The interaction between feedback type and learning in routine grasping with myoelectric prostheses

Wilke, Meike; Hartmann, Cornelia; Schimpf, Felix; Farina, Dario; Dosen, Strahinja

Published in:
IEEE Transactions on Haptics

DOI (link to publication from Publisher):
10.1109/TOH.2019.2961652

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.2019.2961652, IEEE Transactions on Haptics

The interaction between feedback type and learning in routine grasping with myoelectric prostheses

Meike A. Schweisfurth*, Cornelia Hartmann*, Felix Schimpf, Dario Farina, *Fellow, IEEE*, Strahinja Došen, *Member, IEEE*

Abstract— While prosthetic fitting after upper-limb loss allows for restoration of motor functions, it deprives the amputee of tactile sensations that are essential for grasp control in able-bodied subjects. Therefore, it is commonly assumed that restoring the force feedback would improve the control of prosthesis grasping force. However, the literature regarding the benefit of feedback is controversial. Here, we investigated how the type of feedback affects learning and steady-state performance of routine grasping with a prosthesis. The experimental task was to grasp an object using a prosthesis and generate a low or high target-force range (TFR), both initially unknown, in three feedback conditions: basic auditory feedback on task outcome, and additional visual or vibratory feedback on the force magnitude. The results demonstrated that the performance was rather good and stable for the low TRF, whereas it was substantially worse for the high TFR with a pronounced training effect. Surprisingly, learning curve and steady-state performance did not depend on the feedback condition. Hence, in the specific context of routing grasping with a prosthesis controlled via surface EMG, the basic feedback on task outcome was not outperformed out by force-related end-of-trial feedback and hence seemed to be sufficient for accomplishing the task.

This conclusion applies to the context of routine grasping using a myoelectric prosthesis with surface EMG electrodes, which means that the control signals are variable and the feedback is perceived and processed at the end of the trial (motor adaption).

Index Terms— Somatosensory feedback, vibrotactile stimulation, prosthetic grasping, trial-by-trial adaptation

1 INTRODUCTION

The hands are our most important "tools" in life, allowing dexterous grasping and manipulation. The motor control of the human hand is impressive, and it strongly depends on the presence of somatosensory feedback [1]. When a person traumatically loses a hand, the lost functions can be substituted to a certain degree by a myoelectric prosthesis [2], providing a reliable control of simple activities (e.g., opening and closing). However, until today none of the commercial prostheses (except for one recent device [3]) provides the subject with any of the missing somatosensory feedback. Closing the loop by restoring the feedback, thereby mimicking the bidirectional communication characteristic of a normal limb, could improve the utility, facilitate the embodiment, and possibly increase the acceptance rate [4].

To restore somatosensory feedback, the data from sensors embedded in the prosthesis are read online and transmitted to the user through tactile stimulation [4]. The stimulation can be implemented using noninvasive technologies, such as electrocutenous stimulation [5], vibration motors [6], linear pushers [7], rotation motors [8], or pressure cuffs [9], or through invasive approaches based on delivering electrical pulses to the peripheral nerves [10]–[12] or the brain [13]. The invasive solutions might produce more natural sensations, as they can activate the neural structures that were used originally (e.g., the nerves innervating the hand), but they require complicated surgery. Vibration motors and surface electrical stimulation are still the most common methods to provide feedback information [4]. The prosthesis state is communicated by modulating the stimulation parameters. For example, an increase in the prosthesis force can be transmitted to the user by increasing the stimulation intensity/frequency (parameter modulation) and/or changing the location (spatial modulation). Most of the systems presented in the literature provide feedback on the grasping force [4], [14], since this variable cannot be directly perceived using vision (contrary to joint angles). The simplest approach is to transmit the force information by modulating the intensity of a miniature pager motor [15]. In some studies, an array of coin vibrators or stimulation electrodes has been used to communicate force through spatial [16] or mixed coding [17]. Finally, the force feedback has also been transmitted using tactors producing vibrations transversal to the skin [18] or through am-

- **These two authors equally contributed to this publication.*
- *M. A. Schweisfurth is with the Faculty of Life Sciences, HAW Hamburg, Germany (MeikeAnnika.Schweisfurth@haw-hamburg.de)*
- *M. A. Schweisfurth, F. Schimpf, D. Farina, and S. Došen are affiliated to the Advanced Rehabilitation Technology Lab at the Clinics for Trauma Surgery, Orthopedic Surgery and Plastic Surgery (CUOP), University Medical Center Göttingen (UMG), Germany.*
- *C. Hartmann is with the Otto Bock HealthCare SE & Co. KGaA, Duderstadt, Germany.*
- *D. Farina is with the Department of Bioengineering, Imperial College London, UK.*
- *S. Dosen is with the Faculty of Medicine, Department of Health Science and Technology, Center for Sensory-Motor Interaction, Aalborg University, DK-9220, Aalborg, Denmark*

plitude-frequency modulated patterns generated by advanced vibration motors [19], [20].

Natural sensory feedback is instrumental in the motor control of able-bodied humans [1]. It is therefore surprising that the conclusions in the literature regarding the benefits of artificial sensory feedback in prosthetics are rather inconsistent. A consistent outcome is that the supplemental feedback is beneficial when the subjects were deprived of other feedback sources (e.g., blinded and wearing headphones), and this has been demonstrated with invasive [21], [22] and non-invasive methods [17], [23]–[27]. That is, however, an obvious result, as any feedback is likely to be better than no feedback at all. The studies implementing more realistic conditions, however, report conflicting conclusions. For example, the studies using different non-invasive and invasive methods to transmit force feedback have reported no difference in performance [8], [16], [24], [28], clear improvement in an abstract electrotactile tracking task [29], or in functional and real-life tasks [11], [30]–[32] , or improvement in only some subjects and conditions [33], [34].

The rich literature on feedback in prosthetics describes many methods to transmit information to the prosthesis user (e.g. [10], [12], [17], [18], [35]), and some of them are rather simple to implement. However, the conflicting reports on the benefits of feedback imply that there is a general lack of knowledge about the nature and role of artificial somatosensory feedback in prosthetics. This has been recognized in some recent studies demonstrating that the effectiveness of the prosthesis feedback is a complex phenomenon which depends on multiple factors, including user training and experience [34], nature and complexity of the task (robust or delicate grasping) [34], internal models [36], [37], the quality of feedforward control [17], [25], the approach used to provide feedback (e.g., continuous [34], [39], [40] or discrete [41]) etc.

The aim of the present study was to investigate basic mechanisms governing the artificial feedback in prosthetics. Specifically, we assessed how the type of feedback impacts on the learning and performance in a realistic prosthesis control task. Importantly, we focused on a specific but relevant context of prosthesis control, namely, routine grasping. In this paradigm, the grasps are performed fast, by using feedforward myoelectric commands, mimicking the manner able-bodied subjects grasp objects in daily life. This approach can also avoid the need for force adjustments after the prosthesis has closed around the object, which can be a challenging task due to the nature of force control in myoelectric prostheses [54]. In the present study, the subjects received feedback on the generated force, but as the force increases abruptly upon contacting the object, the feedback was useful only to assess the outcome of the grasp (instead of modulating the grasp online). Essentially, this type of prosthesis control represents a motor-adaption [38], [42] paradigm, where the subjects used feedback from the current trial to adjust the feedforward motor command in a subsequent trial.

Three different types of feedback were provided to drive the adaption. In the first condition, auditory feedback transmitted the most basic, ternary information on the task outcome (force appropriate, too high, or too low). In the second and third condition, the feedback interface communicated more information. More specifically, in addition to the force range, the magnitude of the generated force was transmitted using a visual bar (second condition) and vibrotactile stimulation (third condition). We hypothesized that the subjects would be able to interpret the force level more reliably when relying on visual versus (vibro)tactile information. Therefore, in all three conditions (auditory, auditory + visual, auditory + vibration) the subjects were provided with an end-point feedback that could be used for trial-by-trial learning. However, the amount of information transmitted through the feedback interface differed across the conditions.

We expected that all types of feedback would facilitate task learning through improved feedforward control. In addition, we assumed that the vibrotactile and visual conditions (feedback on task outcome plus force magnitude) would outperform the auditory-only condition (feedback on task outcome), and that the visual condition would be better than the vibratory.

2 Methods

2.1 Participants

Twenty-two able-bodied subjects (13 male, 9 female, 24 ± 4 years) volunteered to participate in the study. After receiving oral and written information about the experiment, they gave informed consent. The study was approved by the Ethics Committee of the University Medical Center Göttingen (UMG). The data from one subject were excluded from the analysis for lack of compliance with the protocol.

2.2 Experimental Setup

The participants controlled a prosthesis (Michelangelo Hand, Otto Bock HealthCare Deutschland GmbH, Germany) to repeatedly grasp a dummy object in a realistic scenario. To ensure standardized grasp conditions and object contact, the prosthesis was detached from the subject and placed on the desk. The prosthetic hand was secured using a vice, and a rigid dummy object was permanently attached to the prosthetic thumb, so that when the prosthesis closed it grasped the object using a pinch grasp. The grasping force was measured using a strain gauge sensor embedded in the thumb of the prosthesis. The sensor has a resolution with 255 levels for the force range from 0 to maximum grasping force (~ 100 N). The force was expressed in percent of that maximum force.

An electromyographic (EMG) signal was recorded from the hand/wrist flexor muscles of the left forearm using an active bipolar electrode (13E200 = 50AC, Otto Bock HealthCare Deutschland GmbH, Germany). The prosthetic hand was connected to the host PC via a Bluetooth connection to transmit the sensor data (force and EMG) and receive the normalized velocity commands. The force and EMG were sampled at 100 Hz and 1 kHz, respectively. The root mean square (RMS) value of the EMG was computed over a 128 ms time window and normalized to 90% of the maximum voluntary contraction (MVC). To measure

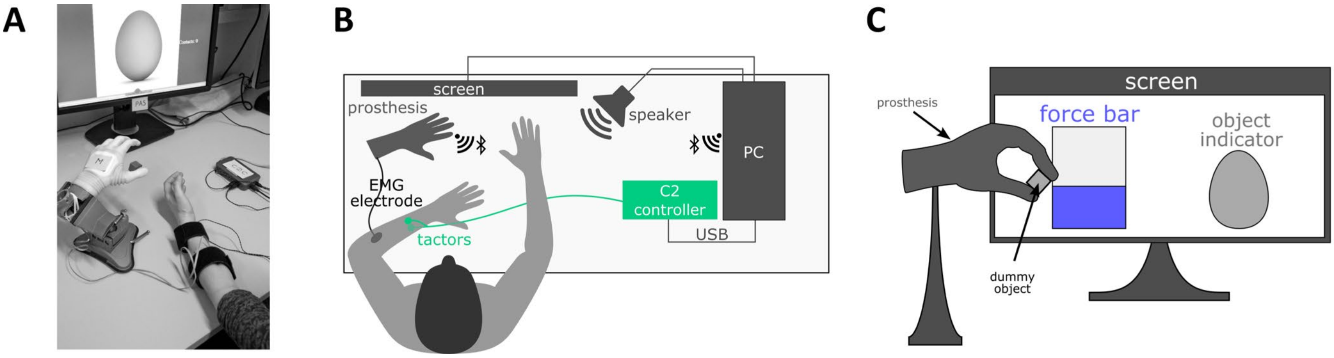

Fig. 1. Setup. A) Picture and B) schematic as seen from above. A bipolar EMG electrode recorded the flexor muscle signal used to control the prosthetic hand. Sensor data from the prosthesis were transmitted to the PC via Bluetooth. Basic feedback on the success of each grasp attempt was provided via the speaker (in all groups). For the vibrotactile feedback group, two C2 tactors were positioned next to the EMG electrode (shown in green) to deliver a vibrotactile representation of the current grasp force. C) Schematic of the view presented to the subject during the experiment. The prosthesis was placed on a stand in front of the screen and grasped a dummy object fixed to its thumb. On a computer screen the target force range (TFR) was indicated by a visual cue (object indicator, here the egg indicating the low TFR). For the visual feedback group, a visual representation of the current grasp force (force bar, shown in blue) was presented on the screen next to the object indicator.

the MVC, the subjects were asked to activate the muscles strongly but to a level that they can comfortably hold for 30 seconds without fatiguing (so called prolonged MVC as in [43]). This command was used to proportionally control the closing speed and hence the resulting grasping force. This is a standard operation of most commercial myoelectric prostheses, where a myoelectric signal is translated into voltage input for the prosthesis motor. Therefore, while the prosthesis moves, the myoelectric command sets the motor speed. When the prosthesis contacts an object and the motor stalls, the command corresponds to the motor torque and thereby applied grasping force. Subjects were informed prior to the experiment about the proportionality between contraction strength, closing velocity, and resulting force. The subject was instructed to close the prosthesis by using one continuous contraction. He/she would activate the muscle to the desired level and then maintain the contraction until the object was grasped, emulating fast routine grasping, as also used in other studies [35], [37], [44], [44]. Of course, this is only an approximation to real life, since able-bodied subjects as well as amputees might use more complex muscle activation patterns to accomplish the same task in daily life. There was no additional adjustment to the grasping force after contacting the object and a single EMG channel was used to control closing. The hand opened automatically 0.65 s after the grasp was accomplished. Note that due to the inertia of the prosthesis, careful force adjustment after contact remains an unmet challenge which goes beyond the current study.

During the experimental session, the subject sat comfortably in front of a desk with the prosthesis, a standard screen, and loudspeakers on top. Tactile feedback was delivered in one condition (see next section) via two C2 tactors (Engineering Acoustics, USA). The full range of the prosthesis force was mapped proportionally to the full range of vibration intensity and frequency, as explained below. A picture and a schematic view of the experimental setup are shown in Fig. 1A and 1B, and the view as seen by the subject during the experiment is represented in Fig. 1C.

2.3 Experimental task

The subjects' task was to quickly grasp the dummy object, generating a force within one out of two target-force ranges (TFR). If the force was outside the desired window, it was deemed too low/high and the object was considered as lost (slipped from grasp) or broken, respectively. The lower TFR was 35 ± 7.5%, simulating a light, fragile object, such as a fresh egg. The higher TFR was 67.5 ± 7.5% and represented the force necessary to handle a heavy, robust object, e.g. a dumbbell. Subjects were initially left ignorant of these target-force ranges. They only received a cue (an egg or a dumbbell, see Fig. 1C) to indicate whether a low or high force was requested. Additionally, different amount of feedback (see next section) was provided about the outcome of the trial. Exploiting this information, the subjects had to discover the correct force range and hence the appropriate amount of muscle contraction on a trial-and-error basis throughout the experiment. The subjects were blinded neither visually nor acoustically, and the prosthesis was placed in the clear view of the subjects, in front of the monitor (Fig. 1A and 1C). This has been done to allow the subjects to exploit the incidental feedback sources that will be anyway available to the prosthesis user in daily life.

In order to monitor the performance over time, the experiment was split into seven blocks with 5-min breaks between blocks. Each block consisted of 50 grasp attempts, out of which 25 trials aimed at the lower and 25 trials at the higher target force. Low and high-force trials were presented in pseudorandom order under the constraint that at least two consecutive trials had the same target force window, to simulate repeated grasping of different objects by a prosthesis user, where some objects can be fragile and some robust.

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.2019.2961652, IEEE Transactions on Haptics

2.4 Feedback conditions

Three types of artificial feedback were tested in this experiment. An auditory feedback on the success of each grasp attempt, and both visual and tactile feedback on the current grasp force exerted by the prosthesis.

In order to investigate the effect of the different feedback (FB) interfaces on the training of a routine grasping task, the subjects were randomly distributed into three groups, where each group performed the experiment under a different feedback condition. All three groups received an auditory feedback on the success of the previous grasp, i.e. whether the achieved force was appropriate for the object, too high (object broken), or too low (object slipped). This was the most basic feedback transmitting only the sign of the force error with respect to the TFR. This information can be also perceived in real-life grasping with prosthetic hands. The group of subjects receiving only this feedback was called the "basic FB" group.

The second group ("visual FB") received a visual feedback about the generated grasping force additionally to the basic auditory FB. The visual feedback was displayed as a bar on the computer screen, where the height of the bar was proportional to the grasp force measured by the prosthesis (see Fig. 1C). We assumed that the visual feedback was the most reliable interface because of its comprehensiveness and intuitiveness, due to its good visual resolution and the human experience in perception and control based on visual feedback. In this condition, therefore, the feedback transmitted more information, i.e., not only the sign of the force error, but also a precise magnitude of the generated force (visual force bar).

The third group ("tactile FB") received vibrotactile stimulation additionally to the basic FB, transmitted via two C2 tactors (Engineering Acoustics Inc., USA) placed on the ventral and dorsal side of the forearm close to the EMG recording electrode (see Fig. 1A and 1B). The two tactors generated vibrations at the same frequency and intensity to improve the perceptibility of the feedback, as demonstrated in [39]. The vibration was modulated to represent the prosthesis' measured grasping force, by mapping both the intensity and the frequency of vibration proportionally to the current force. The minimum detectable force (~1%) corresponded to a 50 Hz vibration at 20% of the maximum amplitude, while the maximum prosthetic force was mapped to a 100 Hz vibration at 100% tactor amplitude. Like in "visual FB", the subjects received the information on the sign of the force error as well as on the force magnitude; however, we assumed that this feedback was likely more difficult to exploit, as the vibration interface was likely characterized by a higher variance compared to the visual feedback [46], [47].

Therefore, in all the conditions, the subjects controlled the prosthesis using feedforward commands to close the hand and generate force. Since the target object was rigid, the grasping force increased abruptly after contact (as shown in Fig. 2) and the subjects then basically perceived a constant feedback value until the hand was opened again. Hence, the subjects could not use the provided visual and vibrotactile feedback to modulate the control during the trial, but only to observe the force that had just been generated. Put differently, although the feedback in principle transmitted the measured force continuously, the force modulation was so brief that the feedback effectively transmitted a discrete message delivered at the end of the trial. The subjects could use this information to decide how to adjust the muscle contraction in the next trial. This experimental paradigm therefore corresponds to trial-by-trial motor adaption based on end-point feedback (e.g. [38], [42]).

Three independent groups of subjects, each performing a single feedback condition, were chosen in order to avoid the uncontrolled interactions between the conditions. If subjects sequentially performed all the conditions, even in randomized order, they would learn the task in the preceding condition, which would affect the performance in the subsequent one(s). Therefore, an objective comparison between the conditions would not be possible.

2.5 Outcome measures

The grasp was deemed successful if the generated force was within the TFR. If the force was too low or too high, this was defined as a slip or break outcome, respectively. The percent rates for success, slip, and break were determined to evaluate the task performance and the reasons for failure. In addition, the median absolute error (MAE) was calculated as the median of the absolute per-trial errors between the generated force and the center of the TFR. This was a more sensitive measure than the success rate, assessing how far the subjects were from an optimal grasp (halfway between slip and breaking forces). To evaluate the consistency in force generation, the inter-quartile range of the generated forces (IQRF) was computed. Outcome measures were determined per subject, block and TFR.

2.6 Statistics

The data were not normally distributed according to Kolmogorov-Smirnov tests, and therefore non-parametric statistics were applied to explore the differences between the across-groups factor "feedback condition" as well as the within-group factors "target-force range" and "block". The significance threshold was set to $p<0.05$.

First, a Kruskal-Wallis test was used to compare the performance between the feedback conditions, separately for each TFR and block. Then, the within-group factors "target-force range" and "block" were explored for each feedback group. For the within-group factor "block", a Wilcoxon signed rank test was applied to compare the performance between the first block and the average of the last four blocks (in which a rather stable performance was observed, see Results), separately for the low and high TFR. For the within-group factor "target-force range", a Wilcoxon signed rank test was used to compare the outcome measures between the low and the high TFR. For this test, the subject results were averaged across blocks. All within-group tests were Bonferroni-Holmes corrected for the three feedback groups (i.e. $p<0.017$ was the margin for significance).

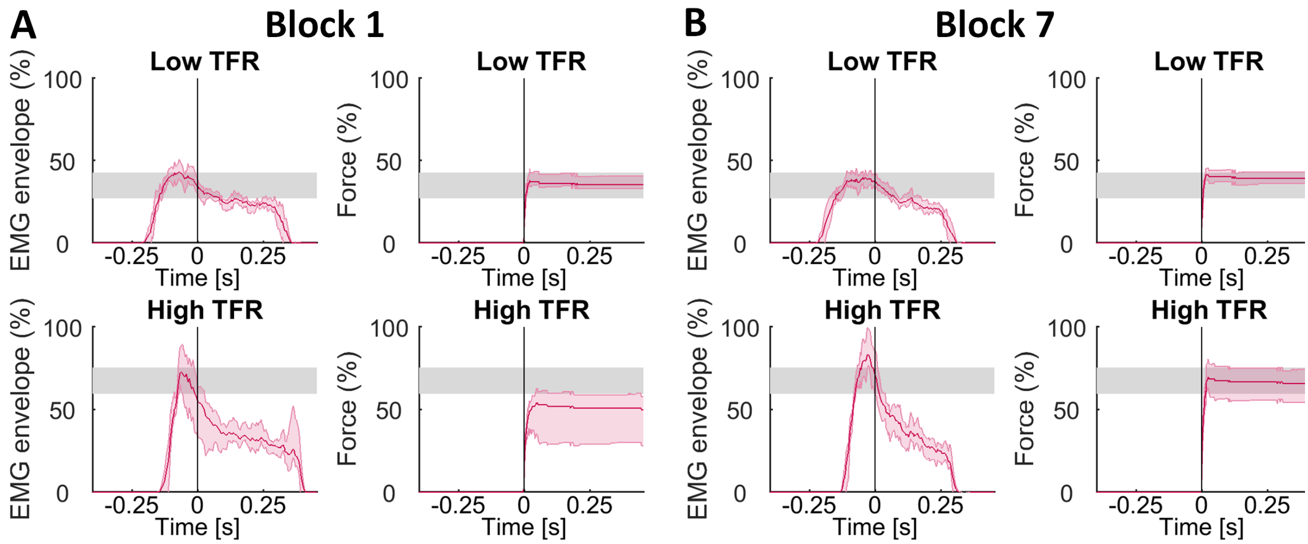

Fig. 2. EMG and force profiles. For one representative subject (of the “basic FB” group), the median EMG and force profiles across trials are shown for the first (A) and last block (B) for the low (upper panels) and high (lower panels) TFR. Error bars indicate the range between the 25th and 75th percentile). The 0s time point denotes the force onset (contact). Grey boxes show the respective TFR.

3 RESULTS

Representative results are shown in Figs. 2 and 3. Fig. 2 depicts the (median) myoelectric commands and resulting grasping forces generated by one representative subject in the first and last block of trials of the “basic FB” condition. At the low TFR, the subjects increased the muscle activation to the desired level (shaded region), maintained this level until the prosthesis closed, and generated the desired force. The prosthesis was non-backdrivable and therefore, the subjects could relax their muscles soon after contact (zero line), while the prosthesis continued to maintain the achieved force. The prosthesis was automatically opened by an external command from the PC at the end of the trial. At the high TFR, the subjects followed a similar control approach. However, since the prosthesis velocity (and force) was proportional to the muscle activation, the time to increase the activation to the proper level was in this case much shorter (on average 174 ms for the high compared to 300 ms for the low TFR). The generated myoelectric signals were more variable, which translated into a higher variability of the generated forces. From the first to the last block, the median force generated in high-TFR trials centered within the desired region and the variability decreased, although it was still higher than for the low TFR. In Fig. 3A the forces generated in the first and last block are depicted for one representative subject per feedback

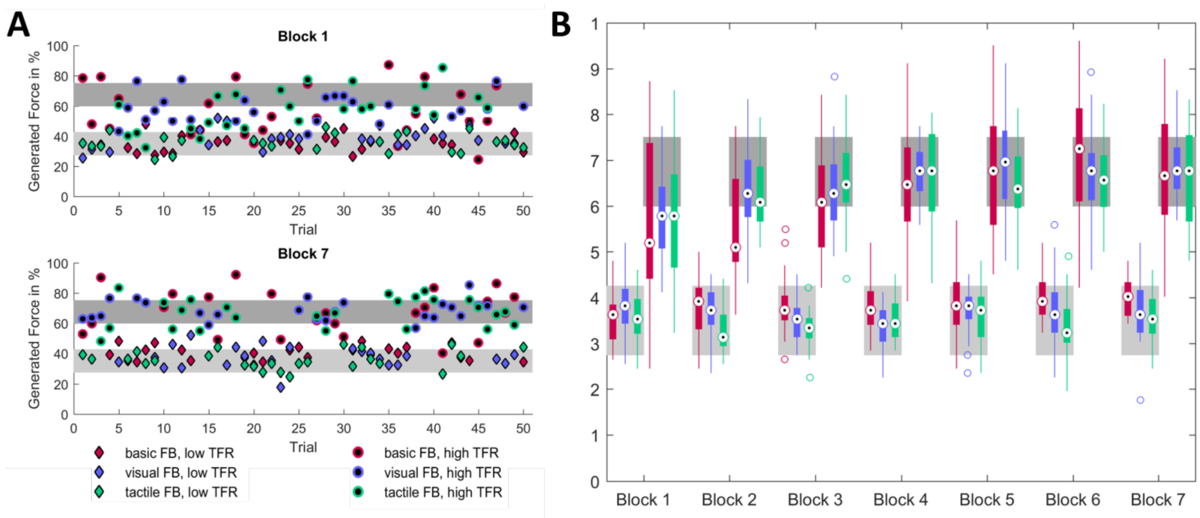

Fig. 3. Generated prosthesis grasping forces. A) The forces generated in each trial of the first (upper panel) and last block (lower panel) are shown for three individual representative subjects (one per feedback group). The color indicates the feedback group (basic FB = red, visual FB = blue, tactile FB = green). Trials with a low target force range (TFR) are given as diamonds and those with a high TFR as circles. Light/dark grey boxes show the low and high TFR, respectively. B) Boxplots (including median, IQR, range, and outliers) of the forces generated in three representative subjects for each block and for the low (first three bars per block, underlaid in light grey) and high (last three bars per block, underlaid in dark grey) TFR. Grey boxes and color code are the same as in A.

group. The forces were concentrated around the respective TFRs, although there were large deviations especially for the high TFR. Fig. 3B shows the distribution of the generated forces across blocks for the three representative subjects, one per feedback condition. The subjects successfully scaled the forces according to the indicated TFR, as there is a clear difference in the medians of the force distributions for the high and low TFR. For the low TFR, the performance was good already in block 1, as most of the forces were within the TFR. For the high TFR, however, the subjects started by significantly undershooting in block 1 and then the performance gradually improved over the blocks.

The summary results for the success rate, slip and break rate, MAE, and IQRF are shown in Fig. 4. All outcome measures except for the break rates exhibited a similar trend. Surprisingly, in all blocks, there were no statistically significant differences between the feedback conditions in any of the outcome measures (Kruskal-Wallis tests). In particular, there was no statistically significant difference in any performance measure at any TFR between groups in block 1 ($p > 0.2$ in all ten Kruskal-Wallis tests), showing that the three groups started off with a similar baseline and are well comparable.

The performance in the low TFR was consistently and significantly better compared to the high TFR in all the outcome measures, apart from the break rate (Wilcoxon signed rank tests). The overall median success rate (70%, 74%, and 65% for the basic, visual, and vibration condition, respectively) in the low TFR was significantly higher ($p = 0.008$ in each group) compared to the high TFR (35%, 39%, and 42%, respectively; Fig. 4A). The median slip rate (Fig. 4C) was substantially lower for the low (6%, 6%, and 13%, respectively) compared to the high TFR (42%, 37%, and 40%, respectively; $p = 0.008$ in each group), whereas the break rates (Fig. 4D) were similar (24% vs. 21%, 20% vs. 25%, and 18 vs. 15% for the basic, visual, and vibration condition in the low and high TFR, respectively; $p = 0.414$, $p = 0.655$, and $p = 0.414$ for the basic, visual, and vibration condition, respectively). The overall MAE (Fig. 4B) was significantly lower for the low (5%, 4%, and 5%, respectively) compared to the high TFR (10%, 10%, and 9%, respectively; $p = 0.008$ in each group). Finally, the forces were more variable ($p < 0.001$, Fig. 4E) for the high (19%, 19%, and 15%, respectively) compared to the low TFR (9%, 8%, and 9%, respectively; $p = 0.008$ in each group).

For the low TFR, all outcome measures were stable across blocks, i.e. they did not differ from the first to the average of the four last blocks. For almost all subjects, the MAE was within the margin of a successful grasp.

For the high TFR, the success rate improved significantly with training from the first to the last blocks, indicating a learning effect. This improvement was significant for the basic condition (26% vs. 41%; $p = 0.016$) and the visual condition (27% vs. 42%; $p = 0.016$); for the vibration condition, significance was not reached (30% vs. 45%; $p = 0.078$). Both the slip and the break rate did not change across blocks in any group. The median MAE was outside the zone of a successful grasp in all the blocks in all groups (except for block 6 in the visual condition). However, for all groups a significant improvement was observed from block 1 to the latter blocks, for the basic condition from 14% to 10% ($p = 0.016$), for the visual condition also from 14% vs. 10% ($p = 0.016$) and for vibration condition from 14% to 9% ($p = 0.031$). While a trend was observed for each group, no significant improvement in grasp-force variability across blocks was shown in any group ($p = 0.047$, $p = 0.031$, and $p = 0.031$ for the basic, visual, and vibration condition, respectively).

5 Discussion

We investigated how the type of feedback the subjects received affected the performance of grasping-force control in a myoelectric prosthesis in the context of routine grasping. The able-bodied subjects learned to generate grasping forces within two TFRs (low and high) through the process of motor adaption driven by three types of feedback. The feedback interface transmitted only the task outcome (basic condition) or additionally the magnitude of the generated force (vibrotactile or visual condition). The subjects were required to locate the two TFRs by adjusting their muscle contraction levels (feedforward commands) based on the end-point feedback, through trial and error, and then consistently generate forces within the indicated TFR.

5.1 Low versus high target-force range

While the subjects successfully learned the task irrespective of the feedback condition, the performance strongly differed between the low and high TFR. The control was substantially better at the low TFR, as the subjects were overall more successful in hitting the target range (Fig. 4A), deviated less from the middle of the range (Fig. 4B), and tended to generated forces more consistently (Fig. 4E). For the lower force range, the performance was high already in the first block (see Fig. 2, 3, and 4) and then remained stable across the session, such that no learning was observed. Contrarily, reaching the high TFR was challenging and the subjects needed more blocks to learn the task. In this case, the effect of training was clearly visible, as the subjects improved the performance across blocks. The subjects demonstrated learning, as the success rate increased (Fig. 4A), while force error (Fig. 4B) and variability decreased (Fig. 4E, only trend) across blocks. However, at the end of the training (after 175 grasping trials), the success rate was still rather low.

Most likely, the difference in performance between TFRs was due to an increased variance of the feedforward control at the higher compared to the lower muscle contraction levels, as demonstrated in [19] and [31]. The grasp-force variability almost doubled from the low to the high TFR. Both TFRs had the same size (15%), and therefore, when hitting the high TFR the subjects were required to hit equally precisely as in the low TFR; however, due to the substantially more variable control signals, the task was more difficult. In addition, the subjects had less time to adjust the muscle contraction to a proper level when aiming at the high TFR, since the prosthesis closes faster for higher activations. For the low TFR, the subjects had enough time to carefully adjust and maintain the muscle contraction

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.2019.2961652, IEEE Transactions on Haptics

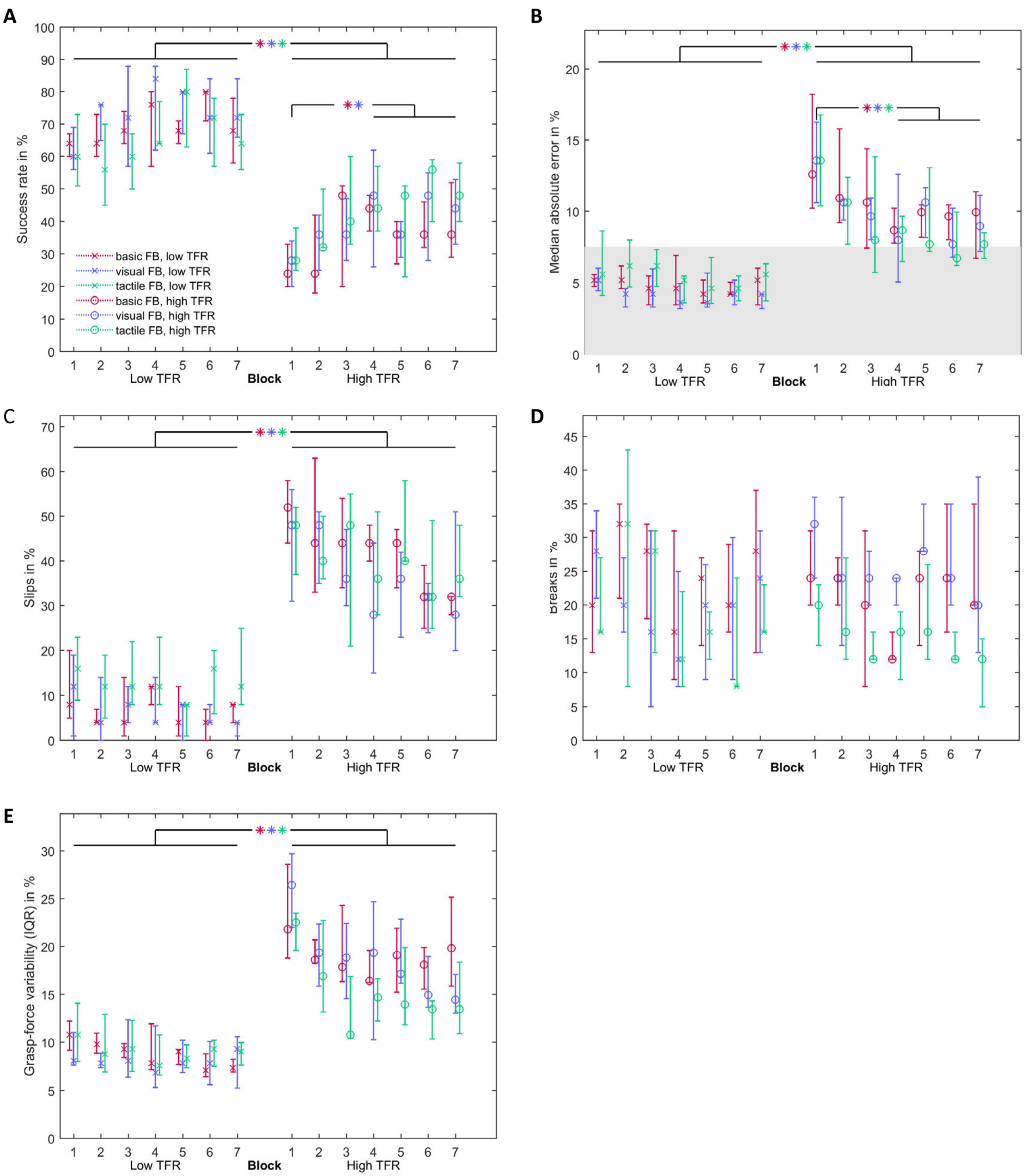

Fig 4. Success rates (A), median absolute error (MAE, B), slips (C), breaks (D), and grasp-force variability (IQRF, E). Each measure is shown as median and IQR per block and TFR (crosses depict low TFR, circles high TFR) for each group (basic FB = red, visual FB = blue, tactile FB = green). Significant performance differences ($p < 0.05$, corrected) between low (left part of each panel) and high (right part) TFR as well as from first to average of last four blocks are marked by asterisks (*) in the respective group color. The light-grey area in (B) indicates the absolute errors that are still within the TFR (successful grasps).

level whereas for the high TFR, the subjects needed to produce essentially a myoelectric impulse of the correct amplitude (Fig. 2). This could have been an additional source of control variability, adding to the intrinsic variability of myocontrol signals. The lower success rate at the high TFR was due to an increased number of slips, while there was no difference in the breaks (Fig. 4C and 4D). This implies that the subjects approached the higher TFR from below, which might reflect an attempt to control the variance. The subjects might have noticed that they can produce lower forces more consistently (with lower variability). Also, it is as well possible that the higher number of slips was due to the shorter closing time for the high TFR. The subjects might not have had enough time to increase the muscle activation to the proper level.

5.2 Influence of feedback condition

The basic condition of the present study demonstrates that supplemental feedback is indeed beneficial for the control of grasping. The transmitted feedback can be simple: the subjects improved their performance across blocks even when receiving just an indication of the force range (too high/low or appropriate). This result is in line with a recent study demonstrating the benefits of a time-discrete feedback communicating only grasp contact and release [32].

Surprisingly, in the present experiment, the second and third condition did not lead to better performance although the feedback interface communicated more information, namely the force magnitude, which is a common approach in prosthetics literature [29], [44], [45], [51]. By using information on the task outcome (auditory feedback), the subjects could determine if they needed to increase or decrease the grasping force (muscle contraction) in the next trial of that TFR. With vibration and visual feedback, the assumption was that the subjects additionally received the information on the force magnitude, from which they would be able to estimate both the direction (increase/decrease) as well as the amount of required force change. Nevertheless, the additional information transmitted through the feedback interface did not improve the rate of learning nor the task performance within blocks.

Several factors might have contributed to this surprising outcome. First, there is a level of uncertainty related to information transmission through the tactile channel. The subjects had to decode the tactile sensation elicited through vibrotactile stimulation and map it to the level of grasping force, relatively (higher/lower than in previous trials) and/or absolutely (percent of grasping force). Initially, the subjects knew only the basic relation, i.e., that the vibration intensity and frequency were proportional to force. From that point of view, the visual bar might have been easier to interpret; however, this was an assumption that was not explicitly tested. Therefore, although the feedback interface in the second and third condition transmitted more information, it is unknown how well this information had been actually received by the subjects. This motivates the need for an explicit systematic investigation on how these factors (i.e., feedback precision and reliability) impact on the closed-loop control of prosthesis grasping. This can be done, for example, through a controlled modulation of the properties (e.g., uncertainty) of a single feedback source (e.g., visual feedback).

Second, the subjects might not have benefitted from the better feedback (vibration and visual) due to inherent limitations of the feedforward interface, i.e., the myoelectric signal variability [52], [53]. The more informative feedback (error magnitude vs. error sign) might have been successfully interpreted by the subjects. However, they could not translate it into better control, because they were unable to modulate the myoelectric signals well enough, especially at the higher forces. Instead, the control was rather rough and, consequently, only the rough feedback information about the direction of change (increase/decrease/hold) could be exploited effectively.

Third, the subjects might have used the intrinsic feedback, which was present in all feedback conditions: the proprioceptive information from the own muscles (sense of contraction) as well as the intrinsic information from the prosthesis (sound and movement) were available in all feedback conditions. As the prosthesis operated proportionally, it closed at markedly different speeds and produced characteristic motor sounds when generating different forces. These cues could have been used by the subjects to estimate the force magnitude, even in the absence of an explicit force magnitude feedback (as in the auditory condition) [53].

Nevertheless, the present study explains why closing the loop in prosthetics through force feedback might not necessarily improve the performance in the real-life application of routine grasping. The prosthesis user normally receives feedback on the task outcome, as he/she can see if the object slipped from the grasp (force too low), broke (force too high), or was grasped successfully (force within the target window). Furthermore, the user receives implicit information coming from the muscles and the prosthesis. As demonstrated by the present experiment, this might already be enough to allow successful accomplishment of simple tasks (low TFR) as well as learning across trials (high TFR).

5.3 Restrictions and Limitations

The present results hold for able-bodied subjects only and likely for novice users of myoelectric prostheses. Experienced users might be more consistent in generating myoelectric commands due to practice and learning, and therefore, they might better exploit the feedback. In addition, the results are specific for the task and feedback approach (force) that have been used in the present study. Another type of feedback could still outperform the simple feedback on the task outcome. For example, promising results have been demonstrated when transmitting the magnitude of the myoelectric signal (prosthesis command) both in our previous work [40], [48] and, more recently, by other groups [49], [50]. Importantly, this is a substantially different approach to closing the loop in prosthesis control compared to conventional force feedback, and therefore, it could lead to different conclusions.

The small subject sample for each condition (N=7) is one of the main limitations of the study, especially considering the relatively large variability across subjects. A power analysis that includes the observed variability indicated a larger sample size (N>20), which could not be measured in the current study. For the small sample measured, no statistical differences were observed, although it cannot be excluded that differences would be detected with larger samples. Nonetheless, we still believe that the lack of significant difference in performance across conditions or even a trend in that direction is a relevant and interesting result, despite the small subject sample.

Finally, the focus in the present study was on fast routine grasping, characteristic for daily life, and we did not examine the performance in delicate grasping tasks where the subjects need to slowly and carefully modulate the force after contact. The impact of feedback in this scenario is yet to be investigated. Nevertheless, also with our current experimental paradigm, the possibility that the subjects used some feedback and online modulation already during the grasping trial cannot be fully excluded. The generated prosthesis movements were short but the subjects might still have had enough time to incorporate feedback (e.g., prosthesis sound/closing velocity) and modulate the myoelectric command towards the end of the trial.

Finally, it should be kept in mind that the implementation of grasping-force feedback might be important due to the positive emotional benefit [55] as well as the potential of feedback to facilitate embodiment and reduce phantom-limb pain [56].

6 Conclusion

The present study demonstrated that while a simple supplemental feedback on the task outcome can be sufficient for task learning, providing additional, task-relevant information (force magnitude) is not necessarily beneficial for prosthetic grasping. This conclusion applies to the context of routine grasping using a myoelectric prosthesis with surface EMG electrodes, which means that the control signals are variable and the feedback is perceived and processed at the end of the trial (motor adaption).

Acknowledgment

This work has been supported by the German Federal Ministry of Education and Research (BMBF) under the project INOPRO (16SV7657) and by the Independent Research Fund Denmark (ROBIN-8022-00243A).

References

[1] T. Brochier, M.-J. Boudreau, M. Paré, and A. M. Smith, "The effects of muscimol inactivation of small regions of motor and somatosensory cortex on independent finger movements and force control in the precision grip," *Exp. Brain Res.*, vol. 128, no. 1, pp. 31–40, 1999.

[2] I. Vujaklija, D. Farina, and O. Aszmann, "New developments in prosthetic arm systems," *Orthop. Res. Rev.*, vol. Volume 8, pp. 31–39, Jul. 2016.

[3] "Vincent Evolution 2 Hand." .

[4] 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, Jan. 2013.

[5] A. Y. Szeto and F. A. Saunders, "Electrocutaneous stimulation for sensory communication in rehabilitation engineering.," *IEEE Trans. Biomed. Eng.*, vol. 29, no. 4, pp. 300–8, Apr. 1982.

[6] K. A. Kaczmarek, J. G. Webster, P. Bach-y-Rita, and W. J. Tompkins, "Electrotactile and vibrotactile displays for sensory substitution systems," *Biomed. Eng. IEEE Trans.*, vol. 38, no. 1, pp. 1–16, 1991.

[7] S. G. Meek, S. C. Jacobsen, and P. P. Goulding, "Extended physiologic taction: design and evaluation of a proportional force feedback system.," *J. Rehabil. Res. Dev.*, vol. 26, no. 3, pp. 53–62, Jan. 1989.

[8] J. D. Brown, A. Paek, M. Syed, M. K. O'Malley, P. A. Shewokis, J. L. Contreras-Vidal, A. J. Davis, and R. B. Gillespie, "An exploration of grip force regulation with a low-impedance myoelectric prosthesis featuring referred haptic feedback.," *J. Neuroeng. Rehabil.*, vol. 12, p. 104, 2015.

[9] P. E. Patterson and J. A. Katz, "Design and evaluation of a sensory feedback system that provides grasping pressure in a myoelectric hand.," *J. Rehabil. Res. Dev.*, vol. 29, no. 1, pp. 1–8, Jan. 1992.

[10] S. Raspopovic, M. Capogrosso, F. M. Petrini, M. Bonizzato, J. Rigosa, G. Di Pino, J. Carpaneto, M. Controzzi, T. Boretius, E. Fernandez, G. Granata, C. M. Oddo, L. Citi, A. L. Ciancio, C. Cipriani, M. C. Carrozza, W. Jensen, E. Guglielmelli, T. Stieglitz, P. M. Rossini, and S. Micera, "Restoring natural sensory feedback in real-time bidirectional hand prostheses.," *Sci. Transl. Med.*, vol. 6, no. 222, p. 222ra19, Feb. 2014.

[11] D. W. Tan, M. A. Schiefer, M. W. Keith, J. R. Anderson, J. Tyler, and D. J. Tyler, "A neural interface provides long-term stable natural touch perception," *Sci. Transl. Med.*, vol. 6, no. 257, p. 257ra138-257ra138, Oct. 2014.

[12] M. Ortiz-Catalan, B. Håkansson, and R. Brånemark, "An osseointegrated human-machine gateway for long-term sensory feedback and motor control of artificial limbs," *Sci. Transl. Med.*, vol. 6, no. 257, 2014.

[13] G. A. Tabot, J. F. Dammann, J. A. Berg, F. V Tenore, J. L. Boback, R. J. Vogelstein, and S. J. Bensmaia, "Restoring the sense of touch with a prosthetic hand through a brain interface," *Proc. Natl. Acad. Sci.* , vol. 110, no. 45, pp. 18279–18284, Nov. 2013.

[14] B. Peerdeman, D. Boere, H. Witteveen, R. Huis in `tVeld, H. Hermens, S. Stramigioli, H. Rietman, P. Veltink, and S. Misra, "Myoelectric forearm prostheses: State of the art from a user-centered perspective," *J. Rehabil. Res. Dev.*, vol. 48, no. 6, p. 719, 2011.

[15] C. Pylatiuk, A. Kargov, and S. Schulz, "Design and Evaluation of a Low-Cost Force Feedback System for Myoelectric Prosthetic Hands," *JPO Journal of Prosthetics and Orthotics*, vol. 18, no. 2, pp. 57–61, 2006.

[16] I. Saunders and S. Vijayakumar, "The role of feed-forward and feedback processes for closed-loop prosthesis control.," *J. Neuroeng. Rehabil.*, vol. 8, no. 1, p. 60, Jan. 2011.

[17] S. Dosen, M. Markovic, M. Strbac, M. Perovic, V. Kojic, G. Bijelic, T. Keller, and D. Farina, "Multichannel Electrotactile Feedback with Spatial and Mixed Coding for Closed-Loop Control of Grasping Force in Hand Prostheses," *IEEE Trans. Neural Syst. Rehabil. Eng.*, pp. 1–1, 2016.

[18] H. J. Witteveen, H. S. Rietman, and P. H. Veltink, "Vibrotactile grasping force and hand aperture feedback for myoelectric forearm prosthesis users.," *Prosthet. Orthot. Int.*, vol. 39, no. 3, pp. 204–212, 2015.

[19] S. Dosen, A. Ninu, T. Yakimovich, H. Dietl, and D. Farina, "A Novel Method to Generate Amplitude-Frequency Modulated Vibrotactile Stimulation," *IEEE Trans. Haptics*, vol. 9, no. 1, pp. 3–12, Jan. 2016.

[20] C. Cipriani, M. D'Alonzo, and M. C. Carrozza, "A miniature vibrotactile sensory substitution device for multifingered hand prosthetics.," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 2, pp. 400–8, Feb. 2012.

[21] S. Raspopovic, M. Capogrosso, F. M. Petrini, M. Bonizzato, J. Rigosa, G. Di Pino, J. Carpaneto, M. Controzzi, T. Boretius, E. Fernandez, G. Granata, C. M. Oddo, L. Citi, A. L. Ciancio, C. Cipriani, M. C. Carrozza, W. Jensen, E. Guglielmelli, T. Stieglitz, P. M. Rossini, and S. Micera, "Restoring Natural Sensory Feedback in Real-Time Bidirectional Hand Prostheses," *Sci. Transl. Med.*, vol. 6, no. 222, p. 222ra19-222ra19, 2014.

[22] K. Horch, S. Meek, T. G. Taylor, and D. T. Hutchinson, "Object discrimination with an artificial hand using electrical stimulation of peripheral tactile and proprioceptive pathways with intrafascicular electrodes," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 19, no. 5, pp. 483–489, 2011.

[23] H. Xu, D. Zhang, J. C. Huegel, W. Xu, and X. Zhu, "Effects of Different Tactile Feedback on Myoelectric Closed-Loop Control for Grasping Based on Electrotactile Stimulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 24, no. 8, pp. 827–836, Aug. 2016.

[24] A. Ninu, S. Dosen, S. Muceli, F. Rattay, H. Dietl, and D. Farina, "Closed Loop Control of Grasping With a Myoelectric Hand Prosthesis: Which Are the Relevant Feedback Variables for Force Control?," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. PP, no. 99, pp. 1–12, Apr. 2014.

[25] H. J. B. Witteveen, L. de Rond, J. S. Rietman, and P. H. Veltink, "Hand-opening feedback for myoelectric forearm prostheses: performance in virtual grasping tasks influenced by different levels of distraction.," *J. Rehabil. Res. Dev.*, vol. 49, no. 10, pp. 1517–26, 2012.

[26] H. J. B. Witteveen, H. S. Rietman, and P. H. Veltink, "Vibrotactile grasping force and hand aperture feedback for myoelectric forearm prosthesis users.," *Prosthet. Orthot. Int.*, vol. 39, no. 3, pp. 204–12, Jun. 2015.

[27] H. J. B. Witteveen, E. A. Droog, J. S. Rietman, and P. H. Veltink, "Vibro- and Electrotactile User Feedback on Hand Opening for Myoelectric Forearm Prostheses," *IEEE Trans. Biomed. Eng.*, vol. 59, no. 8, pp. 2219–2226, Aug. 2012.

[28] C. Cipriani, F. Zaccone, S. Micera, and M. C. Carrozza, "On the Shared Control of an EMG-Controlled Prosthetic Hand: Analysis of User-Prosthesis Interaction," *IEEE Trans. Robot.*, vol. 24, no. 1, pp. 170–184, 2008.

[29] N. Jorgovanovic, S. Dosen, D. J. Djozic, G. Krajoski, and D. Farina, "Virtual grasping: closed-loop force control using electrotactile feedback.," *Comput. Math. Methods Med.*, vol. 2014, pp. 1–14, Jan. 2014.

[30] F. Clemente, G. Valle, M. Controzzi, I. Strauss, F. Iberite, T. Stieglitz, G. Granata, P. M. Rossini, F. Petrini, and S. Micera, "Intraneural sensory feedback restores grip force control and motor coordination while using a prosthetic hand," *J. Neural Eng.*, vol. 16, no. 2, p. 26034, 2019.

[31] P. D. Marasco, J. S. Hebert, J. W. Sensinger, C. E. Shell, J. S. Schofield, Z. C. Thumser, R. Nataraj, D. T. Beckler, M. R. Dawson, and D. H. Blustein, "Illusory movement perception improves motor control for prosthetic hands," *Sci. Transl. Med.*, vol. 10, no. 432, p. eaao6990, 2018.

[32] F. Clemente, M. D'Alonzo, M. Controzzi, B. B. Edin, and C. Cipriani, "Non-invasive, temporally discrete feedback of object contact and release improves grasp control of closed-loop myoelectric transradial prostheses," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 24, no. 12, pp. 1314–1322, 2016.

[33] A. Chatterjee, P. Chaubey, J. Martin, and N. Thakor, "Testing a Prosthetic Haptic Feedback Simulator With an Interactive Force Matching Task," *JPO J. Prosthetics Orthot.*, vol. 20, no. 2, pp. 27–34, 2008.

[34] M. Markovic, M. A. Schweisfurth, L. F. Engels, T. Bentz, D. Wüstefeld, D. Farina, and S. Dosen, "The clinical relevance of advanced artificial feedback in the control of a multi-functional myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, p. 28, 2018.

[35] M. A. Schweisfurth, M. Markovic, S. Dosen, F. Teich, B. Graimann, and D. Farina, "Electrotactile EMG feedback improves the control of prosthesis grasping force," *J. Neural Eng.*, vol. 13, no. 5, pp. 1–16, 2016.

[36] M. Strbac, M. Isakovic, M. Belic, I. Popovic, I. Simanic, D. Farina, T. Keller, and S. Dosen, "Short-and Long-Term Learning of Feedforward Control of a Myoelectric Prosthesis with Sensory Feedback by Amputees," *IEEE Trans. Neural Syst. Rehabil. Eng.*, 2017.

[37] S. Dosen, M. Markovic, N. Wille, M. Henkel, M. Koppe, A. Ninu, C. Frömmel, and D. Farina, "Building an internal model of a myoelectric prosthesis via closed-loop control for consistent and routine grasping," *Exp. Brain Res.*, vol. 233, no. 6, pp. 1855–1865, 2015.

[38] R. E. Johnson, K. P. Kording, L. J. Hargrove, and J. W. Sensinger, "Does EMG control lead to distinct motor adaptation?," *Front. Neurosci.*, vol. 8, 2014.

[39] S. Dosen, M. Markovic, C. Hartmann, D. Farina, C. Hartman, 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. 267–76, Mar. 2015.

[40] M. A. Schweisfurth, M. Markovic, S. Dosen, F. Teich, B. Graimann, and D. Farina, "Electrotactile EMG feedback improves the control of prosthesis grasping force," *J. Neural Eng.*, vol. 13, no. 5, p. 056010, Oct. 2016.

[41] F. Clemente, M. D'Alonzo, M. Controzzi, B. Edin, and C. Cipriani, "Non-invasive, temporally discrete feedback of object contact and release improves grasp control of closed-loop myoelectric transradial prostheses," *IEEE Trans Neural Syst Rehabil Eng*, vol. 4320, 2015.

[42] R. E. Johnson, K. P. Koerding, L. J. Hargrove, and J. W. Sensinger, "EMG versus torque control of human-machine systems: equalizing control signal variability does not equalize error or uncertainty," *IEEE Trans. Neural Syst. Rehabil. Eng.*, pp. 1–1, 2016.

[43] S. Amsuess, S. S. Member, P. Goebel, S. S. Member, B. Graimann, D. Farina, S. S. Member, P. Gobel, B. Graimann, D. Farina, P. Goebel, B. Graimann, and D. Farina, "A multi-class proportional myocontrol algorithm for upper limb prosthesis control: Validation in real-life scenarios on amputees," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 23, no. 5, pp. 827–836, 2015.

[44] S. Dosen, M. Markovic, S. Kelef, B. Graimann, and D. Farina, "EMG Biofeedback for online predictive control of grasping force in a myoelectric prosthesis," *J. Neuroeng. Rehabil.*, vol. 12, no. 55, pp. 1–13, 2015.

[45] M. Markovic, M. A. Schweisfurth, L. F. Engels, D. Farina, and S. Dosen, "Myocontrol is closed-loop control: incidental feedback is sufficient for scaling the prosthesis force in routine grasping," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, p. 81, 2018.

[46] R. Sigrist, G. Rauter, R. Riener, and P. Wolf, "Augmented visual, auditory, haptic, and multimodal feedback in motor learning: A review.," *Psychon. Bull. Rev.*, vol. 20, no. 1, pp. 21–53, Feb. 2013.

[47] S. Micera, J. Rigosa, J. Carpaneto, L. Citi, S. Raspopovic, E. Guglielmelli, A. Benvenuto, L. Rossini, G. Di Pino, G. Cavallo, M. C. Carrozza, C. Cipriani, K. P. Hoffmann, P. Dario, and P. M. Rossini, "On the control of a robot hand by extracting neural signals from the PNS: preliminary results from a human implantation.," *Conf. Proc. IEEE Eng. Med. Biol. Soc.*, vol. 2009, pp. 4586–9, Jan. 2009.

[48] S. Dosen, M. Markovic, K. Somer, B. Graimann, and D. Farina, "EMG Biofeedback for online predictive control of grasping force in a myoelectric prosthesis.," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, p. 55, 2015.

[49] A. W. Shehata, L. F. Engels, M. Controzzi, C. Cipriani, E. J.

Scheme, and J. W. Sensinger, "Improving internal model strength and performance of prosthetic hands using augmented feedback," *J. Neuroeng. Rehabil.*, vol. 15, no. 1, p. 70, 2018.

[50] A. W. Shehata, E. J. Scheme, and J. W. Sensinger, "Audible Feedback Improves Internal Model Strength and Performance of Myoelectric Prosthesis Control," *Sci. Rep.*, vol. 8, no. 1, p. 8541, 2018.

[51] M. Markovic, H. Karnal, B. Graimann, D. Farina, and S. Dosen, "GLIMPSE: Google Glass interface for sensory feedback in myoelectric hand prostheses," *J. Neural Eng.*, vol. 14, no. 3, 2017.

[52] P. Parker, K. Englehart, and B. Hudgins, "Myoelectric signal processing for control of powered limb prostheses.," *J. Electromyogr. Kinesiol.*, vol. 16, no. 6, pp. 541–8, Dec. 2006.

[53] A. Ninu, S. Dosen, S. Muceli, F. Rattay, H. Dietl, and D. Farina, "Closed-loop control of grasping with a myoelectric hand prosthesis: Which are the relevant feedback variables for force control?," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 22, no. 5, pp. 1041–1052, Sep. 2014.

[54] 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, Mar. 2015.

[55] 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," *JPO J. Prosthetics Orthot.*, vol. 28, no. 4, pp. 136–144, 2016.

[56] H. Flor, C. Denke, M. Schaefer, and S. Grüsser, "Effect of sensory discrimination training on cortical reorganisation and phantom limb pain," *Lancet*, vol. 357, no. 9270, pp. 1763–1764, Jun. 2001.

Meike A. Schweisfurth studied mathematics and physics at the University of Göttingen, Germany, and at the University of Cambridge, England, where she completed her mathematical MSc in 2008. She then entered an integrated MSc/PhD program in Neuroscience at the University of Göttingen, from which she obtained her PhD in 2013, focusing on exploration of the tactile modality through functional imaging. Until 2014, she continued working as a post-doctoral researcher at the BiomedNMR at the MPI for Biophysical Chemistry in Göttingen. From 2014 to 2017, she worked in the Research Group for Neurorehabilitation Systems at the University Medical Center Göttingen. Since 2017, she is Associate Professor at the University of Applied Sciences (HAW) in Hamburg, Germany. Her current research focuses on closed-loop-control of upper-limb prostheses, exploring the improvement of forward control as well as the potential benefits and cortical effects of different feedback types.

Cornelia Hartmann received the B.Sc. degree and Diploma in electrical engineering and information technology from Technische Universität München, Munich, Germany, in 2006 and 2008, respectively. From 2008 to 2012 she was with the Research and Development Department of Otto Bock Healthcare GmbH SE & Co. KGaA, Germany. In 2012 she took up a research fellowship at the Department of Neurorehabilitation Systems, University Medical Center Göttingen, Georg August University, Göttingen, Germany, where she also started working towards the and obtained a Ph.D. degree in theoretical and computational neuroscience in 2017. In 2014 she resumed her position with Otto Bock Health Care GmbH SE & Co. KGaA while in parallel continuing work on her doctoral project. Her research interests include closed-loop prosthesis control as well as sensory feedback for prosthesis users.

Felix Schimpf studied applied computer science at the Georg-August University of Göttingen, Germany, where he completed his bachelor degree in cooperation with the Institute of Neurorehabilitation Systems in 2015. He is currently enrolled at the University of Göttingen to earn his master's degree.

Dario Farina (M'01-SM'09) received Ph.D. degrees in automatic control and computer science and in electronics and communications engineering from the Ecole Centrale de Nantes, Nantes, France, and Politecnico di Torino, Italy, in 2001 and 2002, respectively. He is currently Full Professor and Chair in Neurorehabilitation Engineering at the Department of Bioengineering of the Imperial College London, UK. He has previously been Full Professor at Aalborg University, Aalborg, Denmark (until 2010) and at the University Medical Center Göttingen, Georg-August University, Germany, where he founded and directed the Department of Neurorehabilitation Systems (2010-2016), acting as the Chair in Neuroinformatics of the Bernstein Focus Neurotechnology Göttingen. His research focuses on biomedical signal processing, neurorehabilitation technology, and neural control of movement. Within these areas, he has (co)-authored approximately 400 papers in peer-reviewed Journals and has been the Editor of the IEEE/Wiley books "Introduction to Neural Engineering for Motor Rehabilitation" and "Surface Electromyography: physiology, engineering and applications". Among other awards, he has been the recipient of the 2010 IEEE Engineering in Medicine and Biology Society Early Career Achievement Award. Professor Farina has been the President of the International Society of Electrophysiology and Kinesiology (ISEK) (2012-2014) and is currently the Editor-in-Chief of the official Journal of this Society, the Journal of Electromyography and Kinesiology. He is also currently an Editor for IEEE Transactions on Biomedical Engineering and the Journal of Physiology.

Strahinja Dosen (M'05) received the Diploma of Engineering degree in electrical engineering and the M.Sc. degree in biomedical engineering in 2000 and 2004, respectively, from the Faculty of Technical Sciences, University of Novi Sad, Serbia, and the Ph.D. degree in biomedical engineering from Aalborg University, Denmark, in 2009. Between 2011 and 2017, Mr. Dosen was a Research Scientist at the Institute of Neurorehabilitation Systems, University Medical Center Göttingen, Germany. Presently, he is an Associate Professor at the Department of Health Science and Technology, Aalborg University, Denmark. His research interests are movement restoration and control, rehabilitation robotics, sensory feedback and human-machine interfacing for sensory-motor integration.