow Exosuit

The elbow exosuit used in this study, previously described in [10], [13], [33], supports the performance of elbow movements (Fig. 1a). The exosuit assists the elbow joint by pulling a tendon—routed through a Bowden sheath anchored on the superior region of the shoulder—attached to the forearm. The Bowden sheath (M-System Brake Cable Housing, Shimano) is connected to a unit consisting of a DC motor (*EC*-i ∅40, P/N 449469, Maxon Motor) in series with a planetary gearhead (reduction 33:1, P/N 166938, Maxon Motor) connected to a pulley around which the tendon (Kevlar ® Fiber, DuPont) is wound. The wearable part of the device can be divided in: chest strap, adapted from a passive orthosis (Master-03, Reh4mat), which secures the Bowden sheath anchor point on the shoulder; forearm strap, which holds a load cell to which the tendon is connected; back brace, adapted from a commercial back protector (Spine Evc X7, Zandona), which secures the chest strap. Both the chest and forearm straps are padded with neoprene to soften peaks of pressure points.

The exosuit presented here has gone through two primary changes compared to our previous works meant to improve its functionality, wearability and comfort. The Bowden sheath is now anchored on the shoulder instead of the upper arm, increasing the range of motion of the elbow joint. Furthermore, a rigid back brace was added with the purpose of stabilising the chest strap, as in the previous version it would frequently slip and rotate around the chest.

The control paradigm has been extensively described in our previous works [13], [34], to which we refer the reader for more detailed information. The algorithm estimates the position-dependent force necessary for creating an assistive torque at the elbow which cancels out the gravitational torque. The assistive torque is dependent on the angle of the elbow, whereas the gravitational torque is dependent on the angle between the forearm and the direction of gravity. These angles are determined from the 3 IMUs placed on the forearm, upper arm and trunk. The algorithm is also able to estimate the intention of the user: extending the elbow increases the perceived assisted torque, resulting in unwinding of the tendon to reduce the torque at the joint, and vice-versa. For calculating the gravitational torque, the weight of the arm is estimated through anatomical geometric relationships [35]. This method assumes all participants have the same anatomical proportions, therefore, we adapted the model by adding a tunable parameter to account for inaccuracies in using parametric models. The parameter changes the weight of the arm perceived by the algorithm, acting as a gain on the gravitational acceleration. The tuning is done such that, at 90° elbow flexion, the forearm's weight is fully supported.

The motor is velocity-controlled by a motor driver (EPOS2 50/5, Maxon Motor). The motor driver is torque-controlled using a data acquisition board (QPIDe, Quanser) which reads the tension of the tendon from the load cell. The IMUs (STEVAL-STLKT01V1, ST Microelectronics) are connected to a low-energy Bluetooth board (X-NUCLEO-IDB05A1, ST Microelectronics), integrated with the STM32 Nucleo board (ST Microelectronics) and connected to a PC (Windows 7). The control algorithm has been implemented in MATLAB ®.

B. Hand Exoskeleton

The hand exoskeleton assists finger extension and prevents unwanted closure. This is achieved by using structures composed of serially connected 3-D printed elements (MJF Plastic PA 12, HP) that are attached to the back of the index and middle fingers and the thumb (Fig. 1b), previously introduced in [36]. The use of this underactuated hyper-redundant mechanism allows for a high level of adaptability that is expressed in 2 ways: 1) the structures conform to different postures, creating minimal interference in hand motion when the device is not being actuated; 2) the structures are adaptable to different hand sizes, hence allowing for a single device to be used by many users. Another fundamental property of the used structure is that hyperextension is prevented due to mechanical blocks in the design of the elements (refer to Fig. 1b).

A tendon (PE Braided Line, 8 strands, ∅1 mm, max payload 114 Kg, Hercules) is run through each structure, such that pulling it results in straightening of the structure and extension of the finger. The tendons are routed through Bowden sheaths, fixed using anchor points on rigid anchor bases on the back of the hand. The Bowden sheaths are held on the forearm by using a strap, reducing the total weight felt by the user by lowering the interaction forces due to the stiffness of the sheaths. The anchor bases are attached to a soft framework adapted from a commercial wrist brace (Slim Silhouette Wrist

TABLE I
BASELINE DEMOGRAPHIC AND CLINICAL CHARACTERISTICS OF PARTICIPANTS (N = 10)

I.D.	Age [years]	Sex	Handedness	Impaired side	Time since stroke [months]	FMA		MAS					MOCA	Grip strength [kgf]
						SE	WH	S	E	W	F	T		
1	40	M	L	R	18	31	6	0	1	1	2	2	26	6.65
2	59	F	R	R	8	31	9	0	0	0	0	0	25	3.15
3	50	M	R	R	32	34	10	0	0	1	2	1	28	17.7
4	68	F	R	R	34	20	2	2	3	1	1	0	25	9.8
5	29	F	R	R	30	28	2	0	1	1	2	2	22	8.9
6	64	M	R	R	41	29	9	0	1	1	1	0	30	8.78
7	51	M	L	R	6	32	21	0	0	1	0	0	29	7.05
8	54	F	R	L	66	16	2	2	2	3	3	3	26	2.3
9	25	F	R	L	18	22	2	0	1	2	2	1	27	2.75
10	39	M	R	L	9	23	3	0	2	2	2	2	23	5.4
Mean	47.9	50% F	80% R	70% R	26.2	26.6	6.60	0.40	1.10	1.30	1.50	1.10	26.1	7.25
SD	14.4	—	—	—	18.5	5.97	6.04	0.84	0.99	0.82	0.97	1.10	2.51	4.55

FMA: Fugl-Meyer Assessment for the Upper Limb; MAS: Modified Ashworth Scale; MOCA: Montreal Cognitive Assessment; SE: Shoulder-Elbow coordination; WH: Wrist-hand function; S: Shoulder; E: Elbow; W: Wrist; F: Finger; T: Thumb; SD: standard deviation.

Support, FuturoTM, 3M). The anchor points can be moved across the bases to the user's preferred position, making the device customisable to the user's needs and comfort. This also allows for the thumb anchor point to be placed in different positions, modulating the desired level of abduction which results in distinct opening patterns.

The design of the actuation system is inspired from [37] in that it shares the concept of actuating the fingers according to hand postural synergies. Each tendon is wound around different sections of the same spool, whose diameters have been adapted to correspond to the first postural synergy. In this way, using a single actuator, it is possible to actuate multiple fingers in a manner that resembles the natural motion of the hand. The diameters of each section are 40 mm (index finger), 45.5 mm (middle finger) and 20 mm (thumb).

The spool is connected to a DC motor (*EC*-i ∅40, brushless, 50 W, P/N 449464, Maxon Motor) in series with a planetary gearhead (GP 42 C, ∅42 mm, reduction 150:1, P/N 203128, Maxon Motor). A handheld joystick encases a thumbwheel switch (IP67 Pre-wired Thumbwheel Switch, APEM) connected to a single-board computer Beaglebone Black (BBB) (BeagleBoard.org Foundation, MI). The BBB converts a voltage command to a pulse-width modulation (PWM) signal sent to a motor driver (ESCON Module 50/5, Maxon Motor). The motor driver controls the current on the motor windings according to the reference PWM signal. Therefore, rotating the thumbwheel in a defined direction positively increases the current on the motor, creating rotation that pulls the tendons, resulting in finger extension. Conversely, rotating the switch in the opposite direction provides the same proportional response but with negative current values, unwinding the tendons and passively allowing finger flexion. A manual trigger such as the joystick is used to control the hand device due to its robustness and simplicity. All components are stored remotely in a box (Fig.1). The wearable part of the device weighs 215 g.

C. Study Design, Setting and Participants

A feasibility single-centre open label clinical trial was carried out between 1 March 2020 and 31 October 2020. A total of 10 patients with subacute to chronic stroke who had completed their initial phase of inpatient rehabilitation at the Tan Tock Seng Hospital (TTSH) Rehabilitation Centre in Singapore or other similar centres were recruited for this study. Ethical approval was granted by the National Healthcare Group (NHG) domain specific review boards (NHG DSRB 2018/01358) in July 2019 and the trial was registered under www.clinicaltrials.gov (NCT05118321). Prior to enrolment, all participants gave written consent for clinical participation and use of pseudonymised data. The study inclusion criteria included patients with: 1) First-ever confirmed stroke as diagnosed by CT or MRI, who were attending outpatient rehabilitation services at TTSH Rehabilitation Centre @ Centre for Advanced Rehabilitation Therapeutics (CART); 2) Duration post stroke of at least 3 months with stable neurological status; 3) Age between 21 to 80 years, inclusive; 4) Hemiplegic pattern of motor weakness and shoulder abduction Medical Research Council (MRC) [38] motor power $>$ 2/5; 5) Ability to give informed consent; 6) Montreal Cognitive Assessment (MOCA) [39] $\geq$ 22/30.

Patients were excluded based on the following exclusion criteria: 1) Presence of severe cognitive, perceptual (include hemi-neglect), and/or emotional-behavioural challenges; 2) Presence of moderate to severe levels of pain (vertical Numerical Pain Rating Scale (vNPRS) $>$ 5); 3) Presence of unstable or terminal medical conditions which may affect participation or anticipated life expectancy of $<$ 1 year due to malignancy or neurodegenerative disorder; 4) Presence of non-stroke related causes of arm motor impairment; 5) Presence of local factors which may be worsened by arm therapy or device interface: spasticity of Modified Ashworth Scale (MAS)[1] $>$ 3 [40], unhealed skin wounds/rashes, shoulder pain Visual Analog Scale (VAS) $>$ 5/10, active fractures or arthritis or fixed flexion contractures of shoulder, elbow, wrist or fingers incompatible with device interface; 6) Inability to tolerate 90 minutes of therapy session; 7) Pregnancy or breast feeding; 8) Presence of severe sensory impairment to the affected upper limb.

[1]The MAS used in this study has been mapped from [0, 1, 1+, 2, 3, 4] $\rightarrow$ [0, 1, 2, 3, 4, 5] to have an ordinal scale with intervals of 1 between grades.

TABLE II
LIST OF FUNCTIONAL TASKS PERFORMED BY PARTICIPANTS

I.D.	Task name	Task description
1	Drawer	Open drawer at height such that elbow is at 90° flexion with affected hand.
2	Towel	Open drawer with unaffected hand. Retrieve a towel from inside drawer at same height as in Task 1. Grasp the towel and lift it out of drawer.
3	Toothbrush	Stabilize adult toothbrush in affected hand as unaffected hand applies toothpaste.
4	Chair	Pull non-rolling chair by the backrest, grasping with both hands, by 40 cm.
5	Cup	Pick up designated cup by handle with affected hand and bring to mouth level, and hold for 5 s.
6	Biscuit	Bring biscuit in plastic packaging to mouth level and hold for 5 s.
7	Key	Place key in affected hand with help of unaffected hand and put it through a keyhole at a height of 100 cm from the floor.
8	Phone	Pick handphone with affected hand and bring to either ear, and hold for 5 s.
9	Bag	Pick up a designated cloth bag with 1 kg weight from chair with affected hand. Bring it next to body, with elbow fully extended and shoulder at 0° flexion and abduction.
10	Cupboard	Fully open a cupboard door by its handle with the affected hand.

Fig. 2. Example of performance of the "towel" task in the NO-EXO and EXO conditions. The aim was to open/extend the fingers wide enough so that patient is able to actively grasp the towel and bring it to their side. Without the assistance of the robots, the participant was not able to grasp the towel, whereas with the device's assistance, the patient could open the hand wide enough to grasp it.

D. Study Protocol

Each participant completed three 90-minute long sessions at CART over the course of a week. In the first session, screening of the patient was performed, where their demographics and clinical parameters were collected, including the grip strength of the affected hand measured by a hand dynamometer (Jamar ®), and MOCA, Fugl-Meyer Motor Assessment for Upper Limb (FMA) [41] and MAS scores (Table I). Afterwards, a familiarisation phase was conducted, where the participant tried the robotic devices to learn how to use them. This phase also allowed the researchers to adjust the fitting of the robots and tune the control algorithm of the elbow exosuit to the biomechanics of each user. The exosuit's control parameters were heuristically selected by adjusting the assistive torque on such that, at 90° flexion, the elbow joint is stable.

In the two remaining sessions, participants performed a series of 10 functional tasks (Table II). The tasks were performed without assistance (NO-EXO condition, where the devices were not worn) and with assistance (EXO condition) of the robots (please refer to Fig. 2 for an example of the performance of a task in both conditions). The tasks were performed first in the NO-EXO condition to allow the patients to familiarise themselves with the tasks before performing them with the assistance of the devices. The participants were instructed to start every attempt from a position marked on the floor, with their arms fully relaxed, to ensure that variability from unknown factors was minimised. The instruction to commence the task was given in the form of a spoken command triggered by the press of a button. The button was pressed again to signal the end of the task, defined by the following in order of priority: 1) participant decided to stop; 2) the participant dropped the object; 3) therapist stops the task after deciding the participant was experiencing difficulties and further attempts would not contribute to a successful performance of the task; 4) task is successfully completed. Up to five attempts per task were allowed. The two best attempts were selected based on the following procedure: (1) select two attempts that are successful, starting from the last performed; (2) if there are less than two successful attempts, select the last unsuccessful attempt(s). Priority was given to the attempts performed last to allow for learning on how to perform the task. In addition to the performance of functional tasks, the standing horizontal fingertip reaching distance and the elbow Range of Motion (ROM) were measured using a measuring tape and a goniometer, respectively. At the end of the third session, the participants were asked to fill a questionnaire to provide their feedback regarding comfort, ergonomics, ease of use, overall clinical benefit and user satisfaction.

E. Data Collection & Segmentation

The data collected during the performance of the experiments included the EMG data of the upper limb muscles—BB, TB, AD, PM and Extensor Digitorum Communis (EDC)—and kinematic data of the affected arm. The primary outcome of the study was the difference in muscular efforts, measured by the amplitude of the EMG signal, between the EXO and NO-EXO conditions. As a secondary outcome, we used the EMG data to investigate changes in muscular coactivation. In line with this, the kinematic data were used for investigating the influence of the devices on compensatory movements of the shoulder. Sensor data was collected at a sampling frequency of 166 Hz. Data was segmented at the time instances when the button was pressed.

1) Electromyographic Data: The EMG data were obtained from 5 wireless electrodes (Delsys Trigno IM) placed following SENIAM guidelines for electrode placement [42]. At the beginning of each session, the patients performed a Maximum Voluntary Contraction (MVC) which was later used to normalise the EMG signals. Data were processed in an offline analysis: twice notch-filtered (IIR notch-filter with cut off frequency 50 Hz for removing powerline noise, and 1.67 Hz

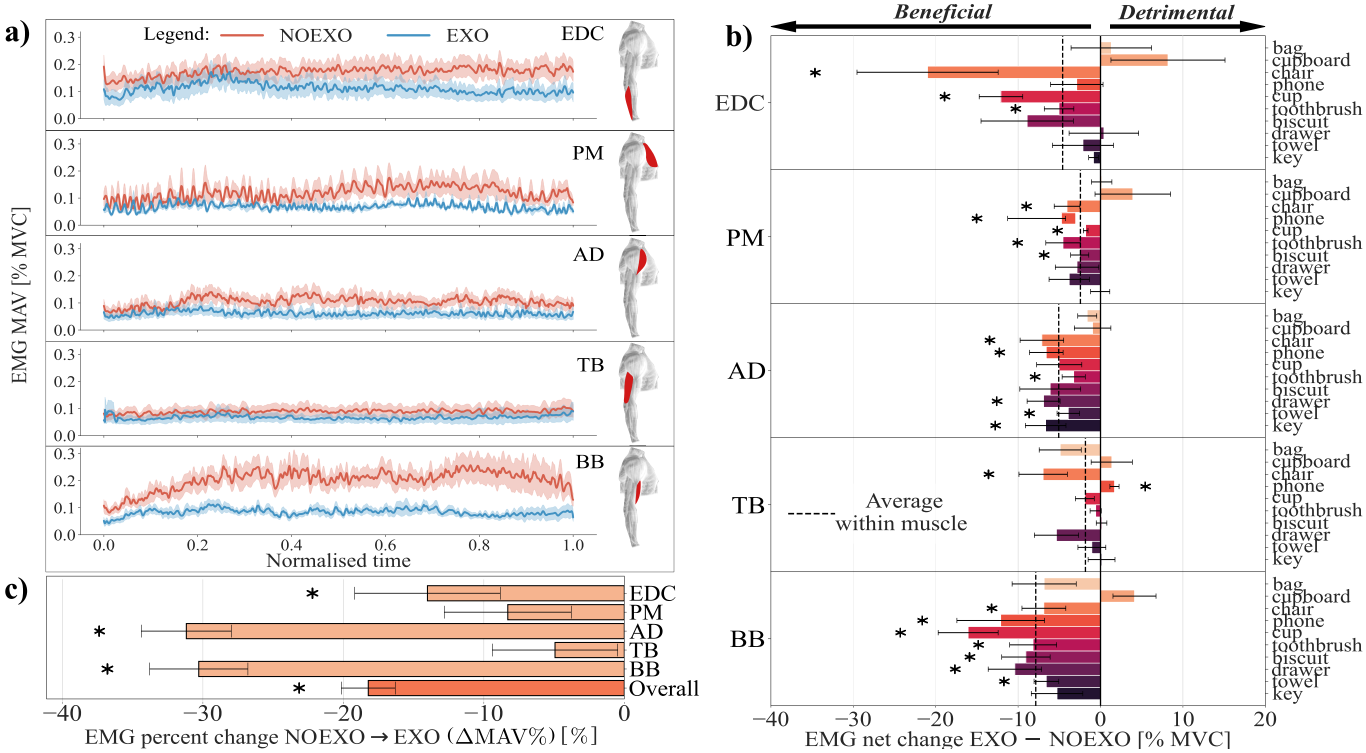

Fig. 3. **a)** Example of EMG profile, as a percentage of the MVC, for each muscle, averaged for all participants, when performing the "toothbrush" task. **b)** Net changes in muscular activation from NO-EXO to EXO condition discriminated per task. **c)** Changes in muscular activation as a percentage of the MAV in the NO-EXO condition, averaged across tasks. Negative values indicate a reduction in muscular activity when using the robots compared to not wearing them. The mean percent change was (mean $\pm$ standard error): BB $-30.3 \pm 3.49\%$ ($p = 5.26 \times 10^{-15}$), TB $-4.93 \pm 4.47\%$ ($p = 0.27$), AD $-31.2 \pm 3.21\%$ ($p = 1.00 \times 10^{-17}$), PM $-8.29 \pm 4.53\%$ ($p = 0.069$), EDC $-14.0 \pm 5.12\%$ ($p = 7.88 \times 10^{-3}$). Total EMG change averaged across all muscles was $-18.2\% \pm 1.92\%$ ($p = 2.81 \times 10^{-20}$). There was a significant reduction in muscular activation for the BB, AD and EDC. Asterisk (*) indicates statistical significance. Error bars indicate standard error of the mean.

for removing heartbeat noise); high-pass filtered (10th order Butterworth filter with cut off frequency 20 Hz); rectified by taking the absolute value; and finally smoothed via a low-pass filter (10th order Butterworth filter with cut off frequency 4 Hz) (inspired by [43]). The final set of values was averaged to obtain the Mean Averaged Value (MAV) for each attempt, and the values of the two best attempts were averaged. Regarding the secondary outcome, muscular coactivation patterns of pairs of muscles associated with pathological synergies were investigated, namely the BB + AD, the TB + PM, which are often the target of rehabilitation efforts [44], [45]; and the BB + TB, which can also be associated to abnormal muscular control in post-stroke participants, albeit less so [44]. Coactivation between muscles associated with healthy synergies was also investigated, namely the pair BB and PM [44]. Muscular coactivation was calculated using Pearson's r, similarly to previous studies [44], [45].

2) Kinematic Data: The kinematic data were obtained from 3 wireless IMUs placed proximally on the forearm and laterally on the upper arm, and on the back of the participant (Fig. 1a). The data was collected according to a 6-Degree of Freedom (DOF) model of the upper limb and trunk: elbow flexion θ_e, shoulder elevation θ_{se} and shoulder azimuth θ_{sa}, trunk flexion θ_{tf}, tilt θ_{tt} and rotation θ_{tr}. The IMUs data were low-pass filtered (2nd order Butterworth, cut off frequency 10 Hz) and calibrated by considering the initial position $\theta_e = 180°$ and $\theta_{se} = \theta_{sa} = \theta_{tf} = \theta_{tt} = \theta_{tr} = 0°$. The 2 best attempts (the same that were used with the EMG data) were selected for both conditions and averaged. Only the elbow and shoulder trajectories were included in the analysis, as these are the joints whose motion is mostly affected by the devices.

3) Questionnaires: The participants were asked to fill out a questionnaire at the end of the third session to give their feedback on the devices and the overall experience with respect to all sessions (Table III). The questions were created based on common usability attributes utilized for analysing user feedback and performance of wearable robotic devices [46]. The questionnaire was divided in 2 parts, one for the elbow and one for the hand exoskeleton, both with the same questions. The answers could be given in a 5-point Likert scale, ranging from "Strongly disagree" to "Strongly agree". The results were mapped to a numeric ordinal scale to allow for easier visualisation, where higher scores correspond to a more positive outcome.

F. Statistical Analysis

A statistical analysis consisting of Linear Mixed Models (LMMs) was conducted considering the response variables as the net change (ΔMAV) and the percent change (ΔMAV%) of the MAV of the EMG from the NO-EXO to the EXO condition, and the difference between the trajectories ($\Delta\theta_i$, with i representing each DOF) for each condition. The condition

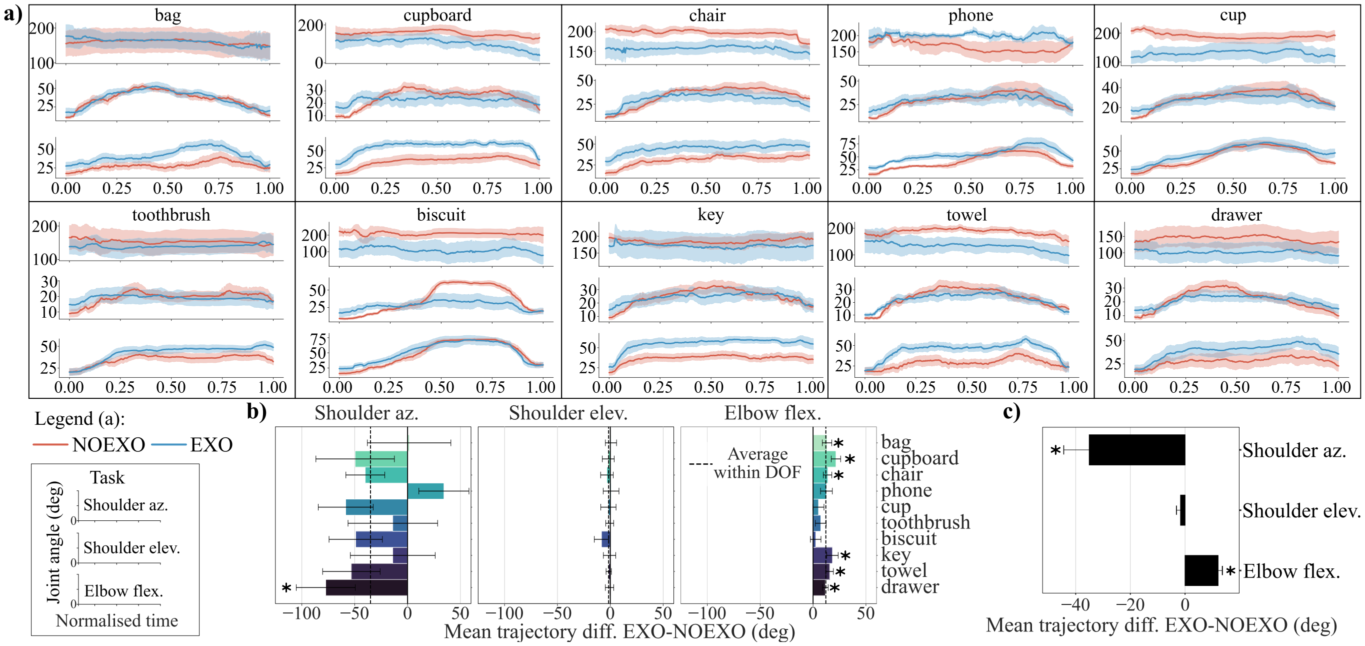

Fig. 4. **a)** Angular trajectories of elbow flexion (bottom), shoulder elevation (middle) and shoulder azimuth (top), per task (divided by panels), averaged across participants, for both EXO (blue) and NO-EXO (red) conditions. The legend illustrates how each panel should be read. **b)**, **c)** Mean difference between the trajectories in both conditions, with values within each DOF averaged across tasks (b) and discriminated per task (c). Negative values indicate a reduction in the mean values for the angular trajectory. The mean difference was (mean ± standard error): elbow flexion 12.1±1.53 ($p = 4.50 \times 10^{13}$), shoulder elevation −1.65±1.50 ($p = 0.273$), shoulder azimuth −35.0±9.32 ($p = 2.53 \times 10^{-4}$). There was a significant decrease in shoulder azimuth and a significant increase of elbow flexion angle. Asterisk (*) indicates statistical significance. Error bars indicate standard error of the mean. Asterisk (*) indicates statistical significance.

and all the parameters indicated in Table I were considered fixed effects, resulting in a total of 15 fixed effects. Once the LMMs were fitted, the only significant factor was the condition (NO-EXO or EXO). The models were then fit a second time dropping all other factors.

We used LMMs to control for non-independence arising from nested structures, such as the performed task or the analysed muscle. This allows us to select random effects depending on the intended grouping of the data. Within each combination of muscle (or DOF) and task, the only random effect considered was the participants. Within each muscle (or DOF) and across tasks, the tasks were added as a random effect. Finally, for the overall effect across all muscles (or DOFs), the muscles (or DOFs) were added as a random effect. Outliers were removed by visual inspection and by removing values larger than 1.5x the interquartile range.

For investigating muscular coactivation, data were tested for normality using Shapiro-Wilk tests, and in non-normal data, a rank-based inverse normal transformation was performed before computing the Pearson's r, as recommended by [47]. A LMMs-based analysis of the correlation coefficients was conducted, and the condition was also found to be the only significant factor. Differences are reported as $\Delta r = r^{\text{EXO}} - r^{\text{NO-EXO}}$. The standing horizontal reaching distance and the ROM of the elbow in both conditions were also compared. Normality tests verified the data were normal, therefore paired samples t-tests were used. Statistical analysis was conducted using RStudio ® (significance level: $p < 0.05$).

III. Results

A summary of the baseline characteristics of the participants can be found in Table I. All 10 participants completed all assessment tasks without adverse events or drop out. There was an increase of the users' reaching distance of 2.13 ± 1.63 cm ($p = 0.906$) and a decrease in the ROM of the elbow of 0.5 ± 4.1° ($p = 0.225$), both non-significant. The average number of attempts for a successful task was 2.75, indicating that at the 3rd attempt, the participants had learned how to use the devices to perform tasks.

A. Effects on Muscular Effort

An example EMG profile is shown in Fig. 3a, where each panel corresponds to a different muscle. The net change in EMG activity of the 5 analysed muscles per task can be seen in Fig. 3b, where each coloured bar represents a task. Most of the 50 possible combinations muscle + task exhibited a decrease in muscular activity, although not necessarily a significant one. The overall percent change in muscular activation for each muscle can be seen in Fig. 3c. There was a significant reduction of 30.3 ± 3.49% ($p < 0.001$) for the BB, 31.2 ± 3.21% ($p < 0.001$) for the AD and 14.0 ± 5.12% ($p < 0.01$) for the EDC. The TB and the PM exhibited non-significant reductions of 4.93 ± 4.47% ($p = 0.27$) and 8.29 ± 4.53% ($p = 0.069$), respectively. Overall, there was a significant reduction in muscular activity of 18.2% ± 1.92 % ($p < 0.001$).

TABLE III

QUESTIONNAIRE AND RESPONSES IN A 5-POINT SCALE. HIGHER SCORES REPRESENT MORE SATISFACTORY OUTCOME. RESULTS ARE PRESENTED AS MEAN ± STANDARD ERROR. Q_E: SCORE FOR THE ELBOW DEVICE; Q_H: SCORE FOR THE HAND DEVICE

Question	Q_E	Q_H
The suit was easy to use.	3.9	3.1
The suit was comfortable.	3.8	3.6
I would use it frequently.	3.5	2.9
I find it unnecessarily complex.	2.3	2.6
I would need the support of a technical person to be able to use this system.	2.7	2.1
The training was enjoyable.	4.4	4.2
The training allows me to be able to do activities that I would have otherwise been unable to.	3.9	4.1
I found the various functions in this system were well integrated.	4.1	3.9
The device allows me to do movements that I was not able to achieve.	4.1	4.3
	1 – 5	1 – 5

B. Effects on Upper Limb Trajectories

The 3 analysed angular trajectories (elbow flexion, shoulder elevation and shoulder azimuth), averaged across participants, can be found in Fig. 4a. Each panel corresponds to a different task, and within the panel, the top, middle and bottom plots correspond to shoulder azimuth, shoulder elevation and elbow flexion (in degrees), respectively. The mean difference between the trajectories in both conditions can be seen in Fig. 4b (separated by task) and in Fig. 4c (averaged across all tasks). Overall, users achieved significantly larger elbow flexion ($\Delta\theta_e = 12.1 \pm 1.53°$, $p < 0.001$) and shoulder azimuth angles ($\Delta\theta_{sa} = -35.0 \pm 9.32°$, $p < 0.001$), while no significant changes were observed in shoulder elevation angles ($\Delta\theta_{se} = -1.65 \pm 1.50°$, $p = 0.273$).

C. Effects on Muscular Coactivation Patterns

The change in muscular coactivations can be seen in Fig. 5. There was a significant overall percent reduction in coactivation for the following pairs of muscles: 14.0 ± 11.5% for the BB + TB ($\Delta r = -0.068$, $p = 0.014$), 14.7 ± 6.92% for the BB + AD ($\Delta r = -0.079$, $p = 0.003$) and 12.8 ± 4.36% for the TB + PM ($\Delta r = -0.051$, $p = 0.029$). There was a non-significant reduction of 9.83 ± 7.13% for the BB + PM ($\Delta r = -0.032$, $p = 0.21$). Observing Fig. 5, we can see that the average changes in Pearson's r per task have more variability for the pair BB + PM, which contributes to why this is the only pair with non-significant results.

D. Questionnaire Responses

The responses to the questionnaires can be found in Table III. Subjective ratings of impairment reduction, functional assistance and user experience were better rated than usability of the prototype. This implies positive initial acceptance of the prototype system.

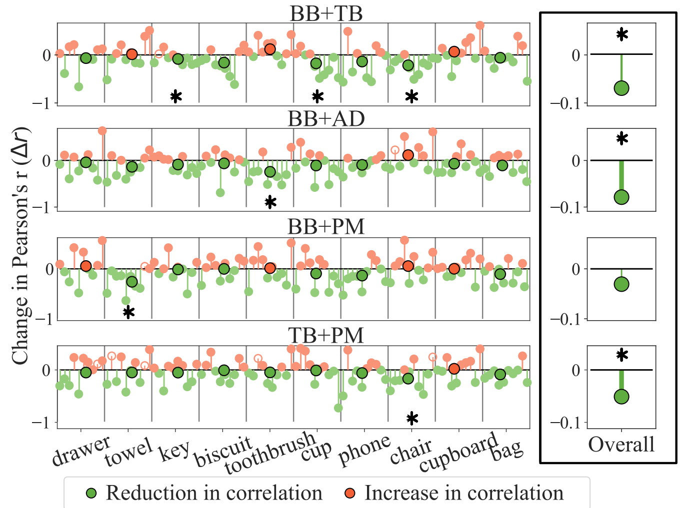

Fig. 5. Changes in EMG correlations between different pairs of muscles. Small circles indicate values for each participant, large circles with black contour indicate mean. Empty circles indicate non-significant Pearson's r. The panels on the right have a different scale to better highlight differences between different pairs of muscles. There was a decrease in average muscular coactivation across tasks for all pairs (mean ± standard error): BB + TB ($\Delta r = -0.068$, $p = 0.014$), BB + AD ($\Delta r = -0.079$, $p = 0.003$), BB + PM ($\Delta r = -0.032$, $p = 0.21$) and TB + PM ($\Delta r = -0.051$, $p = 0.029$). All reductions were significant except for the pair BB + PM. Asterisk (*) indicates statistical significance.

IV. DISCUSSION

A. Benefits of Using the Wearable Robots

One of the observed benefits of using our robots is a reduction of muscular activity of stroke patients. The reduction of 30.3% in the BB activity is in accordance with our previous results when evaluating the former version of the elbow exosuit (24.3% reduction) [13]. We observed a reduction in the AD activity, whereas in our previous study there had been an increase. This happened likely due to two reasons. First, the Bowden cable anchor point was shifted to the shoulder, meaning the assistance can now be delivered to the shoulder muscles as well. Second, the high stiffness present at the elbow joint of stroke patients also contributes to the propagation of the torque delivered at the elbow joint to the shoulder joint. This could be partly responsible for the increased assistance to the AD muscle. The change in the TB effort was, as expected, non-significant, as the exosuit provides a flexion torque on the elbow joint which counteracts elbow extension. Finally, there was a significant reduction of 14% in the activity of the EDC, showing that the hand wearable robot was successful in reducing the muscular effort required to open the hand.

From inspection of Fig. 3b, one can see that, for the PM, the overall trend across tasks is that of reduction in activity with the clear exception of the "cupboard" task. In fact, this is the task for which there is mostly an increase in muscular activity. Ignoring this task in the analysis yields a muscular reduction of 35 ± 3.28% for the BB, 15.1 ± 3.47% for the PM and 17.0 ± 4.82% for the EDC. We believe the reason for this is twofold: first, the cupboard used in this study was spring loaded, demanding more force than in a task with similar movements (e.g. "drawer" task); second, the glove introduced a thickness of about 15 mm on the fingers of the users, making it difficult to loop their fingers in the cupboard handle.

Other studies have also reported on the effects of wearable robotic devices for upper limb assistance on muscular activity (please refer to Sec. I. Introduction), but there was poor comparability to our results. We could not find studies that explore the myographic effects of devices on stroke patients with a sample size larger than one [20]. Thus far, reports on larger samples have been conducted on healthy populations [19], [21], [48]. Importantly, we also did not find studies that use ADLs for assessment in stroke patients, with most studies using very specific tasks or controlled movements.

We have previously shown [10] that the elbow exosuit is capable of reducing muscular activity in healthy participants by amounts similar or superior to those reported in literature. However, the results reported here indicate the muscular activity reduction is lower in stroke patients. A possible reason for this is that there is less motor recruitment in stroke patients than in healthy ones due to upper motor neuron lesions. Nonetheless, it is noteworthy that this study is, to the best of our knowledge, the first to systemically investigate changes in muscular activation in a large sample of post-stroke patients performing ADLs

Another benefit of using the proposed wearable system is the reduction of muscular coactivation typically associated with pathological synergies. There was a decrease in coactivation of the BB and AD, TB and PM, BB and TB. Importantly, healthy muscular coactivation has not been affected: there was no significant change in the coactivation of the BB and PM. We suspect that the reason for an observed reduction in pathological muscular coactivation is related to the provided gravity support. It has been shown that providing gravity support to shoulder abductors can reduce the coactivation of elbow flexors [49]. Although our elbow exosuit does not directly provide shoulder abduction support, it does support the AD muscle, which contributes to shoulder abduction when the hand is away from the body in the frontal plane. In hemiparetic patients, the observation of increased shoulder abduction during task performance is a manifestation of associated reaction related to the increased compensatory effort made by the participant to attempt completion of the task. Associated reactions are effort-dependent phenomena causing involuntary increase in upper limb muscle tone, with awkward and uncomfortable postures [50]. In future studies, other muscles such as the lateral deltoid should be investigated to understand better the effect of our device on shoulder abduction.

A common compensatory strategy adopted by stroke patients when moving their upper limb is to elevate their shoulder and upper arm in order to raise the height of the hand. In our study, we observed no significant change in shoulder elevation, but there was a significant reduction in shoulder azimuth. This could be due to the higher assistance to the elbow joint, as observed by the increase in elbow flexion, which allows for less reliance on shoulder movement. In this way, the necessary hand height can be achieved via elbow movements rather than shoulder ones. This finding further supports the hypothesis that the use of this wearable technology contributes to restoring normal function of the upper limb.

It is also important to comment on the clinical relevance of the results, given that although the reduction in muscular coactivation was statistically significant, the reduction in Pearson's r was small (10$\sim$20%). This could be explained by the nature of training, which did not focus on dissociating pathological synergies, but rather in positively impacting performance of ADLs. In addition, 4 of 10 participants had moderate to severe finger spasticity which may have been difficult to overcome voluntarily. Furthermore, this study is limited to evaluating the orthotic effect of the devices, i.e. the immediate effect observed upon using them, as opposed to the therapeutic effects which would arise over a longer study period. For example, in a longitudinal study where participants moved a cursor using a myoelectric-computer interface to follow targets [45], there were muscular coactivation reductions of up to 100%. We postulate that, if the patients were given a longer period of time to practice with the devices, and had there been more therapy sessions, the effects on muscular effort and coactivations would be amplified. Therefore, a longitudinal study is necessary to draw conclusions on the impact of this wearable robotic system on muscular synergies and overall long-term effect on hemiparetic users.

The modifications to the elbow exosuit from our previous works had a positive impact in the usability of the device. Before the modifications, an adjustment of the fit was required after a short period of time, whereas now the participants were able to wear the device throughout the whole 90-min session without requiring significant adjustments, while also observing acceptable skin tolerability. Furthermore, the rigid components in the hand device prevented the participants' fingers from undergoing hyperextension. Both of these observations indicate the benefits of utilising a hybrid approach to the design of wearable technologies.

B. Limitations and Future Work

The results of this study are encouraging, however, there are some limitations that should be highlighted. Distinguishing the effects of the elbow and hand robots is not straightforward. As mentioned before, the thickness introduced by the glove device hampered the performance of some tasks due to reduced finger tip sensory feedback and higher muscle forces needed to overcome tactile blocking from the glove. Therefore, it could be that there would be even further reduction of muscular activity if the users could better grasp the objects they interacted with. Conversely, the effects observed on the elbow and shoulder muscles are affected by the assistance in the performance of tasks by the hand module. As such, extrapolation of individual effects of each wearable robot should be done with caution.

The tasks were all performed first by the participants in the NO-EXO condition. This could have an effect on muscular activation changes, as learning effects from the first session could mean the users have adjusted their performance to activate their muscles less. Future studies should randomise the order of conditions for each participant.

There was no significant increase in elbow ROM when using the devices. However, this measurement was taken in a short period of time before the trials (<5 s), whereas considering the ROM during the trials, the devices helped the users achieve elbow flexion angles on average 12° larger than without their help. A possible explanation for this could be that the exosuit reduced fatigue (which we have shown before with healthy users [10]), allowing the user to achieve larger ROM for a longer period of time.

Regarding the subjective feedback given by the participants, we can observe that overall, the users were satisfied with the devices and the training. However, even though they were able to perform more activities with the hand otherwise not possible ($\boldsymbol{Q_H} = 4.1$ vs. $\boldsymbol{Q_E} = 3.9$), they would still use the elbow device more frequently ($\boldsymbol{Q_E} = 3.5$ vs. $\boldsymbol{Q_H} = 2.9$). This is not because the hand device is more complex ($\boldsymbol{Q_H} = 2.6$ vs. $\boldsymbol{Q_E} = 2.3$), but likely due to needing the support of a technical person to be able to use it ($\boldsymbol{Q_H} = 2.1$ vs. $\boldsymbol{Q_E} = 2.7$). In fact, participants reported that the current prototypes required multiple adjustments—such as alignment of Bowden cables for optimal functioning—and would prefer automatic control of the hand device. These concerns highlight the importance of developing wearable robotic systems that can be used without requiring assistance and that are simple to operate. In future iterations, these issues should be addressed, hopefully giving more independence to hemiparetic patients using them.

As mentioned above, the participants highlighted that control of the hand device via intention detection methods is preferred. However, from a rehabilitation perspective, using a joystick can present benefits to the user. The use of the joystick requires bimanual coordination of the upper limb, which is an important aspect of skilled arm use necessary for optimal human functioning in performing ADLs [51]. The appropriate timing for the preshaping of the hand is critical during the transport phase to the target object under visual control for successful reach and grasp [52]. Therefore, using the joystick enables the opportunity for bimanual training of the upper limb, which may improve the paretic upper limb function because of spatial and temporal coupling [53], [54]. On the other hand, a literature review on intention detection methods for robotic upper limb orthotic devices [55] mentions that manual triggers are a common strategy used due to their ease of use, reliability, and robustness, as they are not dependent on physiological signal acquisition and processing. Nonetheless, it would be useful to explore alternative methods to control the hand device and what their benefits can be in a real-life scenario such as the one presented in this study.

V. Conclusion

This study has shown the orthotic effects of our robotic system, namely the mitigation of the muscular effort required by the upper limb in hemiparetic users, and the reduction of the presence of pathological muscular coactivation and compensatory movements of the shoulder. These results further contribute to our understanding on the potential roles of arm and hand wearable robots in rehabilitation and assistance of hemiparetic stroke patients, highlighting the positive effects they can have on functional performance in everyday life scenarios.

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