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Closed-Loop Control of a Multifunctional Myoelectric Prosthesis With Full-State Anatomically Congruent Electrotactile Feedback

Martin A. Garenfeld, Matija Strbac[✉], Nikola Jorgovanovic, Jakob L. Dideriksen[✉], and Strahinja Dosen[✉], *Member, IEEE*

Abstract—State-of-the-art myoelectric hand prostheses provide multi-functional control but lack somatosensory feedback. To accommodate the full functionality of a dexterous prosthesis, the artificial sensory feedback needs to convey several degrees of freedom (DoF) simultaneously. However, this is a challenge with current methods as they are characterized by a low information bandwidth. In this study, we leverage the flexibility of a recently developed system for simultaneous electrotactile stimulation and electromyography (EMG) recording to present the first solution for closed-loop myoelectric control of a multifunctional prosthesis with full-state anatomically congruent electrotactile feedback. The novel feedback scheme (coupled encoding) conveyed proprioceptive (hand aperture, wrist rotation) and exteroceptive information (grasping force). The coupled encoding was compared to the conventional approach (sectorized encoding) and incidental feedback in 10 non-disabled and one amputee participant who used the system to perform a functional task. The results showed that both feedback approaches increased the accuracy of position control compared to incidental feedback. However, the feedback increased completion time, and it did not significantly improve grasping force control. Importantly, the performance of the coupled feedback was not significantly different compared to the conventional scheme, despite the latter being easier to learn during training. Overall, the results indicate that the developed feedback can improve prosthesis control across multiple DoFs but they also highlight the subjects' ability to exploit minimal incidental information. Importantly, the current setup is the first to convey three feedback variables simultaneously using electrotactile stimulation while providing multi-DoF

myoelectric control with all hardware components mounted on the same forearm.

Index Terms—Multi-functional prosthesis, closed-loop myoelectric control, electrotactile stimulation, multi-channel electrode, artifact blanking, signal processing, pattern recognition.

I. INTRODUCTION

HAND amputation causes substantial motor and sensory impairments, which have a dramatic impact on the quality of life of the affected person. Myoelectric prosthetic devices for transradial amputees restore the lost motor function by allowing the person to use the same muscles as before the injury to control the movements of the bionic hand. However, the control and ergonomics of the prosthetic hands are far from being on par with the lost limb [1], and despite the developments in the technology, still up to 44% of the users abandon their myoelectric prosthesis [2], [3]. Artificial somatosensory feedback is an additional feature that is missing from mainstream commercial devices, while it has been reported as an important user requirement [4]. Only a few recently developed commercial systems offer simple somatosensory feedback to their users [5], [6], [7]. Without proprioceptive or exteroceptive information, the user needs to rely on visual and auditory feedback, which can be cognitively demanding and might lead to sub-optimal control [8], [9]. Despite some indications that closing the loop can improve user experience [10], [11] and performance [12], [13], implementing effective feedback is still coupled with significant challenges [14]. Feedback can be delivered either invasively to the sensory-motor structures within the body or noninvasively, by stimulating the skin of the residual limb [15]. Invasive methods evoke somatotopic sensations by electrically stimulating peripheral nerves [13], [16], [17]. However, this approach demands additional surgery, which prosthesis users can be reluctant to undergo [18]. Non-invasive methods provide substitution feedback, in which the missing somatosensory information is conveyed to the subject by delivering vibrotactile and electrotactile stimulation [19], [20].

State-of-the-art multifunctional prostheses such as Michelangelo Hand (Otto Bock, Duderstadt, Germany)

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provide opening and closing of the hand plus an active wrist and are often equipped with embedded force sensors. Therefore, three feedback variables (hand aperture, wrist rotation, and grasping force) need to be conveyed to the user to provide the full state of the prosthesis. Conveying multiple variables through the stimulation interface, however, is a challenge as the current tactile stimulation technologies are characterized by a limited bandwidth for information transfer [21]. Accordingly, most studies in the literature focused on transmitting a single feedback variable, usually grasping force [14]. Specifically, only a few studies proposed feedback solutions to communicate at most two variables through either invasive [17], [22], [23], [24] or surface stimulation [25], [26], [27], [28]. In [17] and [22], the information on grasp force and hand aperture were provided through peripheral nerve stimulation by linearly modulating the frequency, while in [23] and [24] intensity modulation was utilized to convey the same variables. Vibrotactile stimulation with spatial and intensity modulation was used to transmit hand aperture and grasp force of a prosthesis placed on the table [25] and a virtual prosthesis controlled by scrolling a computer mouse [26], [27]. In [28], the finger joint angle and grasp force of a prosthesis placed on a table were communicated via a longitudinally placed electrode and transversely placed vibrotactile array on the upper arm. So far, to the best of our knowledge, there is no system in the literature that provided three feedback variables simultaneously to accommodate the full state of a multifunctional prosthesis.

Electrotactile stimulation is particularly convenient for multivariable feedback as it is a flexible interface that can generate versatile and dynamic stimulation profiles modulated in intensity, frequency, and location [29], [30]. Therefore, the use of matrix electrodes that can be printed in different shapes, sizes, and configurations of stimulation pads can increase the information bandwidth for communication through the tactile sense. For instance, the encoding schemes were proposed to convey hand aperture and wrist rotation using spatial and amplitude/frequency modulation across the pads of a flexible array electrode that wraps around the residual limb [31]. However, the approach was not tested with the prosthesis in the loop and the stimulation and recording were performed on the contralateral forearms to avoid interference. Importantly, we recently presented a compact unit with integrated dynamic blanking that can implement simultaneous electrotactile stimulation and EMG recording from the same forearm [32].

In the present study, we build on top of these previous developments and demonstrate the feasibility of the first system for myoelectric control of multifunctional prostheses that simultaneously provides a comprehensive full-state electrotactile feedback with all the components (stimulation, EMG recording, and prosthesis) placed ipsilaterally. We have developed a novel feedback scheme (coupled encoding) that leverages the flexibility of a multi-pad array electrode to provide three feedback variables (hand aperture, wrist rotation, and grasping force) by producing tactile sensations anatomically congruent with prosthesis motions. The effectiveness of the novel scheme was compared to the conventional approach (sectorized encoding) used in our previous studies [31], [32] as well as to the performance without electrotactile feedback, while the

subjects used the prosthesis equipped with closed-loop control to conduct a functional task. Although the coupled feedback was intuitively related to prosthesis motion, we expected that it might be more difficult to interpret by the subjects due to more complex encoding, thereby potentially leading to lower performance. We also assumed that both feedback methods would outperform the condition without electrotactile feedback, possibly at the expense of a longer time to accomplish the task.

II. METHODS

A. Subjects

Ten non-disabled subjects (8 males and 2 females, right-handed, 34.4 ± 11.6 years) and one congenital limb-deficient participant (female, 56 years old) completed the experiment after signing a consent form. The experimental protocol was approved by the Ethical Committee of Region Nordjylland, Denmark (approval number N 20 150 075).

B. Equipment and Setup

The experimental setup is shown in Fig. 1. The EMG recording and electrotactile stimulation were implemented using the MaxSens system (Tecnalia Research & Development, Spain) described in [32]. The stimulation electrode comprised an array of 16 circular active pads and a large, elongated reference pad. This electrode design was used successfully in previous studies to provide tactile feedback [30], [31], [32]. The recording electrode consisted of 8 pairs of circular pads to measure bipolar EMG and three larger reference pads. Both electrodes were produced by printing conductive Ag/AgCl and dielectric inks over $125 \mu\text{m}$ thick PET film. The pads were covered with hydrogel (AG725, Axelgaard, Denmark) to increase conductivity and improve electrode-skin contact. The two electrodes were connected to the stimulation and recording unit (MaxSens) with integrated blanking to prevent interference between electrical pulses and EMG signals [32]. The device generated biphasic symmetric pulses with pulse width and amplitude that could be modulated individually for each pad in the range of $50\text{--}1000 \mu\text{s}$ in steps of $10 \mu\text{s}$ and $100\text{--}10000 \mu\text{A}$ in steps of $100 \mu\text{A}$, respectively. The stimulation frequency was common to all pads and could be adjusted from 1 to 400 Hz in steps of 1 Hz.

Before mounting the equipment, the skin was cleansed using alcohol wipes and a small quantity of conductive and abrasive paste (everi, Spes Medica, IT). Three sizes of recording and stimulation electrodes were produced to accommodate different forearm sizes in line with the anthropometric variations expected for the forearms of healthy non-disabled subjects. The electrodes were placed circumferentially and, for each subject, the size which resulted in the smallest distance between the outermost electrode pads was selected. The recording electrode was positioned on the right forearm proximal to the elbow and the reference pads were placed on the elbow condyles and olecranon. The stimulation electrode was placed next and distally to the recording electrode. The two centrally located pads were aligned with the radial bone on the dorsal side of the forearm. Two soft Velcro straps were used to tighten the stimulation and the recording electrodes and the MaxSens device was attached to the strap. To ensure

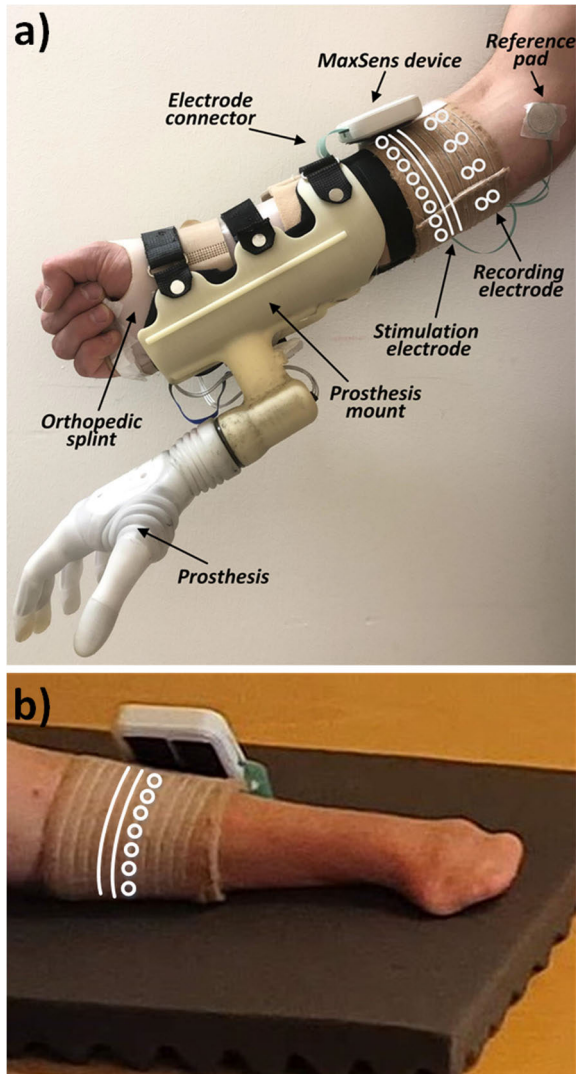


Fig. 1. Image of the closed-loop setup for **a)** non-disabled subjects and **b)** amputee. The *Stimulation electrode* and *Recording electrode* are indicated by white markings (stimulation: circular pads with one large, elongated pad; recording: circular pad pairs) under the Velcro straps. The *Reference pads* were placed on the elbow condyles and fixated with sports tape. The electrodes were connected to the *MaxSens device* via the *Electrode connector*. The non-disabled subjects wore an *Orthopedic splint* under a *Prosthesis mount* to which the *Prosthesis* was connected. The amputee subject wore only the *Stimulation electrode* and the *MaxSens device* attached to the Velcro strap.

isometric contractions, an orthopedic splint was worn by the non-disabled subjects. A hand prosthesis with an active wrist (Michelangelo hand from Otto Bock) was attached to a 3D-printed mount, and the splint was placed within the mount and secured with Velcro straps. MaxSens system and Michelangelo Hand were connected to a laptop PC via Bluetooth. The prosthesis control and the experimental task were programmed in MatLab 2019b (MathWorks, USA).

The range of motion for the wrist rotation was from -160° to 160° with a maximum speed of approx. 3 s for the full rotation. The range of motion for the hand aperture in the palmar grasp was ~ 11 cm and the closing from the fully open position at the maximum speed lasted approx. 0.4 s. The hand closing was therefore much faster compared to wrist rotation and the closing speed was thus limited to 40% of the maximum in the present study. The maximum grasping force was 70 N.

C. Myoelectric Control

The prosthesis control was implemented using linear regression where the 2 DoFs, wrist rotation and hand aperture, were controlled by performing four movement classes, namely, supination/pronation and opening/closing of the hand, respectively. The EMG signal was acquired by the MaxSens device and transmitted to the PC, which computed regression outputs and sent the velocity commands back to the prosthesis.

The EMG was sampled at 1 kHz and segmented into windows of 200 ms without overlap [33], [34]. As observed in [32], some stimulation artifacts still leaked into the EMG recordings despite the blanking, and they were suppressed by post-processing. Specifically, a sliding 10-sample Hampel filter was applied along each window to remove EMG outliers. Additionally, the samples that were four times larger than the standard deviation of the window were eliminated and the segments were filtered with a 50-Hz notch and a comb filter at the current stimulation frequency. Finally, the processed EMG signal was band-pass filtered using a 2nd order Butterworth filter with the cut-off at 15 Hz and 250 Hz.

Multiple linear regression models were trained for simultaneous velocity-based proportional control of two DoFs [33], [35], [36]. In the evaluation task, the subject had to flex the elbow to grasp an object. Therefore, all training data were recorded with the elbow approx. 60° flexed. For each of the four movement classes, a maximum long-term (15 s) voluntary contraction (MLVC) was recorded [37]. To collect the training data, the subjects were asked to perform the indicated movement and track the reference profile shown on the computer screen. They controlled a virtual cursor moving horizontally with time and vertically in proportion to the contraction intensity, estimated by computing the mean absolute value of windowed EMG across all channels. The subjects traced a trapezoidal trajectory with a 3 s incline, a 5 s plateau at 40, 50, and 70% MLVC, and a 3 s decline. In addition, a 15-s recording of the rest class was acquired while the subject was relaxed. To obtain a stable rest despite the stimulation artifacts that leaked into the EMG, the stimulation was delivered during the recording as recommended in [32]. The pairs of pads were activated along the electrode every 2 s at the localization intensity [38], and with the pulse width and frequency set to 400 μ s and 35 Hz, respectively. A multiple linear regression model was fitted for each of the four movements. The mean absolute value of the EMG was computed from each channel in 200-ms windows and used as input into the model. The regression output was the normalized velocity command for each DoF (i.e., 0 – no movement, 1 – maximum velocity). The subjects were informed that they could control the two DoFs simultaneously, but they were not required to do so.

D. Electrotactile Feedback Designs

Two encoding schemes conveying all DoFs of the Michelangelo hand (wrist rotation, hand aperture, and grasp force) were designed. In both schemes, to achieve an intuitive mapping, the proprioceptive feedback (rotation and aperture) was conveyed by moving the location of stimulation (spatial encoding), while the grasping force was mapped to a change in amplitude and frequency (parameter modulation). When the hand was

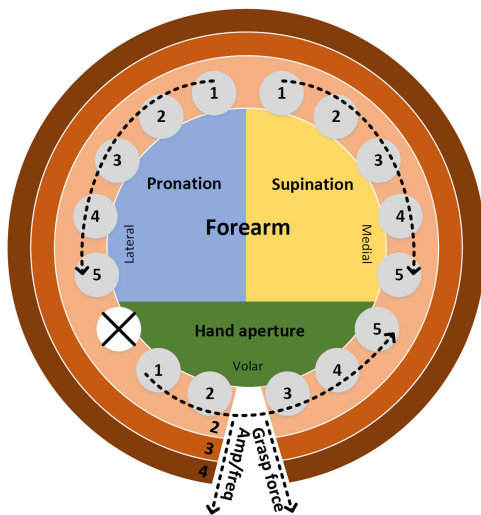


Fig. 2. The sectorized feedback encoding. The electrode array is divided into two sections communicating wrist rotation (dorsal side – blue/yellow area) and hand aperture (volar side – green area). To indicate pronation and supination, the active dorsal pad moves counterclockwise and clockwise, respectively, and for hand aperture the volar pad moves counterclockwise. One pad is unused. Force levels 2, 3 and 4 (red outer rings) are conveyed by concurrently increasing both stimulation amplitude and frequency of the active pad(s).

horizontal and fully opened, no stimulation was provided (reference position).

In the first approach (sectorized encoding), the feedback variables were mapped to the dedicated sectors of the electro-tactile interface [30], [31]. To effectively employ the spatial resolution of the electrode array (16 individually controllable pads), the range of motion of different DoFs (aka feedback variables) was divided into equal intervals to provide discrete feedback. Specifically, the motion ranges of wrist rotation and hand aperture were divided into 11 and 6 levels, respectively, reflecting the relative size of each motion. As illustrated in Fig. 2, 10 dorsal pads were associated with rotation, while 5 volar pads provided hand aperture. To convey pronation, the active pad moved counterclockwise by starting centrally and moving laterally to indicate higher rotation angles, and vice versa for supination. Thus, the stimulation elicited a sensation on the forearm that moved congruently with the rotation of the prosthesis wrist. The hand aperture was conveyed by moving the active pad counterclockwise across the volar side of the forearm.

In the second approach (coupled encoding), the encoding schemes of the two DoFs were “coupled” so that the elicited sensations mimicked the movement of both DoFs of the prosthesis (wrist and fingers). Specifically, the two active pads rotated around the forearm to convey the wrist rotation, counterclockwise for pronation and clockwise for supination, while they separated and moved towards each other on the other side of the forearm to indicate hand closing (Fig. 3). To match the number of levels conveyed for each feedback variable in the sectorized scheme (i.e., 11 for rotation and 6 for aperture), the active pads moved one position per level for the first rotation level and last three aperture levels, while otherwise, they moved two positions along the electrode. An additional reason for the larger spatial “jumps” in the first two aperture levels was to indicate more clearly that

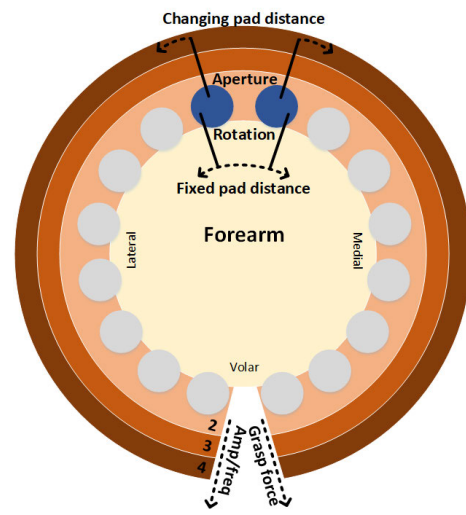


Fig. 3. The coupled feedback scheme. The movement of the two active pads indicate the prosthesis position. The pads rotate together in the counterclockwise/clockwise direction to convey pronation/supination (i.e., distance between the pads fixed). The hand closing is represented by the movement of the two pads towards the other side of the forearm (i.e., distance between the pads changes). Force is conveyed as in the sectorized feedback scheme.

the hand started closing. An example showing the pattern of pad activations associated with the corresponding prosthesis movements is provided in Fig. 4. Although the figure displays the feedback during the sequential movements of the DoFs, similar patterns would be produced during the simultaneous motion of both DoFs.

In both feedback configurations, the grasping force information was provided in 4 levels by modulating the frequency and amplitude of the active pads. The stimulation was delivered at 35 Hz when no force was applied to produce a clear sensation. Object contact and force level 1 was indicated by activating 6 dorsally placed pads for 200 ms. Force levels 2, 3, and 4, were conveyed by increasing the stimulation frequency and amplitude to 50, 65, and 80 Hz and 1.1, 1.2, and 1.3 \times base amplitude (localization threshold), respectively. It has been previously suggested that the simultaneous modulation of both parameters can increase the subjects’ ability to discriminate the elicited sensations [39]. The maximum frequency was set to 80 Hz to avoid the excessive loss of EMG data due to blanking.

E. Experimental Protocol

The experiment was divided into two sessions conducted on two consecutive days. The first session was the training of the encoding schemes while the second was the evaluation of the feedback in a functional closed-loop control task. Specifically, three conditions were tested to assess the subjects’ performance when using sectorized and coupled feedback as well as without electrotactile feedback, respectively. The sessions were approximately 2 and 3 h long, respectively.

1) Feedback Training: The first session was devoted to feedback familiarization and training, and thus, only the stimulation electrode was placed around the subjects’ forearms. The pulse width and frequency were set to 400 μ s and 35 Hz, respectively. The localization threshold, defined as the intensity that is clearly perceivable and yet well localizable below

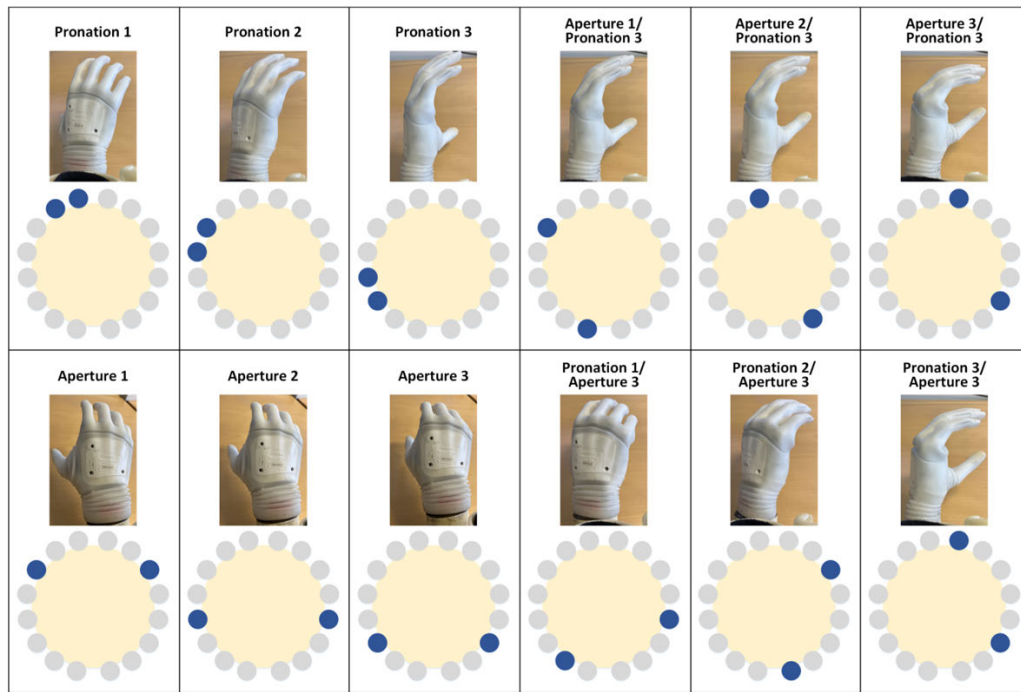


Fig. 4. Examples of the feedback (pad activation patterns) generated using coupled encoding in response to the sequential prosthesis movements. Blue and grey denote active and inactive pads, respectively. The prosthesis reaches the same end position in the two examples (rows) but the order of the DoFs activations is different (rotation and closing in the first row, and oppositely in the second row).

the pad [38], was determined for each pad using the ascending method of limits [40]. The subjects were informed that the test had started, the stimulation was activated starting from 0.5 mA and the stimulation amplitude was increased in steps of 0.1 mA every second. The subjects were asked to report when the localization threshold had been reached. The amplitude was subsequently adjusted to obtain a similar intensity across pads. This was performed by activating the pads sequentially and then asking the subject if the amplitude of some pads was very different from the rest. In this case, the amplitude was slightly increased or decreased.

The proprioceptive feedback was then trained using the same procedure for each scheme, and the schemes were presented in random order. The encoding method was first verbally explained using Fig. 2 or 3 for visual reference. The subjects then underwent familiarization and reinforced learning [31], [32], [38]. During familiarization, the subjects received tactile feedback while observing the movement of a virtual prosthesis (Fig. 5) to associate the elicited sensations with the prosthesis state. The virtual prosthesis moved across the full range of motion for each DoF individually as well as for all DoF combinations. In the reinforced learning, the virtual prosthesis moved from the reference (no stimulation) to a target position while the visual feedback was removed. The prosthesis DoFs were activated sequentially, and the subject was asked to determine the target position by focusing on the electrotactile feedback. The target position was then disclosed to the subject (visually and verbally). Two blocks of 32 “non-trivial” target positions were completed. More specifically, this included all combinations of levels in the two DoFs, as defined in the Methods section, *Electrotactile feedback designs*, excluding neutral and end positions (i.e., neutral wrist, fully opened and closed hand, fully pronated and supinated wrist). Finally, the force feedback was

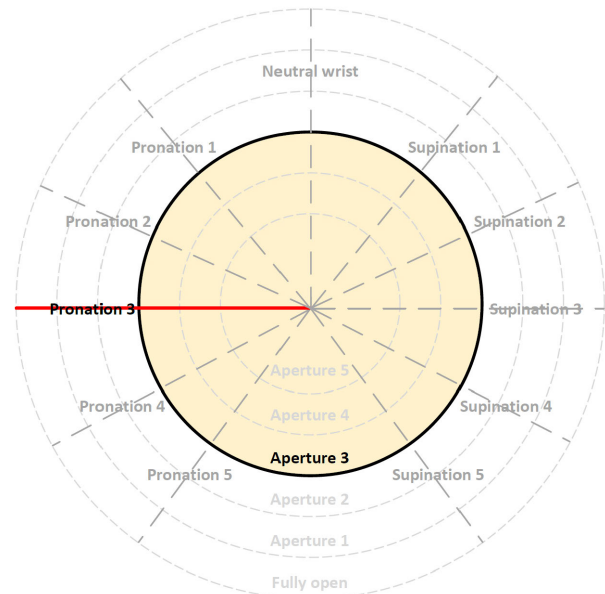


Fig. 5. The visual feedback used during feedback training, closed-loop control training and evaluation tests showing the target position from Fig. 4 (level 3 of pronation and aperture). The size of the yellow circle indicates the level of hand aperture, and the angle of the red line indicates the wrist rotation. The dashed circles and lines are only showed here for representation, as only the current/target levels were displayed during the training/evaluation tests.

explained and demonstrated to the subjects, but no reinforced learning was conducted. This step was left out to reduce experiment duration as the training was deemed less relevant in this case since the encoding was rather simple compared to that used for aperture and rotation. Before removing the stimulation electrode, the electrode outline was marked on the skin to reproduce the same electrode positioning in the second session.

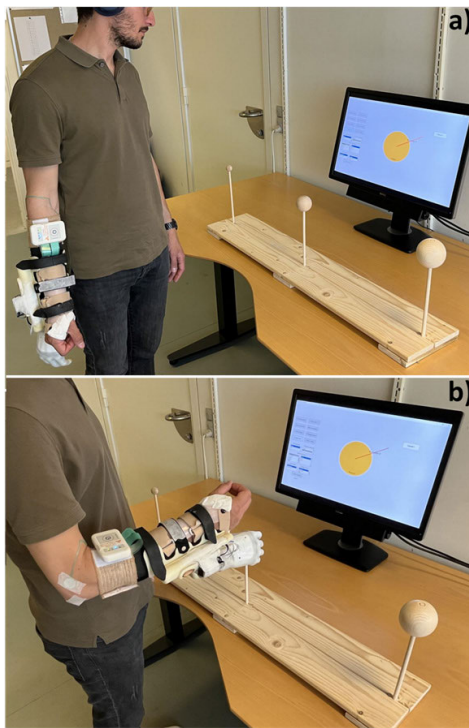


Fig. 6. Experimental setup for the **a)** position and **b)** force control phase of the evaluation test. In the position phase, the subjects adjusted the wrist rotation and hand aperture to reach the target position shown on the screen, while the prosthesis was held next to the body. In the force control phase, the subjects reached for one of the wooden balls, grasped it, and generated the target force indicated on the screen.

2) Closed-Loop Control Assessment: The second session was the assessment of performance during a functional task, and thus, the full setup was mounted on the subjects, as described before. The stimulation parameters were set to the values determined in the first session and adjusted if required. The myoelectric control was then calibrated following the procedure described in the Methods section, *Myoelectric control*.

Before running the functional task, the subjects briefly practiced prosthesis control. The subjects then performed the main experimental task in each feedback condition with the order of conditions pseudorandomized across subjects. Each feedback scheme was first trained in combination with prosthesis control for 5-10 minutes. When the subject was confident with the closed-loop control, the evaluation started. The task for the subject was to adjust the hand to match 32 “non-trivial” target configurations presented in random order and with a 2-minute break after each set of 8 trials to avoid fatigue. Each trial comprised a position and force control phase, and before each trial, the prosthesis was automatically reset to the reference position (hand horizontal and fully opened). During position control, the arm with the prosthesis was relaxed next to the body so that the prosthesis was not visible to the subject while the target position was shown on the computer screen (Fig. 6a). The subject was then asked to control wrist rotation and hand opening/closing to bring the prosthesis into the target position with electrotactile feedback activated or deactivated, depending on the tested condition. When the subjects deemed that the correct position was reached, they indicated this verbally, and the force control phase started by showing the target force on the monitor. The subjects were required to reach

the indicated force level by grasping one of three rigid wooden balls (Fig. 6b) produced to fit into the respective target hand aperture levels (i.e., fully open and level 1 for the largest ball, levels 2 and 3 for the medium ball and level 4 and 5 for the smallest ball). If the subjects generated an incorrect aperture, too small to grasp the target ball, they were instructed to open the hand to match the ball size and then perform the grasp. To maintain the flow of the task, the subject was not penalized if accidentally selecting the incorrect ball or rotating while grasping. The spherical shape (ball) was selected so that the object could be grasped with the hand in any orientation. When the subjects deemed that the correct force was reached, they indicated this verbally, and the trial ended. During the evaluation, the subjects wore headphones playing brown noise to mask the sound of the prosthesis motors. Thus, when electrotactile feedback was not provided, the subjects could not rely on the prosthesis sound, but they could still feel vibrations through the mount as well as rely on feedforward estimation (e.g., the duration of contraction/prosthesis movement). In this condition, therefore, they only received limited incidental feedback. After the experiment, the subjects were asked which of the feedback conditions they preferred (coupled, sectorized, or incidental feedback).

The amputee subject followed a similar experimental protocol. However, it was not possible to place the EMG electrode as the residual limb was too thin even for the smallest electrode size. Therefore, the prosthesis was placed on a table, secured by a vice, and the amputee subject controlled the prosthesis using a joystick where the left/right and up/down push of the stick corresponded to the wrist supination/pronation and hand opening/closing. The tests were conducted during a single session of 2.5 hours comprising familiarization, reinforced learning, closed-loop control training, and evaluation using coupled feedback before repeating the same steps using sectorized feedback. The incidental feedback condition was omitted since the subject did not wear the prosthesis.

F. Data Analysis

The main outcome measures were the positioning and force errors, computed as the difference between reached and target levels, and the time spent completing the position and force adjustments, respectively. For the reinforced learning phase, only the errors were calculated by averaging the two blocks while both outcome measures (time and error) were computed for the functional closed-loop control task. The overall average was then calculated for each subject in each test condition. The amputee subject was not included in this analysis and her results were shown separately as a case study. A Lilliefors test showed that the data were normally distributed in all cases except for rotation error in the functional task with the coupled feedback and reinforced learning phase with sectorized feedback. Therefore, repeated measures ANOVA and Friedman tests were used for multiple comparisons, and if a statistically significant difference was found, the post hoc pairwise comparisons were performed using paired t-test or Wilcoxon sign rank test with Bonferroni correction. The paired t-test and Wilcoxon sign rank test were also applied to compare the performance of the two feedback schemes in the reinforced learning phase. The significance level was set to $p < 0.05$. The reported p-values are after Bonferroni

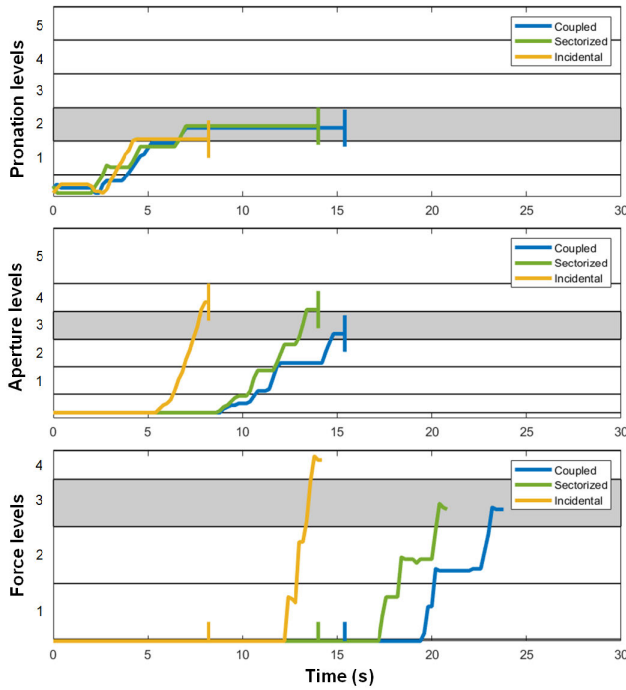


Fig. 7. Illustration of representative trials from each condition from an example subject. The blue, green and yellow lines represent the pronation (top), aperture (middle) and force (bottom) progression over time in the coupled, sectorized and incidental feedback conditions, respectively. The vertical lines in each condition represents termination of position phase. White and gray areas indicate feedback levels and targets, respectively.

adjustment. Despite some data being normally distributed, for consistency, all results are shown using boxplots and reported in the text as median (interquartile range).

III. RESULTS

A. Representative Trials

Representative position and force trajectories generated by an example subject performing the functional task in each condition are shown in Fig. 7. In all the conditions, the subject moved the prosthesis DoFs sequentially, by first adjusting the rotation and then the hand aperture. The hand aperture and wrist rotation trajectories were longer in the electro tactile feedback conditions compared to the incidental feedback condition. This implied more time spent adjusting the prosthesis with clear feedback-driven corrections before reaching the target. More specifically, the trajectories generated when using electro tactile feedback are composed of flat segments aligned with feedback levels and connected by slopes, showing how the subject moved the prosthesis from level to level, guided by the feedback. Such corrections were mostly absent in the incidental feedback condition, where the trajectories were continuous slopes, implying that in this case, the subject relied mainly on feedforward estimation (e.g., moving the prosthesis for a predefined duration). For force trajectories, the corrections (flat segments and slopes) were present in all cases, indicating that in the incidental feedback condition, the subject might have used visual and mechanical (vibrations) cues to adjust the force.

B. Summary Results

The average overall error when adjusting aperture, rotation, and force across conditions is summarized in Fig. 8. The aperture error (median(interquartile range)) when using coupled (0.76(0.34) levels) and sectorized feedback (0.67(0.38) levels) was significantly lower ($p < 0.01$) compared to incidental feedback (1.19(0.44) levels). Similar results were obtained for the rotation error, where coupled (0.31(0.44) levels, $p < 0.01$) and sectorized feedback (0.44(0.31) levels, $p < 0.05$) significantly outperformed the incidental feedback (0.81(0.56) levels). Regarding the force error, however, there was no significant difference between the two electro tactile feedback conditions versus incidental feedback (coupled: 0.34(0.46) levels, sectorized: 0.45(0.34) levels, and incidental: 0.50(0.22) levels). The errors achieved when using the two feedback schemes were not significantly different in any of the tested DoFs.

The time to complete the position and force control phases is shown in Fig. 9. During the position control phase, the total time to adjust two DoFs (aperture and wrist) was significantly smaller with incidental feedback (11.4(4.1) s) compared to coupled (18.3(5.4) s, $p < 0.001$) and sectorized feedback (17.8(6.4) s, $p < 0.01$). In the force control phase, the difference between feedback and incidental feedback exhibited the same trend, but the difference was smaller and not significant. Again, similarly to the position control phase, there was no significant difference in performance between the two feedback encoding approaches.

The results from the reinforced learning phase of the feedback training are shown in Fig. 10. Contrary to the results achieved in the functional task, the two feedback schemes exhibited significant differences. Specifically, the sectorized feedback (aperture: 0.09(0.08) levels, rotation: 0.18(0.05) levels) significantly outperformed the coupled feedback in both DoFs (aperture: 0.34(0.19) levels, rotation: 0.48(0.33) levels).

When asked which of the feedback conditions they preferred, all subjects chose one of the feedback conditions, and 8 out of 10 selected the sectorized scheme.

The results for the amputee subject (see green markings in Fig. 8, 9 and 10) followed the same trend, and she also selected sectorized feedback as the preferred condition. All outcome measures were within the ranges obtained in non-disabled subjects apart from the time to adjust the force, where the amputee participant was somewhat faster.

IV. DISCUSSION

The experimental results demonstrated the feasibility and effectiveness of the developed solution for myoelectric control of a multifunctional prosthesis with full-state electro tactile feedback. Regression-based control combined with multi-variable electro tactile feedback provided significantly better accuracy when adjusting the prosthesis aperture and wrist rotation compared to the condition in which the feedback was not available. Importantly, there was no significant difference in performance between the two encoding schemes when performing the functional task, implying that both methods as well as the system as a whole are viable solutions for potential clinical application. However, the provision of feedback increased the time to perform the task, which is an effect that has been already demonstrated in the literature [31], [32], [41]. Therefore, introducing feedback entails a trade-off between

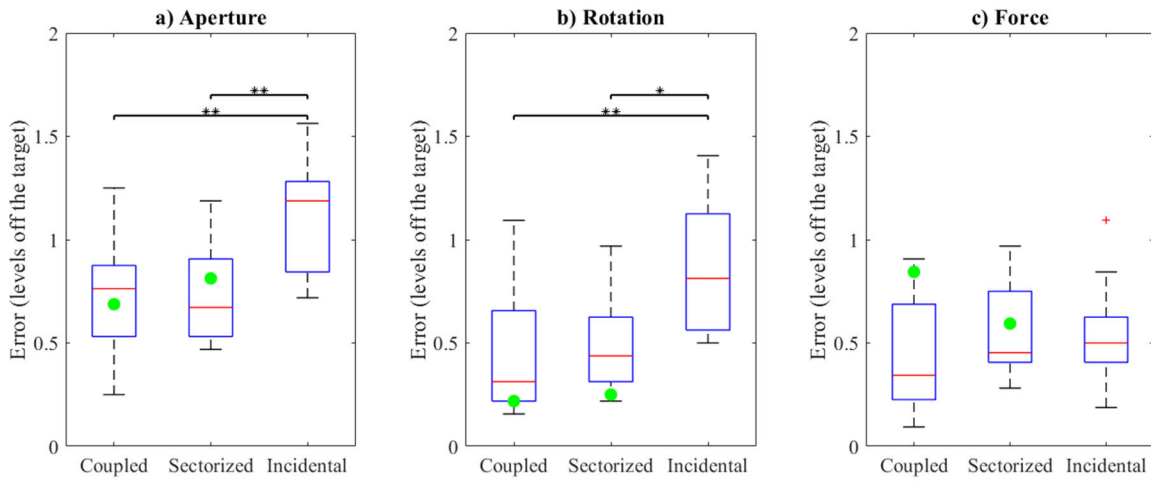


Fig. 8. Boxplot of errors in the functional task for each condition for **a)** aperture, **b)** rotation, and **c)** force. The red lines, blue boxes, black whiskers, and red marks represent median, interquartile range, maximum/minimum values and outliers. The green dots indicate the results of the amputee subject. Asterisks indicate significant differences in the post-hoc pairwise comparison (* is $p < 0.05$, ** is $p < 0.01$ and *** is $p < 0.001$).

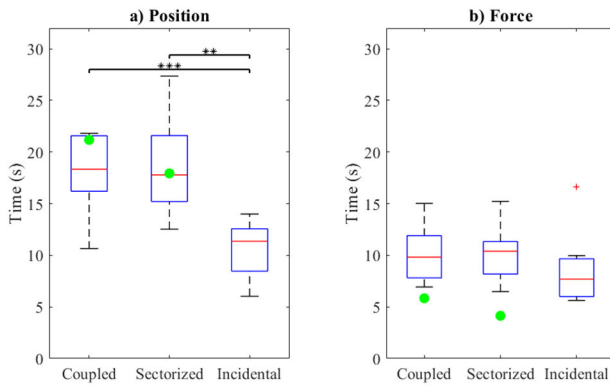


Fig. 9. Boxplots of time spend to reach the target in the functional task in each condition during **a)** position phase and **b)** force phase. The boxplot annotation is the same as in Fig. 8.

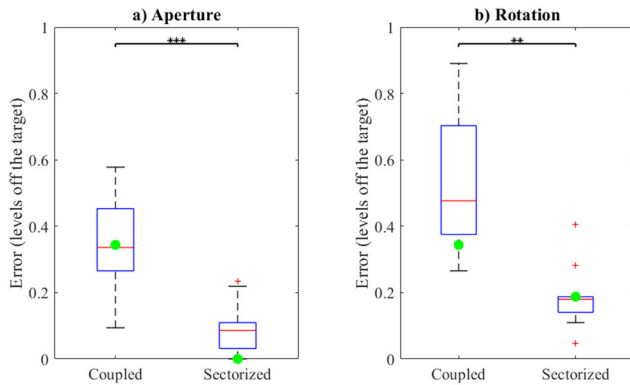


Fig. 10. Boxplots of error from the reinforced learning in the two feedback conditions for **a)** aperture and **b)** rotation. The boxplot annotation is the same as in Fig. 8.

improved accuracy and longer time [42], but the latter is likely to decrease following more extensive training. The main contribution of this work is the technical and perceptual feasibility, while short exposure to the feedback prevented assessing the sensorimotor integration. In essence, the subjects were asked to perform a novel task using a new channel of

feedback information, and it is natural to assume that they would attend to the feedback carefully, thereby increasing the time and cognitive load. The obtained insights are nevertheless valuable as this scenario corresponds to a naïve user interacting with the sensate prosthesis.

One of the key aspects of the present study is that all the main components, including the stimulation and recording interfaces as well as the prosthesis, were mounted on the same forearm. The proportional myoelectric control was robust and reliable, despite concomitant stimulation, the tight fit on the forearm and the fact that the subjects moved the prosthesis to reach and grasp the target object. Now, when the technical platform has been developed and assessed, and the anatomically congruent encoding (coupled scheme) verified as effective, we can proceed and evaluate the impact of closed-loop control on the sensorimotor integration aspects during prolonged and/or more challenging prosthesis use (e.g., introducing disturbance [43], intermittent feedback). Indeed, we assume that the true benefits of anatomically congruent feedback would be expressed in these paradigms, when the subjects have enough time to integrate the feedback into the internal representation of the motor task and develop internal models of prosthesis behavior [14]. The resulting performance will then reflect an interplay between feedforward and feedback control strategies [44]. In general, this is a promising venue for further research, especially in the context of artificial proprioception, where such processes were less investigated compared to controlling the grasping force [14]. However, addressing such questions requires dedicated studies and was out of the scope of the present effort.

As explained in the Introduction, previous studies that investigated multivariable feedback typically conveyed grasping force and hand aperture using parameter modulation (intensity and frequency) [17], [22], [23], [24], [25], [26], [27], [28], [31], [32] but also spatial encoding [25], [26], [27], [28], [31], [32]. Spatial and mixed encoding combining the change in location with parameter modulation has also been employed to provide single variable feedback (e.g., grasping force [30], [45], [46], wrist rotation [30], [47]). The novel schemes proposed in the present study combined a high-resolution

feedback interface and mixed encoding strategy to convey all the DoFs of a multifunctional prosthesis. Many pads (i.e., 16 independently addressable channels) allowed us to sectorize the electrode into segments allocated to individual DoFs, while still maintaining the desired feedback resolution. We have used a similar approach in our previous work [31] but with lower resolution (fewer feedback levels), without the prosthesis in the loop, and with contralateral recording and stimulation.

The coupled scheme, however, is an original approach tested for the first time in the present study. Similar encoding methods have been proposed in [30], but they have been tested only psychometrically, without prosthesis and closed-loop control, and only for individual DoFs. In the present study, those initial ideas were integrated into a novel approach that superimposed the stimulation patterns associated with individual DoFs to translate multi-DoF prosthesis movements into anatomically congruent tactile sensations. The main concern was, however, that such a scheme could be complicated to interpret as the final position of the active pads depends on both DoFs. For instance, different prosthesis motions can lead to the same final active pad configuration, which means that the subject needs to (at least partially) track the transitions between the pads to disambiguate the sensations and decode the prosthesis state. Such “overlap” does not exist in the sectorized feedback, but the sensations are not congruent with prosthesis motion. For instance, the sensation indicating finger position is always “projected” to the dorsal side of the forearm even when the wrist is rotated, and the fingers point upwards. Indeed, the reinforced learning indicated that it was more demanding to initially understand the coupled compared to the sectorized scheme. However, when used during closed-loop control to accomplish the functional tasks, the difference in performance with the two feedback approaches was not significant anymore, indicating that a short training was enough to equalize the subjects’ ability to decode the two mappings. In addition, some subjects reported that the coupled feedback had a more natural feel and would have likely been preferred with more training.

The coupled scheme relied on spatial encoding to convey the full state of the prosthesis. The same variables could have been represented using other encoding schemes, for instance, the intensity/frequency modulation through a single dedicated pad per variable as in [48]. This would reduce the number of channels but also disrupt the anatomical congruency between the feedback and prosthesis movement, which is a trademark feature of the coupled approach. An additional factor considered in our design was that the simultaneous activation of more electrode pads and the use of higher frequencies would cause longer and more frequent blanking intervals and hence less EMG signal per window, which would eventually jeopardize the quality of myoelectric control. Importantly, the proposed encoding could be applied even to dexterous hands, with individually controllable fingers and thereby multiple grasp types in addition to wrist rotation. The simplest approach would be to use the encoding as is, as every functional grasp is characterized by a changing aperture. However, to provide more meaningful feedback, the user could be informed about the active grasp by delivering a brief tactile icon [49] uniquely assigned to each grasp. Alternatively, such information could be easily integrated by expanding the presented encoding scheme. For instance, the

number of activated fingers could be conveyed by activating the same number of electrode pads (instead of only two pads, as in the present study). These considerations further underline that identifying technically feasible and yet effective feedback configurations that maximize user experience is indeed an important topic to be investigated further.

As the control was implemented using regression, the subjects could in principle move both DoFs simultaneously, but they preferred a sequential approach. This is likely due to a high cognitive effort that would be required for the simultaneous control in combination with feedback decoding. In addition, the feedback was trained using only sequential DoF transitions to facilitate interpretation, and this could have also contributed to the choice of the control strategy. In principle, simultaneous control could have been enforced by the experimental design, but such an approach could have overwhelmed the subjects. Nevertheless, if introduced gradually, we would expect that simultaneous control and feedback would indeed lead to the best performance, as already demonstrated for the control alone [50], [51]. The assessment of the simultaneous closed-loop control remains an important future step, which is however effectively enabled by the present work. Note that despite the control being sequential, the feedback always provided information about all three variables, which was interpreted successfully by the subjects. Nevertheless, the spontaneous choice of the conservative (sequential) control strategy highlights the challenge of establishing the user-prosthesis interaction that would be fully simultaneous (in both control and feedback). This likely requires the development of gradual training approaches to facilitate the adoption of the simultaneous closed-loop interface.

Adaptation to stimulation was not an issue in the present study due to the dynamic nature of the profiles and the limited duration of the experimental trial/session. However, during clinical use in daily life, the adaptation could be minimized by automatically turning off the stimulation when there is no change in the hand position and/or grasping force for a given time. This would avoid delivering prolonged constant stimulation when the feedback is anyways likely to be unnecessary (unchanged hand state). However, when the stimulation is then reactivated, it could be more difficult for the subject to interpret the feedback. In this case, they would need to recognize the “absolute” location of active pads, whereas during prosthesis control they can track the transitions between successive pads. The present experiment showed that the latter strategy was indeed effective as the median errors were less than 1 level (adjacent pads) for all DoFs. How well the subject would interpret the feedback when the level transitions are missing due to intermitted deactivation/activation of the feedback should be investigated in future studies. The results in the literature imply that absolute recognition might not be a challenge if the subject is properly trained [52].

The performance of the amputee subject followed the same trend as that of the non-disabled participants. However, it should be considered that the amputee subject used the joystick instead of myocontrol to move the prosthesis, and the former is a more reliable command interface. This inconsistency, however, has only a limited impact. The technical aspects of the developed closed-loop control system have been verified in non-disabled subjects and will function equally well

in amputees. Both sequential control using pattern recognition [53] and simultaneous control with regression [54] have been previously demonstrated in prosthesis users. Therefore, demonstrating the successful utilization of the novel feedback encoding approach was indeed the most relevant aspect in the context of the present study. As explained before, the actual sensorimotor integration including both control and feedback will be addressed in future work.

Contrary to controlling the position, the feedback was not beneficial when controlling the grasping force. In this task, the subjects moved the prosthesis in front to grasp the wooden ball and were, therefore, able to see the prosthesis. It has been demonstrated before that the subjects can use incidental feedback (e.g., muscle proprioception, visual observation of prosthesis closing and grasping) to control the generated force [55], [56], especially when the number of target force levels is small as in the current study (see Fig. 7). Indeed, the results imply that the problem was not that the subjects could not exploit the feedback properly (e.g., due to a lack of training), but that they were already “too good” when using only incidental sources, as the median error was less than 0.5 levels. Increasing the number of levels would likely improve the relevance of feedback [57]. Although the feedback was beneficial for the position control, even in this case, the control without feedback was far from being “blind”. The accuracy was significantly worse than when receiving electrotactile feedback, but the median error was still around 1 for both DoFs. This is a rather remarkable result considering the number of target levels and the fact that in this task, the incidental feedback was reduced to the propagation of motor vibrations through the mount, as the vision and auditory feedback were not available. This insight reinforces the findings reported in the literature regarding the important role of incidental feedback in prosthesis control [58].

In most studies investigating proprioceptive feedback, the incidental sources were partially or fully blocked [25], [28], [59], and this study is no exception. Nevertheless, the task considered in the present experiment is relatable to a practical application, as it demonstrated that the subject could use the feedback to adjust the prosthesis configuration just before reaching to grasp an object and while the prosthesis was outside the field of view (e.g., adjusting the prosthesis while approaching a table to grasp an object). However, the times required to accommodate the adjustment of two DoFs were still rather long. The task of future research will be to establish if the control dexterity and speed will improve with prolonged use, and if the presumed benefits of the proprioceptive feedback (e.g., less visual contact, adjustment while reaching) can be elucidated using some of the recently proposed assessment methods (e.g., gaze tracking [8], [9], [60], [61], motion capture [8], [9], [62]).

V. CONCLUSION

In summary, the main contributions of the present study are: 1) the demonstration of the technical feasibility of an integrated system for multichannel closed-loop prosthesis control in the context of the functional task, and 2) the novel encoding scheme to convey full prosthesis state by generating sensations that are anatomically congruent to prosthesis motion. In addition, the results of the experimental

assessment showed that the developed scheme is as effective as the conventional approach (sectorized encoding) and more effective than incidental feedback. These results are important as they establish the new frontiers of human perception and interpretation of multichannel stimuli, contribute to our understanding of effective methods to encode multivariable information, and show the benefits of such feedback compared to incidental sources. As recognized in the literature [58], and also highlighted in the present study (see results for force control), the latter is not a trivial outcome as the intrinsic power of such incidental sources should not be underestimated. The present study is therefore an important step towards a fully self-contained solution for advanced feedback and control of a multifunctional prosthesis.

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