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Closed-Loop Force Control by Biorealistic Hand Prosthesis with Visual and Tactile Sensory Feedback

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***Abstract*—The ability of a novel biorealistic hand prosthesis for grasp force control reveals improved neural compatibility between the human-prosthetic interaction. The primary purpose here was to validate a virtual training platform for amputee subjects and evaluate the respective roles of visual and tactile information in fundamental force control tasks. We developed a digital twin of tendon-driven prosthetic hand in the MuJoCo environment. Biorealistic controllers emulated a pair of antagonistic muscles controlling the index finger of the virtual hand by surface electromyographic (sEMG) signals from amputees' residual forearm muscles. Grasp force information was transmitted to amputees through evoked tactile sensation (ETS) feedback. Six forearm amputees participated in force tracking and holding tasks under different feedback conditions or using their intact hands. Test results showed that visual feedback played a predominant role than ETS feedback in force tracking and holding tasks. However, in the absence of visual feedback during the force holding task, ETS feedback significantly enhanced motor performance compared to feedforward control alone. Thus, ETS feedback still supplied reliable sensory information to facilitate amputee's ability of stable grasp force control. The effects of tactile and visual feedback on force control were subject-specific when both types of feedback were provided simultaneously. Amputees were able to integrate visual and tactile information to the biorealistic controllers and achieve a good sensorimotor performance in grasp force regulation. The virtual platform may provide a training paradigm for amputees to adapt the biorealistic hand controller and ETS feedback optimally.**

***Index Terms*— Digital twin, virtual prosthesis, biorealistic control, evoked tactile sensation, multi-sensory integration.**

I. Introduction

The human hand can accomplish numerous dexterous manipulations with ease. For forearm amputees, their sensorimotor pathways from brain to the hand have been completely severed after amputation. Myoelectric prostheses are currently a relatively mature solution for amputees to restore motor function of amputated limb and return to normal life [1]. Commercial myoelectric prostheses collect surface electromyography (sEMG) signals from antagonistic muscles, decoding signals through conventional proportional control [2], [3] or emerging machine learning strategies [4], [5], then driving multiple motors to achieve switching the aperture of the artificial hand or manipulating fingers to grasp objects. Hence, commercial myoelectric prostheses are able to furnish basic motor functions to restore feedforward pathway from brain to the artificial hand. However, the majority of commercial prostheses lack sensory feedback mechanisms to offer tactile information to the user [6].

In general, a large variety of neural interface techniques, including invasively and/or non-invasively, have been explored to facilitate the communicating of sensory information to amputees [7]–[11]. Invasive techniques utilize implanted microelectrodes to directly stimulate the cerebral somatosensory cortex [12], spinal cord [13], and peripheral nerves [14]. Mechanical stimulation or transcutaneous electrical nerve stimulation (TENS) non-invasively stimulates receptors and nerve endings under the skin to generate sensory signals [15]–[18]. These sensory feedback techniques have been demonstrated to significantly improve the recognition of object features, such as length [8], [19], size [20], compliance [21], and even texture [22], providing amputees with a sense of 'touch' while grasping objects. However, the question of whether the reconstructed tactile feedback loop can facilitate motor control in the artificial hand remains controversial [18].

Research has demonstrated that the addition of tactile feedback assists amputees in better controlling applied force in force control and daily tasks [18], [23]–[27], while others found the use of closed-loop prosthetic hand does not substantially enhance task performance [28]–[31], especially when vision is available [28], [30], [31]. These distinct outcomes stem from several factors, including the integration of vision and touch, the difficulty of selected tasks, diverse myoelectric control approaches, and the duration of prosthesis training. Vision plays a crucial role in providing feedback to amputees during their daily manipulation of objects [32]. It allows amputees to monitor the grip state and make

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real-time adjustments. However, it cannot completely substitute tactile feedback due to potential visual obstruction or in some scenes to grasp delicate objects [23]. The role of vision in different tasks and the ability of amputees to integrate visual and tactile information for feedforward control are different, which leads to the inconsistencies in the efficacy of tactile feedback for performance enhancement. Furthermore, tactile feedback to enhance motor performance is also limited by myoelectric control strategies. If the control strategy fails to accurately decode motor intentions, it may result in the prosthetic hand moving to inaccurate positions or applying inaccurate forces, rendering the perceived state irrelevant. If the prosthesis hand already provides satisfactory feedforward control with optimal performance, then the potential for motor performance improvement through artificial tactile feedback is constrained [27].

Recently, a biorealistic control approach that emulates neuromuscular reflex control of human upper limb system based on computational models has been used in prosthetic control [33], [34], [35]. The biorealistic control prosthetic hand could recapture human-like neuromuscular properties in the virtual environment [33], and has been verified to have more superior performance in evaluations of force control and functional tasks compared to traditional proportional control, particularly in more delicate tasks [34], [35]. Meanwhile, a unique non-invasive approach using TENS has also been extensively studied in many years [16], [36], [37]. Electrical stimulation on the projected finger map (PFM) area of the amputee's stump could directly induce finger-special evoked tactile sensations (ETS) [36], and elicit the corresponding response in the somatosensory cortex to establish a natural sensory feedback pathway [36]. Here we developed an integrated prosthesis training platform with an ETS-based somatotopic sensory feedback and biorealistic control approach in the virtual environment. The goal of this study was to systematically assess the effectiveness of sensory feedback in enhancing motor control and investigate the roles of tactile and visual feedback in closed-loop force control tasks without prior training. Two representative force control tasks also used in other literature were examined for six forearm amputees to evaluate closed-loop force tracking and holding using the virtual prosthetic hand. These tasks could also serve as a means to learn the biorealistic controller in future training. Preliminary assessments have been published in [38].

II. Materials and Methods

A. Subject Recruitment

Six unilateral forearm subjects were recruited to participate in this study (six males, age range: 48-69). Detailed clinical information of amputee subjects was listed in Table I. The stumps of all subjects were preserved at or above the elbow, allowing for the functionality of their residual flexor carpi ulnaris (FCU) and extensor carpi ulnaris (ECU) muscles. Only Subject 2 and Subject 3 (S2 and S3) had a moderate level of experience in myoelectric prosthesis control. Both S2 and S3 are daily users of myoelectric prosthesis, which S2 wore his myoelectric prosthetic hand for 3-4 hours of the workday while S3 wore his myoelectric prosthetic hand for 5-6 hours a day on weekdays. Despite the difference in the duration after amputation (11 years for S2 and only 3 years for S3), both S2 and S3 have considerable practical experience in grasping objects using their own myoelectric prostheses. All subjects reported they had no neurological disease or cognitive disabilities and had normal or corrected-to-normal vision. In each task under the healthy hand control, each subject grasped the object by pinching it between their thumb and index finger using their healthy hand, except for S3. Since the index finger of S3's healthy intact hand had been injured by a sickle and was unable to perform complex functional tasks, S3 used the middle finger instead of the index finger under the healthy hand control. All subjects were informed about experimental protocols and signed the consent form before joining the study. This study was approved by the Institutional Review Board for Human Research Protections, Shanghai Jiao Tong University (IRB number: E20200211).

TABLE I
Clinical Information of Amputee Subjects

Subject	Age	Sex	Amputation level and side	Cause	Year since amputation	Dominant hand	Myoelectric prosthesis experience
S1	69	Male	Left distal third of forearm	Trauma	16	Right	None
S2	58	Male	Right distal third of forearm	Trauma	11	Right	Yes (moderate level)
S3	48	Male	Right distal third of forearm	Trauma	3	Right	Yes (moderate level)
S4	65	Male	Right distal third of forearm	Trauma	37	Right	None
S5	58	Male	At right wrist	Trauma	43	Right	None
S6	69	Male	Left distal third of forearm	Trauma	47	Right	None

B. Integration of the Sensorial and Biorealistic Virtual Prothesis

The overview of the integrated virtual prosthetic hand training system was shown in Fig. 1. The subject collected sEMG signals from residual muscles to control the virtual hand to grasp an object by biorealistic control in the virtual environment. Contact force information on the fingertip was transmitted to the amputee through the ETS feedback, allowing the subject to adjust the grasp force in real time to complete tasks displayed on the screen.

The virtual prosthetic hand was developed in the MuJoCo (Version 2.0) environment. It was modified from the provided vMPL hand (Johns Hopkins Applied Physics Lab, Laurel, MD), and tendon-driven structure was added to fit the neuromorphic control. There were 14 hinge joints of the virtual hand as knuckles, two on the thumb and three on each of the other fingers. Each hinge joint had only one degree of freedom, with the rotation axis of the joint parallel to the palm side, enabling finger flexion or extension. Each finger had two tendons, starting from the forearm and extending to the tip of each finger, connecting all finger joints according to the anatomy of hand musculature [39]. The calculated neuromorphic muscle forces were acted on

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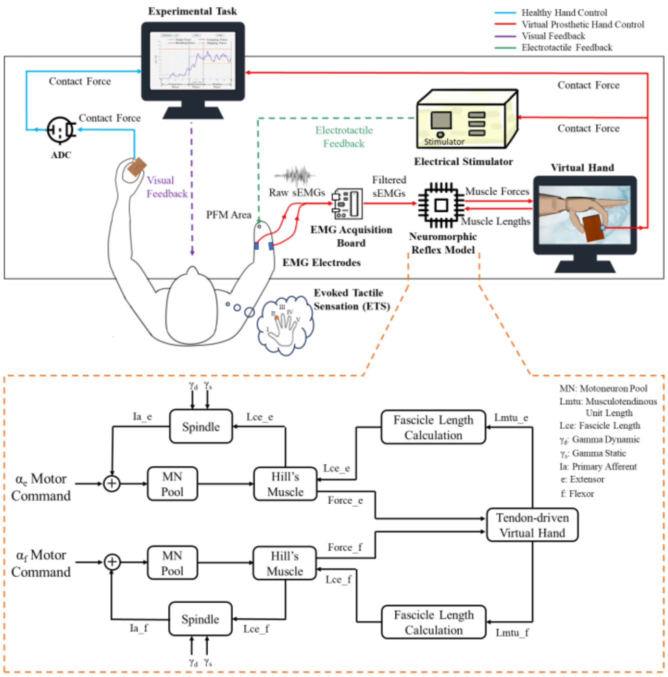

Fig. 1. Overview of the integrated virtual prosthetic hand training system. The top panel presents an amputee controlled the virtual hand to grasp a wooden block using biorealistic control and perceived the grasping force by ETS feedback, or grasped the same sized block using the intact healthy hand in the real scenario to complete the force control task showing in the screen. The bottom panel presents the block diagram of a pair of antagonistic neuromorphic reflex models.

antagonistic tendons to control the movement of the finger. When the flexor tendon was pulled by flexor force, all finger joints rotated together towards the palmar side of the hand, causing flexion of the digit. While the extensor tendon was pulled by extensor force, all finger joints rotated in the opposite direction, causing extension of the digit. Thus, the movement of one-freedom flexion/extension was achieved through the combined action of a pair of antagonistic muscles. The maximum limit of muscle force was set at approximately 200 N. Further details about the construction of the virtual prosthetic hand was provided in our published article [33].

In the control loop of the virtual prosthesis system, two sEMG electrodes were attached at the residual FCU muscle and ECU muscle on the amputee's stump. The signal processing module collected raw sEMG signals by sEMG electrodes, removed stimulated artifacts from TENS in real time by hardware blanking and software filtering [40]. The filtered sEMG signal was imported to the biorealistic controller as an excitatory postsynaptic current, which was superimposed with a Gaussian noise and distributed to several motoneuron pools following Henneman's size principle [41]. The activated motoneuron spiking signals drove skeletal muscles to generate muscle force based on muscle activation dynamics and viscoelastic characteristics [42]. Meanwhile, the mammalian spindle model modulated the sensitivity by gamma static and gamma dynamic fusimotor fibers [43] and accepted the information about muscle lengthening to send Group Ia afferents back to the motoneuron pools to modulate the muscle force [44] (bottom panel in Fig. 1). Each biorealistic controller including spinal reflex models was implemented on a programmable Very-Large-Scale-Integration (VLSI) using FPGA chips (Xilinx Spartan-6) [45] to calculate the real-time muscle force. Finally, the total muscle output forces from a pair of biorealistic controllers were actuated on the antagonistic virtual tendons to control the movement of the virtual prosthetic hand. The biorealistic controller was already validated in the virtual prosthetic hand [33].

In the sensory loop of the virtual prosthesis system, contact force information from contact sensor attached at the fingertip of the virtual index finger, or force sensor (FlexiForce A301, Tekscan Inc, the United States) pressed by the index finger of the healthy hand in the real scenario was encoded into biphasic, charge balanced, cathodic-first, rectangular electrical pulse trains by a self-developed multi-channel electrical stimulator [36] (Fig. 1). The pulse trains were transmitted into amputees via surface stimulation electrodes attached to the PFM area of the stump skin to elicit the evoked tactile sensation of the index finger. When the fingertip contact force was increased, the amputees were able to experience a stronger electrotactile sensation through ETS feedback. The stimulation electrode was a disk Ag/AgCl powered sintered electrode [46] with 10 mm diameter, and the 5-cm diameter non-woven fabric circular electrode (Yancheng Dalun Medical Equipment Co. Ltd, China) was used as the reference electrode, which located near the elbow on the amputation side.

C. Sensory Feedback Encoding

The perception of lost fingers in selected forearm amputees can be elicited by mechanically and electrically stimulation of specific stump regions. We poked the subject's stump skin using a blunt pen with a 2 mm tip diameter to identify the projected finger map (PFM) area by the amputee's verbal report of lost fingers. The most sensitivity points (MSP) of projected finger regions were marked and the electrical stimulation was applied on them for further confirmation that the amputee could feel the ETS at the lost fingers.

The application of electrical stimulation on the MSP with different parameters may elicit various modalities of evoked sensations on amputees, including touch, vibration, buzz, numb, tinging, etc [16]. The buzz sensation modality was not a natural and intuitive force sensation, we chose it to encode the force information because all six amputee subjects were able to feel this modality and it had a wider range of pulse width modulation [37]. The feedback calibration on each individual subject was performed through a fixed calibration protocol [37]. A linear pulse width modulation strategy of electrical stimulation was utilized to map contact force information, with a fixed frequency of 50 Hz and a personalized amplitude. The personalized amplitude was chosen at which the buzz sensation was perceived under the fixed frequency (50 Hz) and pulse width (200 µs). Under the selected amplitude and fixed frequency (50 Hz), the experimenter adjusted the pulse width to gradually increase from 10 µs with a step of 20~60 µs to find the appropriate coding range of pulse width. The smaller step (20 µs) was used to accurately determine the minimum and maximum pulse width, and the larger step (60 µs) was used to traverse rapidly in the intermediate regions. The pulse width was linearly proportional to contact force as follows:

$$PW = \begin{cases} PW = 0, & F < F_{min} \\ \dfrac{PW_{max} - PW_{min}}{F_{max} - F_{min}} * (F - F_{min}) + PW_{min}, & F_{min} \leq F < F_{max} \end{cases} \tag{1}$$

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$$PW = PW_{max}, \qquad F \geq F_{max}$$

Where PW represented the pulse width of electrical pulse trains. PW_{min} represented the minimal pulse width of electrical pulse trains and was set to the value when the subject could just feel the buzz sensation. PW_{max} represented the maximal pulse width of electrical pulse trains and was set to the previous value when the subject could just feel the uncomfortable or pain sensation. F represented the collected force by force sensors. F_{min} was the minimum force set at 0 N, and F_{max} was the maximum force set at 9 N for wider range of pulse width modulation.

D. Force Control Evaluation

Each amputee was seated in a comfortable chair, facing a screen which showed the task scene in front of 60 cm. The amputee was asked to participate in two force control tasks: (1) Force tracking task; (2) Force holding task. In each task, the participant was asked to control the virtual prosthetic hand using the residual muscles on the amputation side to grasp a 7*5*10 cm^3 sized virtual wooden block to generate corresponding fingertip contact forces according to different experimental requirements under various sensory feedback conditions, as illustrated in Fig. 1. In an alternative control condition, the amputee grasped a wooden block of the equivalent size as the virtual item using the healthy hand in the real scenario to complete the corresponding task as the best criterion (Fig .1). The information of grasping force which was generated by the subject was transmitted to the amputee by visual or electrotactile feedback. Each subject was asked to only look at the screen showing the task scene when visual feedback was present. In order to ensure that amputees completed the task with comparable difficulty using the amputated side and the healthy hand, the maximum neuromorphic muscle force of the virtual prosthetic hand could be fine-tuned as necessary so that the maximum grasping force generated on the amputated side was comparable to the maximum force generated by the health hand under the muscle maximum voluntary contraction (MVC). The generated maximum grasping force by the virtual prosthetic hand and the intact hand was shown in Table II.

TABLE II
THE GENERATED MAXIMUM FORCES OF THE VIRTUAL HAND AND THE INTACT HEALTHY HAND BY ALL SUBJECTS

Subject	Force generated by the virtual hand at 100% MVC (N)	Force generated by the intact healthy hand at 100% MVC (N)
S1	15.5	15.7
S2	15.5	15.3
S3	12.4	10.7
S4	11.5	11.8
S5	13.8	14.2
S6	16.0	14.9

Detailed experimental design was shown as follows:

(1) Force tracking task

The force tracking task assesses amputees' ability to promptly adjust grasping force to follow the target. In the force tracking task, the subject was instructed to control the grasping force to pursue the increasing target force by the real-time visual feedback. The scene of the force tracking experiment was shown in Fig. 2(A). The whole task was divided into two phases: the preparation phase and the tracking phase. In the preparation phase, the target force was maintained at 1 N. In the tracking phase, the target force continued to be maintained at 1 N for 1.5 s, then increased linearly from 1 N to 6 N over 6 s, and continued to be maintained at 6 N for more than 1.5 s. The subject needed to control the grasping force to follow the target force in real time.

Considering that the rising trajectory of the target force was relatively clear, we adjusted the ascending node of the target force at 3rd s or 4th s for 50% probability in order to prevent subjects from increasing the grasping force in advance when being familiar with the target trajectory after several trials. The random nature of the ascending node prevented amputees from predicting when to begin tracking the target even if the trajectory was memorized, so visual feedback is necessary in the force tracking task. The force tracking task was performed by amputees under three different experimental conditions, namely biorealistic control with visual and tactile feedback (B_VT), biorealistic control with visual feedback (B_V) and healthy hand control with visual and tactile feedback (H_VT).

(2) Force holding task

The force holding task mimics a scenario of grasping fragile objects in daily lives. In the force holding task, the subject was asked to control the grasping force to maintain a level comparable to the fixed target force (6 N), and not exceeding the breaking force (8 N) or less than the slipping force (4 N). The whole scene of the force holding task was illustrated in Fig. 2(B). In the preparation phase, the target force kept on 1 N and the subject was asked to control the virtual index finger gently touch the wooden block to generate a smaller grasping force. Then the target force immediately stepped to 6 N at the beginning of adjustment phase and kept on the 6 N until the end of the task. The grasping force was adjusted by the subject to rapidly increased from around 1 N to around the target force within 2 s in the adjustment phase and remained around the target force between the slipping force and breaking force for 4 s in the holding phase. The step moment of each trial was randomly assigned at the 3rd and 4th s for 50% probability to prevent the subject from adapting the trajectory of the target force. The whole holding task was conducted under five experimental conditions, including biorealistic control with visual and tactile feedback (B_VT), biorealistic control with visual feedback (B_V), biorealistic control with tactile feedback (B_T), biorealistic control with no feedback (B_N) and healthy hand control with visual and tactile feedback (H_VT). Under the B_T

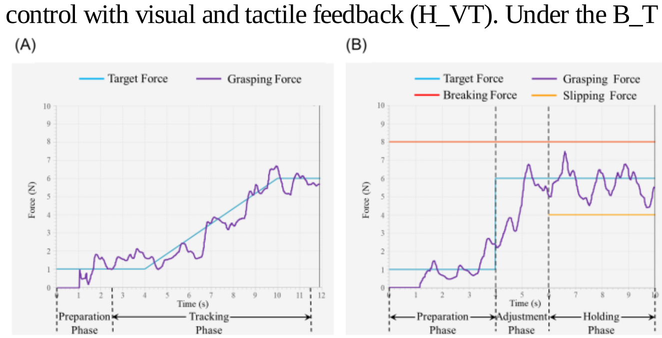

Fig. 2. Task scenes of the force tracking task (A) and force holding task (B), developed in the IntelliJ IDEA (Version 2019.3.3).

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and B_N conditions, the generated grasping force of the subject would be invisible after the preparation phase ended, the subject would not rely on visual feedback to adjust and maintain the grasping force. The trial was marked as a success if the grasping force did not exceed the breaking force throughout the whole process and the grasping force did not fall below the slipping force at the holding phase. At the end of each trial, the amputee was not told whether this trail was successful or failed, so there was no outcome feedback to help the amputee adjust grasping force for the next trial.

Prior to starting the experiment under each experimental condition, the subjects performed a short familiarization session (~5 minutes) to study the sensorimotor integration and familiarize the task requirements. In each task, H_VT condition was easiest to perform first, the order of B_VT and B_V conditions in the force tracking task and the order of B_VT, B_V and B_T conditions in the force holding task was randomized to avoid feedback conditions enabled the subject to create an internal model that can facilitate feedforward control. B_N condition in the force holding task was last to perform because it was hardest and only had feedforward control. Twenty trials were performed under each experimental condition for both force control tasks. The 20 trials were divided into 4 blocks, and each block contained 5 trials. After completing each block, the subject would take a 2-minute break to prevent fatigue. For the start of each block, two or three practice trials would help the participant become familiar with the feedback conditions and then continued performing formal trials.

E. Performance Metrics

In order to quantify the motor performance of the prosthetic hand under different feedback conditions, some outcome metrics were accessed as follows:

(1) Root-mean-square Error (RMSE)

The RMSE between the target force and grasping force was calculated as follows:

$$RMSE(n) = \sqrt{\frac{1}{T}\sum_{t=0}^{t=T}(GF(n,t) - TF(n,t))^2} \tag{2}$$

where GF and TF represented the grasping force and target force, n was the trial number, T denoted the force duration, which referred to the tracking phase in the tracking task and the holding phase in the holding task.

(2) Success Metrics and Success Rate

In the force holding task, each trial from all the subjects could be assessed with a binomial outcome of Success (0-Failed, 1-Succeeded). The success rate of completing the force holding task under different experimental conditions was also calculated.

(3) Force Correlation

The cross-correlation between the generated grasping force under different experimental conditions and mean grasping force under H_VT condition of each subject was computed to quantify the compatibility with the intact hand under different experimental conditions. The computational formula was as follows:

$$GF_{H\_VT}(t) = \frac{1}{N}\sum_{n=1}^{n=N} GF_{H\_VT}(n,t) \tag{3}$$

$$Force\ Correlation(n) = \frac{\sum_{t=0}^{t=T}\left(\left(GF(n,t)-\frac{1}{T}\sum_{t=0}^{t=T} GF(n,t)\right)\cdot\left(GF_{H\_VT}(t)-\frac{1}{T}\sum_{t=0}^{t=T} GF_{H\_VT}(t)\right)\right)}{\sqrt{\sum_{t=0}^{t=T}\left(GF(n,t)-\frac{1}{T}\sum_{t=0}^{t=T} GF(n,t)\right)^2}\cdot\sqrt{\sum_{t=0}^{t=T}\left(GF_{H\_VT}(t)-\frac{1}{T}\sum_{t=0}^{t=T} GF_{H\_VT}(t)\right)^2}} \tag{4}$$

where both GF_{H_VT} and GF characterized the grasping force, the former indicated the grasping force under H_VT condition and the latter indicated the grasping force under any experimental conditions. N and n represented the total number of trials and trial number. T denoted the force duration, which referred to the tracking phase in the tracking task and the period from 1.5 s before the adjustment phase to the end of the holding phase in the holding task.

F. Statistical Analysis

All data processing was done using MATLAB (R2021a, MathWorks Inc., Natick, MA) and the statistical analysis was carried out with IBM SPSS Statistics, version 26.0. Since no dataset passed the one-sample Kolmogorov-Smirnov test, indicating the dataset was not normally distributed, the results are reported as median values.

The RMSE, success rate and force correlation indices for overall and individual subjects were first ranked and then analyzed using Friedman's rank test to evaluate between-group variation under different experimental conditions. Since the success or failure of one trial was a binomial variable, the success matrix under different experimental conditions was analyzed with Pearson's chi-squared test. All statistical and correlation analyses were run with significance as $p < 0.05$. Post hoc comparisons with a Bonferroni-correction assessed the pairwise differences after Friedman's rank test and Pearson's chi-squared test.

III. Results

A. Performance in the Force Tracking Task

The representative average grasping force profiles from S1 under different experimental conditions in the force tracking task were shown in Fig. 3(A)-(C). The shadow area of average grasping force under H_VT condition appeared to be smaller than those under B_VT and B_V conditions (Fig. 3(A)-(C)), which indicated that the variability of the grasping force controlled by the healthy hand was less and the grasping force was closer to the target force in the tracking phase.

The amputees were able to complete the tracking task on the amputation side relatively good, but there was still a certain gap with the healthy hand control in RMSE (Fig. 3(D)) and force correlation (Fig. 3(F)). Separate comparisons of two conditions that used biorealistic control on the amputation side were illustrated in Fig. 3(E) and Fig. 3(G). The results of RMSE under the B_VT and B_V conditions showed that the additional tactile feedback seemed to have no effect on the motor performance across all six subjects when visual feedback was always present (Fig. 3(E)), but for each individual subject, the effect of tactile feedback may be completely different. As seen in Fig. 3(E), RMSE of S1, S2 and S3 under B_VT condition was significantly greater than that under B_V condition, while the opposite was true for S4, S5 and S6. Similar results occurred for the index of force correlation (Fig. 3(G)). There was no significantly different of force correlation across all subjects between B_VT and B_V conditions (Fig. 3(G)). For individual subjects, S1 and S3 had better force correlation under B_V condition, while S4, S5 and S6 did better under the B_VT condition (Fig .3(G)).

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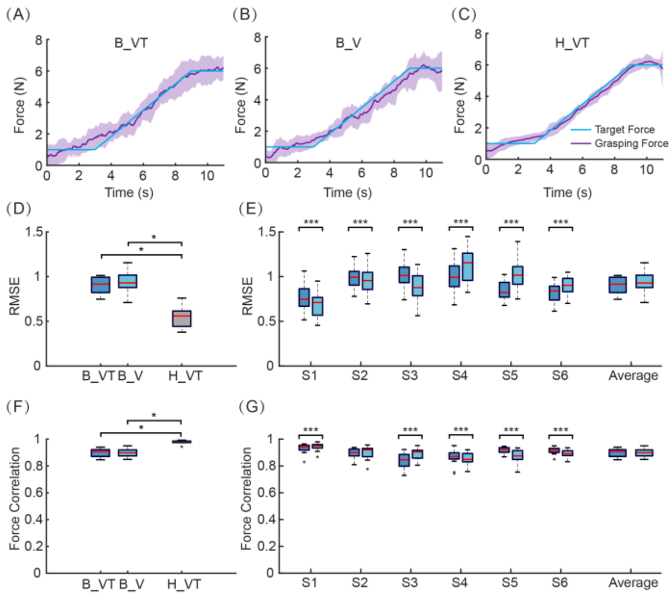

Fig. 3. Motor performance in the force tracking task. Generated grasping forces of 20 trials (mean ± std) by S1 under B_VT condition (A), B_V condition (B) and H_VT condition (C) in the force tracking task. (D). The comparison of RMSE across all subjects under B_VT, B_V and H_VT conditions. (E). The comparison of RMSE under B_VT and B_V conditions. (F). The comparison of force correlation across all subjects under B_VT, B_V and H_VT conditions. (G). The comparison of force correlation under B_VT and B_V conditions. The statistical difference is presented by '*' ($p<0.05$), '**' ($p<0.01$) or '***' ($p<0.001$).

B. Performance in the Force Holding Task

The grasping forces of 20 trials controlled by S1 under different experimental conditions in the holding task were presented in Fig. 4(A)-(E). The shadow area under H_VT condition was smallest in the holding task (Fig.4(C)), which presented that the force stability controlled by the healthy hand may be better than the amputated limb. Fig. 4(E) showed that the shadow area under B_N condition looked larger than that under the other conditions and had a tendency to decrease over time.

Three indexes about RMSE, success rate and force correlation were evaluated in the force holding task. The RMSE of all six subjects under H_VT condition was smaller than that under the other conditions when using biorealistic control on the amputated side, and the Friedman's rank test revealed that there was a significant different between B_T condition and H_VT condition, and between B_N condition and H_VT condition (Fig. 5(A)).

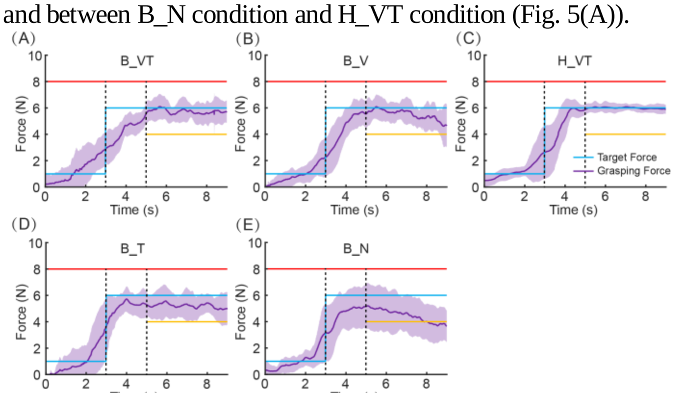

Fig. 4. Generated grasping forces of 20 trials (mean ± std) by S1 under B_VT condition (A), B_V condition (B), H_VT condition (C), B_T condition (D) and B_N condition (E) in the force holding task.

Since the motor performance on the healthy hand control was much better than that used biorealistic control on the amputated side, the effects of different feedback conditions on force control ability on the amputation side were overwhelmed when H_VT condition was present, hence we compared the four conditions that used biorealistic control on the amputated side to evaluate effects of different feedback conditions on force holding. As seen in Fig. 5(B), when visual feedback was present, there was no significant difference across all subjects in RMSE under B_VT and B_V conditions, which indicated that adding tactile feedback did not significantly improve the stability of the grasping force when visual information was available. However, the results for individual subjects were inconsistent with the overall results. Observing individual subjects, we found that both subjects with myoelectric control experience (MCE), namely S2 and S3, as well as S4, had a greater RMSE under B_VT condition than that under B_V condition (Fig. 5(C)), while two of four persons (S1 and S5) who have no MCE presented the RMSE under B_VT condition significantly smaller than that under B_V condition (Fig. 5(C)).

When visual feedback was blocked, RMSE under B_T condition was significantly smaller than that under B_N condition across all subjects (Fig .5(C)). For individual subjects, two of the four subjects without MCE (S1 and S4) and S3 with MCE had a lower RMSE under B_T condition compared to that under B_N condition, and only S2 performed better under B_N condition with a lower RMSE (Fig. 5(C)).

At the same time, we also evaluated the effects of visual feedback and tactile feedback individually on the promoting of the steady grasping force by post hoc pairwise comparisons. Fig. 5(B) showed that the RMSE under B_V condition was significantly lower across all subjects compared to that under B_T condition. For each subject individually, S2, S3 and S6 had a smaller RMSE under B_T conditions compare of that under B_N conditions (Fig. 5(C)).

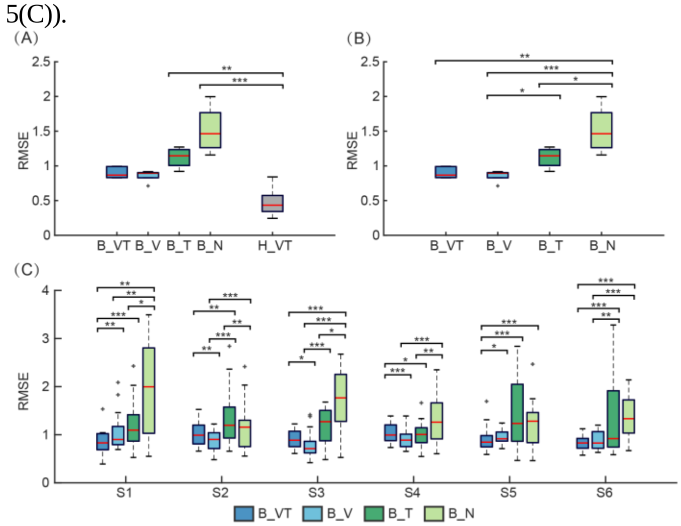

Fig. 5. The RMSE between grasping force and target force in the force holding task. (A). The comparison of RMSE across all subjects under B_VT, B_V, B_T, B_N and H_VT conditions (B). The comparison of RMSE across all subjects under conditions with biorealistic control, including B_VT, B_V, B_T and B_N conditions. (C). The comparison of RMSE for individual subjects under conditions with biorealistic control, including B_VT, B_V, B_T and B_N conditions. The statistical difference is presented by '*' ($p<0.05$), '**' ($p<0.01$) or '***' ($p<0.001$).

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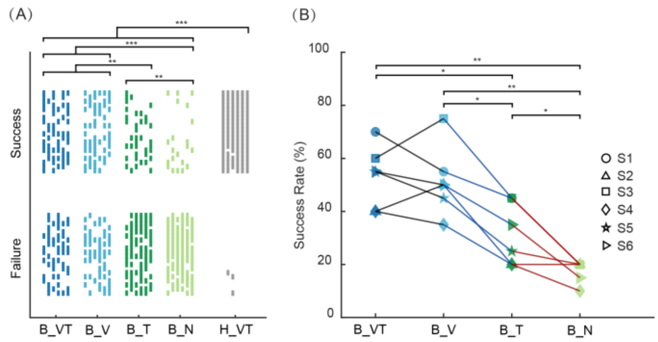

Fig. 6. Motor performance of success metrics and success rate in the force holding task. (A). The comparison of success metrics under B_VT, B_V, B_T, B_N and H_VT conditions. (B). The comparison of success rate under conditions with biorealistic control, including B_VT, B_V, B_T and B_N conditions. The statistical difference is presented by '*' ($p<0.05$), '**' ($p<0.01$) or '***' ($p<0.001$).

The success rate reflected the subject's subjective level of care as well as the objective ability of the force control stability. The results showed that when using the healthy hand control with visual and tactile feedback, all subjects were able to perform the task with almost 100% success rate, except for occasional small error (Fig. 6(A)), whereas for the amputated side control, the best performance of the amputee was only 75% success rate (Fig. 6(B)). The binomial outcomes of successful trials were illustrated in Fig. 6(A). The number of successful trials under H_VT condition was significantly greater than that under all other conditions by Pearson's chi-squared test with Bonferroni correction (Fig. 6(A)). The number of successful trials under B_VT and B_V conditions had the same distribution and they were both significantly more than that under B_T and B_N conditions (Fig. 6(A)). The number of successful trials under B_T condition was also more than that under B_N condition significantly (Fig. 6(A)). Friedman's rank test was used to evaluate the success rate of all subjects under different experimental conditions using biorealistic control on the amputation side. There was no significant difference in success rate between the B_VT and B_V conditions (Fig. 6(B)). The statistical test presented a significant decrease of success rate from B_V condition to B_T condition, as well as from B_T condition to B_N condition (Fig. 6(B)). For individual subjects, all subjects had a decreased success rate from B_V condition to B_T condition, as well as from B_T condition to B_N condition, except for S2 who was remained flat with only tactile feedback and without any feedback (Fig. 6(B)).

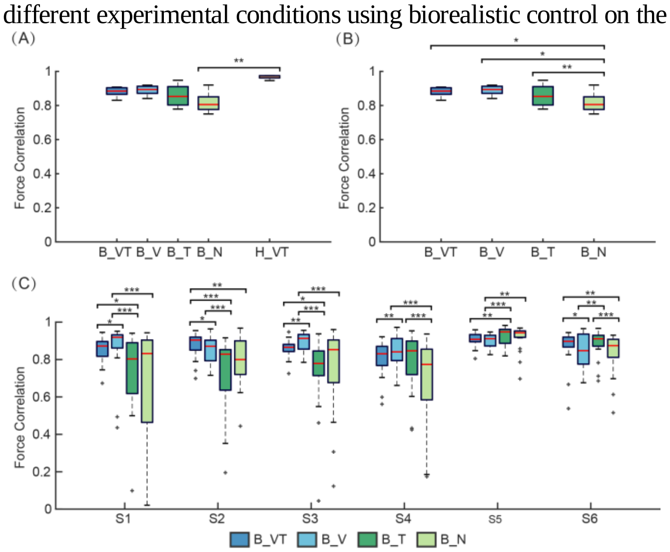

Fig. 7. The force correlation between grasping force and mean force under H_VT condition in the force holding task. (A). The comparison of force correlation across all subjects under B_VT, B_V, B_T, B_N and H_VT conditions. (B). The comparison of force correlation across all subjects under conditions with biorealistic control, including B_VT, B_V, B_T and B_N conditions. (C). The comparison of force correlation for individual subjects under conditions with biorealistic control, including B_VT, B_V, B_T and B_N conditions. The statistical difference is presented by '*' ($p<0.05$), '**' ($p<0.01$) or '***' ($p<0.001$).

The index of force correlation reflected the compatibility with the intact hand under different experimental conditions. There was a significant difference in force correlation between the H_VT and B_N conditions (Fig. 7(A)). Friedman's rank test examined that there was no significant difference in force correlation under B_VT and B_V conditions, and force correlation under B_N condition was smaller than that under B_VT, B_V, B_T conditions (Fig. 7(B)). For individual subjects, the force correlation of S2 and S6 under B_VT condition was significantly greater than that under B_V condition, while the opposite was true for S1, S3 and S4 (Fig. 7(C)). When visual feedback was blocked, the force correlation of S4 and S6 under B_T condition was significantly greater than that under B_N conditions (Fig. 7(C)). In the comparison of B_V and B_T conditions, S1, S2 and S3 had a higher force correlation under B_V condition, while the opposite was true for S5 and S6 (Fig. 7(C)).

IV. Discussion

Prosthetic hand is an important motor rehabilitation system for amputees to complete many complex and dexterous functions. Recently, a novel concept was proposed to improve the functionality of the prosthetic hands by maximizing its neural compatibility with the sensorimotor system of users [47]. Several studies have demonstrated that the sensory or control approaches with improved neural compatibility perform better in their respective prosthetic tasks [34]–[36], [48]. Based on this point of view, in this study, we developed a virtual prosthetic hand training platform with biorealistic control approach and ETS-based somatotopic sensory feedback, evaluated the effect of various sensory feedback in enhancing motor control and investigated respective roles in visual and tactile feedback in the closed-loop force tracking and holding tasks. This training platform could be examined through the basic force control tasks and train amputees to operate the biorealistic controller and ETS feedback optimally.

This study investigated the role of visual and tactile feedback in closed-loop force tracking and maintaining on the integrated virtual training platform. All six forearm amputee subjects were able to complete these two representative force control tasks. In the force tracking task, the subjects were required to generate grasping forces that followed the changed target force in real time. In the force holding task, the target force was constant and the subjects were asked to maintain a force close to the target for 4 s, not exceeding the breaking force or falling below the slipping force. The information of the grasping force was transmitted to subjects by visual or electrotactile feedback. Different feedback conditions were employed for the two force control tasks. The force tracking task was designed to only include experimental conditions when visual feedback was always present, because the random nature of the ascending node from target trajectory prevented amputees from predicting when to begin tracking the target even if the trajectory was memorized. However, adding no feedback condition and tactile-only feedback condition in the force tracking task is closed to the scenes in clinical applications and can make a good comparison. This comparative evaluation should be considered in future studies in relation to training amputee users.

Although the requirements and feedback conditions of these two force control tasks are different. However, in the presence of visual feedback, the results of both tasks with biorealistic control on the amputation side were quite similar. The additional tactile feedback did not enhance the performance of force tracking and holding tasks under visual feedback present in all subjects (Fig. 3(E) and Fig. 5(B)), which is similar to some previous studies [28], [30], [31], yet had different results in individual subjects. On two force control tasks, both subjects with MCE (S2 and S3) had better performance when visual-only feedback was available compared to that when both visual and tactile feedback was present (Fig. 3(E), Fig. 5(C)), whereas the majority of subjects without MCE were in favor of more feedback pathways to promote better performance (Fig. 3(E), Fig. 5(C)). This may indicate that two populations of people in possession of different internal models of control and sensation to deal with force control tasks [49], [50]. For those with extensive MCE of prosthesis, they are accustomed to relying on visual feedback to perform various operations with their customized myoelectric prosthesis in daily use. Thousands of myoelectric prosthetic hand use in daily lives have led to involuntarily establish an internal model of myoelectric control and visual feedback in prosthesis. Hence, amputees with MCE could quickly get the hang of biorealistic control approach using residual sEMG signals and achieve a more favorable performance under visual feedback in force tracking and holding tasks. When the tactile information was added without training, tactile feedback was not effectively integrated to form a new internal model with multi-modality sensory loop, so the redundant tactile information may interfere with the original internal model to reduce the motor performance. This is exactly the opposite reflected in subjects who are without MCE. They wore cosmetic hands in daily lives without functional use. Subjects without MCE could quickly learn the additional tactile feedback and integrated it with visual feedback to apply them into feedforward control in different tasks.

For the force holding task, we also evaluated the subject's motor performance under tactile-only feedback condition and under no feedback condition when visual feedback was blocked. Under no feedback condition, the amputees completed the task in formal experiments by memorizing the muscle activation in the familiarization session. The virtual environment also ruled out the proprioception that can be felt through the wearable socket connected to the prosthetic hand. In the absence of any feedback information, subjects could accomplish the force holding task with an average success rate of 17.5% through biorealistic feedforward control alone (Fig. 6(B)). We found there was a decrease in grasping force over time in some trials at the end of the holding task, as seen in Fig. 4(E), suggesting that it is difficult to maintain a stable muscle contraction for a long time with feedforward control alone [51], [52], which is one of the reasons for the low success rate. When tactile feedback was added without visual information, motor performance and behavioral correlation were enhanced for the overall subjects (Fig. 5(B), Fig. 6(B) and Fig. 7(B)), as reflected in the index of RMSE, success rate and force correlation, except for S2. This suggests that tactile feedback is a reliable source of information [53], [54], amputees can rely on this ETS feedback to facilitate motor performance in the force holding task and enhances the compatibility of the hand prosthesis to the intact hand, which is similar to findings of many studies when visual feedback was blocked [18], [23], [31]. S2 exhibited the higher RMSE and the same success rate under B_N condition compared to those under B_T condition. The anomalous motor performance of S2 may stem from his excellent feedforward myoelectrical control to stabilize the grasping force in long-term daily use, leading him to pay more attention to reproduce memorized muscle activation and intentionally refused to integrate tactile feedback. Some training is needed for him to better integrate tactile feedback to the myoelectric control.

The motor performance and behavioral correlation under visual-only feedback was much better that those under tactile-only feedback for most subjects (Fig. 5(C), Fig. 6(C) and Fig. 7(C)). This finding is in accordance with previous studies on similar tasks [23], [53], which suggests that visual feedback is a more effective feedback loop than tactile feedback in the force holding task. Visual feedback provides high-fidelity information and characterized by higher accuracy and greater timeliness compared to tactile feedback. In the holding task, the visualizing grasping force trajectory provides the most reliable information to the brain to help the amputee to make real-time adjustments. On the other hand, touch itself is a relatively lower quality feedback loop compared with vision, and it is more sensitive to sensory modalities and less sensitive to sensory intensity for normal people. Moreover, electrotactile stimulation feedback needs a delay of several stimulation pulse to perceive the deviation of the current state from the target [53]. Hence, under the same conditions, visual feedback plays a more valuable role in force control tasks [23], [53]. In a recent study, neural evidence also confirms the same conclusion that visual feedback played a predominant role compared to tactile feedback in a similar force control task by electroencephalogram (EEG) recordings on cortical activities [32]. Visual feedback in this study established a benchmark for test and evaluation and we hope it could help the subject to acquire an internal model of the biorealistic controller with electrotactile feedback through training in this virtual training platform.

The generated force trajectories of tracking a ramp or a step target force using biorealistic control were compared with those using the intact hand control. Although average correlations of 0.89 in the tracking task and 0.83 in the holding task (Fig. 3(G) and Fig. 7(B)) were obtained, the high correlation values might be biased by the simple profile of the target trajectory. A target trajectory of more variability may be employed in future experiments to reveal improved neural compatibility in the biorealistic control.

This study is the first time to integrate ETS-based somatotopic sensory feedback with the biorealistic controller in an integrated prosthesis training platform. Two candidate tests of force tracking and force holding were used to evaluate the performance of force control ability in various sensory feedback conditions. The ETS feedback could effective improve the performance of feedforward control when vision was blocked. Visual feedback played a predominant role than the ETS feedback in force tracking and holding tasks, and as a benchmark could help amputees to acquire an internal model of the biorealistic controller. These results were consistent of the findings of others in previous studies [18], [23-28], [30], [31], and validated the virtual platform could be as a training program for amputees to operate the biorealistic controller with the ETS sensory feedback. On the other hand, we also found that the effects of feedback were highly subject-specific, and may be consistent with experience of myoelectric control. This prosthesis training platform in the future is expected to train amputees to create an internal model of the biorealistic control and ETS feedback to maximize performance.

Nevertheless, there were still some limitations in this study. First, it is limited by the number of amputees who have phenomenon of evoked finger sensory, there is not enough to do a group analysis of people with and without myoelectric control experience. Hence, this study only separately analyzes for each individual subject, providing some possible explanations for the effect of myoelectric control experience on motor performance. The validated conclusions would require the recruitment of more amputee subjects in subsequent studies. Second, tactile sensation is not a particularly sensitive sensory modality, and the sensitivity of ETS feedback is different for each amputee. A pilot experiment needs to be carried out to differentiate how many levels of grasping force that the amputee could distinguish, ensuring that amputees can perceive different electrotactile intensities below the slipping force and above the breaking force in the force holding task. Due to the limitation of the experiment duration, all subjects were only interviewed in the familiarization session and all subjects reported that they could perceive the difference in the intensity of the electrotactile at the slipping force, target force, and grasping force in the force holding task.

IV. Conclusion

In conclusion, this study developed a digital twin of tendon-driven prosthetic hand that integrated a pair of antagonistic biorealistic controllers and an ETS-based somatotopic sensory loop in a virtual training platform, and evaluated the respective roles of visual and tactile feedback in the fundamental force control tasks with little training. Results showed that visual feedback played a predominant role than the ETS feedback in force tracking and holding tasks, and the ETS feedback was a reliable sensory loop to effective improve the performance of feedforward control when visual feedback was not present. We also found that the effects of tactile and visual feedback on force control were subject-specific when both feedback information was provided simultaneously. This virtual platform of prosthetic hand may be useful in training amputees to create an internal model of the biorealistic control and ETS feedback to maximize performance.

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