Robert to transfer the nerve signals from the brain to the prostheses

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Current treatment of upper limb amputation restores some degree of functional ability, but this ability falls far below the standard set by the natural arm. Although acceptance rates can be high when patients are highly motivated and receive proper training and care, current prostheses often fail to meet the daily needs of amputees and frequently are abandoned. Recent advancements in science and technology have led to promising methods of accessing neural information for communication or control. Researchers have explored invasive and noninvasive methods of connecting with muscles, nerves, or the brain to provide increased functionality for patients experiencing disease or injury, including amputation. These techniques offer hope of more natural and intuitive prosthesis control, and therefore increased quality of life for amputees. In this review, we discuss the current state of the art of neural interfaces, particularly those that may find application within the prosthetics field. *PM R 2011;3:55-67*

INTRODUCTION

The market for upper limb prosthetic devices is relatively small. In the United States in 2005, the estimated prevalence of arm amputation above the wrist was approximately 41,000, whereas the estimated prevalence of leg amputation above the foot was approximately 623,000 [1]. As a result, progress in the design and control of upper limb prostheses has come relatively slowly. For example, today's body-powered prostheses are largely similar to the original design patented in 1912 [2]. Major advances have primarily been made in the aftermath of wide-scale tragedies. Modern body-powered prostheses were developed in response to the large number of amputations resulting from the American Civil War. They were then improved during World War I and advanced after World War II [3]. Externally powered (eg, hydraulic) prostheses were first invented after World War I and further developed after World War II [4]. Myoelectric prostheses were first investigated in the late 1940s and developed in the late 1950s and early 1960s in response to the thalidomide tragedy, which caused birth defects in more than 10,000 children worldwide. The latest attention to this field has been largely fueled by ongoing conflicts in Iraq and Afghanistan, and by technological and scientific advances that suggest the feasibility of neurally controlled prosthetic arms.

Current Prosthetic Treatment of Upper Limb Amputation

Functional prostheses for upper limb amputees currently fall into 1 of 3 categories: (1) body-powered, (2) myoelectric, and (3) hybrid (Figure 1). Body-powered prostheses are largely mechanical devices. To control them, amputees use remaining shoulder movements to pull on a cable and sequentially operate prosthetic functions such as the elbow, wrist, and terminal device. To switch between functions, users must lock the joints they wish to remain stationary by pressing a switch or using body movements to pull a locking cable. Myoelectric prostheses are motorized and are controlled via surface electromyogram (EMG) signals from residual muscles sites. Control of myoelectric prostheses is generally achieved by recording from 2 independent muscles or by differentiating weak and strong contractions of 1 muscle [5].

Figure 1. Examples of (A) body-powered and (B) myoelectric prostheses for shoulder disarticulation amputees.

Switching techniques such as muscle co-contraction are commonly used to sequentially operate more than 1 joint. Mechanical switches, linear transducers, and force-sensitive resistors also can be used for switching or to control an additional joint. For a typical fitting, a below-elbow amputee will contract wrist flexor and extensor muscles to control terminal device closure and opening. An above-elbow amputee will contract the biceps and triceps muscles to control elbow flexion and extension or terminal device closure and opening. Shoulder disarticulation amputees may use the trapezius, latissimus, pectoral, or deltoid muscles for control of the elbow or terminal device. It is common practice to combine myoelectric control and body-powered operation in a hybrid prosthesis, such as a body-powered elbow combined with a myoelectric terminal device. Mechanical switches, locks, and spring assists can also be included.

Choices of treatment are as unique as the individual, and the rehabilitation team must tailor each prosthesis to the abilities, physiology, and preferences of the patient. Most amputees in the United States continue to use body-powered prostheses [6], likely because of their relatively low cost, light weight, high functionality and reliability, and the sensory feedback they provide through cable forces [7]. Myoelectric prostheses offer their own benefits: they are more self-contained and do not require the complex harnessing of bodypowered prostheses, and they offer a better cosmesis, wider range of motion, and higher grip strength [5,7].

Need for Advancements

In 1996, The Institute for Rehabilitation and Research published the results of a survey of the top research priorities of more than 2,000 upper limb amputees [6]. Transradial and transhumeral body-powered prosthesis users desired additional wrist movements and the ability to perform coordinated movements of multiple joints. Transradial myoelectric users, the largest group of myoelectric respondents, indicated that multifunctional, multiarticulated hands were the most important concern. Less need for visual attention and increased wrist movement were also important concerns for myoelectric prosthesis users.

Work is currently underway to address many of the priorities of myoelectric prosthesis users, yet many challenges remain. There has been a recent increase in the attention paid to the development of multifunctional prosthetic hands and wrists [8-11]. However, there remains a gap in methods to control simultaneous movements and to decrease the need for visual attention. As the level of amputation increases, the number of functions to be replaced by the prosthesis increases, yet fewer muscle sites are available for control. In addition, the muscles used for control of high-level prostheses are not physiologically related to distal arm function. This, along with the lack of sensory feedback from the arm, forces patients to pay close visual attention to their movements and causes a high cognitive burden.

In recent years, research into the control of upper limb prostheses has attempted to go far beyond conventional prosthetic treatment by the use of novel interfaces to the nervous system. In addition, neural interfaces developed for other applications, such as communication and control for paralyzed patients, may find eventual use with upper limb prostheses. Many studies have demonstrated that useful neural movement commands can be intercepted from muscles, nerves, and the brain. Each technique introduces unique promise and challenge to neural control of prosthetic arms.

GETTING MORE FROM MUSCLES

Traditional myoelectric control systems take advantage of the neural information provided by muscle contractions. The electrical signals generated during muscle contractions are the result of movement commands generated in the motor cortex and propagated along peripheral nerves. Because myoelectric prostheses use these signals as control commands, they could be considered simple neural interfaces.

Two major limitations of myoelectric control are the difficulty in recording suitable EMG signals and the shortage of information for control of multiple functions. Many patients are unable to produce isolated EMG signals or have difficulty making repeatable contractions. In addition, shifting electrode locations and changing skin conditions (such as sweat) alter EMG signals and can cause control to be unreliable. The limited amount of control information forces patients to use switching techniques to operate more than one joint. These limitations result in less functional control, which in turn can lead to frustration and prosthesis abandonment [12]. Three developing technologies—implantable EMG electrodes, EMG pattern recognition, and targeted reinnervation—may address some of the problems inherent in traditional myoelectric control.

Implantable Electrodes

Implanted electrodes could potentially alleviate many of the difficulties associated with surface EMG recordings. They would not be subject to many of the environmental factors that affect surface recordings and could therefore provide more stable recordings and more consistent control. In addition, implanted electrodes could record focal EMG signals from several muscles, thereby providing additional prosthesis control sites. Focal EMG recording is currently accomplished in short-term research studies by use of percutaneous intramuscular electrodes [13]. Use of this strategy with prosthetic devices would require long-term recording systems; this would require the development of robust and biocompatible percutaneous connectors or fully implanted sensors and the use of telemetry. Because of the risks associated with long-term percutaneous connectors, research in this field has begun to focus on fully implanted systems.

Two research groups have made notable progress in developing wireless, implantable EMG recording systems for use with prosthetic devices [14,15]. Both systems use implantable sensors to record myoelectric signals from muscles and transmit them to a prosthesis controller (Figure 2). The basic concepts of both designs are similar: the implants are wirelessly powered by a magnetic field that is generated on the same external coil that receives the data. Up to 32 implants can currently be used with the system built by Weir et al [15]. This system, called the implantable myoelectric sensor (IMES) system, has been bench tested and tested in animal models [15]; however, it has not yet received approval from the Food and Drug Administration for clinical testing in humans. Several technological issues—such as the reduction of power consumption and the improvement of telemetry to deeper implants—must be addressed before this technology can move forward and become available to patients.

Pattern Recognition of Myoelectric Signals

EMG pattern recognition is one way to increase the amount of information gleaned from muscles and to alleviate the need for isolated EMG signals. These algorithms look for patterns of muscle activity across one or more muscle sites rather than relying on independent EMG signals. This action has the potential to make control easier and more natural for patients. The majority of pattern recognition studies have been performed on nonamputee subjects [16-20] or below-elbow amputees [17,21-24] because there is insufficient information in residual muscles of high-level amputees for wrist and hand control.

A typical pattern recognition system consists of signal detection (with varying numbers, types, and configurations of electrodes), feature extraction (in which defining characteristics of EMG signals, such as magnitude and frequency measures, are calculated), classification (where features are probabilistically assigned to a “class” of movement), and calculation of movement speed. Many researchers have investigated and published different variations of each of these steps [25]. In addition, the functionality of pattern recognition for real-time control of transradial prostheses has recently been demonstrated [26].

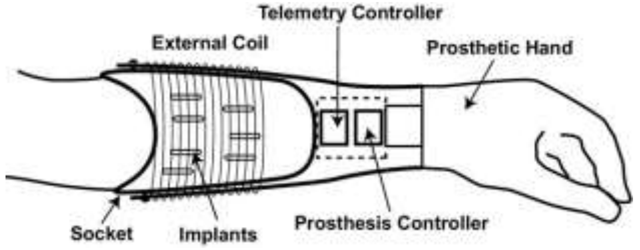

Figure 2. Implantable myoelectric sensor system for a transradial amputee. Adapted from Weir et al [15] with permission.

Pattern recognition systems are not commercially available for patients. To be considered a clinical option, these systems would require a simple and practical method of being trained and refreshed because successful use depends upon the stability of signal patterns. Research is being conducted to improve these systems and to demonstrate their potential advantages over conventional myoelectric control systems. Newer studies focus on feasibility and clinical implementation issues, such as effective methods of training the user, more effective techniques for classifier training, and the design of robust clinical interfaces. These systems could also benefit from the signal stability offered by implantable EMG systems. In addition,

Targeted Reinnervation

Targeted reinnervation increases the amount of information available from muscles, enabling the control of multiple prosthetic functions. This technique is based on the fact that motor commands for the missing limb continue to travel down residual nerves after an amputation. During the targeted reinnervation procedure, these nerves are surgically connected to residual muscles, forming functional connections in 3 to 6 months. Patients can then contract the reinnervated muscles by attempting to move their missing limb. These muscle contractions can be detected by EMG electrodes and used to control a prosthetic limb. Apart from the one-time surgery, the control scheme is entirely noninvasive because surface electrodes are currently used to record the EMG signals. Because of the increased number of myoelectric control sites, patients are able to simultaneously control multiple functions, such as prosthetic hand opening and closure and elbow flexion and extension. Targeted reinnervation is now a clinically available treatment for upper-limb amputees and has been performed on more than 40 patients worldwide.

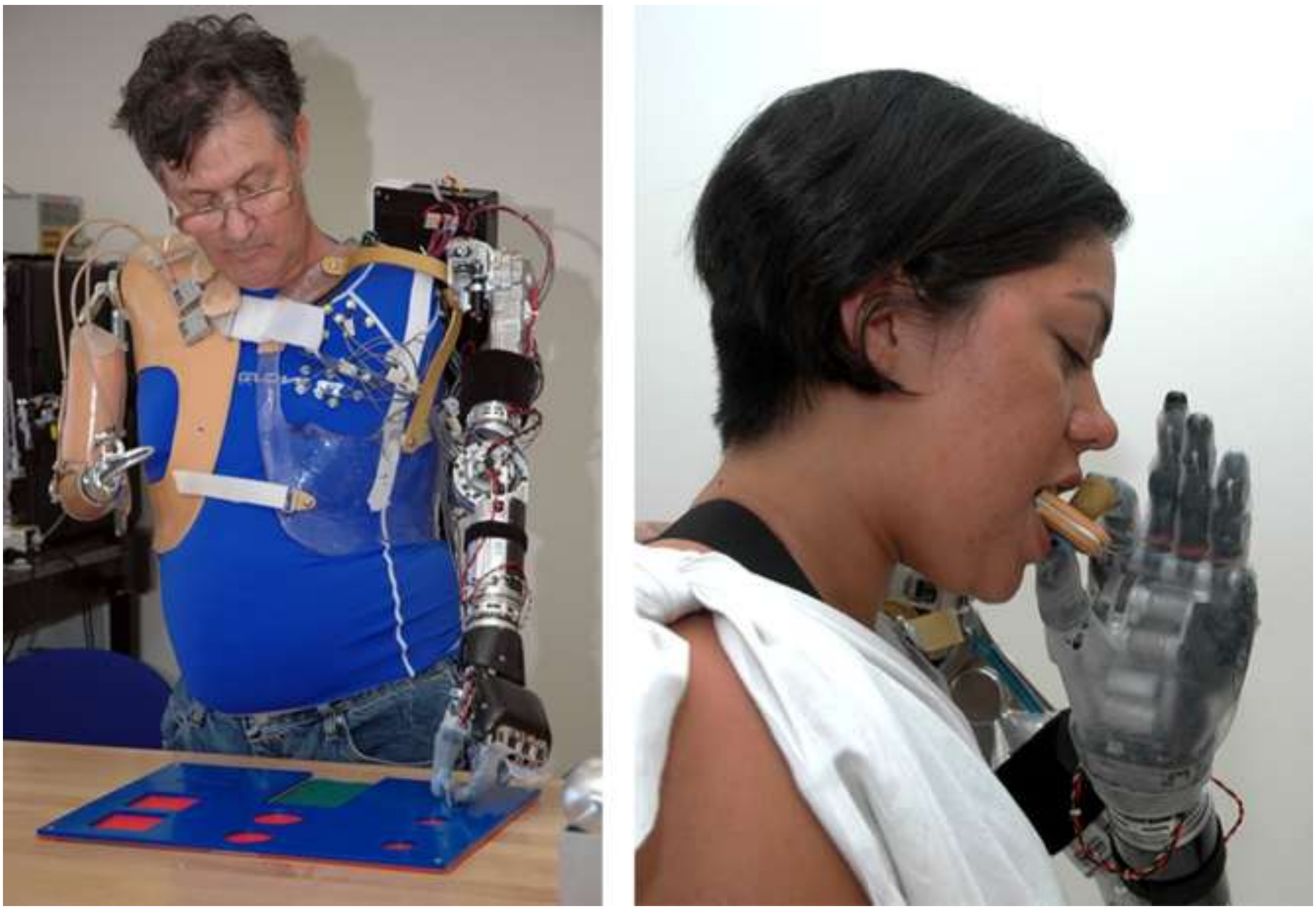

Figure 3. Two targeted reinnervation patients perform functional manipulation tasks with an advanced prosthetic arm. Photos courtesy of the Rehabilitation Institute of Chicago and DEKA Research.

A recent study demonstrated the efficacy of targeted reinnervation in combination with pattern recognition techniques to allow patients with above-elbow amputations to control multiple prosthetic functions in real time [28]. Three patients—2 with shoulder disarticulations and another with a transhumeral amputation—performed several arm and hand movements with a virtual prosthesis and were then fitted with advanced multifunctional prosthesis prototypes.

All patients demonstrated the ability to control elbow, wrist, and hand-grasp movements with both the virtual and physical prostheses (Figure 3).

An additional benefit of targeted reinnervation is the potential it creates for providing cutaneous sensory feedback to amputees. In a number of targeted reinnervation patients, the skin overlying the target muscles has been reinnervated by sensory afferents of the transferred nerves [29]. This circumstance results in the perception that stimuli applied to the reinnervated skin are applied to the missing limb (Figure 4) [29]. Patients can feel light touch, graded pressure, vibration, sharp/dull stimuli, and hot/cold stimuli, with all these sensations perceived on their missing hand or arm [29-31]. This reinnervated skin could be used as a "portal" for applying relevant sensory feedback to the amputee.

Several difficulties associated with myoelectric prosthesis control remain after targeted reinnervation. Current commercial devices rely on EMG magnitude, and it can be difficult to separate the surface EMG signals of different muscles. Once again, successful control depends upon the stability of

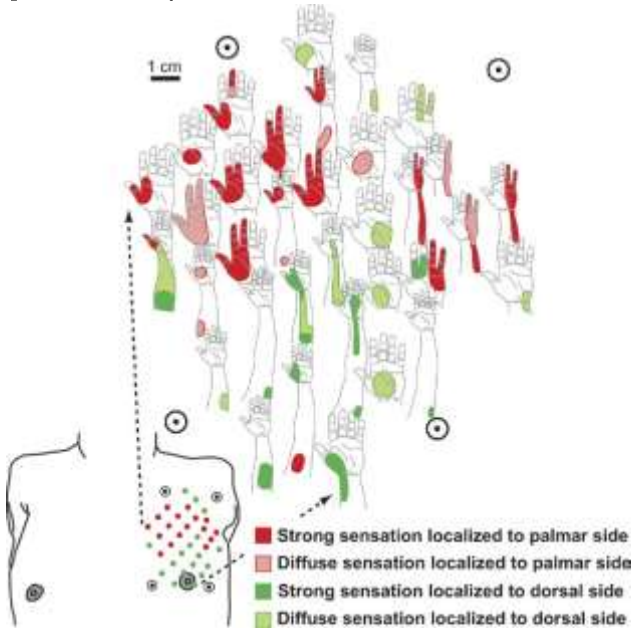

Figure 4. Map of areas on the missing limb in which a patient with targeted reinnervation perceived force (300 g) applied to various points on the reinnervated chest. Reprinted from Kuiken et al [29] with permission. Copyright © 2007 National Academy of Sciences.

EMG signals, whether conventional myoelectric or pattern recognition techniques are used.

INTERCEPTING NERVE SIGNALS

Neural signals also can be intercepted from peripheral motor nerve fibers. These signals are approximately a thousand times smaller than EMG signals generated by muscle contractions but can be detected by electrodes placed inside of or directly adjacent to nerve bundles. In addition, stimulation of afferent nerve fibers may provide sensory and proprioceptive feedback. Although human and animal studies have shown axonal atrophy, motor neuron loss, and a decrease in conduction velocity and fiber diameter after nerve section [3234], the authors of additional studies have demonstrated that motor commands for the missing arm can be generated in the residual nerves of amputees long after amputation has occurred [35-37]. In addition, sensory afferents have been shown to remain viable long term in amputees [36,37]. Researchers have therefore investigated the use of both extraneural and intraneural electrodes for direct nerve recording and stimulation.

Extraneural Electrodes

Extraneural electrodes record and stimulate from outside the nerve. They generally have a low selectivity for recording and stimulating individual axons [38]. One common type of extraneural electrode, the nerve cuff electrode, surrounds the nerve and therefore records and stimulates a few to several combined nerve fascicles, depending on the number and arrangement of contacts (Figure 5A) [39]. An alternative design, the flat-interface nerve electrode, gently flattens the nerve between an array of extraneural electrodes, allowing more direct access to individual fascicles (Figure 5B) [40]. Nerve cuff electrodes are effective for long-term use in functional electrical stimulation systems [41] and are clinically available in systems such as those to correct foot drop (eg, Otto Bock's ActiGait® system [42]). However, they have not yet been widely considered for use in prosthesis control systems.

Intraneural Electrodes

Intraneural electrodes penetrate the nerve, allowing them to record from or stimulate individual or small clusters of nerve

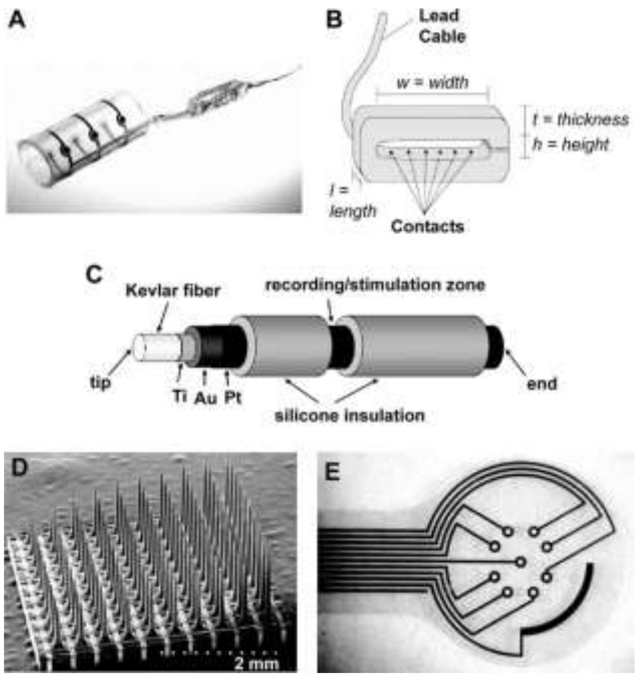

Figure 5. Direct nerve interface electrodes for recording and stimulating nerve fascicles. Extraneural electrodes surround the nerve and record/stimulated from the surface; variations include (A) nerve cuff electrodes (reprinted from Navarro et al [39] with permission from Wiley-Blackwell) and (B) flat-interface nerve electrodes (reproduced from Leventhal and Durand [40] with permission. Copyright © 2009 IEEE). Intraneural electrodes record from inside the nerve bundle; variations include (C) LIFEs (reproduced from Lawrence et al [43], Copyright © 2003, with permission from Elsevier), (D) microelectrode arrays such as the Utah Slant Array (reproduced from Branner et al [46] with permission. Copyright © 2004 IEEE), and (E) regenerative electrodes (reprinted from Lago et al [49], Copyright © 2005, with permission from Elsevier).

fibers. Because of this, intraneural electrodes have a much greater selectivity than extraneural electrodes and require lower stimulus intensities for nerve stimulation.

Longitudinal intrafascicular electrodes (LIFEs) are thin electrodes that are inserted into individual nerve fascicles, parallel to nerve fibers (Figure 5C) [43]. They are biocompatible, and their removal does not require additional surgery, making them a promising option for direct nerve recording and stimulation, particularly for acute or subacute experiments [38]. In a pilot experiment, LIFEs were implanted into the median nerve of 6 amputee subjects. The LIFEs were tested to see whether they were contacting motor or sensory fascicles by electrical stimulation and by recording firing rates during attempted movements (Figure 6) [37,44]. Three subjects then demonstrated the ability to control the prosthetic grip force or the angle of the prosthetic joint, which were proportional to the neuronal firing rate measured from a motor fascicle. Three additional subjects were able to distinguish graded force levels and changing angles applied to the prosthetic limb, conveyed to the patient with varied stimulus intensities applied to a sensory fascicle. This study demonstrated the feasibility of using intraneural electrodes for bidirectional prosthesis control.

Multielectrode arrays contain dozens of electrodes arranged on a rigid base. The arrays are typically inserted into the side of the nerve, piercing the perineurium. Although the large number of recording sites increases selectivity, the rigid structure and the transmission of tethering forces can cause damage to the nerves [38,45]. The Utah Slant Array is probably the most studied array for peripheral nerves (Figure 5D) [46]. This device has been used to record and stimulate motor and sensory fibers in the sciatic nerve and dorsal root ganglion of cats, and has shown promise for future application to neuroprosthetic devices [47,48].

Regenerative (or sieve) electrodes are placed between 2 sides of a transected nerve to selectively record and stimulate nerve fibers as they regenerate (Figure 5E). Studies in animals have shown consistent, although limited, regeneration through sieve electrodes. However, long-term use is significantly challenged by a progressive loss of nerve fibers and

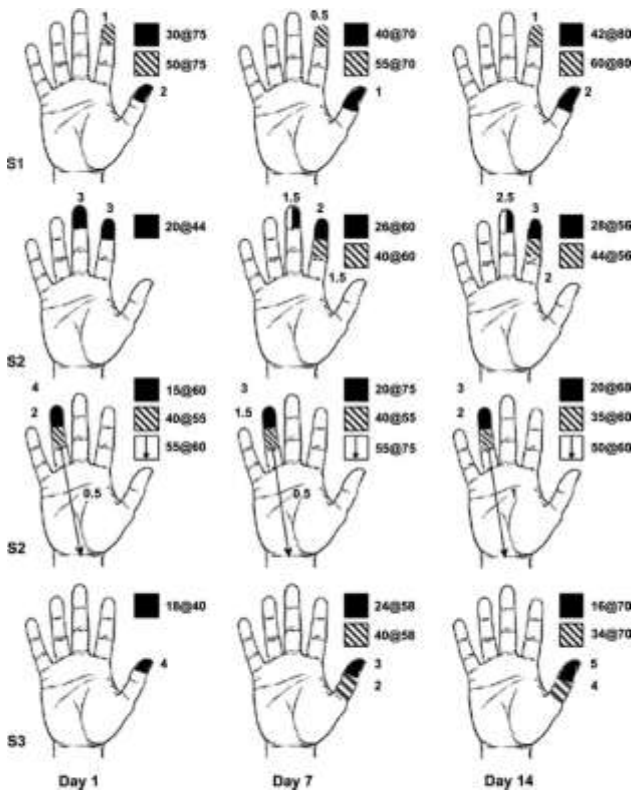

Figure 6. Localization of perceived sensation resulting from electrical stimulation of peripheral nerves in 3 amputee patients (S1-S3) during a 2-week period (days 1, 7, and 10). Stimuli were applied as 300-s pulse trains with intensities (A) and frequencies (pulses/s) listed in each legend. The perceived intensity of the sensation was rated by each subject and appears as a number next to each shaded region. Used with permission from Dhillon et al [44], Journal of Neurophysiology, 2005.

resultant decline in functional recovery [45,49].

Key difficulties of intraneural electrode systems involve the safety and stability of the interface. Because intraneural electrodes record single or multiunit action potentials from only a few of the thousands of individual fibers, it is difficult to obtain desired or repeatable results. In addition, because of nerve penetration, micromotion, and fibrosus, intraneural electrodes have the potential to cause nerve damage [36].

RECORDING DIRECTLY FROM THE BRAIN

A third potential source of control information for prosthetic limbs is the brain itself. The focus of the majority of current research in this area is to provide control and communication for patients who experience high spinal cord paralysis, locked-in syndrome, or severe communication disorders. These approaches may someday hold promise for amputees as technologies advance and the risks of device implantation decrease [50].

The electrical potentials recorded from the brain can be divided into 2 general categories: action potentials and field potentials [50]. Action potentials are the direct electrical signals produced by individual neurons and are the primary communication signals underlying brain activity [51]. Field potentials are voltage changes caused by the summed activity of a few or many combined action potentials. Although neuron action potentials (or spikes) must be measured within the brain tissue, field potentials can be measured inside the cortex (local field potential), on the cortical surface (electrocortigram), and over the scalp (electroencephalogram).

Cortical Spike Recording

Some of the most impressive advancements with brain–machine interfaces have involved spike recordings from the primary motor cortex. The primary motor cortex contains a topographical map of the body, and spiking patterns in the areas related to the hand and arm have been shown in primate studies to relate to movement direction [52], movement velocity [53], grip force [54], and even individual finger movement [55,56]. Investigators have used primate studies to demonstrate the success of these interfaces for control of 2and 3-dimensional cursors [57,58], and control of a robotic arm with a gripper for a self-feeding task [59]. Several types of multisite recording electrodes have been evaluated in animals, including microwires (small insulated wires affixed to the skull), multisite probes (with multiple recording sites along a flat probe), and platform arrays (an array of microelectrodes attached to a rigid base) [60]. Platform arrays and cone electrodes, which make single-unit recordings, are the only electrodes currently approved for use in human studies [60].

Beginning with a simple interface in which a single cone electrode recorded spiking patterns and allowed a subject to control an on–off switch, spike recording interfaces in humans have evolved to systems that have allowed severely disabled patients to control a computer cursor [61,62]. The BrainGate platform array, a 4 4-mm array of 100 electrodes, has even been used with 1 human subject to open and close a prosthetic hand and to control a robotic arm to grasp and move an object (Figure 7) [63]. During training sessions, the subject was asked to imagine using his hand to control a computer mouse to track a cursor on the screen. Single and multiunit recordings from the electrode array were then used

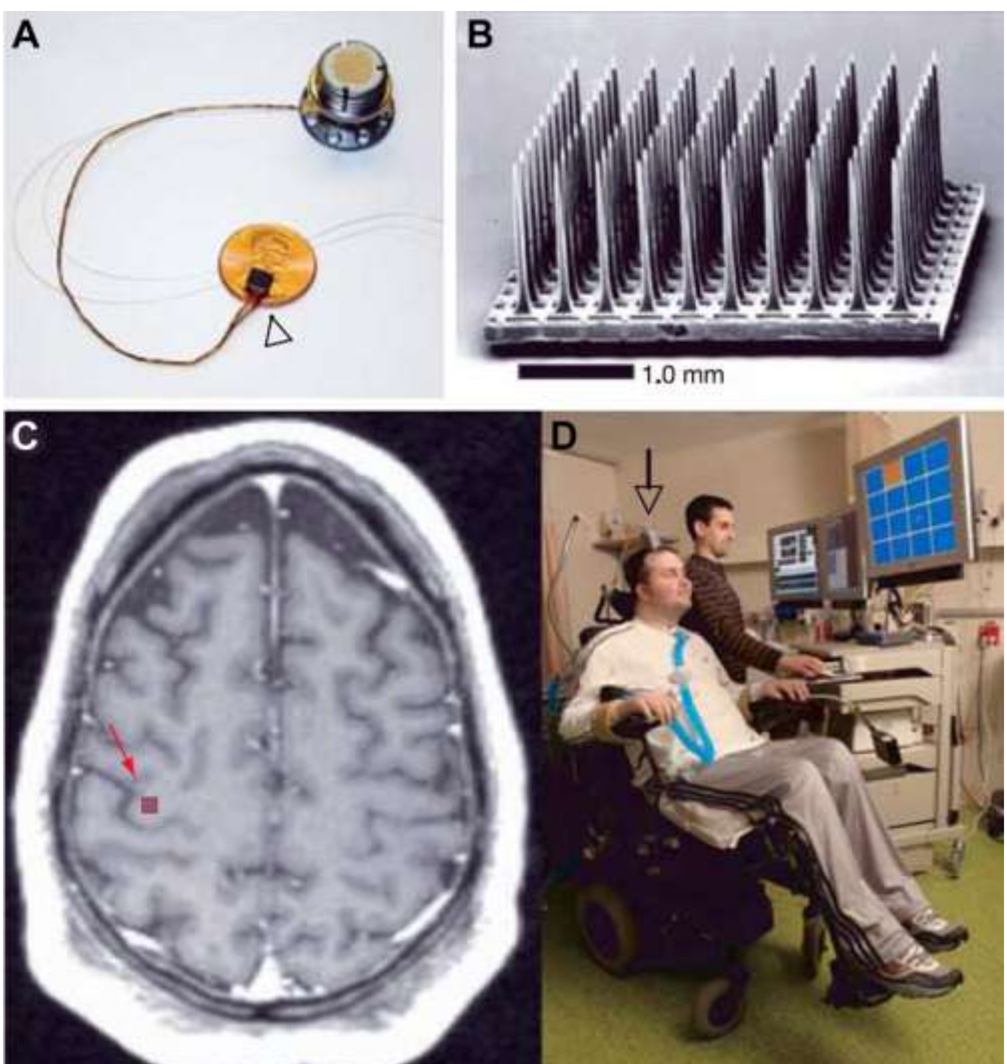

Figure 7. The BrainGate microelectrode array (A, B) was implanted into the primary motor cortex (C) of a patient with tetraplegia (D) and used to control 2-dimensional movement. Adapted by permission from Macmillan Publishers Ltd: Nature, [63], Copyright © 2005.

to create a linear filter that predicted 2-dimensional cursor movement on the basis of spiking patterns. These same outputs were then used for very basic control of a prosthetic hand and robotic arm.

The recording of action potentials requires the use of microelectrodes inserted into the cortex, and consistent performance requires a maintained connection with the recorded neuron. This action is made difficult by the responses of cortical tissue to chronic and acute microelectrode insertion, including neuronal loss and the formation of scar tissue around the electrode tip [64,65]. There are also difficulties maintaining and achieving stable recordings from individual neurons, and there is high variability in neuronal behavior [66]. Additional challenges include the physical failure of the implant, insulation leaks, or the risk of secondary infection [60]. In studies on primates, Donoghue [60] has routinely reported a decrease in the number and signal quality of units recorded over several months [60]. Both cone electrodes and microelectrode arrays have been shown to record reliable signals for a little over a year in monkeys; however, no microelectrode arrays have been verified to reliably record action potentials for more extended periods of time [67-69].

Local Field Potentials

Local field potentials for brain-machine interfaces have begun to receive more attention in recent years. This increased interest stems partially from the belief that they circumvent many of the problems inherent in spike recording stability. Local field potentials are recorded with the same types of penetrating electrodes used for spike detection. These signals are relatively robust because they are the result of the summed activity of several neurons firing in close proximity to the electrode. They are also considered easier to record and more feasible for long-term use than spike recordings [70]. The ability of local field potentials to encode information has been directly compared with that of spike recordings, with some researchers finding decreased performance and others finding comparable performance [71,72]. Local field potentials in monkeys have been shown to encode arm and hand position and velocity [73,74].

Intracortical Stimulation for Sensory Feedback

Efforts have also been made to determine whether stimuli applied directly to the cortex via intracortical electrodes can provide sensory feedback. Penfield and Rasmussen [75] have shown that stimulation of the somatosensory cortex in humans elicits sensations localized to various regions of the body. Similar to the motor cortex, the somatosensory cortex has been shown to contain a topographical map of the body, with larger cortical areas dedicated to body parts with greater sensory innervation (eg, hands and tongue) [76]. Two studies have demonstrated that electrical stimulation of the tactile portion (area 3b) of the primary somatosensory cortex in primates produced the perception of mid-frequency (5-50 Hz) vibration that was to some extent indistinguishable from that produced by mechanical stimuli and sufficient to activate the neural processes involved in vibration discrimination [77,78]. In a later study, London et al [79] showed that monkeys could distinguish different frequencies of electrical stimuli applied to the proprioceptive portion (area 3a) of the primary somatosensory cortex; however this type of stimulation has not yet been demonstrated to provide a perception of the location of a limb.

Electrocorticogram Recording

Electrocorticogram (ECoG) signals are measured on the cortical surface and therefore do not require the penetration of brain tissue (Figure 8). ECoG recordings are generally made over the sensorimotor cortex and comprise rhythmic signals of low (8-13 Hz), medium (13-30 Hz), and high (30 Hz) frequencies, referred to as , , and rhythms, respectively. The amplitudes of these signals decrease during movement or motor imagery, indicating a decrease in the synchronization of the neural signals. ECoG systems have been used to predict 2-dimensional hand movement trajectories in human subjects [80,81]. Recently, ECoG signals have been used for 2-dimensional cursor control and to determine the timecourse of individual finger flexion [66,80]. These studies have demonstrated the potential for ECoG systems to serve as long-term brain-machine interfaces for movement control. The performance of ECoG systems is decreased in comparison with spike recording systems, and ECoG signals have been shown to contain less than one-half of the information content of local field potentials [80]. Because recording electrodes do not penetrate the cortex, there is a smaller risk of brain tissue damage with these systems than with intracortical recording techniques. However, there is still risk involved because these systems require a craniotomy for electrode placement. Most studies in which investigators use ECoG signals are short-term studies on epilepsy patients [73,74] because no ECoG system is currently approved by the Food and Drug Administration for long-term use [60].

Electroencephalogram Recording

Electroencephalogram (EEG) systems are noninvasive and have therefore been a useful means of exploring the control capabilities of neural signals. EEG systems sometimes target slow cortical potentials, including the low-frequency changes in field potentials that appear before the onset of movement and are referred to as readiness potentials (or sometimes as Bereitshaftspotentials, or BP). These signals have been used for spelling devices for locked-in patients [82,83]. It takes months of training for patients to learn to modulate readiness potentials, and in initial reports of a spelling device, patients needed an average of 2 minutes to select one letter [83]. Large-scale, low-frequency changes in field potentials also are observed in response to neural events and can be triggered by external stimuli; these are termed event-related potentials [84,85]. One example, the P300 (which appears as a peak in cortical signals approximately 300 ms after relevant stimuli appear), also has found application in spelling devices and has been used to select desired wheelchair locations and task commands for a wheelchair-mounted robotic arm [8688]. Use of the P300 does not require training, but speeds are around only 1 word per minute [87].

Greater frequency signals, including and rhythms, often are measured from the sensorimotor cortex with EEG systems. These signals have been used for 1- and 2-dimensional cursor control [89-91], control of a hand orthosis (Figure 9) [92], and control of a neuroprosthesis [93]. Most of these systems require several months of training [88,92] and are not very robust: for example, 1 study reported an average accuracy of 76% to 81% for the final 3 sessions of 1-dimensional cursor control by 4 subjects with amyotrophic lateral sclerosis after 3 to 7 months of training [89].

Other drawbacks to EEG systems include noise in the recorded signal and the need to attach recording electrodes to the scalp—both a time issue and an appearance issue [60]. In addition, because EEG signals are filtered through the skull, they are limited in the frequencies of information they can record, which limits the maximum information transfer rate.

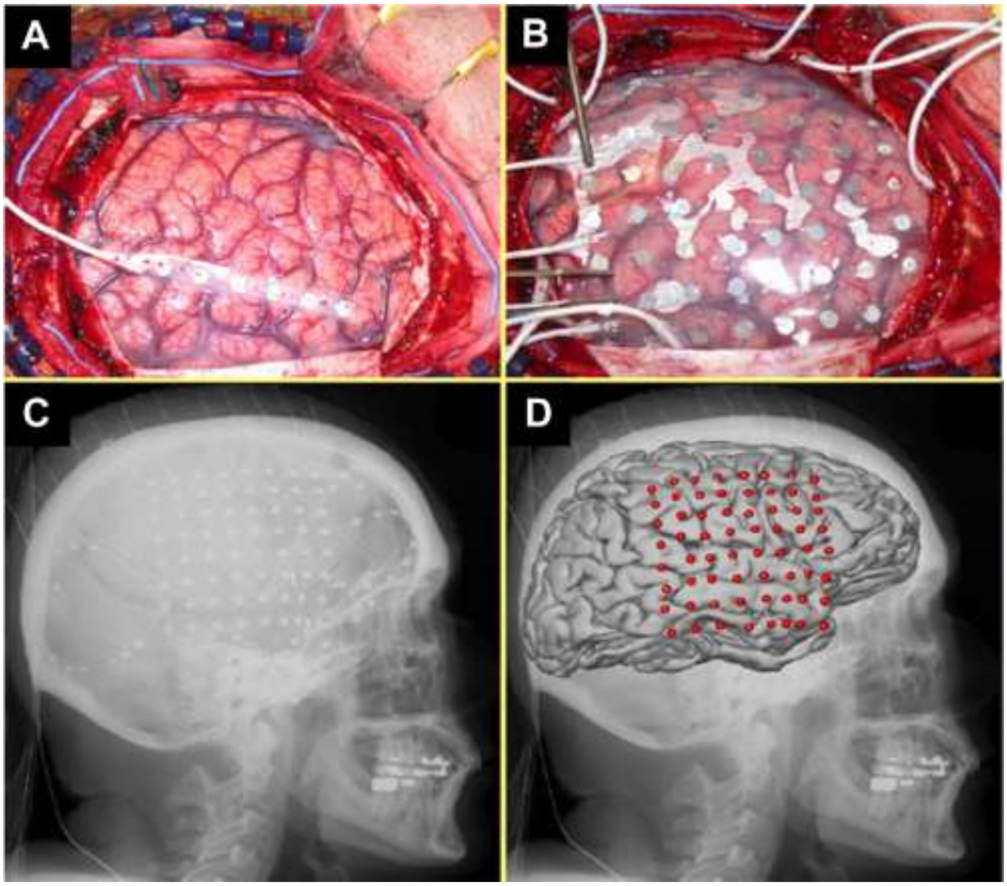

Figure 8. The brain is exposed (A) before placement of the ECoG electrode grid (B). A radiograph (C) shows the location of the grid, made clear by overlaying the electrode locations and an average brain image (D). Reproduced from Schalk et al [80] with permission from the Institute of Physics (Great Britain), IOP Publishing.

DISCUSSION

Each of the approaches outlined previously requires highly specialized technology, presents unique advantages and challenges, and may eventually find a unique application. However, use for prosthetic control requires a delicate balance of several important factors, including information transfer rates, the ability to provide sensory feedback, signal stability, system robustness, and relative risks.

Many control signals must be recorded, transmitted, and processed in a short amount of time to operate a multifunc-

Figure 9. An EEG system recording mid-frequency signals is used to control a hand orthosis. Reprinted from Pfurtscheller et al [92], Copyright © 2002, with permission from Elsevier.

tion prosthesis. Today's standard myoelectric control systems provide only enough information for control of 1 or 2 joints. Advanced muscle interface techniques, such as targeted reinnervation and pattern recognition, offer increased information content and more intuitive control. When combined, targeted reinnervation and pattern recognition have demonstrated the ability to provide control of a limited number of hand grasp patterns, but are still unable to provide truly dexterous control. Direct nerve interfaces have demonstrated the capability in humans to control simple joint movement (elbow flexion/extension) or control of grip force, but require further study to understand their full capabilities. As for cortical interfaces, both spike-recording and ECoG systems have demonstrated the potential for more complex hand control by their ability to extract information related to individual finger movement, but have yet to demonstrate this control in real-time systems. EEG interfaces currently appear to provide inadequate information and transfer rates for multifunctional prosthesis control. Thus, there is much room for improvement in all of the neural interface systems developed to date.

Even if sufficient motor command information were available for multifunction prosthesis control, adequate control would still require some degree of sensory feedback. Both tactile and proprioceptive feedback play an important role in volitional movement. Tactile feedback allows for modulation of grip forces and hand postures for grasping and object manipulation [94,95]. Proprioceptive feedback is essential for accurate motor control and joint coordination [96,97]. Muscle interfaces are currently unable to provide any form of sensory feedback for prosthesis control, although several attempts have been made [98,99]. Current myoelectric systems rely on visual feedback, which causes difficulty and large cognitive burdens for control of relatively simple devices (3 joint movements at most). In this respect, bodypowered prostheses could be considered superior to myoelectric prostheses because position and force information are transferred to the user through cable forces [4]. Targeted reinnervation provides the potential for tactile feedback that feels natural to the user because it is attributed to the lost limb, but it cannot provide proprioceptive feedback [29,30,100]. Nerve interfaces that use LIFEs have shown rudimentary ability to provide tactile and proprioceptive feedback by eliciting perceptions of joint angle and grip force. Similarly, direct stimulation of cortical somatosensory neurons appears to have the potential to provide both tactile and proprioceptive feedback. Further work remains to refine these systems by optimizing stimulus profiles and to demonstrate closed-loop performance with human subjects.

Practical issues of safety and stability are the final hurdles to the implementation of advanced neural control techniques. Surface technologies (EMG and EEG) are noninvasive and therefore provide little risk. However, the biggest promise for dexterous control lies with the more selective and stable, and therefore more invasive, technologies. Work to date on intramuscular EMG recordings, direct nerve recordings, and invasive cortical recordings has relied on percutaneous tethered systems. The continuous risk for infection and tissue damage have caused many to conclude that fully implantable, telemetered systems are necessary before these technologies can become viable clinical options. The task of remotely powering an implant and wirelessly transmitting data is further complicated by the high information content and data transfer rates required for multifunction prosthesis control. As of yet, no fully tested, viable solution to this problem has been presented.

CONCLUSION

Although upper limb prosthetic technology has advanced slowly during the last century, there has been a relatively recent surge in technological development and scientific understanding that has paved the way for the realization of neural interfaces for artificial arm control. How much longer will individuals with limb loss have to wait before these technologies become safe and commercially available? Because of the demonstrated efficacy and limited risks of the 1-time surgery, targeted reinnervation is now a clinical reality for amputees. Pattern recognition may also become a clinical option in the near future. Direct nerve and brain-machine interfaces require much additional investigation and development before attempts can be made to build marketable systems for prosthesis control. Although these challenges may appear daunting, the rate at which these technologies continue to advance should provide hope for those waiting for the realization of dexterous prosthetic arm control.

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