

REVIEW | *Biology of Neuroengineering Interfaces*

Neurophysiology and neural engineering: a review

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Prochazka A. Neurophysiology and neural engineering: a review. *J Neurophysiol* 118: 1292–1309, 2017. First published May 31, 2017; doi:10.1152/jn.00149.2017.—Neurophysiology is the branch of physiology concerned with understanding the function of neural systems. Neural engineering (also known as neuroengineering) is a discipline within biomedical engineering that uses engineering techniques to understand, repair, replace, enhance, or otherwise exploit the properties and functions of neural systems. In most cases neural engineering involves the development of an interface between electronic devices and living neural tissue. This review describes the origins of neural engineering, the explosive development of methods and devices commencing in the late 1950s, and the present-day devices that have resulted. The barriers to interfacing electronic devices with living neural tissues are many and varied, and consequently there have been numerous stops and starts along the way. Representative examples are discussed. None of this could have happened without a basic understanding of the relevant neurophysiology. I also consider examples of how neural engineering is repaying the debt to basic neurophysiology with new knowledge and insight.

neural engineering; neuroengineering; neural stimulators; neuroprostheses; cochlear implants; brain-computer interface; functional electrical stimulation

IT COULD BE ARGUED that neural engineering commenced in the 1750s, when Jan Swammerdam enclosed a muscle in a glass tube containing water and activated it by tugging on its nerve, in order to determine whether its volume changed. In the same decade, Leopoldo Calvani electrically stimulated nerves with the use of an electrostatic machine (Brazier 1984). By the end of the eighteenth century, Giovanni Aldini and others had begun stimulating the nervous system in humans with the use of surface electrodes (Brazier 1988). First, Aldini performed what would now be called proof-of-principle experiments on the freshly severed heads of people who had just been guillotined, placing electrodes on the face, on the brain stem, or in the mouth. In the mid-nineteenth century Duchenne de Boulogne used electrical stimulation to study human facial expressions (Duchenne de Boulogne 1862). Duchenne's photographs were used by Charles Darwin in his arguments regarding the evolutionary nature of the expression of emotions (Darwin 1872).

Electrical stimulation was soon adopted by clinicians to elicit muscle twitches, presumably to impress their patients rather than to treat their disorders (Licht 1971; McNeal 1977). By the mid-nineteenth century, induction coils enabled trains of electrical current pulses to be delivered, which elicited smoother muscle contractions. In the 1850s, the German inventor Isaac Pulvermacher marketed a belt comprised of a

chain of batteries that delivered mechanically switched pulses of electrical current to the wearer's abdomen. Pulvermacher's chain was initially supported by the medical community (Bird 1851). Electrical belts and other garments became enormously popular, selling in their hundreds of thousands by the 1880s. However, vendors began making outlandish medical claims, with the result that by 1920 the field had become thoroughly discredited, as evidenced by articles in leading journals such as the *Lancet* and the *British Medical Journal* (British Medical Journal 1893).

By the mid-twentieth century, electrical stimulation of the nervous system had seen a resurgence and it had become an accepted clinical modality for pain mitigation, muscle strengthening, and rehabilitation. In 1956, with the development of the transistor, it became feasible to manufacture stimulators that could be implanted inside the body. This led to a remarkable burst of innovation and experimentation in the subsequent two decades. Devices that have their origin in this period include the cochlear implant (Chouard 2015; Djourno et al. 1957; Djourno and Eyries 1957), the cardiac pacemaker (Greatbatch 1962), the spinal cord epidural stimulator (Shealy et al. 1967), sacral nerve and root stimulators (Brindley and Lewin 1968), phrenic nerve and diaphragm pacers (Anagnostopoulos and Glenn 1966; Johnson and Eiseman 1971; Van Heeckeren and Glenn 1966), the foot-drop stimulator (Jeglic et al. 1970; Liberson et al. 1961; Waters et al. 1975), and the intraspinal stimulator (Friedman et al. 1972; Nashold et al. 1971b).

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This article focuses mainly on neural engineering devices designed to treat nervous system dysfunction. To illustrate the relationship between such devices and the neurophysiology that underpins them, let us consider two examples: neurostimulators (NSs) and brain-computer interfaces (BCIs).

NSs, also known as neuroprostheses, are devices that activate neural tissue to help restore lost function. Figure 1 gives an idea of how rapidly this field is expected to grow over the next few years.

The best-known and most successful implantable NS is the cochlear implant, arguably the most advanced neural engineering device in existence. As of 2012, cochlear implants had been implanted in over 320,000 people with severe sensorineural hearing loss worldwide (<https://www.nidcd.nih.gov/health/statistics/quick-statistics-hearing>). Sound vibrations are picked up by a microphone, typically located in an earpiece. The vibrations are transduced into voltage signals that are amplified, filtered, digitized, and resolved into components by a microprocessor. The microprocessor encodes this information in a radio-frequency (RF) signal and wirelessly transmits it to a pulse generator implanted under the scalp, which in turn delivers interleaved or simultaneous trains of stimuli to the sensory endings of the auditory nerve via a delicate multielectrode array inserted into the scala tympani of the cochlea. The evoked neural activity results in the hearing and discrimination of complex sounds such as speech and music. Parameters in the computer algorithms that convert the incoming signals into patterns of stimulation are fine-tuned through trial-and-error testing, whereby the recipients report the sounds they hear. The development of the cochlear implant relied on a detailed understanding of the neurophysiology of the auditory system, the physics of sound signals, the components of speech, electronics, computer processing, RF transmission, and electrode-tissue interactions. From an engineering point of view, a large variety of technical problems had to be overcome, not least those involving the multielectrode array (Chouard 2015; Clark 1978; House 1976). Unexpected societal barriers that soon arose are discussed below.

In a BCI, electrical activity is recorded from specific parts of the brain, via penetrating microelectrode arrays, electrode matrices resting on the brain surface, or electrodes attached to the scalp. The recorded signals are amplified, digitized, and resolved into components by a computer processor. For example, in a person with paralyzed arms neural activity signaling the

intent to grasp and move external objects is recorded from motor areas of the cerebral cortex. The signals are decomposed analytically, and the resultant intention-related signals may be used to perform functional tasks with a robotic arm or to emulate a computer mouse, allowing the paralyzed person to use a computer. In the most recent research, the signals recorded from the brain have been used to generate trains of electrical stimuli delivered to muscle nerves via surface electrodes, eliciting movements of the user's own arm. In this case the system is a combination of a BCI and a NS and has been called a "neural bypass" (Bouton et al. 2016; Sharma et al. 2016).

The above systems have two basic things in common: they sense one or more inputs and use this information to control one or more outputs. The same is true of most neurophysiological control systems. The inputs used in neural engineering devices either can come from artificial sensors such as accelerometers, strain gauges, and photoelectric devices or can come from the body itself, in the form of electrical activity of muscle and neural tissues. The outputs can range from computer displays and robotic arms to the recipient's own limbs.

After the cochlear implant, the second most widely deployed implantable NS is the spinal cord stimulator. This is a fully implanted device that delivers trains of electrical pulses through an electrode array that is implanted on the surface of the dura mater, the membrane covering the spinal cord, or on the surface of the spinal cord itself (Iwahara et al. 1991, 1992). Epidural stimulation is used to reduce chronic pain, improve bladder control, and reduce spastic hyperreflexia. In a typical epidural stimulator a therapist sets the stimulation parameters with a wireless external programmer, and after that the device delivers steady trains of pulses for long periods of time. There is no continuous feedback control, and therefore there are no sensors other than wireless or magnetic on-off switches. The same is true of noninvasive transcutaneous electrical stimulators (TENS units) for pain mitigation, transcutaneous neuromuscular electrical stimulators (NMES) for exercising and strengthening muscles, implanted sacral nerve stimulators for bladder control, implanted phrenic nerve stimulators for respiration, implanted vagus nerve stimulators for epilepsy and depression, and deep brain stimulators for various motor and psychiatric disorders, intractable pain, epilepsy, and Alzheimer's disease. The mechanism of action in some of these cases is assumed to be neuromodulation, an indirect effect mediated by intracellular second messengers. This is in contrast to the generally more rapid action of direct electrical excitation of neurons or classical chemical neurotransmission via ligand-gated postsynaptic receptors.

Sensing the Inputs

Artificial sensors. The number of artificial sensors that can be deployed in a practical NS or BCI is insignificant when compared to the vast number of biological sensory receptors in the body. For example, foot-drop stimulators (NSs that activate the pretibial muscles to lift the foot at the onset of the swing phase of the locomotor step cycle) have at most two sensors, a force transducer or switch signaling under-heel pressure and an accelerometer signaling leg tilt. Another example is provided by grasp-release stimulators, in which signals from a shoulder or wrist sensor are used to control stimulation of muscles that

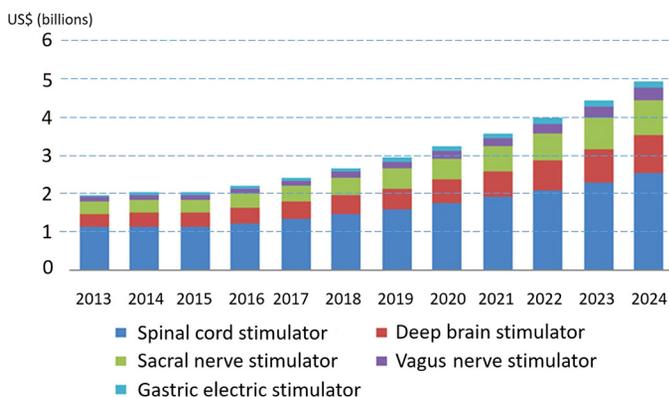


Fig. 1. Global neurostimulation devices market in US, by device, 2013–2024. Reproduced from Grand View Research (2016) with permission.

open or close the hand (Peckham et al. 1980; Prochazka et al. 1996). This is in contrast to the body's thousands of muscle, joint, connective tissue, and cutaneous receptors that provide highly detailed, multivariate information to the central nervous system (CNS) on the biomechanical state not only of the moving limb but also of the entire body (Prochazka 2015a). For example, in relation to proprioception (the sensing of the body's own movements), there are 25,000–30,000 muscle spindles in the human body, including ~4,000 in each arm and 7,000 in each leg (Hulliger 1984; Voss 1971), nearly as many Golgi tendon organs, and many more deep pressure receptors and cutaneous receptors. In fact, it is interesting that NS devices such as foot-drop stimulators and grasp-release stimulators with only one or two sensors can significantly improve motor function after the nervous system has been damaged. This is because in some cases the sensing of simple events such as unloading of the leg during the locomotor step cycle (Liberzon et al. 1961) or extension of the wrist during manual grasp-release actions (Prochazka et al. 1996) can suffice to trigger transitions from one phase of a movement to the next. It also shows that a single artificial sensor, when combined with an electronic analyzer, can provide information that is equivalent to the input from many biological sensory receptors. Again, the cochlear implant provides a good example. Sound vibrations impinging on a microphone are resolved electronically into many frequency components, just as the sound vibrations transmitted from the tympanic membrane (eardrum) to the cochlea are resolved into many frequency components by the 3,000 or so inner hair cells that line the basilar membrane of the cochlea. In this example a single artificial sensor and an electronic analyzer can replicate the function of thousands of biological sensors, albeit with a lower-frequency resolution and amplitude range.

What are some other limitations of artificial sensors compared with biological sensors? Artificial force and displacement sensors can match individual cutaneous and intramuscular receptors in their sensitivity and amplitude range, but it has proven difficult to attach them to body parts in ways that are convenient, cosmetically acceptable, and reliable over the long term. Under-heel switches and force sensors are generally held within shoes, which rules out their use when walking barefoot. In some functional electrical stimulation devices sensors are held in the same enclosures as the stimulators (e.g., wrist motion sensors, accelerometers, and leg tilt sensors; Prochazka 1997; Prochazka et al. 1997a; Stein et al. 2006). Tooth click sensors (Prochazka 2003) and head motion sensors (Prochazka 2016) built into earpieces are used to trigger hand grasp and release (e.g., Rehabtronics ReGrasp). Accelerometers and EMG amplifiers have been used experimentally in closed-loop tremor suppression systems (Gillard et al. 1999; Khobragade et al. 2015; Prochazka et al. 1992).

It took many years for the hardware of hearing aids and cochlear implants to be miniaturized to the point that they are unobtrusive and function for many years. In the cochlear implant, microphones and sound processors are generally external to the body. It was only recently that a fully implanted device that included a microphone and sound processor entered clinical trials (Briggs et al. 2008). Fully implantable hearing aids that monitor incoming sounds and apply corresponding mechanical vibrations to the ossicles have recently become available commercially (Barbara et al. 2011; Bruschini et al.

2016; Martin et al. 2009). In a few NSs, mechanical sensors have been implanted under the skin, for example, the microphones just discussed and wrist joint displacement sensors used with the Freehand upper limb device (Bhadra et al. 2002).

Visual systems in animals operate over an enormous range of light intensities, though some artificial optical sensors surpass them in terms of the range of detectable wavelengths (infrared to X-ray). Visual NSs have been developed recently that operate either by stimulating the visual cortex through electrode arrays implanted on the cortical surface or via electrode arrays attached to the back of the retina (see Table 2 for references). These devices receive inputs from miniature video cameras attached to goggles or glasses worn by the user. As in cochlear implants, the signals are processed electronically to derive trains of electrical pulses for delivery through multielectrode arrays. Learning algorithms are used to optimize this process through postimplant trial-and-error testing with the recipients.

Artificial sensors tend to be energy intensive, so regular battery charging or replacement is required. While this is not a major problem for sensors external to the body, it becomes a significant barrier for fully implanted systems. The approach currently taken is to implant rechargeable batteries that are charged through the skin by electromagnetic coupling. The external charger must regularly be held on, or close to, the body for significant periods of time, which can be inconvenient.

Sensing the body's own neural signals. Inputs to NSs and BCIs may be obtained from the body's own neural input and output signals, for example, by monitoring the activity of neurons in the cerebral cortex (Schwartz 2016). Signals entering the spinal cord from muscle, joint, and skin receptors can be intercepted at the dorsal root ganglia with the use of implanted semimicroelectrode arrays (Loeb et al. 1977; Prochazka et al. 1976; Weber et al. 2006). This has been proposed as a means of providing intraspinal microstimulation devices with the kinematic feedback needed to activate paralyzed limbs (Holinski et al. 2013; Weber et al. 2007) and also as a means of restoring sensation by using the monitored signals to control stimulation of somatosensory regions of the brain (a "sensory neural bypass") (Flesher et al. 2016). However, the technical barriers to long-term recording from peripheral nerves and dorsal root ganglia are formidable.

It is easier to record neural outputs indirectly, for example, by recording the electromyogram (EMG) of muscles that are still under voluntary control or even more simply by recording the movements generated by these muscles (Peckham et al. 1980; Prochazka et al. 1996). EMG control was first exploited in the AutoMove, a stimulator commercialized in the 1970s (van Overeem Hansen 1979). In this device, the EMG of a muscle under weak voluntary control is monitored through surface electrodes. When a preset threshold is reached, the device switches to stimulation mode, activating the same muscle through the same electrodes. This device, now known as the NeuroMove, is used to this day in some rehabilitation centers to strengthen paretic muscles after stroke and spinal cord injury (Francisco et al. 1998; Knutson et al. 2015).

Another approach is to use the EMG from a muscle under volitional control to continuously control functional electrical stimulation of a paretic or paralyzed muscle (Graupe et al. 1983, 1989; Hincapie and Kirsch 2007; Mercier et al. 2017; Williams and Kirsch 2015). In another variant, a pair of

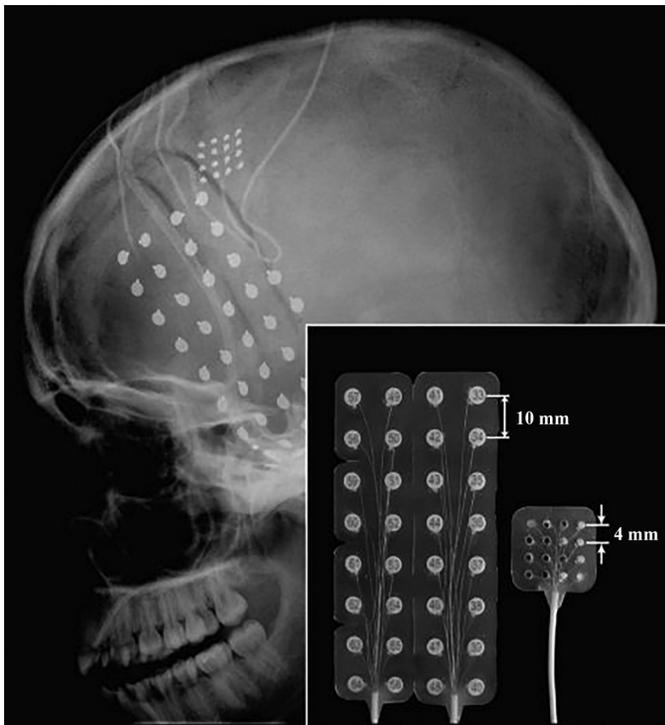


Fig. 2. Head X-ray (lateral view) showing the locations and sizes of implanted ECoG electrodes: one micro-ECoG grid (16 contacts), one regular ECoG grid (32 contacts), and two 6-contact regular ECoG strips. Inset: side-by-side comparison of a regular ECoG grid and a micro-ECoG grid showing the center-to-center electrode spacing (reproduced from Wang et al. 2009 with permission).

stimulating electrodes is attached to the skin over a paretic muscle. A pair of recording electrodes is attached at right angles about halfway between them (Hodgson 1986; Thorsen et al. 2001). This configuration minimizes artifacts in the EMG signal when the muscle is stimulated. A blocking amplifier eliminates any residual artifacts, making it possible to record EMG from the muscle between stimulating pulses, thus boosting voluntary contractions. When recording and stimulating

electrodes are implanted, the stimulus currents and therefore the stimulus artifacts are smaller (Hincapie and Kirsch 2007; Liu et al. 2014).

Brain activity can be recorded with surface electroencephalogram (EEG) electrodes (McFarland and Wolpaw 2003; Wolpaw et al. 1991), implanted electrocorticogram (ECoG) electrodes (Fig. 2) (Branco et al. 2017; Flint et al. 2017; Hotson et al. 2016; Leuthardt et al. 2006; Vansteensel et al. 2016; Vinjamuri et al. 2009; Wang et al. 2009), or implanted penetrating microelectrode arrays (Bouton et al. 2016; Sharma et al. 2016). Surface EEG signals tend to be small, variable, susceptible to artifacts (especially eyeblinks), and poorly localized because the sources (ensembles of cortical neurons) and the recording sites are separated not only by a layer of cerebrospinal fluid and the meninges but also by the skull and the scalp. Nonetheless, EEG-based BCIs were highly represented at the 2016 Society for Neuroscience annual meeting. Furthermore, in a recent study feedback-controlled functional electrical stimulation driven by EEG signals enabled able-bodied subjects to perform a tracking task with their forearms (Vidaurre et al. 2016). In theory, ECoG electrode arrays implanted subdurally on the cortical surface, or epidurally on the dural surface, should provide more localized multichannel signals than surface EEGs. Research is underway to compare these modalities (Flint et al. 2017).

Multiple single-cell recordings with penetrating intracortical silicon microelectrode arrays provide the most localized and information-rich signals, but this approach is also the most invasive. With current devices the recordings degrade within a year or so (Wodlinger et al. 2015), though it is possible to extract useful information even after this has occurred (Perge et al. 2013, 2014; Simeral et al. 2011). In 2013 an important failure analysis of silicon microelectrode arrays was published (Barrese et al. 2013). Of 78 arrays implanted in 27 monkeys, the recording duration ranged from 0 to 5.75 yr, with a mean of 387 days and a median of 182 days. Fifty-six percent of failures occurred within a year, mechanical failures involving leads and connectors being the most common (48%) (Fig. 3).

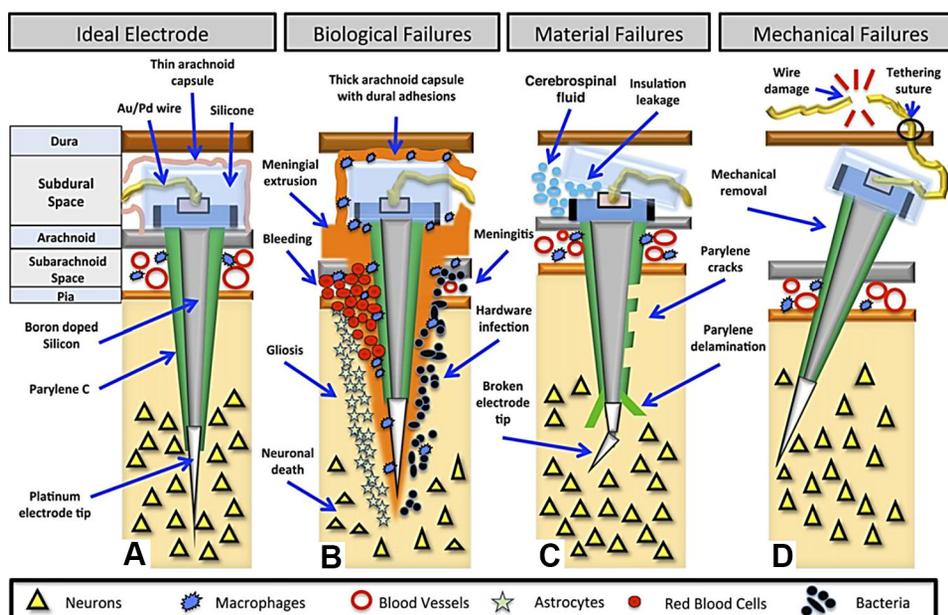


Fig. 3. Major failure modes of microelectrode arrays. A: ideal placement, 1–1.5 mm into cortex. The top of the electrode becomes encased in a thin layer of arachnoid, which helps to stabilize the array. B: biological failures: bleeding, cell death, hardware infection, meningitis, gliosis, and meningeal encapsulation and extrusion. Macrophages originating in the subarachnoid space may mediate the encapsulation response. C: material failures: broken electrode tips, insulation leakage, and parylene cracks and delamination. D: mechanical failures: wire breakage and connector damage (reproduced from Barrese et al. 2013 with permission).

Meningeal reactions that separated the array from the parenchyma were the most common cause of biological failure (14.5%). Recordings showed a slow, progressive decline in spike amplitude, noise amplitude, and number of viable channels, which predicted complete signal loss by ~8 yr. The authors concluded that these arrays could potentially record for many years with the use of implantable wireless systems, better control of meningeal reactions, and improved insulation materials.

The slow deterioration of signal quality has been attributed to tissue encapsulation, inflammatory responses, rupture of the blood-brain barrier, and the death of neurons resulting from the insertion, micromotion, or physiological motion of microelectrodes (Kozai et al. 2010, 2012; McCreery et al. 2016). Smaller microelectrodes with coatings of dielectric and bioactive materials have been shown to result in significant improvements in this regard (Fattahi et al. 2014; Kozai et al. 2016). Syringe-injectable microelectronics are also on the horizon (Liu et al. 2015). Given the breadth of research underway in this field, it seems likely that the quality and reliability of signals recorded from the CNS will see major improvements over the next few years.

Activating the Outputs

Nerves innervating limb muscles are relatively accessible, allowing pulses of electrical current that activate them to be applied through the skin via self-adhesive, conductive gel electrodes, conductive rubber electrodes coated with gel, or metal plate electrodes with an intervening layer of spongy material that is moistened with water. The gel and water conform to skin contours and distribute the flow of current evenly, thus avoiding hot spots of high current density that would cause skin discomfort, inflammation, and burns. Transcutaneous stimulation is commonly used in surface stimulators such as TENS and NMES units and functional electrical stimulators. When the target stimulation sites are deep or close to the CNS (e.g., cochlear nerves, sacral nerves, and the vagus nerve), or when external components cause inconvenience, the stimulator and leads are implanted. The most common implantable stimulators, cardiac pacemakers, only deliver a single pulse every second or so and can therefore be powered by an implanted nonrechargeable battery that is surgically replaced every few years. In contrast, the implantable NSs in Tables 1 and 2 have to deliver stimuli at rates as high as 18,000 pulses/s. As mentioned above, such devices contain batteries or capacitors that are recharged either intermittently or continuously by electromagnetic coupling from an outside coil. The body-worn sound processors and transmitters in cochlear stimulators have high energy needs and incorporate batteries that require recharging every few hours. In an alternative approach where only a single channel is needed, an external stimulator delivers current pulses through the skin between two surface electrodes and some of the current flows through a fully implanted lead to a target nerve (the StimRouter) (Deer et al. 2010, 2016; Gan et al. 2011; Prochazka 2005). In this system, as the stimulator is external to the body, it can be recharged and serviced easily and pulse parameters and modes of control can be modified and updated as the technology improves.

Electrodes in implanted NSs generally consist of insulated leads with conductive terminals implanted on or adjacent to

neural tissues. The terminals are made of a biologically compatible and corrosion-resistant material such as stainless steel, platinum, or silicon. They may be individually inserted, they may be attached to a shaft to form a linear array [e.g., deep brain stimulation (DBS) leads], or they may be mounted on a substrate to form a matrix (e.g., spinal epidural leads, ECoGs). In the case of penetrating microelectrode arrays, the impedance of the conductive tips of the electrodes is reduced with a variety of conductive and dielectric coatings (Campbell et al. 1991; Maynard et al. 1997).

From a theoretical point of view, the excitation of whole nerves or individual nerve axons and cell bodies with electrical stimulation is well understood (Stein and Prochazka 2009). In practice, the main problem is to activate just those whole nerves or groups of axons within nerves that elicit the desired action and not others. For example, when muscle nerves are stimulated via surface electrodes, afferents in cutaneous nerves are nearly always activated too, and this can cause discomfort. If the targeted motor nerves are deep lying, other motor nerves may be activated, producing unwanted components of movement. Implanted electrodes such as epimysial, intramuscular, and nerve cuff electrodes with terminals adjacent to, or attached to, the targeted nerves are much more selective, but they are subject to mechanical failure in the long term. When targeted and untargeted nerve terminals are in close proximity, as in the cochlea, selectivity is a major problem. This is currently the subject of intensive research and development (Roche and Hansen 2015).

Selectivity is also a crucial issue in implants targeting structures within the CNS, where ensembles of neurons with opposite functions may be located less than a millimeter apart. A good example is provided by attempts to develop intraspinal microstimulation systems to restore limb movements after spinal cord injury. Although it has occasionally been possible to activate single limb muscles or groups of synergists by stimulating through individual electrodes implanted in or near spinal motoneuron pools (Mushahwar et al. 2000, 2002), the more common result, particularly in the weeks and months after implantation, is a coactivation of multiple muscles (Moritz et al. 2007; Tai et al. 2003). As a result, intraspinal microstimulation may elicit whole limb synergies or stiffen the limb rather than moving it in a controlled manner (personal observations).

Voluntary control of paretic muscles can be boosted with nonspecific epidural or transcutaneous stimulation of the spinal cord (Angeli et al. 2014; Carhart et al. 2004; Dimitrijevic et al. 1998; Gerasimenko et al. 2002, 2015; Harkema et al. 2011; Herman et al. 2002; Lu et al. 2016; Sayenko et al. 2015), presumably by activating sensory axons in dorsal roots and the dorsal columns (Gaunt et al. 2006; Musienko et al. 2012; Rattay et al. 2000, 2003). Voluntary contractions of different muscles may also be boosted by intraspinal microstimulation at levels insufficient to activate any of the muscles in the absence of volitional drive (Prochazka et al. 2002b) (see Fig. 4). Given that intraspinal microstimulation recruits branches of sensory afferents before alpha motoneurons (Gaunt et al. 2006), the mechanism of boosting in this case may be the same as that for transcutaneous stimulation of the spinal cord.

Another example of the importance of selective activation of neuronal populations is provided by systems designed to treat bladder-sphincter dyssynergia resulting from spinal cord in-

Table 1. Neurostimulators in approximate descending order of numbers deployed worldwide

Device	Sensor	Sensed Variable	Signal Processing	Pulse Generator, Electrodes, Target
TENS and NMES stimulators, external	N/A	N/A	N/A	External PG and electrodes
Cochlear implant*	Microphone	Air pressure variations	Digitizing, filtering, compressing, encoding	Implanted RF-controlled PG, multichannel linear electrode array, auditory nerve endings
Spinal cord epidural stimulator implant*	N/A	N/A	Parameter adjustment via external wireless programmer	Implanted RF-controlled PG, multichannel epidural array, dorsal roots, columns
Sacral nerve stimulator implant*	N/A	N/A	Parameter adjustment via external programmer	Implanted RF-controlled PG, sacral spinal nerve root
Foot-drop stimulator (external)*	Under-heel switch or force sensor or tilt sensor	Under-heel pressure or leg tilt	On-off detection or digitizing, filtering, threshold detection	External PG and electrodes in garter, common peroneal nerve
Vagus nerve stimulator implant*	N/A	N/A	Parameter adjustment via external programmer	Implanted or external PG, vagus nerve
Deep brain stimulator implant*	N/A (EMG and accelerometers: experimental)	N/A (tremor: experimental)	Parameter adjustment via external programmer	External PG, midbrain nuclei
Phrenic nerve and diaphragm pacers (implanted)*	N/A	N/A	Parameter adjustment via external programmer	Implanted or external PG, phrenic nerve cuff or intramuscular electrodes in diaphragm
Sacral anterior root stimulator implant*	N/A	N/A	Parameter adjustment via external programmer	Implanted RF-controlled PG, sacral anterior nerve root
Grasp-release stimulators (external)*	Push-button or earpiece	Taps, tooth clicks, head nods	Digitizing, filtering, threshold detection	External PG and electrodes in brace or garment, nerves innervating hand muscles
Grasp-release stimulators (implant)†	Goniometers or implanted Hall effect sensors	Shoulder or wrist movements	Digitizing, filtering, proportional control of outputs	Implanted RF-controlled PG, epimysial electrodes, nerves innervating hand muscles
StimRouter implant*	N/A or earpiece	N/A or tooth clicks	Parameter adjustment via external programmer or digitizing, filtering, threshold detection	Implanted lead, external stimulator, skin electrodes, nerves suppressing pain, overactive bladder, or restoring hand function
Foot-drop stimulator (implant)§	Under-heel force sensor or nerve cuff	Under-heel force, afferent activity	Digitizing, filtering, threshold detection	Implanted RF-controlled PG, nerve cuff electrodes, common peroneal nerve
Microstimulator implant§	N/A	N/A	Parameter adjustment via external programmer	Implanted BIONs, various nerves
Visual cortex implant§	Video	Visual field	Image analysis, stimulus encoding	Implanted RF-controlled PG, multichannel electrode array, visual cortex
Retinal implant§	Video, eye position sensor	Visual field, eye position	Image analysis, stimulus encoding, eye position compensation	Implanted RF-controlled PG, multichannel electrode array, retinal ganglion cells
Somatosensory cortex implant§	Force or strain sensors	Pressure or movement	Digitizing, filtering, threshold detection	External PG, intracortical electrode arrays or ECoGs, cortical cells
Spinal cord epidural stimulator implant (spatiotemporal)§	Video motion capture, cortical electrode arrays	Limb position or cortical commands	Image analysis, digitizing, filtering, compressing, encoding	Implanted RF-controlled PG, multichannel epidural array, dorsal roots, dorsal columns
Intraspinal microstimulation (limb movement)§	External goniometers	Limb position	Digitizing, filtering, threshold detection	Implanted penetrating semimicroelectrode arrays
Intraspinal microstimulation (bladder control)§	Bladder and sphincter pressures	Urethral pressure sensor	Parameter adjustment to effect	Implanted penetrating semimicroelectrode arrays
Intraspinal microstimulation (respiration)§	EMG	Genioglossus EMG	Parameter adjustment to effect	Implanted penetrating semimicroelectrode arrays

PG, pulse generator; N/A, not applicable. *Commercially available, †discontinued commercially, §experimental.

jury. Here the descending drive from the pontine micturition center is absent, and a form of spasticity develops in which instead of relaxing when the bladder contracts to expel urine the external urethral sphincter (EUS) also contracts, thus preventing voiding. Experiments in animals with spinal cord transections showed that sacral ventral root stimulation could elicit voiding in such cases provided that reflex contractions of the EUS were abolished by transecting the dorsal roots (Brindley 1977; Heine et al. 1977). The problem of selectivity in this case is that ventral root stimulation activates not only the

parasympathetic preganglionic axons that elicit bladder contractions but also the larger axons that innervate the EUS. Fortunately, upon cessation of stimulation, the EUS relaxes more quickly than the bladder, allowing a few seconds of poststimulus voiding. In clinical practice, brief trains of stimuli are repeated until sufficient voiding has occurred (Brindley et al. 1982; Tanagho et al. 1989). This is clearly not the normal physiological mechanism of voiding (McGee et al. 2015), yet the Finetech-Brindley System (Finetech Medical, Welwyn Garden City, UK) has been implanted in >2,500 people

Table 2. *Neurostimulators: studies*

Device	References
TENS, NMES	TENS: Catley et al. (2015); Johnson et al. (2015) NMES: Sheffler and Chae (2007)
Cochlear implant*	Clark (1999); Gates et al. (1995); Roche and Hansen (2015)
Spinal cord epidural stimulator (neuromodulation)*	Crucchi et al. (2016); Health Quality Ontario (2005); Kumar et al. (2007); Wolter (2014)
Sacral nerve stimulator	Al Asari et al. (2014); Gupta et al. (2015); Li et al. (2016); Moya et al. (2016)
Foot-drop stimulator (external)*	Bethoux et al. (2014); Liberson et al. (1961); O'Dell et al. (2014); Stein (1998); Stein et al. (2006)
Vagus nerve stimulator*	Kennedy and Schallert (2001); Révész et al. (2016); Yuan and Silberstein (2016a, 2016b, 2016c)
Deep brain stimulator*	Benabid et al. (2006); Khobragade et al. (2015); Mazzone et al. (2016); Nardone et al. (2014)
Phrenic nerve and diaphragm pacers*	Romero et al. (2012); Santangeli and Marchlinski (2015)
Sacral anterior root stimulator implant*	Brindley (1974, 1977); Heine et al. (1977); McGee et al. (2015); Tanagho et al. (1989)
Grasp-release stimulators (external)*	Alon et al. (2007); Buick et al. (2016); Nathan (1994); Prochazka (2002); Prochazka et al. (1996, 1997a)
StimRouter implant*	Deer et al. (2010, 2016); Gan et al. (2011); Prochazka (2005)
Grasp-release stimulators (Freehand implant)†	Knutson et al. (2012); Peckham et al. (1980, 1981); Peckham and Kilgore (2013)
Foot-drop stimulator (implant)§	Haugland et al. (1994); Hoffer et al. (2005); Jeglic et al. (1970); Martin et al. (2016, 2017); Taylor et al. (2016); Waters et al. (1975)
Implanted microstimulators§	Burridge and Etherington (2004); Loeb et al. (2004, 2006); Richmond et al. (2000); Weber et al. (2005)
Visual cortex implant§	Brindley (1974); Brindley and Lewin (1968); Dobbelle et al. (1976); Lewis et al. (2015)
Retinal implant§	Barry et al. (2012); da Cruz et al. (2016); Rachitskaya and Yuan (2016)
Somatosensory cortex implant§	Flesher et al. (2016); Kim et al. (2015); Shaikhouni et al. (2013)
Spinal cord epidural stimulator (spatiotemporal)§	Capogrosso et al. (2016); Wenger et al. (2016)
Intraspinal microstimulation for bladder control, limb movement and respiratory pacing§	Bladder: Gaunt and Prochazka (2006); Grill et al. (1999); Jonas et al. (1976); Nashold et al. (1971b, 1981); Pikov et al. (2007); Tai et al. (2004) Limb movement: Holinski et al. (2013); Mushahwar et al. (2000); Prochazka et al. (2001) Respiration: Mercier et al. (2017)

*Commercially available, †discontinued commercially, §experimental.

(Rijkhoff 2004), the majority of whom were able to void using their stimulator with residual volumes <30 ml and consequently were freed from catheter usage, which reduced the incidence of urinary tract infections (Martens et al. 2011; Van Kerrebroeck et al. 1993). Some recipients reported stimulator-driven erections and defecation (Egon et al. 1998).

The problem of activating neuronal populations additional to those intended was also encountered in numerous studies evaluating intraspinal microstimulation to treat bladder-sphincter dyssynergia (Gaunt and Prochazka 2006; Grill et al. 1999; Jonas et al. 1975a, 1976; Jonas and Tanagho 1975; Nashold et al. 1971a; Tai et al. 2004; Tanagho 1988). In the initial studies in lightly anesthetized dogs and cats, partial voiding could be elicited by stimulation within the sacral spinal cord, but bladder contractions were usually accompanied by EUS contractions (Friedman et al. 1972; Jonas et al. 1975a, 1975b). It was recognized that intraspinal microstimulation at separate sites

would be needed to activate the bladder and at the same time inhibit the EUS. Poststimulus voiding (see above) was proposed as an intermediate solution (Jonas and Tanagho 1975). Between 1970 and 1980, 27 people with spinal cord injury were implanted with pairs of sacral intraspinal electrodes (Nashold et al. 1982). They remain the only recipients of intraspinal electrodes in the world. In most cases, stimulation produced some voiding, but the bladder-sphincter coactivation seen in the animal experiments remained a problem. Partial sphincterotomies improved voiding in some of the implanted people. The detailed outcomes of these human trials have been reviewed elsewhere (Gaunt and Prochazka 2006; Nashold et al. 1981).

Studies of intraspinal microstimulation for bladder control resumed in a number of centers in the late 1990s. In lightly anesthetized cats, interleaved pulse trains were delivered via implanted microelectrodes to the sacral parasympathetic nucleus to produce bladder contractions (de Groat 2006; Grill et al. 1999) and to the dorsal gray commissure to relax the EUS (Blok and Holstege 1998; Grill et al. 1999). The present author and his colleagues were involved in a related study in awake animals (Gaunt and Prochazka 2008). Although the combined, interleaved stimulation of the above sites sometimes elicited the desired combination of increased bladder pressure and decreased EUS pressure, it rarely resulted in complete voiding. More commonly, coactivation of the bladder and EUS occurred, and so the method was deemed insufficiently reliable by the present author and colleagues to justify clinical trials (Gaunt and Prochazka 2008). In retrospect, it was known from neurophysiological experiments that some interneurons in the dorsal gray commissure increase their firing during voiding and inhibit EUS motoneurons, while others decrease their firing

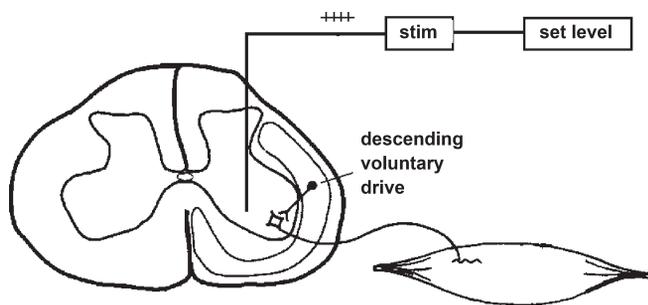


Fig. 4. Schematic showing how steady intraspinal microstimulation at a level close to threshold for activating motoneurons that elicit muscle contractions could boost descending commands to those motoneurons and muscles in a person with spinal cord injury (reproduced from Prochazka et al. 2002b with permission).

during voiding and excite EUS motoneurons (Buss and Shefchyk 2003). Presumably, when intraspinal microstimulation in the dorsal gray commissure activated more of the latter than the former interneurons the EUS contracted, exacerbating bladder-sphincter dyssynergia rather than reducing it.

In a separate study, coordinated bladder contraction and EUS relaxation was elicited in cats lightly anesthetized with propofol by intraspinal microstimulation in, or just dorsal or lateral to, the dorsal gray commissure (Pikov et al. 2007). Voiding was elicited in two-thirds of the implanted animals, though in nearly all cases the voiding was incomplete. Stimulation at closely adjacent sites could elicit EUS contraction rather than relaxation, as found in the previous studies. The effects on the bladder and the EUS could reverse when the bladder filled. Three of the animals were tested after a low-thoracic spinal cord transection. The voiding responses to intraspinal microstimulation were maintained. Unlike humans, however, spinalized cats do not develop sustained bladder-sphincter dyssynergia, and therefore it is unclear whether intraspinal microstimulation would have the same effects in humans with spinal cord injury. Setting aside the difficulties and uncertainties, it may be possible in the future to use arrays of multicontact microelectrodes to select a small number of specific stimulation sites that produce reliable voiding in spinal cord-injured people.

Other, less invasive approaches to this problem that target the pudendal nerve have been explored. The pudendal nerve contains motor axons that innervate the EUS and axons that transmit sensory input from the pelvic floor, urethra, and external genitalia to the sacral spinal cord. The sensory input can either inhibit or facilitate bladder contractions, depending on its source (McGee et al. 2015). Selective stimulation of the sensory and motor branches of the pudendal nerve with specific amplitudes, rates, and patterns has been used either to inhibit the EUS and facilitate bladder contraction for voiding or to inhibit the bladder and activate the EUS to maintain continence (Boggs et al. 2006a, 2006b; Lee and Creasey 2002; McGee et al. 2015; Yoo et al. 2007; Yoo and Grill 2007). High-frequency blockade of the pudendal nerve to relax the EUS is another promising line of investigation (Bhadra et al. 2006; Boger et al. 2008; Kilgore and Bhadra 2004; Tai et al. 2007a, 2007b). It was shown recently in chronically spinalized animals that activation and blockade of the pudendal nerve can be achieved with the StimRouter system (Gaunt and Prochazka 2009). This may provide a low-cost type of neuroprosthesis for either maintaining continence or eliciting voiding, as only one or two leads would need to be implanted.

Failures and Successes

Most therapeutic interventions in biology are not always completely successful, and neural engineering interventions are no different. Some of the obstacles have been discussed above. To summarize, the most common problems are 1) to connect the recording or stimulating device to the target neurons and only those neurons; 2) to maintain this connection over long periods of time; 3) to achieve these outcomes without damaging the target or surrounding tissues; 4) to process the recorded signals and the stimulus trains so that they lead to useful outcomes; 5) to overcome or compensate for connective tissue encapsulation; and 6) to avoid infection. Failure reports

on commercial NSs are available in the Manufacturer and User Facility Device Experience (MAUDE) database of the US Food and Drug Administration (Food and Drug Administration 2017).

What has been the impact on people's lives of neural engineering devices? Cardiac pacemakers are not strictly neural prostheses, but they activate highly specialized cardiac muscles that have nerverlike properties and many aspects of their design are common to those in neural prostheses. The benefits of cardiac pacemakers are well known: they have extended the lives and quality of life of hundreds of thousands of people.

The cochlear implant has provided functional hearing in over 300,000 deaf people. The history of its development over the last 60 years, its clinical adoption in many countries, and the controversies that have swirled around it are excellently reviewed in a book written by a sociologist faced with deciding whether his two deaf children should receive implants (Blume 2010). In the 1950s and 1960s there were concerns about the safety of implanting electrodes inside the cochlea, in relation to infection and the possible destruction of residual hearing, but as the devices and implant surgery improved the rate of adverse events has declined to below 5% (Farinetti et al. 2014). Nonetheless, lifetime follow-up is needed to detect and treat complications (Terry et al. 2015). In the early 1970s, neurophysiologists, finding that auditory nerve axons responded to sound differently than to electrical stimulation, were skeptical that electrical stimulation through one or more electrodes would ever produce useful hearing, and they recommended against proceeding in the absence of further basic research (Kiang and Moxon 1972). However, the clinicians, engineers, and manufacturers involved pressed ahead vigorously. Various electrode designs, speech encoding algorithms, and stimulation patterns were developed and tested in successive generations of devices (Roche and Hansen 2015). The current state of play, after a half century of technical development and clinical experience, is that a sizable proportion of implant recipients show remarkable hearing performance, including open-set speech recognition, acquisition of language skills, and even music appreciation (Roche and Hansen 2015).

To the surprise of those not directly involved, segments of the deaf community opposed the cochlear implant, particularly in relation to its application in children. Their greatest concern was that the cochlear implant undermined Deaf Culture. To quote Blume: "membership in the Deaf Community is psychologically and socially beneficial in itself. . . . parents of children newly diagnosed as deaf may not take the trouble to learn to sign language. However much or however little the child might ultimately profit from the implant, it would be deprived of access to language for a vital period of its early life." The contrary point of view is that a deaf child not in receipt of a cochlear implant would be deprived of a level of hearing that in many cases has led to the acquisition of language through auditory means. The normal procedure is to implant a device on one side first, which reduces the risk of abolishing residual hearing. Furthermore, sign language and lip reading may still be taught, as recommended for example by the French National Consultative Ethics Committee on Health and Life Sciences (CCNE).

The lesson here is that it is important for neural engineers to understand not only the neurophysiology but also the relevant

societal and psychological factors when developing novel interventions. At one of the first conference presentations on brain-machine interfaces, an audience member with quadriplegia told the present author “I need brain electrodes like I need a hole in the head.” But actually it was neither the trephination nor the implants that he was averse to, it was the large connector attached to the recipient’s head. If the device were fully implanted and invisible, he felt it might be acceptable and useful. Most users of assistive devices do not want to attract attention to their disability. Recognition of this factor has led to a focus on the design of less obtrusive and more cosmetically appealing neural prostheses (Kilgore et al. 2001).

Needless to say, the cost-benefit ratio is crucial and this goes well beyond purely monetary costs. For example, stimulator garments can improve upper limb function to some extent in people living with spinal cord injury, but if they take longer than a minute or two to don, they fall into disuse (personal observations). People with hemiplegia learn to perform many tasks one-handed, so any improvement in function of their affected hand has to be very substantial to result in a favorable cost-benefit ratio (Buick et al. 2016). Questionnaires that measure user satisfaction with assistive devices can provide useful information in this regard (Day et al. 2002; Demers et al. 2002; Jutai and Day 2002).

Neurophysiological Insights Arising from Neural Engineering Research and Development

Given the close relationship between neurophysiology and neural engineering, it was only to be expected that experience gained from the practical use of NSs and BCIs would lead to fundamental neurophysiological insights. Let us consider a few interesting examples.

Information conveyed by the relative timing of sensory inputs. Interaural time differences of stimulation of $\sim 100 \mu\text{s}$ with bilateral cochlear implants improve speech and music perception, showing that the detection and decoding of minute differences in timing, presumably by neurons in the cochlear nuclei, lateral geniculate body, and auditory cortex, is an important mechanism, not only for directional discrimination of sound sources but also in higher-level processing of complex sounds (Laback et al. 2015).

Information content, modulation, and feedback effects of proprioceptive signals. It is still not entirely clear how proprioceptive input from mammalian muscle spindles is modulated and used by the CNS to control normal voluntary movement. Neurographic recordings in awake humans have resulted in a number of different ideas about the way the CNS uses gamma fusimotor action to modulate spindle afferent stretch sensitivity. There is evidence for alpha-gamma coactivation, which serves to maintain spindle afferent firing during active muscle shortening (Hagbarth and Vallbo 1969; Matthews 1970; Proske and Gandevia 2012). There is also evidence for predictive gamma biasing action to compensate for future muscle length changes (Dimitriou and Edin 2010; Ellaway et al. 2015; Taylor et al. 2006). Neurography needles are easily dislodged, which has limited the freedom of movement in the human data (Prochazka and Hulliger 1998). Recordings from dorsal root afferents during a wide range of rapid, unrestricted movements in animals have tended to support a simpler, kinesthetic role for muscle spindles (Cody et al. 1975; Cody and Taylor 1973;

Goodwin and Luschei 1975; Prochazka et al. 1976), whereby in addition to there being an alpha-linked component of gamma activity there is an additional component that is related to the difficulty, novelty, and context of the motor task (“fusimotor set”) (Prochazka 2015b; Prochazka et al. 1985; Prochazka and Ellaway 2012).

The latest recordings from sensory afferents in freely moving animals were obtained in the context of using dorsal root recordings as feedback signals to control motor NSs. It was found that signals from as few as 10 afferents enabled a fairly accurate reconstruction of the kinematics of a whole limb during free locomotion (Rigosa et al. 2011; Weber et al. 2006, 2007). Most of the relevant information was derived from the signals from muscle spindle afferents, which supports the idea that muscle spindles provide kinematic information. In addition, it became apparent from modeling studies that a crucial role for proprioceptive feedback is to switch the locomotor pattern generator from generating stance to generating swing and back at just the right time in the step cycle (Cruse 1990; Cruse and Warnecke 1992; Prochazka 1993; Prochazka and Yakovenko 2007a, 2007b). Brain control of balance and postural reactions has been shown to depend in part on proprioceptive input, and recent experiments in humans subjected to postural threats indicate that under these circumstances fusimotor input sensitizes muscle spindles, which increases the sensitivity of postural reflexes (Davis et al. 2011; Horslen et al. 2013).

Positive force feedback. Another role for proprioception is to assist in load compensation, for example in supporting the weight of the body during the stance phase of locomotion. There is some debate as to the relative contributions of stretch reflexes and centrally generated drive in this case (Pearson 2004; Prochazka et al. 2002a). The Pearson laboratory has provided evidence that during locomotion reflexes mediated by extensor Golgi tendon organ afferents may generate significant levels of load-bearing extensor force (Donelan and Pearson 2004). This implies positive force feedback control, a mechanism supported by earlier data showing that whereas in static postures extensor motoneurons are inhibited by input from homonymous tendon organ afferents, during locomotion there is a reflex reversal such that tendon organ feedback becomes excitatory (Donelan and Pearson 2004; Pearson 2004). In technology, positive feedback is usually avoided, because it can lead to instability. On the other hand, when muscles shorten they produce progressively less force in response to a given increase in neural drive, and this has the effect of reducing the loop gain of positive force feedback to below the level that results in instability. From a theoretical point of view, positive force feedback to limb extensor muscles is in fact an effective load-compensating mechanism unless muscle length is constrained, as in isometric contractions (Prochazka et al. 1997b, 1997c, 2002a).

Spatiotemporal activation of spinal motoneuron pools during locomotion. In attempts to elicit locomotion with arrays of intraspinal electrodes (Mushahwar et al. 2000), the need arose for an accurate three-dimensional map of the location of motoneuron pools in the lumbosacral spinal cord. Previously published histological data (Vanderhorst and Holstege 1997) were digitized, and a spatiotemporal animation of the activation of motoneuron pools was developed by combining this model with the known EMG activity of the main hindlimb

muscles in normal locomotion (Yakovenko et al. 2002). This approach has since been used to model motoneuronal activation in human and rodent spinal cords (Ivanenko et al. 2006; Wenger et al. 2016). The animation revealed a rostrocaudal oscillation of motoneuronal activity within the spinal cord during the locomotor step cycle. The first impression was of a wave of activity propagating smoothly up and down the spinal cord (Yakovenko et al. 2000). Waves of activity traveling along the neuraxis had previously been suggested as controlling the orderly sequencing of muscles in locomotion in different species, particularly in limbless animals that move by undulating their bodies (Orlovsky et al. 1999). It was further suggested that the neural network producing these traveling waves had been conserved in the evolutionary transition from aquatic to terrestrial locomotion (Bem et al. 2003; Ijspeert et al. 2007). On closer analysis of the spatiotemporal model, it became evident that the rostrocaudal shifts of activity at the transitions between the swing and stance phases of locomotion were quite abrupt (Yakovenko et al. 2002). This suggests that the evolutionary modification of the spinal locomotor network from limbless to limbed animals must have involved a speeding up of the migrations of activity at the swing-stance and stance-swing transitions. A further insight came from neuro-mechanical modeling, which indicated that the stance phase of the locomotor step cycle lasts longer than the swing phase for biomechanical reasons and that the central pattern generator may be tuned to produce this relationship by virtue of persistent inward currents in timing interneurons (Prochazka et al. 2017; Prochazka and Yakovenko 2007a, 2007b; Spardy et al. 2011a, 2011b).

Hidden layer neurons and causality in corticospinal signals. Regarding the cortical control of movement, in BCI research the firing of ensembles of motor cortical cells has been used to decode a variety of behavioral states: intention to perform a mouse click, attention, change in target direction, forward/backward walking direction, hold/release periods, grasp type, and movement onset (Velliste et al. 2014). When rest and active hold states in monkeys performing center-out arm movement tasks were compared, it was possible to distinguish these states from the firing of ensembles of neurons but not from those of single neurons. Furthermore, when neuronal population vectors were used to move displayed cursors to targets, on occasion the monkey rested its arm while still controlling cursor movement (Schwartz AB, personal communication). This is in line with a previous study that showed that the firing of neurons in motor areas of the cerebrum could be dissociated from actual movement performance, indicating that some, perhaps most, neurons in the cortex do not directly control spinal motoneurons (Alexander and Crutcher 1990). Rather, they might be the equivalent of “hidden layer” neurons in artificial neural nets, which are indirectly involved in controlling input-output behavior but whose activity is not obviously related to either inputs or outputs.

What is activated by electrical stimulation of CNS structures? For many years it was assumed that epidural spinal cord stimulators activated axons in the dorsal columns and interneurons within the gray matter of the spinal cord. In fact, these devices were originally called dorsal column stimulators. It was then shown that epidural spinal stimulation activates axons in dorsal root filaments at lower amplitudes than axons in the dorsal columns (Ladenbauer et al. 2010; Rattay et al. 2000,

2003). Similarly, the identity of the neurons stimulated by DBS electrodes is unclear, which has impeded progress in understanding the mechanism of the therapeutic effects of DBS (McIntyre et al. 2004; Murrow 2014; Zhang and Grill 2010). Intraspinal microstimulation, originally assumed to activate alpha motoneurons and interneurons within a short distance of the electrode tips, was found to antidromically activate nearby branches of sensory axons at lower stimulus intensities and thereby to synaptically excite motoneurons in motoneuron pools spanning several spinal segments (Gaunt et al. 2006). This may explain how focal intraspinal stimulation can spread to activate many of the motoneurons of a synergistic muscle group.

What variable(s) are feedback controlled during movement? BCI recordings in human tetraplegic participants have provided insight into cortical feedback control (Willett et al. 2017). In center-out cursor-control tasks, neural population activity gradually declined as the cursor approached the target from afar, then decreased more sharply as the cursor came into contact with the target. Predictive corrections to the cursor's velocity were made during movement, and feedback corrections continued after the cursor reached the target. These observations are relevant to the basic question “what variable(s) does the nervous system control in limb movements?” (Stein 1982). Increasing the amplitude of electrical stimulation in the midbrain of the decerebrate cat increases the velocity of locomotion (Shik et al. 1966). Computer simulations have supported the idea that locomotor velocity is the basic command sent from supraspinal areas to the spinal locomotor pattern generator (Prochazka and Ellaway 2012). The BCI results suggest that velocity is also the controlled variable during arm movements, and that there may be a switch to positional or even force control once a target is reached.

Intermingling of neurons with opposite functions in the CNS. The conclusion from the experiments involving intraspinal microstimulation to elicit bladder voiding, namely that neurons within the same small region of the spinal cord can have very different and even opposite functions (Gaunt and Prochazka 2006), has a broader implication. Most imaging methods that show the “lighting up” of regions of the CNS do not have the resolution to differentiate between functionally different subpopulations, and therefore should be interpreted with caution.

Therapeutic carryover effects. Therapeutic electrical stimulation and functional electrical stimulation in people living with stroke and spinal cord injury have been shown to have carryover therapeutic effects (Andrews and Wheeler 1995; Shealy 1975; Vodovnik 1981) especially when performed in association with voluntary exercise training (Cauraugh et al. 2005; Cauraugh and Kim 2002, 2003; de Kroon et al. 2002; Kowalczewski et al. 2011; Musienko et al. 2012; Page et al. 2012; Popovic et al. 2006). Carryover effects lasting a few hours may result from short-term changes in the energetics of neuromuscular activation, whereas carryover effects lasting weeks or months have been attributed to muscle strengthening, neural plasticity, or both (Field-Fote 2004; Stein et al. 1992; Thomas and Gorassini 2005). Long-lasting carryover effects are also seen after TENS and sacral nerve stimulation for pain control, posterior tibial nerve stimulation to treat overactive bladder (Peters et al. 2010), and vagus nerve stimulation for epilepsy and depression (Yuan and Silberstein 2016a, 2016b, 2016c). The neurophysiological mechanisms involved are un-

clear and not necessarily the same in each case. In recent years they have been equated with neuromodulation, a process in which the activation of intracellular second messengers changes the responses of populations of neurons to subsequent inputs. Neuromodulation has a slower onset than direct ionotropic synaptic activation of neurons and is therefore associated with metabotropic synaptic transmission (Follesa et al. 2007; Yuan et al. 2016). It is interesting that despite TENS having become a standard therapeutic modality worldwide, the evidence for its efficacy in randomized controlled trials is still deemed inadequate and its presumed neurophysiological mechanisms of action remain uncertain (Catley et al. 2015; Johnson et al. 2015).

Future

In recent years several ambitious, multicenter neural engineering projects have been launched, with much publicity and large amounts of funding. These include the multiagency BRAIN Initiative (<https://www.braininitiative.nih.gov>), the Targeted Neuroplasticity Training Program (<http://www.darpa.mil/program/targeted-neuroplasticity-training>), and the European Human Brain Project (<https://www.humanbrainproject.eu/en>). These initiatives have been criticized by some as having unrealistic goals, ignoring basic neurophysiology, and depriving mainstream neuroscience researchers of funding. On the other hand, the goals are inspirational and will probably attract many young, innovative researchers to get involved in projects, some of which will fail and others will succeed, not always in the manner anticipated. Time will tell whether this shift in the organization and funding of neural engineering and neuroscience, which is taking place worldwide, will have been beneficial or not.

This also raises the interesting question, At what point of neurophysiological knowledge (or ignorance) is human implantation of devices justifiable? Recall that the cochlear stimulator was being implanted in deaf people in the early 1970s, when some expert neurophysiologists had concluded that “the present state of basic knowledge and technical competence argues strongly for additional preliminary work on animals” (Kiang and Moxon 1972). And yet within a few years, recipients were able to comprehend speech without visual input (Scott 2006). Recall too that deep brain stimulators have been implanted in tens of thousands of people without a clear understanding of the mechanism of action, yet they are often highly effective in suppressing tremor and unlocking rigidity (Birdno et al. 2014; Deeb et al. 2016). Perhaps the answer is that when the need is sufficiently pressing, when there is a reasonable chance of success based on the existing knowledge, and when the risks are judged to be acceptable, human trials are justified.

Conclusions

Neural engineering is a fast-growing, multidisciplinary field that has its foundations in neurophysiology and electrical engineering. The dawn of the transistor age in the late 1950s triggered an explosion of innovation, particularly with respect to implanted devices. In the intervening half century, there has been much progress in the development of materials, miniaturization, computerization, wireless communication, surgery, and, last but not least, the understanding of the underlying neurophysiological mechanisms. NSs are used clinically in

their hundreds of thousands and in some cases have become the standard of care for specific neural disorders. None of this would have been possible without a detailed knowledge of basic neurophysiology. Neural engineering is repaying the debt with new basic knowledge and insight. Regarding the future, some developments are predictable. For example, there will no doubt be further miniaturization of electronics and improvements in the biocompatibility of implanted materials, in signal acquisition and processing, in surgical implantation, in selectivity of stimulation, and in the reliability and longevity of devices. Many problems await solutions. Past history indicates that some of these solutions will come from unexpected directions. The growth predictions for NSs indicate a bright and fascinating future for this domain of applied neurophysiology.

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I have a financial interest in Rehabtronics Inc., which markets two of the devices mentioned in this article (StimRouter and Regrasp).

AUTHOR CONTRIBUTIONS

A.P. drafted manuscript; A.P. edited and revised manuscript; A.P. approved final version of manuscript.

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