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Sensory feedback in upper limb prosthetics

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Abstract:

One of the challenges facing prosthetic designers and engineers is to restore the missing sensory function inherit to hand amputation. Several different techniques can be employed to provide amputees with sensory feedback: sensory substitution methods where the recorded stimulus is not only transferred to the amputee, but also translated to a different modality, modality matched feedback which transfers the stimulus without translation and direct neural stimulation which interacts directly with peripheral afferent nerves. This paper presents an overview of the principal works and devices employed to provide upper limb amputees with sensory feedback. The focus is on sensory substitution and modality matched feedback with the principal features, advantages and disadvantages of the different methods are presented.

Introduction

Losing a limb after amputation is a devastating event with the inevitable loss of both motor and sensory function resulting in a disability with possibly enormous consequences for activities of daily living and quality of life [1], [2]. The grasping function can be primitively restored using an active prosthetic hand, either a body-powered or battery-powered prosthesis. In body-powered prostheses, the person uses gross movements of their own body to control the prosthesis. A Bowden cable is used to transmit the forces developed by the body movements to a terminal device: either a split hook or a mechanical hand (cf. Fig. 1A). In technologically advanced battery-powered prostheses, electromyographic (EMG) signals from residual muscles, are tapped through surface electrodes and electrically processed to functionally operate an electromechanical hook/hand with electrical motors (Fig. 1B) [3]. While an acceptable level of grasping function is often gained through active prostheses, the restoration of sensory function remains one of the open challenges in this field, which contributes preventing a wider acceptance and ease of use of these devices, as reported by surveys among amputees [1], [2].

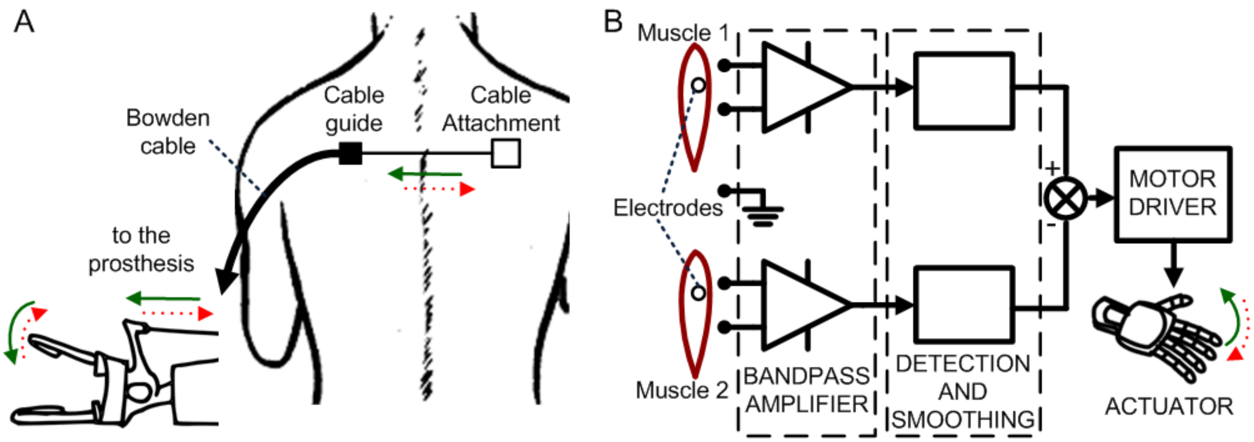

Figure 1 – Conventional powered prostheses. A) In a body-powered prosthesis the movement developed by a body part (in this case biscapular abduction) is transmitted to the terminal hand/hook by means of a cable running into a guide (Bowden cable). The cable in turn transmits reaction forces from the hand/hook to the body, hence providing a sensory awareness of the prosthesis. B) In a battery-powered myoelectric prosthesis the signals recorded from antagonist muscles are processed to open/close the various degrees of freedom of the prosthesis. Besides incidental stimulation (motor sound, socket pressure and vibration, etc.) there is no built-in sensory feedback to the individual.

The control of grasping and manipulation largely relies on tactile feedback, in humans [4]; hence a common thought is that prostheses would function better if they used closed-loop control, making use of both exteroceptive and proprioceptive information [5], [6]. To achieve this goal the prosthesis should not only be able to detect physical interactions with the environment (i.e. exteroception), and sense the proper state like its joint angles (proprioception); it should be able to convey such information to the user in a perceivable and possibly effortless manner. Nowadays none of the prostheses used in clinical practice, have purposely designed closed-loop controllers. Control is achieved by the individual by means of visual feedback and incidental stimulation (audition, socket pressure, harness, etc.) but not often through design intention. With body-powered devices, users can sense the prosthesis state and grip strength, through the reaction forces transmitted by the control cable and harness on their skin/body. Even though this type of feedback is limited, it together with the low weight and cost most likely contributes to the selection of body-powered prostheses over myoelectric ones. Paradoxically, in fact, the more sophisticated myoelectric prostheses lack this sensory awareness that cable and harness attached to the body provide to the wearer, the user must rely on vision to regulate his or her activity with significant cognitive effort. However, technological improvements are making myoelectric robotic multi- fingered prosthesis more reliable, and they may become more popular in the future, hence it is important to identify ways of providing feedback to the user of prosthesis state and grip strength.

Various approaches to and designs of sensory feedback systems have been presented over the years, but none has yet been convincingly proven usable and thus been made commercially available. In this paper we review the important features describing sensory feedback in upper limb prosthetics as well as summarize significant work carried out in the field. Since an exhaustive review cannot be accomplished here, we wish to apologize to those researchers engaged in important work that were not mentioned.

Sensing information and feeding it back

Somatic receptors in the upper limb are divided in cutaneous and subcutaneous mechanoreceptors, muscle and skeletal mechanoreceptors, nociceptors and thermal receptors. This complex sensory system encodes and transmits, to the central nervous system (CNS), information about four major modalities: touch, proprioception, pain and temperature. After amputation, i.e. after the loss of receptors and interruption of the physiological channels, there are two potential ways to elicit sensory feedback: 1) invasively, by interfacing directly to physiologically relevant neural structures in the peripheral nervous system (PNS) or the CNS or 2) non-invasively, by providing feedback to intact sensory systems (e.g., tactile stimuli on the residual limb, chest, etc.). In both cases, the subject should be trained to associate stimuli to physical events occurring at the prosthesis (exteroception) or to states of the prosthesis (proprioception).

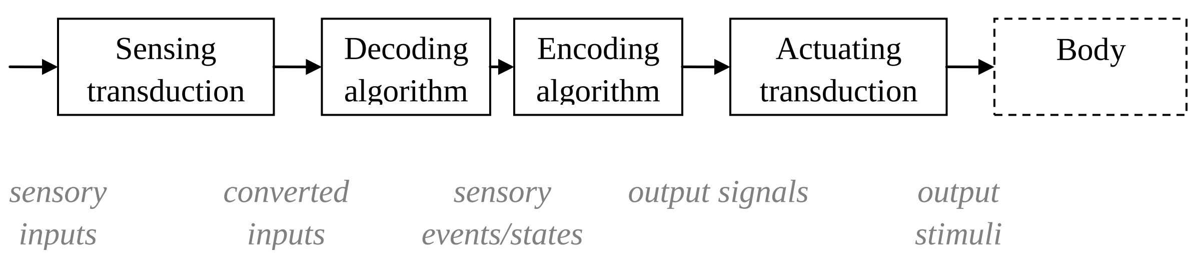

Figure 2 – General architecture for providing sensory feedback. The input stage, namely sensing transduction converts physical stimuli (*sensory inputs*) from one form of energy (typically mechanical) to another form of energy (*converted inputs*, typically electrical) which is more appropriate for processing. Then a decoding algorithm identifies important *sensory-events/states* (e.g. contact or angular position of a joint), and an encoding algorithm transforms them into *output signals* that can be interpreted by the CNS as if they were a substitute of the *sensory input*. The output stage, i.e. actuating transduction, converts the output signals into the appropriate form of energy to be applied on body sensory systems (*output stimuli*) either invasively or non-invasively.

The general architecture of providing sensory feedback could be described by four blocks in cascade that comprise: 1) sensing transduction; 2) decoding algorithm; 3) encoding algorithm; and; 4) actuating transduction, as detailed in Fig. 2 (and caption). Body-powered prostheses represent a very simple yet smart example of the above scheme, in which the whole process is executed mechanically and the feedback comes from the control interface, which is under direct control of the CNS. No signal processing is required and sensing/actuating transducers are substituted by mechanical couplings. Battery-powered myoelectric prostheses lack this sensory awareness, and sensory feedback can be attained and provided by means of sensors (for a review of sensors in artificial hands see [7]) and actuators. In a prosthesis interfaced to the afferent nerves through implanted electrodes [8], the actuating transduction block is likely to be an electronic amplification/conversion stage like a voltage to current converter (e.g. [9], [10]).

Timing Aspects

In addition to the sensory modality (*what* is fed back), another crucial aspect is the timing of sensory feedback (*when* is fed back). The latency between a variation of the output stimulus corresponding to a sensory input variation should be as short as possible for an effective volitional use of such information. In the unimpaired human sensory system, tactile stimuli take 14-28 ms to reach the cuneate nucleus [4]. Therefore it is reasonable to believe that to avoid increasing this value significantly, artificial sensory feedback should be delivered to the individual in a fraction of that time (e.g. 3-5 ms).

Short latencies are also important for the brain to develop a sense of embodiment (body ownership) of the prosthesis. Indeed, the attribution of a visible hand to the self depends on a match between the afferent somatic signals and visual (and eventually audio) feedback from the hand [11]. Self-attribution occurs with temporal delays up to 300 ms, as reported by Shimada et al. [12].

The prevalence of stimulation, i.e. how continuous sensory feedback is to be provided is still a debating point within the field. Traditionally, researchers have implemented systems that presented the sensory feedback in a continuous fashion. However, continuous feedback yields to adaptation, meaning that the stimulation is no longer or just barely perceived by the individual after a short while. This is especially true when sensory substitution (see paragraphs below) is applied and is likely to be caused by the fact that the brain interprets the feedback as irrelevant information containing little variation. Other approaches, like that proposed by Johansson and Edin [13], build on the fact that motor tasks in humans are organized by means of discrete sensory "events" that delimit functional task phases (e.g., object contact, lift-off, etc,) and hence suggest that sensory feedback should be provided in a discrete sensory-event related fashion. They argue that although transferring such events requires good temporal precision, the band-width requirement is evidently much smaller compared to traditional solutions and no adaptation would occur.

Closed-loop control

A prosthesis with sensory feedback can exploit the concept of closed-loop control [5]. D. Childress originally proposed the division of the sensory information from the prosthesis to the individual, into three pathways as shown in Fig. 3 [14]. Pathway A consists of visual or auditory feedback signals that are present by default in all prosthetic systems. Users rely on viewing the terminal device (and to experience gained with usage) while grasping an object, to roughly regulate the grip strength. Furthermore, *incidental* auditory cues like the sound produced when the hand/hook touches the object, as well as the motor sound (in battery-powered prostheses) while grasping are often exploited for control. Pathway B consists of somatic sensory signals (e.g. tactile, proprioception, temperature, vibration) which are directed either to the skin, or directly to the afferent nervous system. Incidental somatic signals like hand vibrations or pressures transmitted to the residual limb by the socket are also included in pathway B and seldom used for control. Pathway C consists of feedback that is intrinsic to the prosthesis control system, e.g. using sensors in the prosthetic hand to automatically adjust the grip force, or to avoid slippage of objects.

There are several sensor options possible for integration in myoelectric hands, capable of measuring a wide range of information like: contact, force, pressure, position, slip and temperature. In commercial devices (like the Sensor-Hand, i-Limb, BeBionic [101], [102], [103]) sensors are used to automatically regulate the grip force (Pathway C in Figure 3) without having the user in the loop. As an example, in the newly marketed multi-fingered hands (i-Limb and BeBionic) motor current sensors are used to stall the different fingers independently, in order to achieve an adaptive grasp. Pathway C is important from the engineering perspective; it also resembles peripheral mechanisms existing in the natural unimpaired hand that prevent slippage of objects [15]. It seems reasonable that these processes should proceed without requiring directed attention (i.e. as automatic subroutines) in order to avoid interference with the overall goal of the manipulative task [5], [14], [15].

To enable the user to regulate the grip force/joint position according to what is sensed, the loop needs to be closed through pathway B. In the following paragraphs we describe systems and techniques to implement pathway B that have been attempted by researchers.

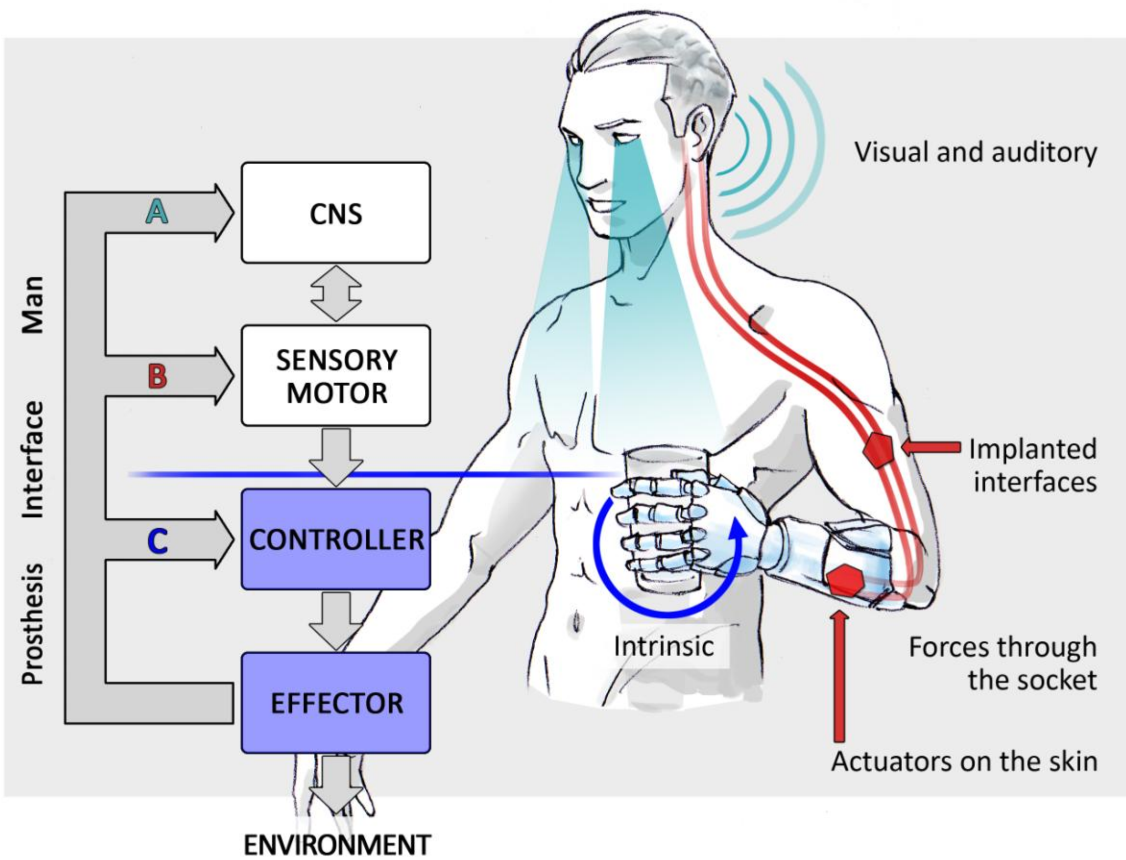

Figure 3 – Diagram of man-prosthesis system showing the three possible information pathways proposed by Childress [14]. Different colors correspond to different pathways. Pathway A (in light green) is related to sensory information directly fed back to the Central Nervous System (CNS), like visual and auditory feedback. Pathway B (in red) refers to sensory information conveyed to functional sensory motor systems either invasively or non-invasively. Pathway C related to the intrinsic feedback within the prosthesis is sketched in navy blue.

Sensory-substitution Feedback

Sensory substitution is a method to provide sensory information to the body, through a sensory channel different from that normally used (e.g. substitute touch with hearing), or through the same channel but in a different modality (e.g. substitute pressure with vibration) [16]. Most of the feedback systems proposed for battery-powered upper limb prosthetics exploit this idea, due to the challenging task of restoring the physiologic sensory information, when the original mechanoreceptors are no longer available. Prevalent techniques have been vibrotactile and electrotactile sensory substitution which uses either a mechanical vibration or an electric current presented to the skin in order to code information relative to the grasp, joint position or direction.

Vibrotactile Sensory Substitution

Vibrotactile stimulation is evoked by a mechanical vibration of the skin, at frequencies ranging between 10 and 500 Hz [16]. The two main features of the stimulus are vibration amplitude and frequency, but other features like pulse duration, shape, and duty cycle, can be modulated to convey different kinds of information [17], [18]. The amplitude discrimination threshold depends on several parameters, including the frequency and location on the body. Lower thresholds are found on glabrous skin as compared to hairy skin, and at frequencies in the range 150-300 Hz. Highest thresholds are found in the abdominal and gluteal regions (4–14 µm at 200 Hz), whereas lowest thresholds are measured on the fingertips (0.07 µm at 200 Hz) [19].

In prosthetics vibrotactile feedback was firstly proposed by Conzelman et al., in 1953 [20]; since then it has been explored due to its compatibility with EMG control and higher acceptability compared to electrotactile stimulation [16], [21]. Bach-Y-Rita and Collins [22] proposed concepts where arrays of vibrotactile stimulators would convey proprioceptive information on the back or on the residual limb of the amputee. Mann et al. [23] detailed a system where the elbow angle of the Boston Arm was fed back exploiting vibrotactile stimulators. Early devices were quite bulky and power consuming. Today’s vibrators can be low-power, unobtrusive and potentially embeddable within the socket [17], [24], [25]. In the recent years vibrotactile systems have been used in research with the Otto Bock, Motion Control and iLimb myoelectric prostheses [25], [26], [27], and with the Cyber- [28], MANUS- [29], Fluid- [30] and Smart- [24] hands. Generally, the use of vibrotactile feedback improves user performance through a better control of grip force and by lowering the number of errors in task execution. However, a comparison between direct haptic feedback (force) and vibrotactile feedback showed that task execution time was lower with direct feedback [31].

Electrotactile Sensory Substitution

Electrotactile (or electrocutaneous) stimulation evokes sensations within the skin by stimulating afferent nerve endings through a local electrical current. The sensation is not necessarily confined to the zone under the electrodes but elicited sensations could spread if these are placed near nerve bundles [16]. Electrotactile stimulators can be designed to be either current- or voltage-regulated. With current-regulated stimulation, the current is not affected by changes in the tissue load and impedance at the electrode interface; on the other hand, voltage-regulated stimulation minimizes the possibility of skin burns owing to high current densities. Current amplitude, pulse waveform (biphasic/monophasic, rectangular/sinusoidal), pulse frequency and duration, and duration of pulse bursts are the principal features of the stimulation. Typical currents range within 1-20 mA, with pulse frequencies ranging from 1 Hz to 5 kHz; biphasic pulses produce more comfortable sensations compared to monophasic pulses [32]. Saunders [33] reported that the average threshold charge for pulses repeated at 60 and 200 Hz was 62 nC, and that this value was constant across a range of current amplitudes (1-20 mA), pulse durations (1-100 us), and frequencies (60-200 Hz). Subjects described electrotactile sensations qualitatively as a tingle, itch, vibration, touch, pressure, pinch, and sharp and burning pain depending on the stimulating voltage, current and waveform, as well as on the electrode size, material and contact force, and the skin location, hydration, and thickness [16]. Typically electrotactile devices consume less power and respond faster than vibrotactile systems as there are no moving mechanical parts. The main drawback is interference with myoelectric control when the stimulation sites are close to the EMG electrodes; frequency or time multiplexing techniques were investigated to address this pitfall [34], [35].

In prosthetics several researchers investigated ways to convey grip or position information by modulating the amplitude of the pulses [36], the frequency [37], or the rate of pulse bursts [36],

[38] to a single or multiple [39] stimulation sites. Amplitude modulation was shown to be superior to pulse rate modulation in conveying grip intensity, however pulse rate modulation was less susceptible to adaptation [16]. Electrodes with a small area are required when only a limited area is available; however larger electrodes provide a more comfortable sensation [16]. Hence, a trade-off between electrode size and skin area should be identified when performing multi-site stimulation.

Other Sensory Substitution Principles

Other sensory substitution principles have been employed to provide information about the state of the prosthesis and what it is touching. Auditory sensory substitution can be achieved e.g. by having the force of a grasp modulating the pitch, timbre or volume of auditory signals. Lundborg et al. [40] and more recently by Gonzales et al. [41] demonstrated, by conveying the information relative to the movement of different fingers by different sounds, that it is possible to employ auditory feedback to convey artificial proprioceptive and exteroceptive information.. Wheeler et al. [42] developed a new sensory substitution device that stretches the skin in order to convey the position and motion of prosthetic joints.

Modality-matched Feedback

Feedback is said to be modality-matched when the output stimulus is felt in the same modality as the sensory input (cf. Fig. 2); e.g. touch on the prosthesis felt as a touch. Reproducing pure modality-matched feedback has been a continuing challenge for researchers in this field, since the training to associate output stimuli to sensory inputs would be virtually effortless to the amputee, and the stimuli would be felt as natural. Unfortunately this represents a nontrivial task, and modality matched feedback is practically difficult to achieve. In his interesting work R. Riso [43] reviewed the potential strategies to provide modality matched feedback to amputees.

In theory it is possible to regain modality-matched touch sensations (contact, normal and shear force/pressure, vibration, texture, temperature) using non-invasive electromechanical devices coupled with thermoelectric devices (e.g. Peltier cells) [44] applied onto the skin, e.g. of the residual limb [24], [45], the chest [46], [47] and other body parts [48]. Modality-matched proprioception is intrinsically a trickier exercise for engineers, since the joint angle (e.g. of the fingers, or of the wrist) needs to be transferred to another sound joint, in order to match the modality. This concept grounds one of the principles of prosthetic design: Extended Physiologic Proprioception (EPP) introduced by D. C. Simpson [49]. EPP means that the human-machine interface is configured such that the body's own physiological mechanisms are directly related to the activation and sensing of the device being controlled. Weir et al. [50] conveyed prosthesis state by interfacing the prosthesis directly with cineplastized muscles on residual limb of amputees, whereas Gillespie et al. [51] employed an exoskeleton to provide the grip reaction force as a reaction torque about the elbow. The techniques were demonstrated to increase the efficacy of control. A different approach to modality-matched proprioception is that discovered by Goodwin et al. [52], lately exploited by Roll and Gilhodes [53] which demonstrated that if a localized vibratory stimulus (in the range of 40-80 Hz) is applied over the tendons at the wrist, normal subjects perceive movement of the wrist joint when in fact there is no motion. To explain these effects, it has been theorized that vibrating the flexor tendons induces neural activity in stretch receptors; hence such findings suggest that if the appropriate spindle afferents were stimulated in an amputee's sensory nerve, then feedback of hand position might be possible.

Mechanotactile Stimulation

Mechanotactile feedback is accomplished when a force normal to the skin is applied by a pusher to convey sensory information. This represents a modality-matched sensory feedback paradigm, when force sensors in the fingers are used to detect the sensory input (cf. Fig. 2). The main features of this class of stimulators are accuracy (how accurately the output force/pressure resembles the sensory input), precision, range, resolution and bandwidth (or its inverse concept, i.e. response time). Drawbacks like complex miniaturization, weight, and energy consumption hamper the exploitation of current haptic/robotic technologies in portable systems. In addition, since these systems are usually constituted by moving parts with a mechanical inertia, another limit is the response of the system that in many cases is slow.

One of the most clever examples of mechanotactile feedback in prosthetics, is the concept proposed by Rosset (1916) in which a pneumatic system composed of pressure pads and a tube transmitted pressure from the fingers of the prosthesis to the residual limb, directly [6]. This concept was further investigated and implemented using the phantom hand map as the target for sensory feedback by Antfolk et al. [45]. Other examples are the works by Meek et al. [54], which embedded force sensors in a cable driven Dorrance hook to provide force feedback using a single motor-driven pusher, on the skin of the forearm, and by Patterson et al. [55] that employed a pneumatic cuff to display single-site pressure feedback on the arm. Antfolk and colleagues proposed the concept of a multi-site mechanotactile system and exploited it to investigate localization and pressure level discrimination on the residual limbs of transradial amputees [57]. Panarese et al. [48] targeted a different body area: they investigated the effects of modality-matched force feedback on the glabrous skin of the big and second toe, demonstrating that individuals were able to integrate it in their control. A state of the art device (and reading) is the work by Kim and colleagues [56]; they developed a display able to regain modality-matched contact, pressure, vibration shear force, and temperature sensations. Sensinger et al. [46] and Marasco et al. [47] investigated the perception threshold and sense of embodiment by employing such device on the chest of amputees having undergone targeted reinnervation surgery, a method by which skin near or over the targeted muscle is reinnervated with afferent fibers of the remaining hand nerves, hence, when this skin is touched, it provides the amputee with a sense of the missing arm or hand being touched..

Direct-neural Stimulation

Since action potentials travel in the form of electrical signals, in pure theory it could be possible to recover all somatic sensory information (i.e. touch, proprioception, and temperature) with modality matching, using invasive neural electrodes implanted in the PNS, in afferents originally serving the fingers and palm (radial, median, ulnar nerve) (a complete review on neural interfaces is the work by Micera and colleagues [8]). Experimental evidence demonstrated that this goal is still far from being achieved due to technological limitations of neural interfaces and consequently the relatively still poor knowledge on how the hand encodes stimuli and conveys them to the brain. Direct nerve stimulation showed to be a viable means for eliciting proprioceptive and touch sensations in several studies. Different type of electrodes were employed to convey the information recorded from the prosthesis: cuff electrodes, where electrode wires are wrapped around the nerve [58]; longitudinal intrafascicular electrodes (LIFE), where electrodes placed in the nerves longitudinally, [9], [10], [59]; but the quality of the sensation perceived by individuals was frequently a foreign feeling, resembling e.g. paresthesia, vibration, tapping or flutter on the skin. These unnatural feelings are thought to be due to the low selectivity of available neural interfaces (that contact bundle of afferents all together), which causes many different types of cutaneous afferents to be activated all at once rather than with precise timing conditions [4]. Techniques to elicit referred tactile or proprioceptive sensations through direct-nerve stimulation are similar to those used in non-invasive electrotactile stimulation, although the electrical charge threshold to elicit sensations is one to two orders of magnitude lower (in the order of 0.1-10 nC [10]).

Expert commentary

Research in the field mainly focus on sensory substitution in battery-powered prostheses, and can be roughly classified in three different kinds of studies: a) works where new devices or stimulation techniques were investigated in **psychophysics studies** (e.g. [17], [57], [10]); b) works in which sensory feedback systems were used in **matching force or position** paradigms (e.g. [27], [30], [42]); c) research that focused on applied scenarios studying if feedback aided performance in **functional grasping** tasks (e.g. **Errore. L'origine riferimento non è stata trovata.**[25], [28], [51]). Table 1 summarizes some of these studies. Results from psychophysics studies corroborated the idea of providing sensory feedback through pathway B; as an example, the number of just-noticeable differences was estimated at from 6 to 59 levels for electrotactile stimulation and 15 levels for vibrotactile [16]. Moreover matching-force/position studies generally demonstrated improved control accuracy when extended feedback is added to vision alone. Finally, outcomes from studies that investigated functional grasps using actual prostheses and sensory substitution have been positive (i.e., subjects considered the feedback important and/or the performance slightly improved), but however, not of the expected significance. It appeared that when visual feedback could be used, the additional (modality mismatched) sensory feedback was not practically exploited. We argue that known sensory substation techniques can be perceived only at the conscious level, and they require a cognitive load far beyond the one needed for processing visual input, due to the modality-mismatch that the CNS has to interpret. To overcome this limitation we suggest to present sensory-substitution feedback in more physiological ways: one possibility is to stimulate afferent pathways that served the limb, as in the works by Marasco et al., where feedback was delivered on the chest of targeted reinnervated amputees [47], or by Antfolk et al., that exploited referred phantom finger maps on the residual limb [24], [45]. In both cases the stimulation was felt as if it was coming from the phantom fingers, or in other words somatotopically matched. Under such hypotheses we expect that physiologically relevant feedback would aid the control of functional grasps also with modality mismatch stimulation. An alternative way to bring artificial feedback to the unconscious level is to get inspiration from human sensory-motor mechanisms and try to substitute them, as proposed by the discrete sensory-event driven control technique described by Johansson and Edin [13]. Direct neural stimulation and mechanotactile displays can also represent viable ways to present functionally-usable feedback, but from our point of view they still have to overcome challenging clinical and technological limitations before becoming of practical use. Ironically after decades of research in upper limb prosthetics, robotics, haptics and applied neuroscience, it is the very simple architecture of the body-powered prosthesis dated 1912, that remains the only device coming close to providing physiologically correct and acceptable sensory feedback to the user.

Five-year view

- Advanced robotic hands or multi-fingered prostheses are nowadays available to a wide audience of researchers. Therefore it is expected that studies using physical devices and focusing on the contribution of sensory feedback to the performance of functional activities, will be carried out more frequently. These studies should focus on the identification of which information, collected by the sensors embedded in the prostheses, are important for the control of the prosthesis by choosing the information to deliver to the user and comparing the task performance.New research on direct neural stimulation of the PNS using more selective neural interfaces should be carried out in humans, in order to provide important perspectives on its clinical viability, as well as improve knowledge of hand-brain communication, that is particularly important for the realization of devices capable to provide modality matched sensations.

- Currently available vibrotactile motors are very small, cheap and low-power. They already entered the prosthetic-components market (in 2002 the Boston elbow was featured with a vibrotactile unit to acknowledge the user of the state of the prosthesis [104]), and it is expected that they will spread more and become available with multi-fingered hands equipped with motor-torque sensors, hence providing contact and/or force feedback. In this way the effect of feedback devices in daily use in the control of prosthesis could be evaluated on amputees.

Key issues

- Since natural grasping largely relies on tactile feedback, a common thought is that prostheses would function better if they used closed-loop control, making use of both exteroceptive and proprioceptive information.

- None of today's prostheses have purposely designed sensory feedback. Body-powered prostheses provide awareness of the terminal device through the control cable. Battery-powered prostheses provide only incidental clues. Hence users largely rely on vision.

- Sensory feedback can be elicited invasively, by interfacing directly the nerves or non-invasively, by providing feedback on the skin. Training by the individual is always required.

- Sensory substitution was often investigated in prosthetic research. Prevalent techniques have been vibrotactile and electrotactile. Although the sensation is substituted, these devices are low-cost and a low-power and can be actually be integrated in the prosthetic socket.

- Modality-matched feedback is achievable with non-invasive mechanotactile devices; however these are bulky, power-consuming and expensive.

- Direct nerve stimulation is a viable means for eliciting proprioceptive and touch sensations, but the quality of the sensation is usually a foreign feeling, due to technological limitations of today's neural interfaces.

- Timing of feedback is important. Delay from the sensory input should be in the order of milliseconds. Discrete sensory feedback (i.e. non-continuous) could be a viable technique for making feedback effective and avoiding adaptation.

Table 1. Summary of studies involving sensory feedback in upper limb prosthetics

Study (year)	Input	Output	Stimulation position	Main findings	Ref
Mann et al (1970)	Elbow joint position	V	Residual limb of upper limb amputee	Improvement of precision and accuracy in positioning tasks with sensory feedback	[23]

Pylatiuk et al (2006)	Grasp force	V	Residual limb of transradial amputee	A significant reduction of the applied grasping force with feedback	[30]
Cipriani et al (2008)	Grasp force	V	Upper arm of healthy subjects	Subjective reports stated the feedback system would be useful during grasping task, however, no statistical difference was found between using the system or not	[28]
Chatterjee et al (2008)	Grasp force	V	Upper arm of healthy subjects	Visual feedback improved performance at all force levels. Training is needed to fully utilize vibrotactile feedback	[27]
Saunders et al (2011)	Grasp force	V	Forearm of healthy subjects	With feedforward controller uncertainty, after training, either visual or vibrotactile feedback enabled successful task performance	[25]
Scott et al (1980)	Grasp force	E	Residual limb of transradial amputees	Subjects reported satisfaction with the electrotactile feedback	[35]
Wang et al (1995)	Grasp force	E	Forearm of healthy subjects	Users can differentiate the appropriate gripping force for a wide class of activities	[37]
Lundborg et al (1998)	Finger force	E	Upper arm of nerve injury patients and amputees	With feedback, users were able to discriminate location and regulate grip force to predefined levels	[36]
Meek et al (1989)	Grasp force	M	Forearm of healthy subjects	Successful manipulations of the test's object increased with the use of sensory feedback	[54]
Panarese et al (2009)	Grasp force	M	The first and second toe of healthy subjects	Participants incorporated sensory feedback received on the foot in their sensorimotor control of a robotic hand	[48]
Sensinger et al (2009)	Force	M	Reinnervated chest area of amputees	Feedback area had near-normal force sensitivity compared to contralateral normal skin	[46]
Antfolk et al (2012)	Passive hand touch	M	Residual limb of transradial amputees	Subjects are able to discriminate feedback sites and pressure levels using a completely passive system	[45]
Weir et al (2001)	Grasp force and finger position	O (EPP feedback)	Residual limb of amputees with cineplasty	Cineplasty and harness-based body powered control showed similar performance with lower variability using cineplasty suggesting a higher consistency	[50]
Wheeler et al (2010)	Elbow joint position	O (skin stretch)	Upper arm of healthy subjects	Targeting errors in blind movements with the feedback device were lower with feedback.	[42]
Gonzales et al (2012)	Hand configuration	O (auditory)	-	The usage of an auditory display to monitor and control a robot hand improves the temporal performance greatly, and reduces mental effort.	[41]
Clippinger et al. (1974)	Grasp force	N	Medial nerve of amputees	The patient correlated the grasp force to the sensation provided by direct nerve stimulation	[58]
Dhillon et al (2005)	Elbow joint position and finger force	N	Median nerve of amputees	Participants could discriminate force applied to the thumb sensor and static position of the elbow joint of the artificial arm	[9]
Rossini et al. (2010)	Finger movement	N	Medial and ulnar nerve of amputees	Phantom limb syndrome was alleviated after 4 weeks of use	[10]
Horch et al (2011)	Finger force and position	N	Ulnar and median nerve of amputees	Tactile and proprioceptive feedback is needed when discriminating object size and stiffness	[59]
Patterson et al (1992)	Grasp force	V,M	Upper arm of healthy subjects	Vision together mechanotactile feedback produced the lowest error rates	[55]
Marasco et al (2011)	Passive hand touch	V,M	Reinnervated area of the upper arm of amputees	Physiologically appropriate sensory feedback appears to elicit an incorporation of a prosthetic limb into the self-image.	[47]
Antfolk et al	Passive hand	V,M	Forearm of healthy	Placement of feedback devices on a complete	[57]

(2012)	touch		subjects and residual limb of forearm amputees	phantom map improves multi-site sensory feedback discrimination	

The letters E, V, M, N and O indicate respectively electrotactile, vibrotactile, mechanotactile direct nervous stimulation and other. In the last case, the type is indicated between the parentheses.

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