Examination of Biofeedback to Support the Use of Upper-Extremity Exoskeletons Under Proportional Myoelectric Control

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***Abstract*—Exoskeletons have the potential to assist individuals in completing daily tasks and augment industrial workers in labor-intensive jobs. While previous studies have shown the capability of powered upper limb exoskeletons to reduce muscle effort and maintain task performance in continuous cyclical movements, their effectiveness in natural movements that contain both dynamic and static tasks remains uncertain. This study aimed to investigate the impact of visual and haptic electromyography (EMG) biofeedback on participants ($n = 36$) while they performed a target position matching task with a powered upper limb exoskeleton. Our hypothesis was that users could benefit from the biofeedback to minimize muscle effort and use the exoskeleton more effectively. However, the results indicated that the biofeedback did not reduce muscle effort in participants, but it had a positive impact on the smoothness of participants' extension movements. The challenge of reducing muscle effort appeared to stem from participants experiencing difficulty in relaxing their muscles, even when the exoskeleton provided support for the task or maintained the desired posture. Nevertheless, participant feedback supported that biofeedback might enhance their satisfaction with exoskeleton usage, which is a crucial factor in promoting long-term acceptance. These findings provide a foundation for future research in user training methods and controller development for exoskeletons.**

***Index Terms*—Powered upper limb exoskeleton, EMG biofeedback, human-exoskeleton interaction.**

I. Introduction

POWERED robotic exoskeletons have shown great potential in assisting elderly or disabled individuals with daily activities or augmenting healthy individuals in performing labor-intensive tasks [1], [2], [3], [4], [5], [6]. Both passive and active exoskeletons have been developed to achieve these goals. Lower-extremity exoskeletons have demonstrated effectiveness in reducing energy expenditure [7], [8], [9], increasing balance [10], [11], [12], [13], and accelerating walking speed [14], among other benefits. However, successful implementations of upper limb exoskeletons are primarily limited to passive exoskeletons in specific tasks, such as repetitive lifting or overhead working [15], [16], [17]. Despite their simplified designs, passive exoskeletons have some disadvantages. They are less adaptable than powered devices since they can only offer fixed mechanical properties, which are not adjustable during use [18], [19]. Passive exoskeletons are typically designed for specific tasks and movements, which may not be suitable for more complex tasks or activities requiring a greater range of motion and functionality. Additionally, passive exoskeletons lack built-in intelligence or sensing capabilities, which limit their ability to adapt to changing conditions or provide real-time assistance. In contrast, active exoskeletons offer greater adaptability and potential due to their capacity to provide a wide range of torque-time profiles and introduce net positive mechanical work [20]. Therefore, evaluating how powered upper limb exoskeletons are utilized by users in task-specific environments and understanding their limitations in such settings is vital for future development efforts.

Numerous studies have investigated the development of powered upper-extremity exoskeletons with the primary goal of designing a human-robot interface that can understand the user's intentions and provide required assistance promptly. One common approach to achieve this goal involves using electromyography (EMG). While EMG is not directly related to joint angle [21], it can be used to estimate the joint torque needed for the desired movement with a constant fraction of the torque provided through robotic assistance [22], [23], [24], [25], [26]. However, precise EMG-based torque estimation necessitates complex and time-consuming user- and session-specific calibrations that require additional technical equipment. A more straightforward approach is to use proportional myoelectric controllers where assistance is proportional to the muscle activation intensity. This method does not provide assistance as a constant fraction of the muscle force, but has proven to be effective in the rehabilitation field, aiding in movement restoration, motor control enhancement and muscle strength improvement [27], [28], [29]. Lenzi et al. [30] have shown that obtaining an accurate estimate of muscle torque may not be necessary, even as an assistive device, since the human central nervous system can compensate for the lower torque accuracy of the robot. They found that

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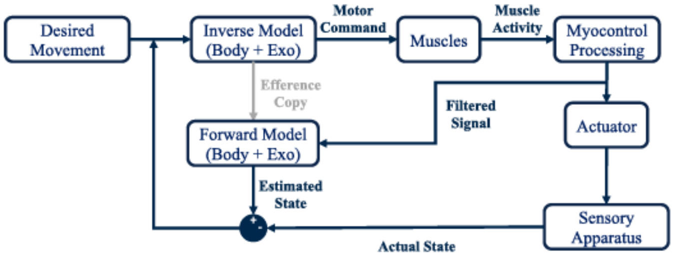


Fig. 1. The internal model includes both forward and inverse models containing the dynamics of the body and the exoskeleton. EMG biofeedback supplies the forward model with the actual filtered signal input to the exoskeleton actuator, enhancing the accuracy of state estimation.

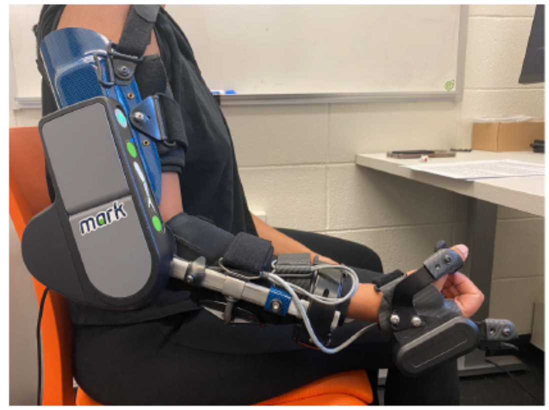


Fig. 2. The participant wore a Myomo exoskeleton on their right arm and performed the experiment while seated on an unarmed chair.

users could maintain movement accuracy while reducing effort using a simple proportional EMG controller. However, their experiment involved only continuous cyclical flexion/extension movement, which is less complex than natural movements that often comprise both dynamic and static tasks, such as holding positions. Although active devices generally provide more assistance compared to passive devices, they may be less preferred in static tasks [31]. Users might perceive the assistance to be too strong and impeding their movements if the same control strategy used for dynamic tasks is also employed for static tasks. Therefore, the effectiveness of active upper limb exoskeletons in complex scenarios should be examined.

The natural and precise movement of human biological limbs is attributed to the closed-loop control implemented in the sensorimotor system (Fig. 1), wherein a well-established internal model anticipates state changes based on knowledge of body dynamics and corrects for motor noise through feedback mechanisms [32], [33]. However, when wearable devices are introduced, a new internal model needs to be learned and stored [34]. Within this closed-loop control model, the motor command used by the forward model to predict the state deviates from actual conditions as the actuator input is a post-processed filtered signal, prone to time-dependent shifts and sensor placement errors. This misalignment undermines the reliability of the forward model, leading to the movement accuracy being more influenced by the corrective component that lags the actual movement. Biofeedback could potentially bridge this gap by providing filtered EMG signals to users to support the forward model estimation. Dosen et al. [35] reported that visual EMG biofeedback allowed myoelectric prosthesis users to control force with less variability and finer resolution. Prior research has also demonstrated that visual or haptic feedback can assist users in modulating ankle torque, walking speed, and gait profiles, leading to improved cooperation with exoskeletons [36], [37], [38]. Therefore, the incorporation of visual or haptic feedback could potentially enhance the use of powered upper limb exoskeletons in complex tasks.

In this study, we aimed to investigate whether visual and haptic EMG biofeedback could enhance user control of powered upper limb exoskeletons during tasks that involved transitory and hold periods. We hypothesized that participants receiving biofeedback would demonstrate reduced muscle effort while maintaining task performance compared to participants without feedback. Additionally, we collected participant perceptions to reflect their preferences regarding the integration of visual or haptic feedback during task performance. Examining how individuals use powered upper limb exoskeletons in complex movements and with EMG biofeedback will support the development of training methods and embedded controllers.

II. Methods

A. Participants

Forty-three participants consented to participate in the study, and thirty-six ($n = 36$; 18F, 18M; age: 23.3 ± 3.5 years; mass: 68.9 ± 16.1 kg; height: 1.69 ± 0.10 m, mean $\pm$ standard deviation) completed the study. All participants were healthy, had no limitations in arm mobility, and had not experienced any arm injuries in the past 6 months. All participants were right-handed and had no prior experience using upper-extremity exoskeletons. Prior to the study, written informed consent was obtained from all participants, approved by the University of Michigan Institutional Review Board (HUM00213716).

Seven participants withdrew from the study, with two reporting discomfort while wearing the exoskeleton and five experiencing technical issues with the EMG sensors during the session. Data from these incomplete cases were not included in the analysis, resulting in a final sample size of 36 participants.

B. Exoskeleton

In this study, we used a portable EMG-based exoskeleton designed for the right arm (Model: Mark, Myomo, Inc., Boston, MA) (Fig. 2). The exoskeleton weighed approximately 2 kg and was equipped with two motors, one for the hand and one for the elbow. For the purpose of this study, we only used the elbow motor. Prior to wearing the exoskeleton, participants were instructed to clean the skin by wiping it with alcohol to support sensor contact. The onboard electronics of the device amplified and processed the EMG signals in real-time. Non-invasive dry EMG sensors (bipolar electrodes with integrated ground reference) were placed on the biceps and triceps to record the corresponding muscle electrical signals.

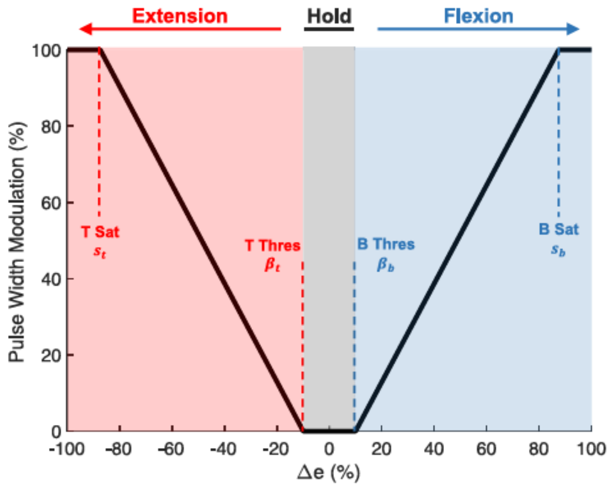


Fig. 3. The scheme of proportional myoelectric control. The Δe was proportionally mapped to the percentage of PWM between the threshold (β) and the saturation level (s).

The EMG signals were sampled at a rate of 1 kHz, followed by a smoothing using a 4th order bandpass filter from 100 to 200 Hz. Subsequently, the signals were rectified and processed using a default Kalman filter with $Q = 0.001$, $R = 20.0$. The filtered signals were then multiplied by biceps gain (k_b) and triceps gain (k_t), respectively. To suspend the exoskeleton on the arm, participants wore a shoulder harness, which partially offloaded the exoskeleton weight and protected the shoulder structure. Data from the exoskeleton were recorded on a laptop via a USB cable (data streaming rate: 20 Hz).

The exoskeleton featured four distinct control modes: *Standby*, *Biceps*, *Triceps*, and *Dual*:

- *Standby*: No motor activation occurred regardless of the status of the recorded EMG signals. However, the motor was backdrivable.
- *Biceps*: The motor provided assistance with flexion when the user contracted their biceps and with extension when the user relaxed their biceps.
- *Triceps*: The motor provided assistance with extension when the user contracted their triceps and with flexion when the user relaxed their triceps.
- *Dual*: The direction of movement was determined by which muscle had a stronger scaled contraction. The motor would hold if there was no difference in the scaled muscle contractions.

In single muscle modes (*Biceps* and *Triceps*), the motor speed remained constant, and the direction of movement was determined by the relative value of the EMG signal and a preset threshold. The exoskeleton could move to fully flexed or fully extended orientations but could not maintain intermediate positions. In the *Dual* mode, both the biceps and triceps were monitored, and EMG signals were converted to EMG effort (e) by dividing by the theoretical maximum EMG signal possible on the relevant channel:

$$\text{EMG Effort } (e) = \frac{\text{EMG Signal}}{\text{Max EMG}} (\%) \tag{1}$$

The EMG effort represented the muscular exertion as a percentage of its theoretical maximum and would be truncated to 100 if it exceeded the maximum value. The Max EMG was set to 75 for both muscles (default manufacturer setting). The difference between the EMG efforts of the biceps and triceps muscles (Δe) was directly proportional to the percentage of PWM (Pulse Width Modulation) (Fig. 3), which, in turn, was proportional to the elbow motor speed ($\dot{\theta}$) under no loading condition. A saturation threshold (s) was applied such that a maximal control signal was generated to achieve maximum motor speed once Δe exceeded the saturation level. The maximum speed of the elbow motor under no load was approximately 165°/s. To maintain the current position, Δe should remain within the deadzone. Within this range, the motor speed $\dot{\theta}$ remained constant at 0, with a PID controller active to maintain the position. Between the biceps/triceps threshold (β) and the saturation level, the control signal was directly proportional to Δe. In this study, $\beta_b = 10\%$, $\beta_t = -10\%$, $s_b = 84\%$, $s_t = -84\%$ as determined during pilot testing.

C. EMG Biofeedback

EMG biofeedback was provided to targeted groups of participants through a visual and haptic display.

1) Visual Feedback: The participants were provided with visual feedback in real-time, which displayed the difference between the EMG efforts (Δe) (Fig. 4a). The desired direction of Δe was highlighted, enabling participants to adjust their muscle activation during the task. For instance, if the task required elbow extension, participants should generate a Δe that was below the triceps threshold. Utilizing real-time and desired Δe information, participants might enhance their control of the device and complete the task with increased accuracy and effectiveness.

2) Haptic Feedback: Haptic feedback conveyed real-time Δe information by vibrating the corresponding haptic motor at 230 Hz (Fig. 4b). For example, the biceps haptic motor would vibrate if the biceps signal was much larger, and the exoskeleton was going to flex. No vibration would occur if Δe was within the deadzone. The haptic motors (Model No.307-103, Precision Microdrives Ltd., London, U.K.) were attached to the forearm and wire-connected to a Raspberry Pi (Model: 4B, Cambridge, U.K.). The Raspberry Pi received data from the laptop at a rate of 20 Hz. To minimize any effect of vibration on the EMG data, the haptic motors were positioned on the forearm while the EMG sensors were placed on the upper arm.

D. Experimental Protocol

1) Familiarization: Participants were given a simplified explanation of how the exoskeleton works and were instructed on how to wear the device. The experimenter checked the position of the EMG sensors and ensured that the participant's elbow was aligned with the motor axis, and the harness saddle was in the middle of the shoulder.

Participants started by practicing in *Biceps* and *Triceps* modes, flexing and extending their arms several times to learn how to control the device by contracting and relaxing the corresponding muscles. Then they proceeded to the *Dual*

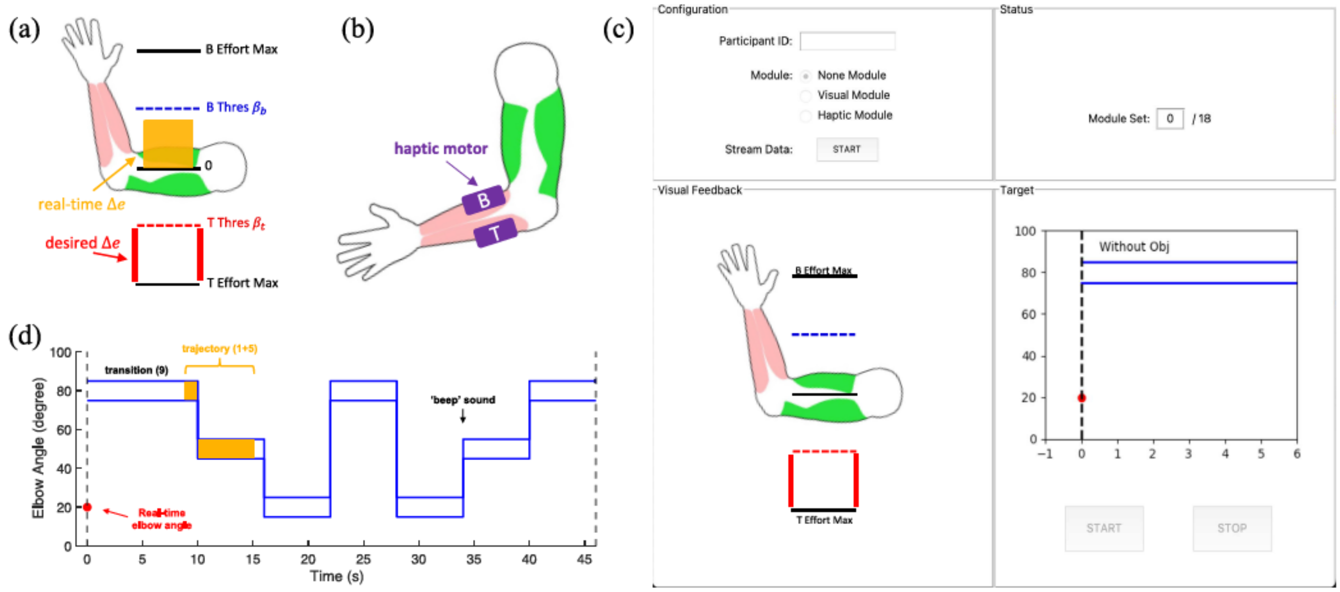


Fig. 4. (a) Visual feedback displayed real-time differences between EMG efforts (Δe) alongside the desired Δe, with bar height represented in a logarithmic scale. The Δe bar utilized distinct colors for clarity (above β_b - blue; between β_b and β_t - orange; below β_t - red). (b) Haptic feedback conveyed the Δe through vibrations in corresponding haptic motors, thus indicating the status of the exoskeleton. (c) The interface was displayed on the screen to all participants. Visual feedback from (a) was presented in the lower left section for participants in the *Visual* group, while it remained blank for participants in other groups. The Target section displayed the task as depicted in (d). (d) In the target position matching task, participants were instructed to match their elbow angle (red dot) to the target position (blue region).

mode and practiced the cyclic movement of the elbow under nominal settings ($k_b = k_t = 7$). Participants were asked if they experienced any difficulties initiating movement and which direction (flexion or extension) was more challenging. The experimenter adjusted the control parameters using the following rule: if extension was more challenging, increased k_t or decreased k_b, whereas if flexion was more challenging, increased k_b or decreased k_t. Participants repeated the cyclic elbow movements to assess if the current setting was satisfactory. This loop continued until the parameter adjustment had no effect, and participants were satisfied. This process typically took 3-5 loops and lasted approximately 5-10 minutes.

Finally, participants were trained on how to relax their muscles at different elbow angles to let the exoskeleton maintain the position. The experimenter guided the participants to reach a fully relaxed state at each elbow angle by holding their forearms and gradually releasing their hands, reminding them not to exert any force. If a participant was unable to relax their muscles at a certain angle, the process was repeated until they could do so at least one time voluntarily.

2) Group Assignment: The participants were randomly assigned to one of three groups: *None*, *Visual*, or *Haptic*. Each group consisted of 12 participants (6F, 6M). All participants were provided with information about the task details, as described in Section II-D3. Participants in the *Visual* and *Haptic* groups received instructions on how the respective feedback modalities worked (Section II-C), but did not undergo any biofeedback practice before the actual task.

3) Target Position Matching Task: The *Dual* mode was used in this task. Participants were instructed to perform a 4-module target position matching task, where they had to match their elbow angle to a target displayed on the screen (Fig. 4c). A red dot represented their real-time elbow angle, and they were instructed to keep it within the range of two blue lines (±5 degrees). Participants were directed to move to a new position after the vertical blue line, which was accompanied with a beep sound indicating the start of the movement. Each module contained 18 sets, half of which were hand-free ('Without Obj' sets), and the other half with a 2 lb dumbbell ('With Obj' sets). Each set contained six unique trajectories generated from three pre-selected elbow angles (20°, 50°, 80°), starting from each one to the other two. Each trajectory consisted of 1 second of initial position and 5 seconds of final position, allowing enough time to initiate the movement, reach the target, and hold the position. While the sequence of trajectories within each set was randomized, it was essential that the final position of one trajectory corresponded to the initial position of the subsequent one. An additional 9 seconds were given at the beginning of each set for the transition between the 'Without Obj' and 'With Obj' sets. A sample set was shown in Fig. 4d.

The participants were asked to:

- Track the target as quickly and accurately as possible.
- Minimize muscle efforts throughout the experiment.

The participants in each group completed 4 modules based on their assigned group. The task was identical in all modules. All participants had no feedback in their first and third modules ($M1$, $M3$). In the second and fourth modules ($M2$, $M4$), participants received the feedback for their assigned group. Each module lasted approximately 15 minutes with 5-minute intervals between modules. The sequence of modules completed by participants in each group was as follows, with N, V, and H corresponding to No feedback, Visual feedback, and Haptic feedback, respectively:

- *None*: $M1$(N) - $M2$(N) - $M3$(N) - $M4$(N)
- *Visual*: $M1$(N) - $M2$(V) - $M3$(N) - $M4$(V)
- *Haptic*: $M1$(N) - $M2$(H) - $M3$(N) - $M4$(H)

4) Post-Study Survey: At the end of the study, a survey (Appendix) was administered to each participant to obtain their perceptions on using the exoskeleton and biofeedback (when present).

E. Data Analysis

1) Electromyography: The collected data (*i.e.,* EMGs, EMG efforts, elbow motor angle) were split into sets. The initial 9 seconds of each set, which represented the transition time, were excluded from analysis. The integrated EMG (iEMG) was calculated for each set and separated based on whether participants were holding objects or not, as the additional weight could affect the EMG values. Therefore, each module had 9 sets for each condition, and the mean iEMG of each module was calculated. The data were further split into trajectories, and for each participant and module, the average EMG profile was calculated for each trajectory.

To assess participants' utilization of the exoskeleton during the hold periods, we measured the mean EMG values during the final second of each trajectory's hold period, referred to as hold EMG. In an ideal scenario, users should be able to rely entirely on the exoskeleton to maintain a certain position without engaging their muscles for support. In such conditions, the hold EMG value should approach zero, indicating optimal exoskeleton utilization. The hold EMG value, when not close to zero, offers valuable insights into the extent to which users achieved the intended exoskeleton usage.

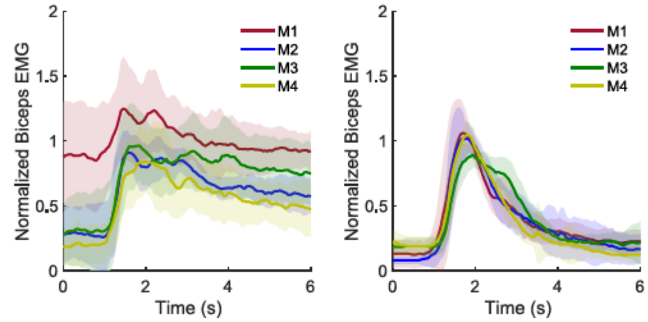
All the EMG data were normalized for each participant and muscle separately using the average peak value of the largest trajectories in the 'Without Obj' sets in $M1$ (*i.e.,* from 20° to 80° for biceps and from 80° to 20° for triceps). Those trajectories were chosen as they required the greatest muscle activation. The iEMGs were further normalized by the average iEMG value of all 'Without Obj' sets in $M1$ so that all participants started with iEMG $= 1$ in the first module.

2) Task Performance: Task performance was evaluated using two measures: movement smoothness and task accuracy.

For movement smoothness, we implemented spectral arc length (SPARC) [39] due to its validity, sensitivity, reliability and practicality. SPARC quantified the negative arc length of the amplitude and frequency-normalized Fourier magnitude spectrum of the speed profile. The elbow speed was obtained by smoothing and differentiating each trajectory.

Task accuracy was determined as the percentage of time participants successfully matched the target throughout the experiment (the red dot located within the blue region).

3) Survey: Survey questions with discrete scales were analyzed by summarizing the number of counts for each response and calculating the percentage of participants who provided that response. For open-ended questions, we coded and categorized the provided comments.



(a) Participant 12 (b) Participant 39

Fig. 5. The biceps EMG profiles of two sample participants while holding objects and moving from 20° to 80°, with the mean (solid line) and standard deviation (shaded region) averaged within modules.

F. Statistical Analysis

The collected data were analyzed using MATLAB (MathWorks, Natick, MA, USA). Four separate ANOVA models were fit, one for each metric. The hold EMG values were log-transformed to meet the assumptions of normality and homogeneity for the ANOVA residuals. The models included:

- *iEMG Analysis:* A four-way mixed ANOVA with three within-subjects factors (module, muscle, condition) and a between-subjects factor (group).
- *Hold EMG Analysis:* A five-way mixed ANOVA with four within-subjects factors (elbow angle, module, muscle, condition) and a between-subjects factor (group).
- *SPARC Analysis:* A four-way mixed ANOVA with three within-subjects factors (trajectory, module, condition) and a between-subjects factor (group).
- *Task Accuracy Analysis:* A three-way mixed ANOVA with two within-subjects factors (module, condition) and a between-subjects factor (group).

These ANOVA models considered various factors, such as module ($M1$, $M2$, $M3$, $M4$), trajectory (20° $\rightarrow$ 50°, 20° $\rightarrow$ 80°, 50° $\rightarrow$ 20°, 50° $\rightarrow$ 80°, 80° $\rightarrow$ 20°, 80° $\rightarrow$ 50°), elbow angle (20°, 50°, 80°), muscle (biceps, triceps), condition (Without Obj, With Obj), and group (*None*, *Visual*, *Haptic*).

Post-hoc pairwise comparisons were conducted, and 95% confidence intervals (CI_d: 95% confidence interval on the difference in means) were calculated with false discovery rate (FDR) correction (corrected significance level (α): iEMG: 0.021; hold EMG: 0.031; SPARC: 0.039; task accuracy: 0.018). Cohen's d effect size [40] was calculated for each pairwise comparison: small effect ($0.2 < |d| < 0.5$), medium effect ($0.5 < |d| < 0.8$), and large effect ($|d| > 0.8$).

III. Results

A. Electromyography

Participant had a significant effect on all metrics (Table I), indicating that individuals exhibited distinct usage patterns when using the exoskeleton to perform the task. Fig. 5 displays the biceps EMG profiles of two sample participants in the 20° $\rightarrow$ 80° trajectory. Participant 12 initially struggled to

TABLE I
MULTI-FACTOR ANOVA RESULTS FOR IEMG, HOLD EMG, MOVEMENT SMOOTHNESS, AND TASK ACCURACY

Source	iEMG		Hold EMG (log)		SPARC		Task Accuracy	
	F	p	F	p	F	p	F	p
Module	61.0	<0.001	67.1	<0.001	51.0	<0.001	44.7	<0.001
Trajectory	–	–	–	–	360.7	<0.001	–	–
Elbow Angle	–	–	98.5	<0.001	–	–	–	–
Muscle	110.5	<0.001	1069.0	<0.001	–	–	–	–
Condition	458.4	<0.001	160.0	<0.001	4.8	0.029	20.5	<0.001
Group	1.0	0.380	0.3	0.760	0.4	0.650	0.9	0.415
Participant (Group)	9.7	<0.001	30.8	<0.001	14.6	<0.001	29.7	<0.001
Module × Trajectory	–	–	–	–	3.0	<0.001	–	–
Module × Elbow Angle	–	–	8.3	<0.001	–	–	–	–
Module × Muscle	1.0	0.390	3.5	0.015	–	–	–	–
Module × Condition	1.9	0.123	0.3	0.797	4.4	0.005	0.2	0.929
Module × Group	2.4	0.028	2.7	0.014	2.0	0.059	3.4	0.003
Trajectory × Condition	–	–	–	–	11.5	<0.001	–	–
Trajectory × Group	–	–	–	–	7.9	<0.001	–	–
Elbow Angle × Muscle	–	–	337.6	<0.001	–	–	–	–
Elbow Angle × Condition	–	–	12.3	<0.001	–	–	–	–
Elbow Angle × Group	–	–	5.1	<0.001	–	–	–	–
Muscle × Condition	156.7	<0.001	51.2	<0.001	–	–	–	–
Muscle × Group	3.6	0.030	15.7	<0.001	–	–	–	–
Condition × Group	0.8	0.446	0.7	0.499	4.4	0.013	0.8	0.473
Elbow Angle × Muscle × Condition	–	–	7.0	0.001	–	–	–	–
Elbow Angle × Muscle × Group	–	–	3.6	0.007	–	–	–	–

* The table displays all main effects and significant interaction effects observed on at least one metric. Interaction effects that do not appear in the table were found to have no significant impact on any metric. For metrics where certain effects are not applicable, they are denoted by −.

use the exoskeleton at $M1$, exhibiting a high degree of variation and biceps contraction. However, as they became more familiar with the device and the task, their behavior became more consistent and their muscle contraction decreased. In contrast, participant 39 quickly adapted to the exoskeleton, displaying highly consistent biceps patterns across all four modules. These two sample participants represent two categories: individuals who can quickly adapt to exoskeletons and those who may require more practice and assistance. However, it should be noted that individuals may fall between these two categories, without a clear boundary delineating them.

A significant interaction effect between Muscle - Condition was observed on iEMG (Table I). Holding a weight significantly increased the iEMGs for both biceps (CI_d: $[0.382, 0.443]$, $|d| = 2.616$) and triceps (CI_d: $[0.086, 0.130]$, $|d| = 0.946$). Significant interaction effects were also observed between Module - Group. IEMGs significantly decreased from $M1$ to $M2$ (CI_d: $[-0.230, -0.150]$, $|d| = 0.914$) and remained relatively stable from $M2$ to $M4$ ($0.070 < |d| < 0.163$). Different groups exhibited similar trends with Module with no significant differences observed across groups for each module ($0.008 < |d| < 0.370$).

Hold EMG demonstrated a significant three-way interaction among Elbow Angle - Muscle - Condition (Table I). Specifically, hold EMG for the biceps increased as the arm flexed under each condition, showing large significant differences across 20°, 50°, and 80° ($0.850 < |d| < 1.543$) (Fig. 7(a)). Conversely, for the triceps, hold EMG stayed highest at 20° and decreased to lower values at 50° and 80° under each condition, with small to medium significant differences observed between 20° to 50°, and 20° to 80° ($0.383 < |d| < 0.593$). Additionally, weight significantly increased hold EMG for the biceps at all elbow angles ($0.500 < |d| < 1.767$), with small to medium effects on

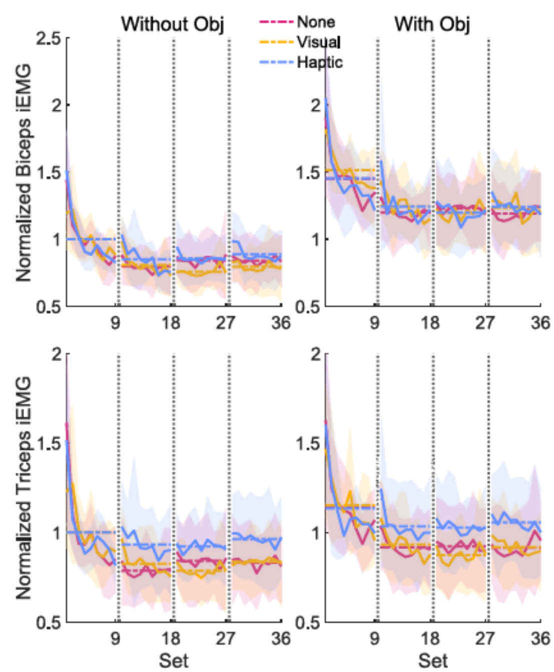
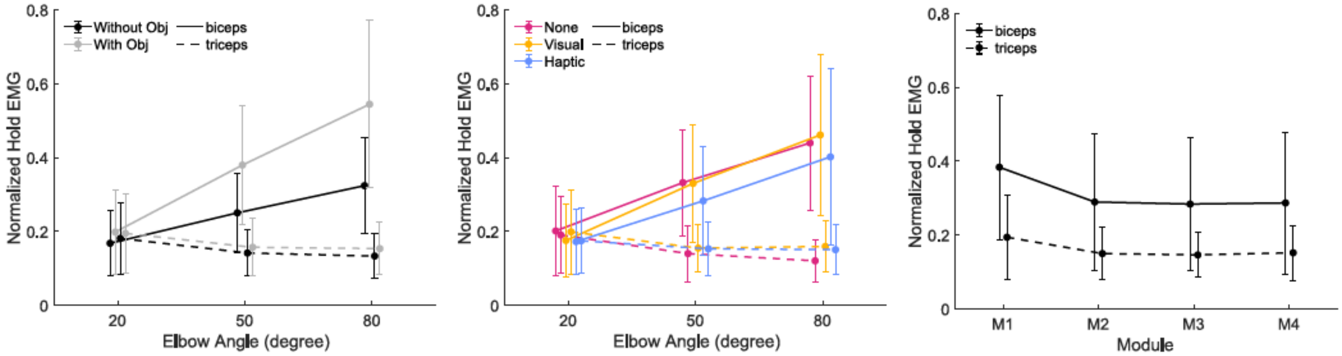


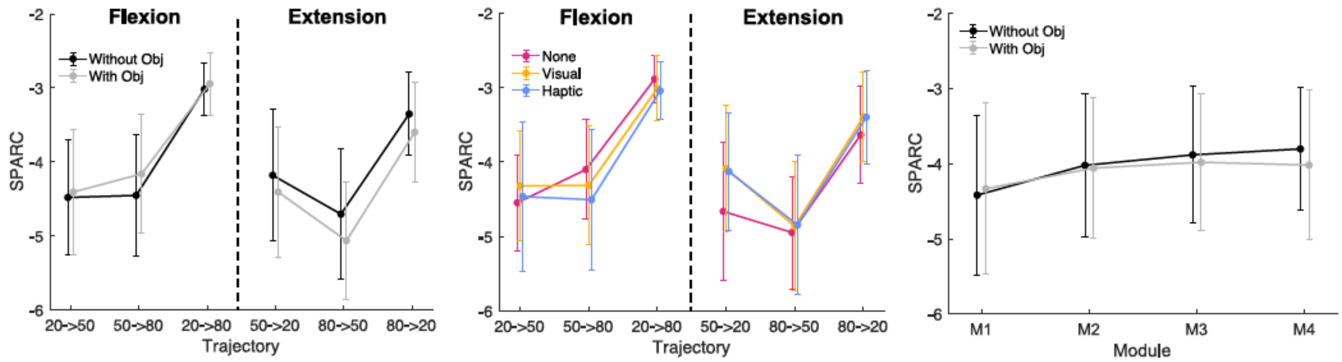
Fig. 6. Normalized iEMGs of biceps and triceps for each group. Mean values (solid line) are averaged across participants, with shaded regions representing standard deviation. Horizontal lines mark average values for each module.

the triceps ($0.212 < |d| < 0.654$). Significant interaction effects were also observed among Elbow Angle - Muscle - Group (Fig. 7(b)), but the different groups exhibited similar trends. Significant differences were only observed between the *None* and *Haptic* groups in the biceps at 50° (CI_d: $[0.004, 0.095]$, $|d| = 0.339$), and between the *None* group and the other two groups in the triceps at 80° ($0.485 < |d| < 0.609$). Additionally, a significant interaction was



(a) Elbow Angle - Muscle - Condition (b) Elbow Angle - Muscle - Group (c) Module - Muscle

Fig. 7. Significant interaction effects on Hold EMG. (a) Three-way interaction: Elbow Angle - Muscle - Condition; (b) Three-way interaction: Elbow Angle - Muscle - Group; (c) Two-way interaction: Module - Muscle. The ideal usage of exoskeletons should have hold EMG = 0.



(a) Trajectory - Condition (b) Trajectory - Group (c) Module - Condition

Fig. 8. Significant interaction effects on movement smoothness. (a) Two-way interaction: Trajectory - Condition; (b) Two-way interaction: Trajectory - Group; (c) Two-way interaction: Module - Condition. Less negative SPARC values indicate smoother movements.

observed between Module - Muscle (Fig. 7(c)), where hold EMG had a large decrease from $M1$ to $M2$ for the biceps (CI_d: $[0.078, 0.110]$, $|d| = 0.853$) and a small decrease for the triceps (CI_d: $[0.029, 0.060$, $|d| = 0.424]$) and then both were not significantly different after $M2$ ($0.046 < |d| < 0.118$).

B. Task Performance

The ANOVA results supported significant interactions between Trajectory - Condition and Trajectory - Group for SPARC (Table I). Our observations indicated that smoother movements were associated with larger distances covered (Fig. 8(a)). The $20° \rightarrow 80°$ and $80° \rightarrow 20°$ trajectories exhibited the least negative SPARC values compared to the other trajectories ($0.609 < |d| < 2.554$). The extra weight contributed to a small improvement in flexion in the $50° \rightarrow 80°$ trajectory (CI_d: $[0.132, 0.451]$, $|d| = 0.318$), while it resulted in small decreases in smoothness across all extension trajectories ($0.254 < |d| < 0.436$). In extension movements, integrating biofeedback resulted in less negative SPARC scores (Fig. 8(b)), with a medium effect size noted in the $50° \rightarrow 20°$ trajectory ($0.625 < |d| < 0.646$) and a small effect size in the $80° \rightarrow 20°$ trajectory ($0.372 < |d| < 0.387$). Flexion movements presented mixed outcomes, with no or small differences when biofeedback was introduced across trajectories ($0.099 < |d| < 0.496$). A significant interaction was also noted between Module - Condition (Fig. 8(c)). SPARC scores became less negative from $M1$ to $M2$ ($0.297 < |d| < 0.439$) and remained stable from $M2$ to $M4$ ($0.054 < |d| < 0.177$).

For task accuracy, there was a significant interaction effect observed between Module - Group (Table I). Participants from both the *None* and *Haptic* groups exhibited large improvement in task accuracy from $M1$ to $M4$ (*None*: CI_d: $[0.045, 0.085]$, $|d| = 1.695$; *Haptic*: CI_d: $[0.040, 0.083]$, $|d| = 1.483$). However, participants from the *Visual* group did not demonstrate this enhancement (CI_d: $[-0.008, 0.049]$, $|d| = 0.372$). Task accuracy was significantly higher in the *Haptic* group compared to the *Visual* group in $M4$ (CI_d: $[0.001, 0.083]$, $|d| = 0.731$), with no significant group differences observed in other modules ($0.015 < |d| < 0.516$). The average task accuracy increased from 70.9% in $M1$ to 75.8% in $M4$. Additionally, holding an object during the task significantly reduced task accuracy compared to performing the task without any objects (CI_d: $[-0.020, -0.009]$, $|d| = 0.548$).

C. Survey

The results of the study showed that a higher percentage of participants in the *Haptic* group (58.3%) perceived the exoskeleton as an aid in achieving and maintaining a target position, compared to the *Visual* and *None* groups

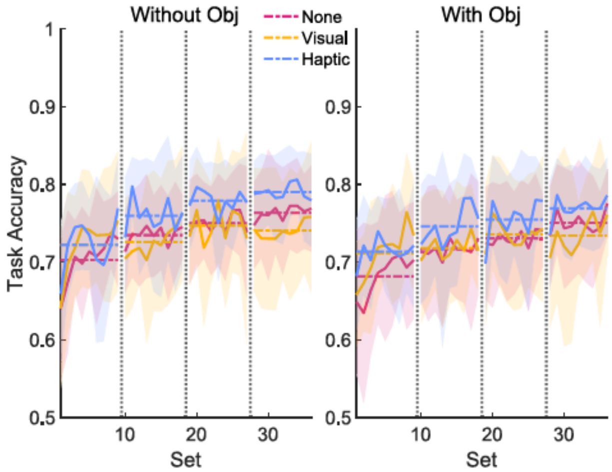


Fig. 9. Task accuracy for each group presented as mean values (solid line) averaged across participants, with shaded regions representing standard deviation. Horizontal lines denote average values for each module.

(8.3% and 16.7%, respectively) (Fig. 10(a)). Between the two biofeedback groups, more participants in the *Visual* group (91.7%) perceived a positive or very positive effect of the feedback on their target matching task performance, compared to the *Haptic* group (83.3%) (Fig. 10(b)). Negative effects of the feedback were reported by only one participant in each feedback group. The majority of participants in the *Haptic* group (75%) believed that they *often* understood the information provided by the haptic feedback, while 8.3% *always* and 16.7% *sometimes* understood it (Fig. 10(c)).

Regarding the use of feedback information (Question 4 and 7), 83.3% of participants in the *Visual* group and 50% of the *Haptic* group reported primarily using the feedback during the hold periods, where their attention was not as necessary to achieve the task and they had sufficient time to interpret the provided feedback. In addition, 8.3% of participants in the *Visual* group and 41.7% of participants in the *Haptic* group reported using the feedback to predict the movement of the exoskeleton and improve the precision of the matching task. Only one participant (8.3%) in each feedback group chose not to use the feedback information, as they considered it a distraction and did not pay much attention to it.

Participants provided suggestions for enhancing the use of the exoskeleton in open comments ($n = 36$). These suggestions included reducing resistance in the extension direction ($n = 10$), designing a lighter and more balanced distribution of weight for unilateral use (participants reported fatigue on their right shoulder due to the harness offloading exoskeleton weight on that shoulder) ($n = 8$), improving the personalized fit of the exoskeleton ($n = 6$), reducing response delay to movement ($n = 4$), improving fine set point control (participants reported that they could go to general positions, but getting to a more specific position was difficult) ($n = 3$), and reducing usage time to alleviate fatigue ($n = 2$).

IV. Discussion

In this study, we aimed to explore the potential benefits of incorporating biofeedback into the use of EMG-based upper limb exoskeletons. We evaluated participant performance in a target position matching task to assess two key aspects: (1) the exoskeleton's ability to reduce the required effort for participants to achieve and maintain a specific position, as measured by *Electromyography*, and (2) the smoothness and accuracy of movement, as measured by *Task Performance*. We also gathered participant feedback through a survey to gain insights into their perceptions.

A. Electromyography

Integrated EMG (iEMG) is a performance metric for assessing muscular effort [41]. We observed a rapid decrease in iEMG within $M1$ for all participant groups, indicating a quick learning process consistent with the exponential model commonly observed in motor learning [42], [43], [44], [45]. The iEMG remained stable after $M2$, suggesting that participants had adapted to the exoskeletons in terms of muscular effort. Similar trends were observed for hold EMG.

Contrary to our hypothesis, biofeedback did not result in a decrease in muscle activation when using the exoskeleton compared to the condition without feedback. No significant differences in iEMG were observed across participant groups. Most participants in the feedback groups reported using the feedback information primarily to help and check muscle relaxation during the hold periods. As instructed, participants exhibited a primary focus on tracking the target during movement, with minimizing muscle effort considered a secondary priority. Participants then directed their attention towards relaxing their muscles after completing the movement when they had sufficient time to interpret the provided feedback information. During this phase, the time constraints associated with matching the target were less demanding, enabling participants to allocate attention to the secondary priority. However, despite the increased attention during the hold periods, no differences were observed in terms of hold EMG across groups. Participants continued to exert muscle effort during the hold periods, despite being aware of their muscle status and the necessity to decrease muscle activation.

There are several potential explanations for the difficulty participants had in decreasing muscle effort during the hold periods with biofeedback. First, there was a discrepancy between what participants perceived visually or haptically and what they perceived proprioceptively. A few participants reported feeling as though they were relaxing, yet the feedback suggested otherwise. This contradiction could have potentially impacted the participant's situation awareness. Endsley's three levels of situation awareness encompass perception of the environment, comprehension of the current situation, and projection of future status [46]. Biofeedback can support level 1 perception, but achieving level 2 comprehension requires an understanding of the signals within the context of relevant goals. The additional input from biofeedback, when considered alongside their proprioceptive cues, may have led to a breakdown in their comprehension. This breakdown could have limited their ability to appropriately plan their actions and determine how to execute them, which corresponds to level 3 projection and relies on the presence of level 2 comprehension.

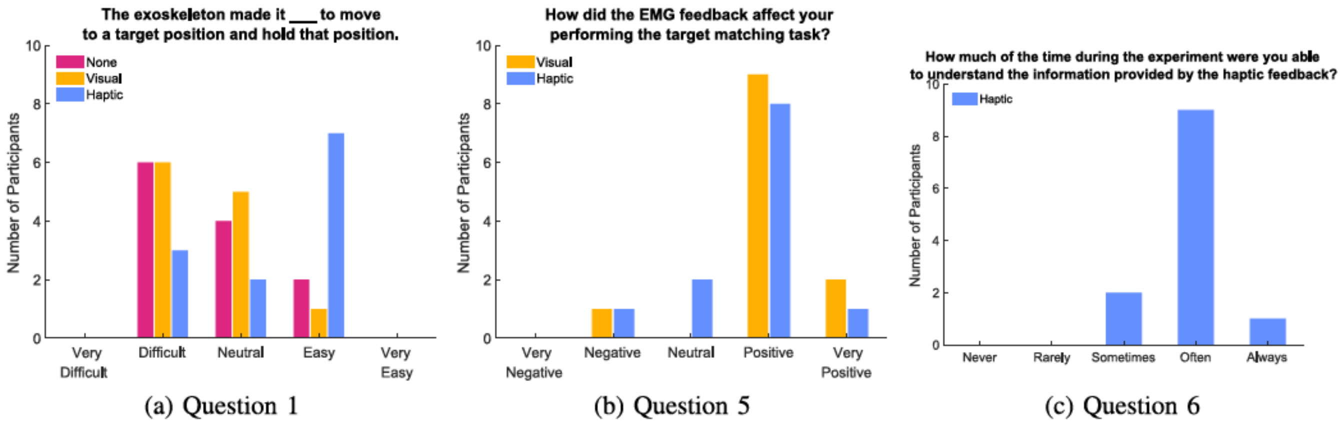


Fig. 10. Distribution of responses to three Likert scale questions. (a) Participants in the *Visual* and *None* groups reported that the exoskeleton hindered their movement, while participants in the *Haptic* group reported that the exoskeleton was helpful. (b) Both *Visual* and *Haptic* groups had an overall positive attitude towards the effect of EMG biofeedback. (c) The majority of participants in the *Haptic* group reported that they could often understand the information provided by the haptic feedback. Note that Q5 was specifically for biofeedback groups and Q6 was only for the *Haptic* group.

Training to understand the causes of discrepancies between proprioceptive feedback and biofeedback could support the interpretation of biofeedback and its more effective utilization.

Second, there was a discrepancy between participants' intentions and capabilities. Some participants attempted to relax their muscles when observing high EMG signals, but they found it challenging to achieve the desired relaxation. This discrepancy could be related to the failure in forming the intended internal model while interacting with the exoskeleton (Fig. 1). It is worth noting that a new internal model did show signs of formation and storage, as participants gradually adapted to the initial use of the exoskeleton during $M1$. However, further progress was not achieved, which likely indicates a fundamental disparity between the biological motor program (the way individuals coordinate movements without the exoskeleton) and the exoskeleton motor program (how people coordinate movements with the exoskeleton). For example, normally individuals must contract their biceps to maintain their arms at 80° against gravity, but this contraction is unnecessary when utilizing the exoskeleton as designed. The motor program adopted when holding a biological limb without the exoskeleton involves increasing hold EMG for the biceps as elbow angle increases. On the other hand, when using the exoskeleton, there does not need to be an increase in the participant hold EMG as the elbow angle increases. However, this expected outcome was not observed among most participants. Despite their awareness of the need to decrease muscle activation, they struggled to exert conscious control over this process. Moreover, holding an object could intensify this challenge, as it requires biceps relaxation while the forearm remains contracted. The inherent muscle synergy between the upper and lower arms could complicate this desired muscle decoupling.

Third, individuals exhibited different muscle activation patterns during the adaptation process. While most participants demonstrated a gradual increase in their familiarity with using the device (sample participant 12 in Fig. 5(a)), a few participants exhibited a high level of proficiency from the outset, showing little progress thereafter (sample participant 39 in Fig. 5(b)). This variation between individuals has been noted in other studies, such as those involving ankle exoskeletons [47], and could mask any potential differences attributable to the biofeedback. To control for this participant variability in future studies, a more accurate prediction of the participant exoskeleton proficiency level is needed to improve randomization for between group studies. Although there have been anecdotal observations suggesting that exoskeleton proficiency is not related to anthropometric parameters or athletic experience, there have not been studies specifically assessing these predictors. Initial pilot testing indicates that cognitive factors, such as reaction time, may serve as indicators of exoskeleton usage and could be useful in future studies [48].

Finally, due to the deformability of human skin, some displacement between exoskeletons and biological limbs is inevitable, particular when the fitting between users and wearable devices is not ideal. This displacement can lead to a perception of a slight drop in the exoskeletons when users attempted to rely on the exoskeleton to hold the position, but the actual elbow angle remains unchanged. Nevertheless, users may experience a decrease in their trust in the exoskeleton's ability to maintain a position, resulting in a lack of confidence in fully relaxing their muscles.

The development of exoskeleton motor programs typically requires learning and training [34], [42]. Although participants practiced relaxing their muscles at different elbow angles, they still encountered difficulty relaxing their muscles during the target-matching task. Moreover, the act of holding additional weight significantly increased the hold EMG, indicating that the skills acquired during training did not readily transfer to different conditions. To attain the desired outcomes, it may be necessary to introduce more comprehensive training regimen involving a wider range of movement tasks, possibly over an extended duration. For example, participants may need to practice reaching and holding, which demands both muscle engagement and disengagement. Extending the training duration could also be advantageous, as the experiment allowed only 5 seconds for the hold periods, which might have constrained participants' ability to relax their muscles effectively at flexed elbow angles within the given time frame.

B. Task Performance

The addition of the weight had a greater impact on extension than flexion in terms of movement smoothness. The difficulty in extending the arm during the experiment could be attributed to the alignment of the arm with respect to gravity. When performing the task against gravity, users did not need to contract their triceps to extend the arm as gravity would extend it. This may have resulted in weaker activation of the triceps muscle, making it easier for Δe to fall into the deadzone and cause the exoskeleton to stop moving. The presence of added weight further exacerbated this effect. Furthermore, we noted smoother movements for longer-distance trajectories ($20^\circ \rightarrow 80^\circ$, $80^\circ \rightarrow 20^\circ$), which aligned with participant perceptions that precise set point control (a smaller corrective movement) was challenging. The corrective movements could have a less pronounced effect when normalized due to the larger primary movement. Additionally, longer-distance trajectories provided users more time to anticipate and adjust movements based on sensory-guided feedback, mitigating overshoots and leading to smoother outcomes.

Biofeedback contributed to smoother movements in certain extension movements ($50^\circ \rightarrow 20^\circ$, $80^\circ \rightarrow 20^\circ$). However, it did not exhibit the same level of effectiveness in the $80^\circ \rightarrow 50^\circ$ trajectory. The exception observed in the latter trajectory may be attributed to the increased difficulty in relaxing the biceps at a more flexed elbow angle. Individuals were able to voluntarily contract their muscles but struggled with voluntary muscle relaxation. At smaller elbow angles, users simply needed the indication from biofeedback to actively contract their triceps. However, at higher elbow angles, relaxation of the biceps was necessary, which could not be solely achieved through biofeedback. Biofeedback aids users in perceiving the differences in motor programs, and supplemental training is needed with this feedback to support the motor program learning.

Similar to the observed adaptation process in muscle effort, movement smoothness improved during $M1$, and stabilized from $M2$ through $M4$. However, in terms of task accuracy, participants from the *Visual* group did not exhibit the same improvement seen in the other two groups between $M1$ and $M4$. This discrepancy might arise from the visual feedback's potential distractions, diverting attention from the target matching task. Participants still had room for improvement. The SPARC scores could be up to -1.6 for point-to-point reaching tasks for a transverse plane task [39], and the task accuracy could exceed 85% based on pilot tests conducted by an expert user.

C. Survey

Participants in the feedback groups expressed a positive attitude towards its impact on their task performance. This positive attitude is important because it has the potential to increase users' confidence in using the device in task-specific settings, given that dissatisfaction with wearable devices is a primary contributor to device abandonment [49], [50], [51] and perceived usefulness is also a critical factor in the adoption of technology [52], [53], [54]. Moreover, the use of biofeedback can help users develop an internal model of exoskeleton dynamics by being aware of and able to predict the device's status, which can further improve their embodiment of the system [55], [56]. However, the challenges observed in developing exoskeleton motor programs suggest that supporting their development may require additional time and potentially more comprehensive feedback strategies.

It is noteworthy that participants in the *Haptic* group found it easier to move to and hold a target position with the exoskeleton than those in the *Visual* and *None* groups. While this difference could be attributable to more adept users assigned to the *Haptic* group, it may also indicate that haptic feedback has some advantages over visual feedback. This finding is further supported by a greater number of *Haptic* group participants using the feedback to predict the exoskeleton's movement, rather than solely to verify muscle relaxation during hold periods (as reported in open response questions). In our study, the potential advantages of visual feedback may be offset by increased cognitive load, as users had to switch their attention between visual feedback and the target-matching task. Ernst and Banks [57] showed that the nervous system might combine visual and haptic information in a statistically optimal fashion based on variance estimation. Additionally, Koritnik et al. [58] found superior performance in the haptic modality compared to the visual modality in a stepping-in-place task for rehabilitative applications, and even better outcomes when combining both modalities. Future research could investigate the potential benefits of integrating both modalities in exoskeleton usage.

D. Limitations and Future Work

In this experiment, we used the Myomo exoskeleton prototype in a manner consistent with the company's typical operation. However, unlike Myomo's approach of molding customers' arms to ensure a personalized fit, we were unable to customize the exoskeleton for each participant in our study. It should be noted that improper sizing can impact exoskeleton performance [59], [60], and overall system performance could improve with further personalization. Future exoskeleton hardware designs should consider reducing the total weight of the system and balancing weight distribution for unilateral use cases. Some participants reported accumulating fatigue during the experiment, which may have affected their performance, limiting the increase in accuracy and movement smoothness. Though the proportional myoelectric controller is not the only controller used on upper limb exoskeletons, we used it as a platform to study the efficacy of biofeedback. The outcomes may differ for alternate controllers. While more advanced methods exist, they often involve more complex calibration processes and require additional equipment. It is worthwhile to begin with a simpler approach to investigate whether a simple controller can achieve acceptable performance with the added guidance of biofeedback.

In future studies, additional training methods should be investigated to help users acquire the exoskeleton motor program. The acquisition of exoskeleton motor programs has significance beyond reducing muscle effort. Improper use of exoskeletons may result in inaccurate prediction of users' intentions, leading to performance inconsistencies in the human-exoskeleton team and potentially compromising the system's effectiveness. For instance, if a user significantly activates their biceps, the exoskeleton may misinterpret it as an intention to flex the arm, even if the user simply intends to hold a heavy object. Additionally, improving the exoskeleton controller is crucial to minimize response delay to movement, improve movement smoothness, and enhance fine set-point control, as indicated by feedback from participants. Although this study extended the task to encompass different magnitudes of motions and hold periods, evaluating the system's performance in more complex tasks, including a transfer test to assess skill transfer to other activities, is necessary for a more comprehensive representation of real-life scenarios. Furthermore, it is worth noting that biofeedback is typically used during training but is often unavailable after training. While this study was designed to assess the benefits when biofeedback was removed, it is important to acknowledge that biofeedback did not demonstrate its efficacy when it was presented. Consequently, the study did not yield results related to any residual effects of biofeedback. However, this topic remains worthy of exploration in future research endeavors.

V. Conclusion

In this study, we examined the impact of EMG biofeedback on the use of EMG-based powered upper limb exoskeletons in a tracking task. The results indicated that neither visual nor haptic EMG biofeedback led to large differences in muscle effort reduction (overall or during hold periods) or task accuracy, which could be attributed to the need for individuals to have additional support to acquire an appropriate exoskeleton motor program and learn how to effectively utilize the biofeedback information. However, it was observed that biofeedback could enhance extension smoothness, particularly at lower elbow angles, indicating that users were able to perform the desired actions when they had the capability to do so. Nevertheless, participants' perceptions suggest that biofeedback may improve their satisfaction with exoskeleton usage, which is a crucial factor for encouraging long-term use.

In light of these results, future studies should investigate training methods to further help the acquisition of appropriate exoskeleton motor programs and enhance the exoskeleton controller to support user intention classification. Furthermore, it is necessary to conduct additional assessments with more complex tasks to simulate daily activities.

Appendix
Survey

See Table II.

TABLE II
The Post-Study Survey

1. The exoskeleton made it ___ to move to a target position and hold that position.
a. Very difficult
b. Difficult
c. Neither difficult or easy
d. Easy
e. Very easy
2. (if a,b) Could you briefly describe what make it difficult to use the exoskeleton?
3. (if a,b,c) What do you think can help you use the exoskeleton more effectively?
4. (*Visual*) During the task, when did you typically look at the EMG visual feedback section? Why did you choose to look or not look at that section?
5. (*Visual*, *Haptic*) How does the EMG feedback affect your performing the target matching task?
a. Very negative
b. Negative
c. Neutral
d. Positive
e. Very positive
6. (*Haptic*) How much of the time during the experiment were you able to understand the information provided by the haptic feedback?
a. Never understand
b. Rarely understand
c. Sometime understand
d. Often understand
e. Always understand
7. (*Visual*, *Haptic*) Please briefly describe how EMG biofeedback affect your performance on the target matching task (*e.g.* how do you use the feedback, how does it help/not help your performance).
8. Do you have any final comments on the experiment?

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