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Published in:
IEEE Transactions on Haptics

DOI (link to publication from Publisher):
10.1109/TOH.2020.2985962

Publication date:
2020

Document Version
Accepted author manuscript, peer reviewed version

This article has been accepted for publication in a future issue of this journal, but has not been fully edited. Content may change prior to final publication. Citation information: DOI 10.1109/TOH.2020.2985962, IEEE Transactions on Haptics

Closed-loop Control using Electrotactile Feedback Encoded in Frequency and Pulse Width

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***Abstract*— Sensory substitution by electrotactile stimulation has been widely investigated for improving the functionality of human-machine interfaces. Few studies, however, have objectively compared different ways in which such systems can be implemented. In this study, we compare encoding of a feedback variable in stimulation pulse width or stimulation frequency during a closed-loop control task. Specifically, participants were asked to track a predefined pseudorandom trajectory using a joystick with electrotactile feedback as the only indication of the tracking error. Each participant performed eight 90 s trials per encoding scheme. Tracking performance using frequency modulation enabled lower tracking error (RMSE: Frequency modulation: 0.27±0.03; Pulse width modulation: 0.31±0.05; p<0.05) and a higher correlation with the target trajectory (Frequency modulation: 83.4±4.1%; Pulse width modulation: 79.8±5.2%; p<0.05). There was no significant improvement in performance over the eight trials. Furthermore, frequency-domain analysis revealed that frequency modulation was characterized with a higher gain at lower error frequencies. In summary, the results indicate that encoding of feedback variables in the frequency of pulses enables better control than pulse width modulation in closed-loop dynamic tasks.**

***Index Terms*— Electrotactile stimulation, closed-loop control sensory feedback, sensory substitution**

I. INTRODUCTION

INCORPORATING tactile sensory feedback in human-machine interfaces has been proposed for a wide range of applications. In virtual reality, tactile stimulation is used to elicit a sensation of touch when interacting with virtual objects [1]. In systems for replacing vision and audition, visual cues captured by a camera or sound recorded by a microphone are translated into tactile stimulation profiles delivered to the belly, back or tongue [2], [3]. For guidance, tactile stimulation can indicate direction to move, while in training, it can be used as an augmented feedback to supplement the information received through natural extero- and proprioceptive sense [4]–[7].

An additional, particularly popular application of tactile feedback is in myoelectric prosthetic hands [8], [9]. Since tactile and proprioceptive sensory feedback is an integral part of motor control of biological limbs [10], [11], it is commonly assumed that prosthesis controllability [8], [9] and embodiment [12] can be improved by sensory feedback. However, only one commercially available prosthesis includes a system to partially replace such sensations using a single vibration motor [13].

For prostheses, it has been proposed to transmit grasp force by a proportional pressure applied by tactors on the residual arm [14], [15], thereby making the feedback more intuitive [16], [17]. Such modality matching, however, is not possible for all human-machine interfaces (e.g., replacing visual feedback). Instead, a more generally applicable method to transmit feedback information is sensory substitution [18]. Here, the relevant information about the system state is transmitted to the user by eliciting tactile sensations. The stimulation can be implemented using vibration motors [19] or low-intensity electrical stimulation [20]. In the latter, the stimulation can be delivered non-invasively using surface electrodes placed on the skin [21] or invasively by stimulating peripheral nerves [22], [23]. The feedback information is communicated by modulating the parameters of stimulation using a selected coding scheme. For example, the grasping force generated by a prosthesis can be coded through the stimulation intensity or frequency (intensity or frequency modulation, respectively), i.e., higher grasping forces are related to stronger or faster stimulation [24]. Typically, a prosthesis user learns to associate the properties of such sensations to the prosthesis state after a brief training [25].

However, the modulation schemes are selected heuristically in each application, based on the preference of a researcher, available equipment and/or limitations of the stimulation methods. For example, in case of eccentric vibration motors, the stimulation parameters are coupled and they cannot be modulated independently [26], whereas in electrical stimulation they can be freely adjusted. The coding schemes are rarely compared under same conditions to assess objectively which scheme is better. In the late 1970's, coding scheme efficiencies were compared using psychometric

This work was funded by the Independent Research Fund Denmark, case-numbers 8022-00243A (ROBIN) and 8022-00226B.

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Manuscript received September 3rd 2019.

assessments [27] or open loop tracking [28], [29], showing small and inconsistent differences between intensity and frequency modulation.

More recently, frequency and amplitude modulation was used to provide feedback on prosthesis grasping force using an implanted neural interface [30]. The two coding schemes were compared in a number of closed-loop tasks in two amputee participants operating a myoelectric prosthesis. In object location, shape and compliance recognition, and force control tasks, amplitude and frequency modulation resulted in similar performance. Nevertheless, the amplitude coding led to better control in the task with staircase force control, where the participants were required to increase or decrease the grasping force. This task comprised a simple modulation of only three distinct levels of force that were self-selected by the participants. Many applications of electrotactile feedback, however, may require a higher resolution.

To effectively implement electrotactile stimulation in prosthetics but also in other applications, the basic properties of closed-loop control using tactile input need to be investigated. To this aim, a standardized test bench with the human in the loop is required that can be used to assess the control independently of the constraints of a particular implementation (e.g., a specific prosthesis).

In the present study, a closed-loop compensatory tracking task was used to compare frequency and amplitude modulation schemes under same conditions in a pool of participants. The participants were asked to track a reference trajectory using a joystick while electrotactile feedback communicated the momentary tracking error using the modulation of amplitude or frequency. Contrary to the open loop tracking [28], which assesses only the quality of perception, for successful closed-loop tracking the participant needs to perceive, interpret and react to feedback with an appropriate control action. This task has been used routinely in the past to evaluate the quality of human manual control using visual feedback [31]. It has been also applied to investigate how properties of electrotactile feedback, namely precision [32] and dimensionality of the task [33], affect the quality of control. In the present study, this experimental paradigm was used to compare the control performance of amplitude and frequency modulation.

II. METHODS

A. Participants

Thirteen healthy able-bodied participants (5 males and 8 females; age: 29 ± 7 yrs.) were recruited for the experiment. The experiment was conducted according the declaration of Helsinki and all participants signed informed consent forms prior to the experiment. The protocol was approved by The North Denmark Region Committee on Health Research Ethics (N-20160021).

B. Experimental Setup

The experimental setup consisted of a joystick (APEM HF22X10U) and a multichannel electrical stimulator (TremUNA, UNA systems) connected to a PC via USB. The joystick allowed the movement around two axes but only single axis (left – right) has been used in the present experiment. The stimulator is fully programmable and includes eight channels that generate current-controlled biphasic compensated pulses. The pulse width and amplitude can be independently controlled for each channel, while the pulse rate is a common parameter. Two channels were used in the present study and the stimulation was delivered via two self-adhesive concentric electrodes (Spes Medica, 50mm x 50 mm) positioned on the forearm of the dominant hand. One electrode was positioned on the ventral side of the forearm halfway between the elbow and the wrist. The other electrode was positioned on the dorsal side, one third of the length of the forearm distally from the elbow. These positions were selected through pilot tests to provide the feedback intuitively, as explained later, while generating clear sensations on both sides of the arm. The stimulation parameters were controlled in real time by using a toolbox for closed-loop human manual control [34] running on a standard PC. The same toolbox was used to implement the compensatory tracking task with visual and electrotactile feedback. The visual feedback was provided on a computer screen, as explained in the next section.

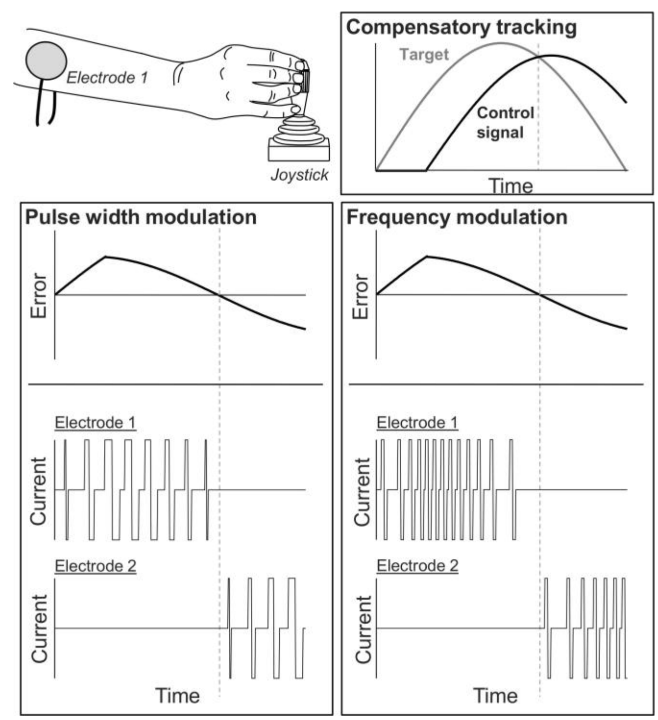

Fig. 1. Simplified representation of the experimental setup. Using a joystick, the participant attempted to match a predefined target (here illustrated as a single sine wave for simplicity, whereas the sum of multiple sine waves was used in the experiment). The output of the joystick (control signal) was proportional to the position of the joystick in one degree of freedom (left/right). The error (difference between target and control signal) was signaled to the participant using electrotactile feedback via two concentric electrodes positioned on the lower arm in two different ways: Pulse width modulation, where the magnitude of the error was proportional to the pulse width (frequency: 70 Hz) and frequency modulation where the error was proportional to the stimulation frequency (pulse width: 80% of pain threshold). The two electrodes signaled positive and negative errors, respectively. Note that the illustrations of the stimulation pulses are not in scale.

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C. *Experimental Protocol*

The participant was seated comfortably in a chair in front of a table with a computer screen placed approximately 50 cm from the participant. The skin on the forearm was cleaned and moisturized by a wet cloth, and the stimulation electrodes were placed. Next, the detection and pain thresholds for the stimulation pulse width were determined using the methods of limits [35]. The pulse width was incremented in steps of 50 us while the frequency and amplitude were set to 70 Hz and 3.5 mA, respectively. For each electrode, the thresholds were determined three times and the average value was then computed.

Figure 1 summarizes the experimental task. The participant was asked to track a predefined target trajectory with a control signal that reflected the movement of the joystick in a proportional way. The target trajectory was a pseudorandom multi-sine obtained by the summation of 9 sine waves with random phases. This type of reference trajectory is often used for the assessment of human manual control [36]. The frequencies of the sine waves were selected by logarithmically dividing the range between 0.2 and 4 rad/s. The trajectory was constructed by repeating the same 30-s segment three times, hence 90 s in total. The basic 30-s segment was generated anew in each trial and it was long enough for the participants not to notice the repetition and therefore experience the trajectory as essentially random. The amplitudes of the five sine waves with lowest frequencies (≤1 rad/s) were twice as high as the amplitude of the other four sine waves. The highest frequencies were attenuated in order to decrease the task difficulty while still maintaining the desired signal bandwidth. The amplitude of the target trajectory was normalized to the range [-0.9, 0.9]. In the compensatory tracking, the task of the participant was to track the desired reference trajectory using a joystick. The feedback transmitted to the participant provides the momentary tracking error, which the participant needs to compensate (nullify). This paradigm ensures that the participant relies on feedback-driven control. Since the reference trajectory is pseudorandom and the participant does not “see” the reference but only the momentary deviation, he/she cannot use anticipatory control. Therefore, this task is designed to objectively measure the participant’s reaction to the electrotactile stimulus. This also allows identifying the frequency profile and the transfer function of the human controller, as explained below.

First, the participant was familiarized with the tracking task by providing visual feedback on a computer screen. A green marker deviated from the red vertical line proportionally to the normalized tracking error. If the tracking error reached extreme values of 1 and -1, the marker would hit the limits of the tracking area. The participant was instructed to move the joystick in order to cancel the “disturbance” (tracking error) and maintain the green marker, as good as possible, at the reference line. To this aim, the participant was supposed to move the joystick in the opposite direction to the movement of the marker proportionally to the magnitude of the deviation (so called position-controlled system). Each participant performed two trials with visual feedback.

In the next step, the electrotactile feedback reflecting the tracking error was provided via the two electrodes simultaneously with the visual feedback. The aim was to teach the participant to properly interpret the elicited tactile sensation. In electrotactile feedback, the two electrodes communicated the sign (electrode) and magnitude (stimulation intensity/frequency) of the normalized tracking error. The activation of the electrode positioned on the ventral side of the forearm indicated negative tracking error, while the electrode on the dorsal side signaled positive tracking error. The magnitude of the normalized error in the range [-1, 1] was linearly encoded by modulating pulse width (PM) or frequency (FM) of stimulation. For PM, the pulse width was modulated in the range from the detection to the pain threshold while the stimulation frequency was fixed at 70 Hz. During FM, the pulse width was set to 80% of the pain threshold, while the frequency was changed in the range between 7 and 63 Hz. While tracking with simultaneous electrotactile and visual feedback, the participants learned to relate the movement of the marker to the pattern of electrical stimulation delivered through two electrodes, as explained above.

After the training trials, the participants started closed-loop tracking using only electrotactile feedback. The strategy for successful tracking was similar to that during visual feedback: if the participants felt the stimulation on the right (left) electrode, they were instructed to move the joystick on the opposite side with the magnitude of the joystick movement proportional to the intensity (PM condition) or frequency (FM condition) of stimulation. During the experiment, the participant completed eight validation trials per condition without visual feedback. The order of the encoding schemes was randomized. Before starting the eight validation trials for each condition, two training trials with visual and electrotactile feedback with the relevant encoding scheme were given (see details above). Between each trial and between the conditions a break of at least 1 minute and 5 minutes was given, respectively.

D. *Data Analysis*

The main outcome measures in the time domain were correlation coefficient (peak of cross-correlation function), time delay (delay of peak of cross-correlation function) and root mean square error (RMSE). The outcome measures were calculated for each trial. First, the time delay between the generated and reference trajectory has been estimated, and then, the correlation coefficient and RMSE were computed between the time-shifted generated trajectory and the target trajectory. This was done to eliminate the impact of the time delay on the RMSE. The data were normally distributed, as determined using Shapiro-Wilk test. Therefore, the outcome measures were analyzed using two-way ANOVA (factors: trial and modulation scheme). The level of significance was set to $p < 0.05$. Effect size was estimated using partial η^2. 0.01, 0.06 and 0.14 were adopted as lower limits for small, medium and large effect size, respectively.

Furthermore, the frequency characteristics of the human controller were determined using the approach presented in [36], a method that is commonly applied to model human behavior during manual control tasks [37]. In this approach, the human is modeled as a quasi-liner system embedded within a closed-loop control scheme. Therefore, the frequency profiles $H(j\omega)$ describing the human controller are obtained as the ratio of the cross-spectrum of the reference trajectory and joystick movement (human output) to the cross-spectrum of the reference trajectory and the tracking error (human input):

$$H(j\omega)=\frac{S_{ru}(j\omega)}{S_{re}(j\omega)} \quad (1)$$

where r is the reference trajectory (pseudorandom multisine), u is the joystick movement generated by the participant during tracking, and e is the tracking error. In a more general setup, the joystick signal would be a command input into a dynamic system (e.g., an integrator) controlled by the participant. In the present study, however, the system is a "simple tracker", with no dynamic element, hence $e = r - u$. The term S_{xy} represents the cross-spectrum defined as $S_{xy}(j\omega) = X^*(j\omega)\cdot Y(j\omega)$, where $X(j\omega)$ and $Y(j\omega)$ denote Fourier transform of the signals x and y, and the star symbol indicates conjugation. A detailed explanation of the methodology can be found in [36].

This provides the estimation of the gain and phase of the human controller for each of the nine frequencies present in the target trajectory. The amplitude and phase diagrams were computed for each trial. To increase the reliability of estimation, the frequency profiles were computed for each 30-s segment of the trajectory and then averaged. Next, the profiles were averaged across all trials of an individual participant to estimate the overall, participant-specific gain and phase characteristics. The profiles were then pooled for PM and FM and compared between the conditions at each frequency using paired t-test. Finally, the transfer function of the human controller in the two encoding schemes was estimated from the frequency profiles across all participants. The form of the transfer function was assumed as a first-order system with a time delay:

$$G(s)=e^{-T_D\cdot s}\frac{K}{T\cdot s+1} \quad (2)$$

The parameters of the transfer function, i.e., gain (K), time constant (T) and time delay (T_D) were estimated using constrained nonlinear optimization, minimizing the difference between the analytic and experimentally obtained frequency profiles ("procest" function in Matlab).

III. Results

Figure 2 shows representative examples of compensatory tracking from the first and last trial in one participant using PM and FM, respectively. Although the performance improved with training for both modulation schemes, tracking based on FM was superior with respect to the correlation and RMSE between the target and the generated signal. With PM, in particular, the participant had difficulties in tracking the full range of the target trajectory, visibly undershooting the peaks of the trajectory. This tendency was confirmed by the average results across all participants (Fig. 3). The correlation was significantly higher for FM (FM: 83±4%; PM: 79±5%; $p<0.05$; partial $\eta^2=0.08$ (medium effect size)), while there was no significant training effect across the repeated trials ($p=0.22$) although the correlation increased by 4% and 8% for FM and PM, respectively. Furthermore, there was no interaction effect between these factors ($p=0.98$). Similarly, there was a significant effect of encoding scheme for RMSE (FM: 0.27±0.03; PM: 0.31±0.05; $p<0.05$; partial $\eta^2=0.15$ (large effect size)), but no effect of learning ($p=0.63$) or interaction ($p=0.99$). Finally, there was no difference across modulation schemes in time-delay ($p=0.52$; learning: $p=0.96$; interaction:

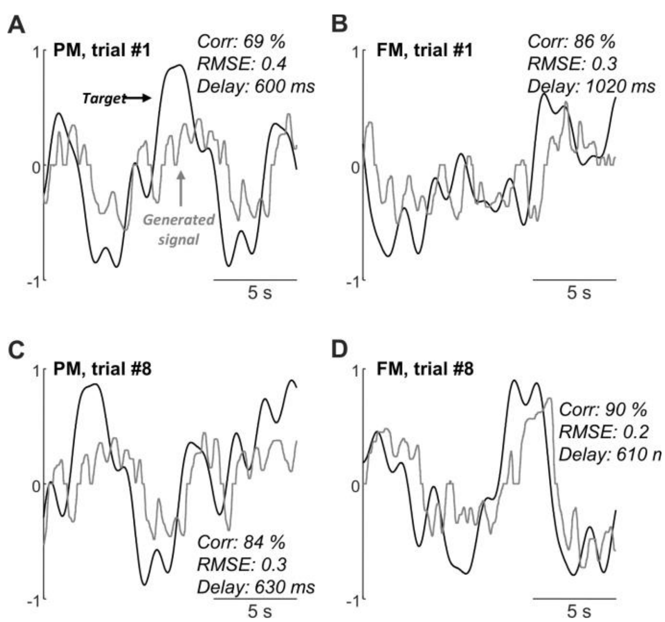

Fig. 2. Representative examples of tracking from the first (A, B) and 8th (last; C, D) trial of one participant for pulse width modulation (A, C) and frequency modulation (B, D). The black lines indicate the target and the grey lines the control signal. Note that the parameters (Correlation, RMSE and time delay) indicated in each panels reflect the full 90 s trial and not the 15 s shown here.

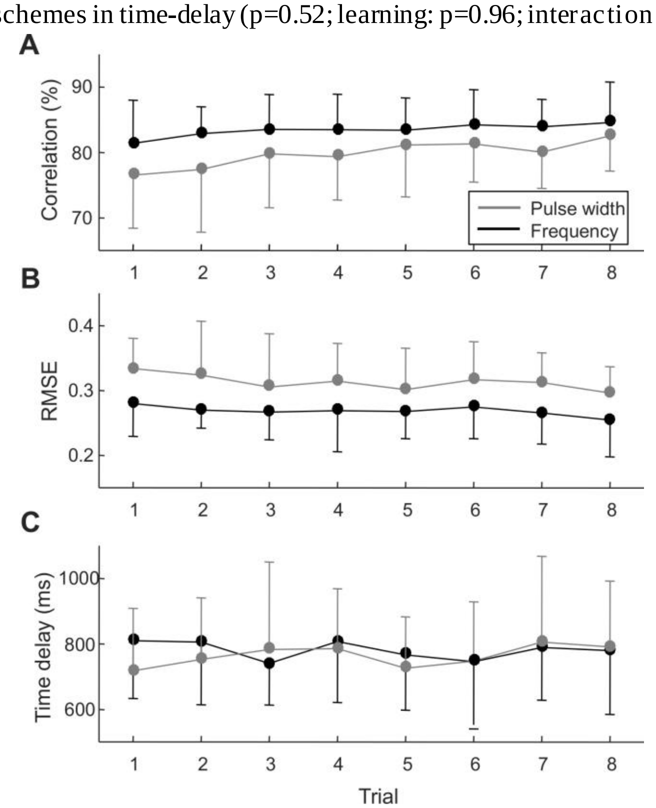

Fig. 3. Correlation (A), RMSE (B), and time delay (C) between target and control signal for pulse width modulation (grey lines/circles) and frequency modulation (black lines/circles) across the eight trials. Error bars (shown only in one direction for clarity) indicate standard deviation.

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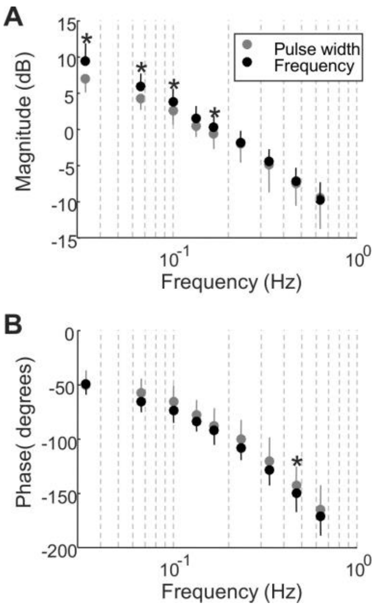

Fig. 4. Average transfer functions during pulse width modulation (grey circles) and frequency modulation (black circles) across all participants for all trials. Each circle represents the magnitude (A) and phase delay (B) for each of the 9 frequencies composing the target signal. Error bars (shown only in one direction for clarity) indicates standard deviation. * indicates significant difference ($p<0.05$).

$p=0.94$).

Figure 4 illustrates the average frequency characteristics of the human controller. Here, the magnitude in FM was significantly higher than for PM for four of the five lowest frequencies (Fig. 4A). Conversely, there was little difference for the phase delay across modulation schemes. The transfer function (Eq. 1) fitted to this data had the following parameters. T_D: 0.32 s (FM) and 0.28 s (PM); K: 4.0 (FM) and 2.9 (PM); T: 4.0 (FM) and 3.3 (PM). These values reveal a higher gain (K) for FM indicating better responsiveness across error frequencies for this encoding scheme.

IV. DISCUSSION

In this study, we have compared the quality of closed-loop control with electrotactile feedback encoded in stimulation pulse width and frequency, respectively. These schemes have been commonly investigated in the literature [8]; however, they are rarely compared objectively in the same conditions. Here, we used the same control setup, namely, closed-loop compensatory tracking, where the tracking error has been communicated through two electrodes using PM and FM coding. The participants used the feedback to manipulate the joystick in order to compensate the error, hence the performance reflected not only feedback perception (as in open-loop tracking [28]) but also interpretation and feedback-driven motor action.

The results have shown that in a dynamic control task, FM significantly outperformed PM. More specifically, FM resulted in better tracking of the reference trajectories in both shape (correlation) and absolute magnitude (RMSE). In the frequency domain, this could be seen through the higher gains in the frequency characteristics and equivalently through the higher gain of the fitted transfer function. The timing parameters, on the other side, were similar for the two conditions. There was no significant difference in the overall time delay between generated and reference trajectory, transport delay of the human controller and time constant of the transfer function. Accordingly, the phase characteristics for FM and PM were largely overlapping.

These results are somewhat different from those reported in two previous studies [29], [30] where electrotactile stimulation was used for feedback restoration in hand prostheses. The studies slightly favored amplitude modulation, and this may be explained by a number of methodological differences between this and those studies.

First, regarding the task, the assessment in [29] was performed using open-loop tracking; hence the stimulation did not change interactively depending on the input from the participant as in the present study. In [30], the task included closed-loop prosthesis control, but the participants had to gradually increase and decrease the grasp force to reach three self-selected levels. This implies that the control task was slow (> 10 s per force level) and coarse (3 levels), unlike the present study that required continuous control. Such control arguably imposes higher demands on feedback information content, while it can be speculated that experienced prosthesis users would be able to produce three grasp force levels with reasonable accuracy even without feedback (feedforward control) and therefore did not depend exclusively on the feedback for task execution.

A second difference across the studies is the range of frequencies employed for FM. Whereas this study and [29] encoded the feedback in frequencies <80 Hz, [30] used frequencies up to 1 kHz. Frequencies above 100 Hz are more difficult to differentiate [28], likely since they produce fused sensation of similar quality, and are more prone to adaptation [20], [30]. This might be also responsible for the lower ability to discriminate the feedback levels using frequency versus amplitude modulation that was reported in [23].

On the other hand, the task in the present study is general and does not directly reflect a realistic application, like prosthesis control in [30]. Specifically, the objective to nullify the sensation (to minimize tracking error) does not reflect typical prosthesis control, although the ability to perceive low-level sensations is highly relevant when manipulating fragile objects with very low levels of force. In addition, the closed-loop compensatory tracking task is directly related to the application of tactile feedback for training and guidance. Here, the electrotactile stimulation is often used to communicate the deviation from the desired performance (e.g., limb position error) [38], [39]. Finally, during FM in [30], the stimulation intensity was set to the sensation threshold producing thereby a weak sensation. In the present study, on the contrary, the intensity during FM was set to 80% of the pain thresholds to generate a more robust sensation. Together, these observations and the results of the present study suggest that FM is superior when the control loop is closed, when the control is continuous in terms of time and target levels, and when the range of frequencies and carrier intensity are appropriately selected. And indeed, we have previously shown [40] that the carrier frequency has substantial impact on the effectiveness of amplitude modulation. In the present study, the carrier intensity could have been an important reason for the better control performance with FM. When using FM, the tracking error was always communicated using a clear sensation located between the sensation and pain thresholds, regardless of its magnitude. With PM, however, small errors were associated to weak sensations due to stimulation near the sensation threshold, which might have compromised the perception and interpretation of feedback. In some participants and trials, it could be observed that the participants consistently undershot the reference trajectory (see e.g. representative examples in Fig. 2A and 2C), which might be due to insufficiently clear perception. This could be particularly important in delicate tasks, for example, when grasping a sensitive object using small forces. To improve perception with PM, one could increase the lowest intensity, but this would also decrease the overall range available for feedback modulation.

An additional advantage of FM is that it is easier to set up and less likely to cause discomfort. For PM, both thresholds need to be determined. Ideally, the intensity should be changed between the two thresholds in order to exploit the full range of sensations for information transmission. This can lead to uncomfortable sensations when the intensity is high and thereby close to the threshold for pain and discomfort. The intrusiveness of stimulation might have also affected the quality of tracking with PM. Furthermore, if the values of the two thresholds are close to each other (i.e. discomfort/pain begins at pulse widths only slightly wider than those required for sensation), the range in which the tracking error can be encoded would be small, likely decreasing control accuracy. In case of FM, the only requirement is to set the intensity so that the elicited sensation is clearly perceived. Therefore, it is not necessary to establish the pain threshold, and the elicited sensation can be maintained well below the discomfort zone all the time during online control using a wide encoding-range of frequencies. Although we did not investigate sensory adaptation, we believe that it had little influence on any of the conditions in our experiment. Stimulation intensity varied continuously, and each electrode was rarely active for more than a few seconds at the time. Under such conditions, we would not expect sensory adaptation to be a concern [30] and we did not observe any indication of adaptation during the experiments.

In conclusion, this study showed that tracking error encoded in stimulation frequency enables superior control with respect to PM. Consequently, FM is likely the preferred encoding scheme for sensory substitution in human-machine interfaces.

The experimental paradigm used in the present study is rather general. In that sense, the results (RMSE and correlation) cannot be directly translated to a specific application (e.g., prosthesis grasping). Nevertheless, this approach can provide insights into the basic characteristics of closed-loop control using tactile feedback, which might be relevant across several applications [6], [7], [21]. And indeed, a similar approach has been used extensively in the past to investigate human control with visual feedback [31], [41]. A particular advantage of this experimental paradigm is that it can be used to explore different facets of tactile feedback by customizing the task parameters (e.g., frequency bandwidth of the reference trajectory, inserting system dynamics in the control loop).

References

[1] D. S. Pamungkas and K. Ward, “Electro-Tactile Feedback System to Enhance Virtual Reality Experience,” *Int. J. Comput. Theory Eng.*, vol. 8, no. 6, pp. 465–470, 2016.

[2] J. Loomis, R. Klatzky, and N. Giudice, “Sensory substitution of vision: Importance of perceptual and cognitive processing,” in *Assistive Technology for Blindness and Low Vision*, R. Manduchi and S. Kurniawan, Eds. Boca Raton, Fl, USA: CRC press, 2012.

[3] H. C. Stronks, E. B. Mitchell, A. C. Nau, and N. Barnes, “Visual task performance in the blind with the BrainPort V100 Vision Aid,” *Expert Review of Medical Devices*. 2016.

[4] M. Kitigawa, D. Dokko, A. Okamura, and D. Yuh, “Effect of sensory substitution on suture-manipulation forces for robotic surgical systems,” *J. Thorac. Cardiovasc. Surg.*, vol. 129, no. 1, pp. 151–158, 2005.

[5] D. Powell and M. K. O’Malley, “The task-dependent efficacy of shared-control haptic guidance paradigms,” *IEEE Trans. Haptics*, 2012.

[6] R. Sigrist, G. Rauter, R. Riener, and P. Wolf, “Augmented visual, auditory, haptic, and multimodal feedback in motor learning: A review,” *Psychonomic Bulletin and Review*. 2013.

[7] E. Van Breda, S. Verwulgen, W. Saeys, K. Wuyts, T. Peeters, and S. Truijen, “Vibrotactile feedback as a tool to improve motor learning and sports performance: A systematic review,” *BMJ Open Sport Exerc. Med.*, 2017.

[8] C. Antfolk, M. D’Alonzo, B. Rosén, G. Lundborg, F. Sebelius, and C. Cipriani, “Sensory feedback in upper limb prosthetics.,” *Expert Rev. Med. Devices*, vol. 10, no. 1, pp. 45–54, 2013.

[9] B. Stephens-Fripp, G. Alici, and R. Mutlu, “A review of non-invasive sensory feedback methods for transradial prosthetic hands,” *IEEE Access*, vol. 6. pp. 6878–6899, 2018.

[10] R. S. Johansson and K. J. grasping, “Sensory-motor coordination during manipulative actions,” *Curr. Opin. Neurobiol.*, pp. 2–815, 1992.

[11] J. C. Rothwell, M. M. Traub, B. L. Day, J. A. Obeso, P. K. Thomas, and C. D. Marsden, *Manual motor performance in a deafferented man.*, vol. 105 (Pt 3). 1982, pp. 515–542.

[12] P. D. Marasco, K. Kim, J. E. Colgate, M. A. Peshkin, and T. A. Kuiken, “Robotic touch shifts perception of embodiment to a prosthesis in targeted reinnervation amputees,” *Brain*, vol. 134, no. 3, pp. 747–758, 2011.

[13] “http://vincentsystems.de/en/prosthetics/vincent-evolution-2/,” *Vincent systems*. .

[14] K. Kim and J. E. Colgate, “Haptic feedback enhances grip force control of sEMG-controlled prosthetic hands in targeted reinnervation amputees,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 6, pp. 798–805, 2012.

[15] K. R. Schoepp, M. R. Dawson, J. S. Schofield, J. P. Carey, and J. S. Hebert, “Design and Integration of an Inexpensive Wearable Mechanotactile Feedback System for Myoelectric Prostheses,” *IEEE J. Transl. Eng. Heal. Med.*, 2018.

[16] S. Fani, K. Di Blasio, M. Bianchi, M. G. Catalano, G. Grioli, and A. Bicchi, “Relaying the High-Frequency Contents of Tactile Feedback to Robotic Prosthesis Users: Design, Filtering, Implementation, and Validation,” *IEEE Robot. Autom. Lett.*, vol. 4, no. 2, pp. 926–933, 2019.

[17] C. Antfolk *et al.*, “Artificial redirection of sensation from prosthetic fingers to the phantom hand map on transradial amputees: Vibrotactile versus mechanotactile sensory feedback,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, 2013.

[18] P. Bach-y-Rita and S. W. Kercel, “Sensory substitution and the human-machine interface,” *Trends in Cognitive Sciences*, vol. 7, no. 12. pp. 541–546, 2003.

[19] K. A. Kaczmarek, J. G. Webster, P. Bach-y-Rita, and W. J. Tompkins, “Electrotactile and vibrotactile displays for sensory substitution systems,” *IEEE Transactions on Biomedical Engineering*, vol. 38, no. 1. pp. 1–16, 1991.

[20] A. Y. J. Szeto and F. A. Saunders, “Electrocutaneous Stimulation for Sensory Communication in Rehabilitation Engineering,” *IEEE Trans. Biomed. Eng.*, vol. BME-29, no. 4, pp. 300–308, 1982.

[21] P. Svensson, U. Wijk, A. Björkman, and C. Antfolk, “A review of invasive and non-invasive sensory feedback in upper limb prostheses,” *Expert Review of Medical Devices*, vol. 14, no. 6. pp. 439–447, 2017.

[22] E. D'Anna *et al.*, "A closed-loop hand prosthesis with simultaneous intraneural tactile and position feedback," *Sci. Robot.*, vol. 4, no. 27, 2019.

[23] J. A. George *et al.*, "Biomimetic sensory feedback through peripheral nerve stimulation improves dexterous use of a bionic hand," *Sci. Robot.*, 2019.

[24] T. Rosenbaum-Chou, W. Daly, R. Austin, P. Chaubey, and D. A. Boone, "Development and real world use of a vibratory haptic feedback system for upper-limb prosthetic users," *J. Prosthetics Orthot.*, vol. 28, no. 4, pp. 136–144, 2016.

[25] M. Markovic *et al.*, "The clinical relevance of advanced artificial feedback in the control of a multi-functional myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, 2018.

[26] M. Azadi and L. A. Jones, "Vibrotactile actuators: Effect of load and body site on performance," in *IEEE Haptics Symposium, HAPTICS*, 2014, pp. 351–356.

[27] A. B. Anani, K. Ikeda, and L. M. Körner, "Human ability to discriminate various parameters in afferent electrical nerve stimulation with particular reference to prostheses sensory feedback.," *Med. Biol. Eng. Comput.*, vol. 15, no. 4, pp. 363–373, 1977.

[28] A. Y. J. Szeto and J. Lyman, "Comparison of codes for sensory feedback using electrocutaneous tracking," *Ann. Biomed. Eng.*, vol. 5, no. 4, pp. 367–383, 1977.

[29] A. B. Anani and L. M. Körner, "Afferent electrical nerve stimulation: human tracking performance relevant to prosthesis sensory feedback.," *Med. Biol. Eng. Comput.*, vol. 17, no. 4, pp. 425–434, 1979.

[30] G. Valle *et al.*, "Comparison of linear frequency and amplitude modulation for intraneural sensory feedback in bidirectional hand prostheses," *Sci. Rep.*, vol. 8, no. 1, 2018.

[31] D. Mcruer and D. H. Weir, "Theory of Manual Vehicular Control," *IEEE Trans. Man-Machine Syst.*, vol. 10, no. 4, pp. 257–291, 1969.

[32] T. R. Schori, "Tracking Performance as a Function of Precision of Electrocutaneous Feedback Information," *Hum. Factors J. Hum. Factors Ergon. Soc.*, vol. 12, no. 5, pp. 447–452, 1970.

[33] H. P. Schmid and G. A. Bekey, "Tactile Information Processing by Human Operators in Control Systems," *IEEE Trans. Syst. Man Cybern.*, vol. 8, no. 12, pp. 860–866, 1978.

[34] S. Dosen, M. Markovic, C. Hartmann, and D. Farina, "Sensory feedback in prosthetics: A standardized test bench for closed-loop control," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 23, no. 2, pp. 267–276, 2015.

[35] F. Kingdom and N. Prins, *Psychometrics: A practical introduction*. Academic Press, 2009.

[36] D. T. McRuer and E. S. Krendel, "The human operator as a servo system element," *J. Franklin Inst.*, vol. 267, no. 5, pp. 381–403, 1959.

[37] J. I. Elkind and C. D. Forgie, "Characteristics of the human operator in simple manual control systems," *IRE Trans. Autom. Control*, vol. AC-4, no. 1, pp. 44–55, 2013.

[38] E. Tzorakoleftherakis, T. D. Murphey, and R. A. Scheidt, "Augmenting sensorimotor control using 'goal-aware' vibrotactile stimulation during reaching and manipulation behaviors," *Exp. Brain Res.*, 2016.

[39] A. R. Krueger, P. Giannoni, V. Shah, M. Casadio, and R. A. Scheidt, "Supplemental vibrotactile feedback control of stabilization and reaching actions of the arm using limb state and position error encodings," *J. Neuroeng. Rehabil.*, 2017.

[40] L. Paredes, S. Dosen, F. Rattay, B. Graimann, and D. Farina, "The impact of the stimulation frequency on closed-loop control with electrotactile feedback," *J. Neuroeng. Rehabil.*, vol. 12, no. 35, 2015.

[41] R. Jagacinski and J. Flach, *Control Theory for Humans: Quantitative Approaches to Modelling Performance*. CRC press, 2018.