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Design and Control of the Portable Upper-limb Elbow-forearm Exoskeleton for ADL Assistance

Hilary HY Cheng
Department of Biomedical Engineering
National University of Singapore
Singapore
hiuyee.cheng@u.nus.edu

Thomas M. Kwok
Integrative Sciences and Engineering Programme,
NUS Graduate School
National University of Singapore
Singapore
thomasm.kwok@u.nus.edu

Haoyong Yu
Department of Biomedical Engineering
National University of Singapore
Singapore
bieyhy@nus.edu.sg

***Abstract*—This paper proposes a portable Unilateral Elbow-Forearm Exoskeleton (UEFE) for assisting chronic stroke patients in the activities of daily living (ADL). UEFE provides users 2DoFs assistance: elbow flexion/extension (eF/E) and forearm pronation/supination (eR). Other than eF/E, forearm rotation is equally crucial for completing ADL tasks; however, limited exoskeletons provide eR assistance for users in ADL. Even though some existing exoskeletons are equipped with eR joints, those devices are not practical to use in ADL due to their heavy weight, insufficient assistance, and obstructing handles. In UEFE, both active joints are actuated by Series Elastic Actuators (SEA) through Bowden cables to ensure safe interactions and a lightweight solution. The design of UEFE is based on the ADL requirements, which include the range of motion (ROM) and torque. The handle-free feature also allows users to pick up objects in ADL. A pill-taking task with impedance control demonstrates the feasibility of using the UEFE in ADL assistance.**

***Index Terms*—Activities of daily living, stroke, Upper-limb exoskeleton,cable-driven robot.**

I. Introduction

One in four adults above 25 years old will have a stroke in a lifetime [1]. After a stroke, patients are likely to develop upper limb motor impairments such as muscle weakness [2], [3]. A study mentioned that even four years after the stroke, 42% of patients retained arm disabilities and were merely possible to grasp and perform gross movements [4]. Along with the upper limb deficits, some stroke survivors are unable to perform activities of daily living (ADLs) [5], including self-care and social activities [6], [7].

With the aid of technology, robotic systems can potentially assist ADLs or even take over the functionality when patients' muscular functions are totally lost [8], [9]. Coming along with the increase in stroke incidence [1], the upper limb exoskeleton for the disabled has become a hot research topic since 2000 [10]. However, not all devices are practical for ADL uses.

For activities of daily living, elbow assistance is essential for chronic stroke patients. First, the elbow movement provides a larger reachable workspace than the wrist and hand anatomically. In addition, most ADL tasks, including drinking and feeding, require 80° or more of elbow flexion [11]. Thus, elbow flexion is significant in ADL.

Other than elbow flexion, a majority of ADL tasks require large forearm supination [11], for instance with a max 53° during perineal care; while pronation is equally crucial for bringing the forearm back to its natural position as well as doing some other tasks, like pronating 13° to don and zip pants [11].

In the current literature relating to upper limb ADL exoskeletons, there are three major design limitations that make the exoskeletons not ADL-friendly. First, some of the ADL exoskeletons are not solely portable since they are designed to be mounted on wheelchairs. Yet, a study shows that 74% of stroke survivors regain their capacity to walk without assistance through rehabilitation in 2 years [12]. Therefore, these wheelchair-mounted upper limb exoskeletons might not be helpful for those who can walk in daily life.

Second, ADL exoskeleton designers often neglect forearm pronation and supination assistances though these rotations are important for accomplishing ADL tasks. Foreshadowed in the last paragraph, both wheelchair-mounted upper limb exoskeletons do not possess the pronation/supination joint. In addition, other portable ADL exoskeletons, for instance, the commercialized device Myomo MyoPro [13], the soft Exosuit [14] developed in NTU Singapore, and Cable-Driven Upper-Limb Exoskeleton [15], are not equipped with forearm rotation joint as well.

In general, for effective force transmission, the user must be tightly strapped to the exoskeleton in order to synchronize motions and prevent joint misalignment. For those exoskeletons that miss the forearm joint, to ensure better elbow joint alignment, the forearm is usually strap-fixed in natural pose (0°), which limits pronation and supination. However, a study shows that restricted Forearm pronation or supination leads to unwanted compensatory movement [16]. The compensatory movement, like excessive trunk and shoulder motion, may cause tiredness and muscle fatigue to stroke survivors.

Third, among the existing exoskeletons that provide pronation/supination motion, there are two major problems that hinder stroke users' from achieving ADL tasks. The first problem is the handle obstructing users from manipulating ADL objects [17]–[22]. In literature [17] to [22], handles are installed at the end-effector to provide better torque transmission at the forearm. Nevertheless, holding the handle refrains users from handling and holding ADL objects, like water cups or dining utensils. Thus, for real-life ADL applications, it is better to adopt a handle-free design that enables users to grasp and hold external objects to accomplish ADL tasks.

The second problem is the forearm assistive torque given by the existing exoskeletons may not be sufficient for all stroke survivors. In the field, there are inadequate robot testing and human experiments concerning the forearm rotation joint. Without testing the forearm rotation motions with the existence of a human subject [23]–[25], the forearm assistive capability cannot be justified.

In addition, there are existing soft exoskeletons that enable forearm rotation [26], [27]. Li et al. tested the system with a dummy with unknown arm weight and joint stiffness [27], but no torque information is reported. As the system is soft and cable-driven, the force transmission is poor due to the unfixed cable attachment points given by the fabric sliding nature. Thus, it is not persuasive that the device can provide adequate forearm rotation assistance for stroke users with different body sizes. In Su et al.'s study [26], the maximum provided torque for forearm rotation is around 846mNm, while a healthy human requires 0.6Nm (600mNm) to perform ADL tasks [28]. It is undoubtedly that the device is able to assist some subjects; nonetheless, stroke survivors may need extra torque to rotate their joints due to joint stiffness and spasticity. Therefore, it would be better to increase the maximum assistive torque for stroke users' sake.

Taking the aforementioned limitations into consideration, this paper proposes a lightweight and portable exoskeleton, Unilateral Elbow-Forearm Exoskeleton (UEFE), that fits the ADL requirements for assisting stroke patients in forearm and elbow rotation. To ensure the practicality in ADL, UEFE guarantees sufficient torque assistance despite of its handle-free design. The details of the exoskeleton, including mechanical structure, actuation, sensing, and potential control method, will be covered. Related experiments are done to show that the exoskeleton can perform the required motion and assist the subject complete an ADL task.

In what follows, we first describe the details of the UEFE in Sec. II and III, which includes the ADL requirements, robot design, and control. The experimental setup and results are presented in Sec. IV. Sec. V presents the significance of the robot through discussion.

II. Robot Design and Actuation

Unilateral Elbow-Forearm Exoskeleton (UEFE) is a portable exoskeleton that assists chronic stroke patients daily anywhere, as shown in Fig.1a. Because of the mentioned limitations, such as robot portability, missing assistance for the forearm (pronation/supination), end-effector handle obstructing grasping, and insufficient assistive torque for ADL, most existing upper limb exoskeletons cannot assist users' ADL anywhere, and not even at home. Our UEFE can overcome the limitations with three unique features: (1) robot design considering forearm pronation/supination, (2) forearm brace fixing and driving the forearm joint, and (3) cable-driven mechanism for force transmission.

A. Robot Design

The UEFE has two active degrees of freedom (DoF) assisting in elbow flexion/extension (eF/E) and forearm pronation/supination (eR), as shown in Fig.1b. Both active joints have connected to a linear series elastic actuator (SEA) in the backpack with a pair of Bowden cables, as in Fig.1c.

Since the SEAs are connected through cables, if needed, changing a SEA with larger torque output is easy and does not need any modification on the exoskeleton structure. The modular design, SEAs and exoskeleton, is helpful for stroke patients with various body weights and spasticity conditions.

Moreover, the UEFE can fit patients with different anthropometries. As the rotational axis of the exoskeleton's elbow and forearm joint are orthogonal to each other, the length between the two active joints does not need to change according to the user's arm length. Therefore, link length adjustment is not required for this design. Furthermore, given the flexibility of Velcro straps attachment, no size alterations are needed for fitting different users. Therefore, this free-size design can be useful when delivering the exoskeleton to patients for daily assistance outside clinics.

B. Forearm brace design

Holding a handle ensures efficient forearm torque transmission. In spite of that, grasping the handle obstructs the user's ADL by occupying his/her hand. Besides, removing the handle and tying the user's forearm to the exoskeleton link using a soft Velcro strap [23], [25] cannot prevent joint misalignment and in turn decreases the torque transmission efficiency.

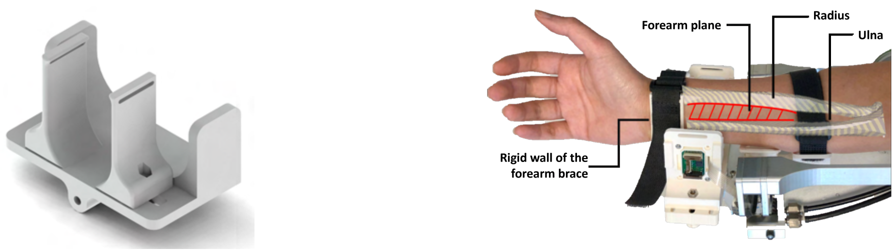
In our solution, a rigid U-shape brace, connecting to the UEFE forearm joint, with orthogonal alignment (Fig.2a) is designed to improve torque transmission by considering the forearm bone anatomy. Forearm rotation is determined by the forearm plane, the plane formed between the radius and ulna, relative to the elbow joint. As shown in Fig.2b, the two rigid walls in parallel to the forearm plane allow efficient torque transmission as rotating the two walls drives the forearm in-between to rotate. The width between the two walls is adjustable, so users with different widths can fit. A Velcro strap going through the two slots of the walls helps hold the forearm in place while avoiding wall bending during rotations.

C. Actuation and Transmission

The UEFE has the linear series elastic actuator (SEA) to actuate active modules to ensure a compliant and safe human-robot interaction. Many examples of upper-limb exoskeleton

(a) (b) (c)

Fig. 1: The Unilateral Elbow-Forearm Exoskeleton (UEFE). (a) A subject wearing the fully portable UEFE. (b) Elbow joint and Forearm joint design. (c) UEFE with the backpack.



(a) (b)

Fig. 2: The forearm brace for fixing and driving user's forearm. (a) The U-shape forearm brace. (b) Rigid walls in parallel to the forearm plane.

TABLE I: A Comparison of the range of motions (ROM) between the UEFE and activities of daily living (ADL) [28], [31]

Joints (Motion)	ROM of the UEFE	ROM of ADL
Elbow(Flexion/Extension)	156° (153°/3°)	156° (149°/7°)
Forearm(Pronation/Supination)	150° (75°/75°)	157° (90°/67°)

TABLE II: A Comparison of the torque/force between the UEFE and activities of daily living (ADL) [28], [31]

Motion	UEFE Max Torque	ADL Torque
Elbow(Flexion/Extension)	28.274 Nm	3.6/2.5 Nm
Forearm(Pronation/Supination)	33.929 Nm	0.047/0.06 Nm

using SEA can be found in [29]–[35]. The advantage of linear SEA over rotary SEA is to limit the range mechanically, ensuring a safe range of motion (ROM) for the exoskeleton's joint. The linear SEA design has been shown in our previous work [36], but different springs are used to achieve the torque requirement for ADL. Specifically, springs with 16.34 N/mm were applied for both eF/E and eR, achieving a bandwidth of 33.5 Hz. As a result, with the linear SEA, the UEFE can achieve the most ROM of ADL (Table I) and provide sufficient joint torque for ADL assistance (Table II).

As mentioned, the UEFE uses a cable-driven mechanism to transmit the force and reduce the exoskeleton weight. The lightweight exoskeleton is essential because it will not burden patients' arms significantly during ADL, so patients can freely mobilize their arm and use it for an extended period.

TABLE III: The Weight of the UEFE frame, estimated by Solidworks

Module	Weight (kg)
Elbow joint	0.635
Forearm rotation joint	0.26
Total	0.895

Furthermore, cable-driven mechanisms enable a lightweight exoskeleton design since the heavy actuators can be placed in other locations, like the backpack in Fig.1a or other movable platforms. Thus, both ambulant and wheelchair-seated stroke patients can use the device. The literature shows that a similar exoskeleton without cable-driven mechanisms, such as the Elbow-Wrist Exoskeleton (3 kg) [29], is heavier than the UEFE (0.895 kg), as shown in Table III. Hence, the cable-driven mechanisms have the advantage of reducing the exoskeleton weight.

III. Control of the UEFE

In this paper, the desired ADL trajectory was pre-set by the human demonstration method. To record the desired trajectory, an experimenter wore the UEFE to perform the ADL task, e.g., pill-taking task, under zero-force robot control. The recorded angle feedback over time were smoothened by a moving window filter and then used as the desired trajectory for that specific ADL motion.

Joint impedance control was implemented to track the desired ADL trajectory yet allow the user to freely adjust the

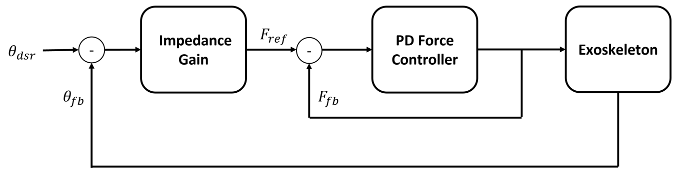


Fig. 3: Controller of the UEFE.

motion and complete the task safely. A PD force controller regulated the assistive forces at the lower level. The SEA force command (F_{cmd}) for exoskeleton actuation is calculated as follows:

$$F_{cmd}[\kappa] = K_p e[\kappa] + K_d \frac{e[\kappa] - e[\kappa-1]}{T_s} \quad (1)$$

$$e[\kappa] = F_{ref}[\kappa] - F_{fb}[\kappa] \quad (2)$$

$$F_{ref}[\kappa] = K(\theta_{dsr}[\kappa] - \theta_{fb}[\kappa]) \quad (3)$$

Here, κ represents the discrete time domain, K_p and K_d are the proportional and derivative gains of the PD force controller, respectively, and T_s is the time step in seconds. The error force, e, is the difference between the reference force (F_{ref}) and the feedback force (F_{fb}). The impedance gain is denoted by K, and the desired joint angle and angle feedback from the encoder are represented by θ_{dsr} and θ_{fb}, respectively. The control block diagram is shown in Fig.3.

IV. Experiments and Results

A. System hardware

A 22.2 Volt 3300mAh Lipo battery powers the UEFE system. An ARM microcontroller, STM32, is used to control the system with controller area network (CAN) communication protocol in general. As there are two active joints in the UEFE system, two SEAs, mentioned in [36], are used to actuate the exoskeleton; and two rotary encoders, analog absolute encoder RLS RM08 at the elbow joint; incremental encoder RLS RM08 with timing belt design at forearm joint, are installed for obtaining joint angle feedback.

B. Exoskeleton System Performance

1) Bandwidth: To secure the performance of the exoskeletal system, the system bandwidth should be large enough to cover the range of human arm achievable frequencies [28]. The UEFE system bandwidth is found by an experimental test at the elbow joint. A constant magnitude (100N) chirp signal, with linearly increasing frequency starting from 0.1Hz, was injected into the system as the force command. As seen in Fig.4, the bandwidth, determined at -3dB magnitude, is around 33.5Hz.

2) Force tracking performance: Repeating sine waves, with 100N magnitude, were injected into the system's elbow joint as the force command for force tracking assessment. Fig.5 shows the force tracking performance over time. The force tracking root-mean-square error (RMSE) at 0.2Hz (Fig.5a) and 2Hz (Fig.5b) were 2.19N and 9.48N, respectively.

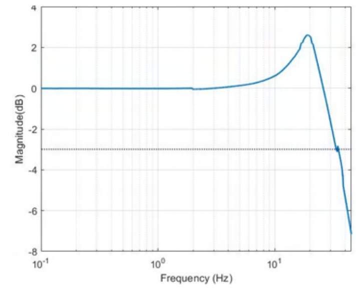


Fig. 4: The Bode Diagram of the UEFE.

3) Joint impedance control performance: Joint stiffness: To validate that the joint stiffness is acting accordingly to the input impedance gain (K), robot feedback data, including joint angles and SEA cable forces, was collected from a fixed angle (0°) tracking experiment. The experimental data were used to estimate the actual joint stiffness given the following relationships.

$$\tau = K\theta = Fd \quad (4)$$

, where τ is the joint torque; K is the impedance gain (stiffness); θ is the joint angle; F is the cable force estimated by the SEA; d is the joint radius.

In this experiment, two different impedance gains ($K = 0.16$ and $K = 0.06$) were chosen to investigate the difference between the experimental elbow joint stiffness and the impedance gain itself (K). During the experiment, the exoskeleton was commanded to track a fixed elbow joint angle while the experimenter kept flexing and extending the exoskeleton.

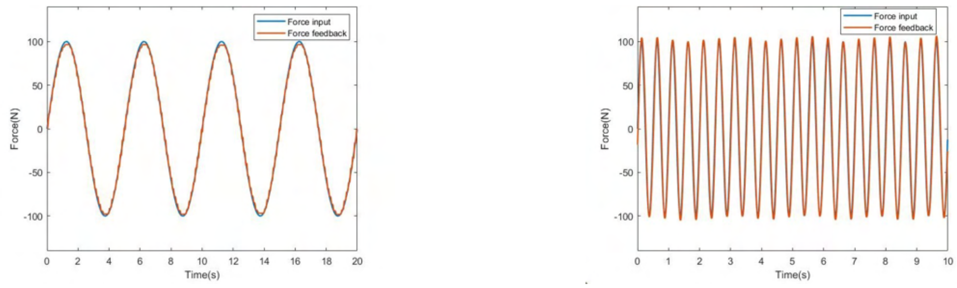
In Fig.6, linear fitting was done based on the experimental feedback represented in scattered dots. The slopes of the fitted lines is the experimental joint stiffness. When the impedance gain was $K = 0.16$, the experimental joint stiffness (K_{exp}) was around 0.13; When the impedance gain was $K = 0.06$, the K_{exp} was also around 0.06. As the stiffness error was less than 0.03 in both cases, the actual robot stiffness was in accordance with the input impedance gain.

Response to disturbance:

In ADL, there may be unpredicted disturbances during the exoskeleton operation. Impedance control allows the existence of disturbances and helps bring the robot back on track once the disturbances are removed. In the angle tracking experiment, a repeated sine wave with constant magnitude was injected into the impedance-controlled system as the desired angle trajectory at the elbow joint. In Fig.7, at around 30s and 60s, an experimenter deliberately disturbed the joint tracking by pushing or pulling the robot against the ongoing movements. Even though there were disturbances, the impedance controller could still react fast and keep tracking the desired trajectory.

C. Experiments on ADL tasks

Taking medication is an important ADL task for post-stroke patients. Picking up a pill from the table involves pronation,



(a) (b)

Fig. 5: Sine wave force tracking at different frequencies. (a) Force tracking at 0.2Hz sine wave. (b) Force tracking at 2Hz sine wave.

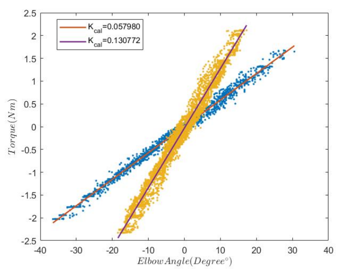


Fig. 6: The experimental joint stiffness.

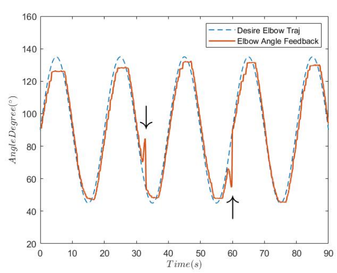


Fig. 7: The UEFE response to disturbances. The Arrows represent the time of input disturbances.

while putting the pill into the mount involves elbow flexion and supination. Fig.8a shows the ADL motion from rest to picking up the pill to bring the pill to the mouth with the exoskeleton.

In this experiment, the experimenter first donned on the exoskeleton and performed the medication-taking motion with zero-force control at normal speed. The exoskeleton feedback was recorded throughout the motion. The recorded encoder data was then smoothened by a moving window filter and resampled to become the desired trajectory for this ADL task.

To increase the task completion rate yet ensure safe interaction, impedance control was implemented to assist the user in performing the ADL task. The gain of impedance control should be tuned according to the user's condition and preference. This experiment examined the trajectory tracking performance of both elbow and forearm joints given different conditions and impedance gains ($K = \begin{bmatrix} K_{elbow} \\ K_{forearm} \end{bmatrix}$). Four conditions were tested in total: 1. Gain $K_1 = \begin{bmatrix} 0.16 \\ 0.25 \end{bmatrix}$ with the exoskeleton alone, 2. Gain K_1 with a subject wearing the exoskeleton and arm fully relaxed, 3. Gain K_2 (80% of K_1) with a subject wearing the exoskeleton and arm fully relaxed, 4. Gain K_3 (60% of K_1) with a subject wearing the exoskeleton and arm fully relaxed. In each condition, 15 trials were performed for analysis.

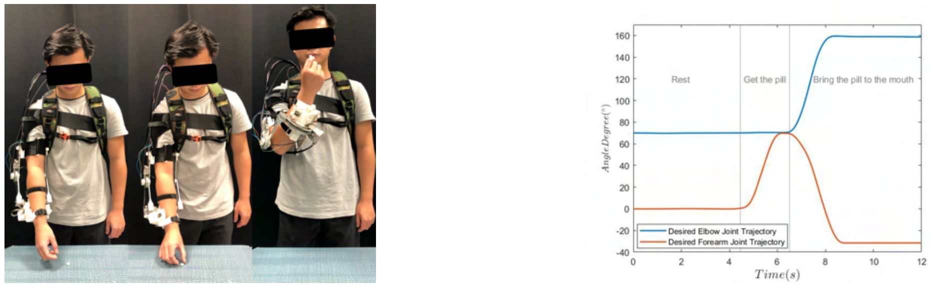
Fig.9 shows the mean and SD for the 15 trials in each condition. The RMSE of the elbow and forearm joint under the four conditions are shown in the box plots in Fig.10. The error increases with disturbance (the gravity of the human arm) and the decrease of impedance gain. Moreover, the mechanical friction, especially at the forearm rotation joint, brings along tracking imperfection. Regardless of the trivial joint friction, the impedance controller was able to assist a fully relaxed arm to follow the desired trajectory as long as the chosen impedance K was befitting.

V. Discussion

UEFE is a fully cable-driven, handle-free exoskeleton with active elbow and forearm rotation joints. Elbow flexion/extension and forearm pronation/supination are inevitable motions in ADL. In addition, restricted forearm motion causes unwanted compensatory motions that may harm post-stroke patients. Thus, the forearm rotation joint should be considered when designing an upper limb exoskeleton for ADL application.

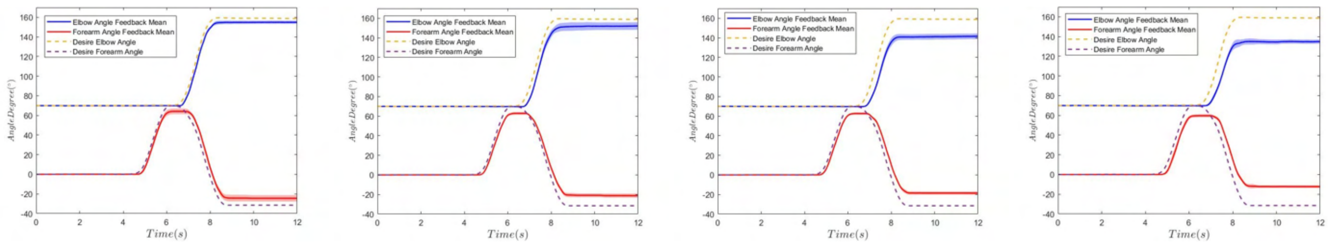
The UEFE is suitable for ADL use due to its lightweight and handle-free design. It also provides sufficient range of motion and assistive torque. Impedance control can be used to assist users in performing ADL tasks. Using different impedance gains give different tracking performance. The impedance gain can be tuned according to the user's condition and will of task engagement.

Regarding the control framework, recording trajectory through human demonstration for ADL assistance is not at always practical as there are numerous ADL tasks with unlimited motion variety. Pre-recording one or several fixed trajectories is not enough to cater to the diversity of ADL tasks. To accomplish various tasks in ADL, intention-sensing algorithms can be adopted to pick up users' intentions and



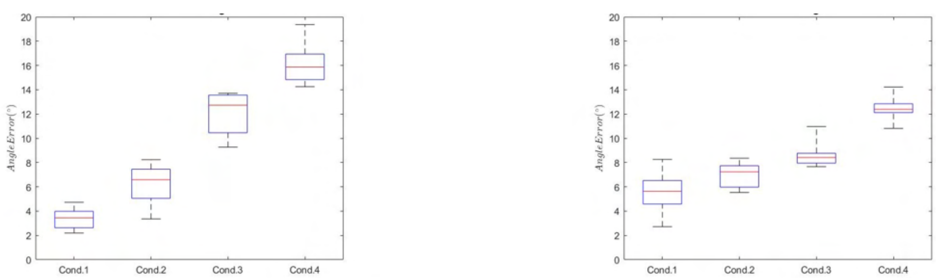
(a) (b)

Fig. 8: The ADL task of the experiment: Medication taking. (a) The ADL motion starts with 1) rest, then 2) get the pill, and 3) bring the pill to the mouth. (b) The elbow and forearm trajectory corresponding to the motions.



(a) (b) (c) (d)

Fig. 9: The tracking performance in different conditions with feedback mean and standard deviation of every 15 trials. (a) Cond1: K_1 with UEFE alone. (b) Cond2: K_1 with UEFE and fully relaxed arm. (c) Cond3: K_2 with UEFE and fully relaxed arm. (d) Cond4: K_3 with UEFE and fully relaxed hand



(a) (b)

Fig. 10: The RSME of joint tracking under different conditions. (a) The RMSE of Elbow joint tracking. (b) The RMSE of Forearm joint tracking.

then modify or generate desired trajectories online [37]–[39]. Thus, having an intention sensing and control framework may improve the ADL assistance capability.

In addition, due to the handle-free design, users are allowed to grasp ADL objects and manipulate tools. Considering that some post-stroke patients with poor grasping ability, an external hand or finger assistive module can be added to the front-most region, given the advantage of this handle-free design.

VI. Conclusions

This paper presents a novel UEFE that assists the user's elbow and forearm motion in ADL. The design rationale of UEFE is based on the ADL requirements. This lightweight UEFE achieves most of the range of motion in ADL tasks. And due to the rigidity of the robot structure, a forceful assistant can be offered to patients with different conditions and body mass. Keeping the design handle-free allows users to grasp objects and use tools. For those with hand or finger disabilities, it is optional to add on a hand assistive module as there is no hindrance caused by a handle.

Several experiments are done to show the control ability of the exoskeleton. The medication-taking experiment demonstrated the feasibility of ADL assistance using this robot.

In the future, we will focus on developing an intention sensing and control framework that assists post-stroke patients in various ADL motions instead of just one specific task.

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