Brain-Computer Interfaces Using Electrocorticographic Signals

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Clinical Application Review

***Abstract*—Many studies over the past two decades have shown that people and animals can use brain signals to convey their intent to a computer using brain-computer interfaces (BCIs). BCI systems measure specific features of brain activity and translate them into control signals that drive an output. The sensor modalities that have most commonly been used in BCI studies have been electroencephalographic (EEG) recordings from the scalp and single-neuron recordings from within the cortex. Over the past decade, an increasing number of studies has explored the use of electrocorticographic (ECoG) activity recorded directly from the surface of the brain. ECoG has attracted substantial and increasing interest, because it has been shown to reflect specific details of actual and imagined actions, and because its technical characteristics should readily support robust and chronic implementations of BCI systems in humans. This review provides general perspectives on the ECoG platform; describes the different electrophysiological features that can be detected in ECoG; elaborates on the signal acquisition issues, protocols, and online performance of ECoG-based BCI studies to date; presents important limitations of current ECoG studies; discusses opportunities for further research; and finally presents a vision for eventual clinical implementation. In summary, the studies presented to date strongly encourage further research using the ECoG platform for basic neuroscientific research, as well as for translational neuroprosthetic applications.**

***Index Terms*—Brain-computer interface (BCI), Brain-machine interface (BMI), electrocorticography (ECoG).**

I. Introduction

OVER the past decade, electrical recordings from the surface of the brain [i.e., electrocorticography (ECoG)] have become recognized as a promising signal platform for brain-computer interface (BCI) research and application. ECoG is acquired by placing electrodes underneath the skull, either above (epidural) or below (subdural) the dura mater, but not within the brain parenchyma itself (see Fig. 1). Compared to signals acquired from the scalp [electroencephalography (EEG)] and intraparenchymal single neuronal recordings, ECoG recordings have characteristics that make them especially suited for basic neuroscience research and resulting translational opportunities. These characteristics include high spatial resolution and signal fidelity, resistance to noise, and substantial robustness over long recording periods. ECoG recordings appear to strike an ideal balance between fidelity and clinical practicality.

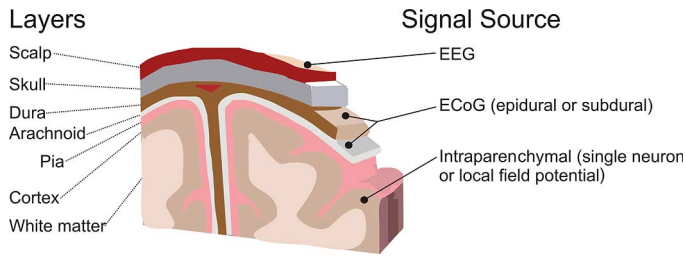

Fig. 1. Recording domains: from single-unit recordings within the brain to EEG recordings on the surface of the scalp. From [1].

Surface cortical potentials were first recorded from animals and humans in the late 19th century [2]. More recently, in the past several decades there has been a renewed scientific interest in ECoG signals in a variety of animal studies (particularly in rats, rabbits, cats, and pigs) (e.g., [3]–[13]). Because placement of ECoG electrodes requires an intracranial surgery, research experience in humans has been more limited. Thus far, the majority of human studies occurred with patients with intractable epilepsy who are candidates for invasive monitoring to localize their seizure focus and to identify eloquent cortex. A smaller minority of studies have also used patients undergoing an awake craniotomy for the treatment of tumors adjacent to motor and speech cortex. Very rarely, patients have been experimentally implanted [14] for research purposes. Because the vast majority of ECoG electrode arrays are surgically placed for clinical indications, the configuration and location of the electrodes, as well as the duration of the implant, are determined solely by clinical requirements and without any regard for research needs. The clinical electrodes are typically platinum electrodes with a diameter of 4 mm (2.3 mm exposed), and are configured in either a grid (e.g., 8×8 electrodes) or strip (e.g., 4 or 6 electrodes) configuration with an interelectrode distance of usually 10 mm (see Fig. 2). They are generally implanted only for periods of several days to 1–2 weeks. More recently, FDA-approved microelectrode arrays have also been implanted concurrently with the more traditional (i.e., macro) electrode grids.

During the one to two week periods of invasive monitoring, there are brief windows of time that these patients can participate in experiments. These experimental sessions are often constrained by the person's willingness to participate, baseline neurologic function, medical condition, and ongoing medications. Early studies engaged these patients to pursue basic neu-

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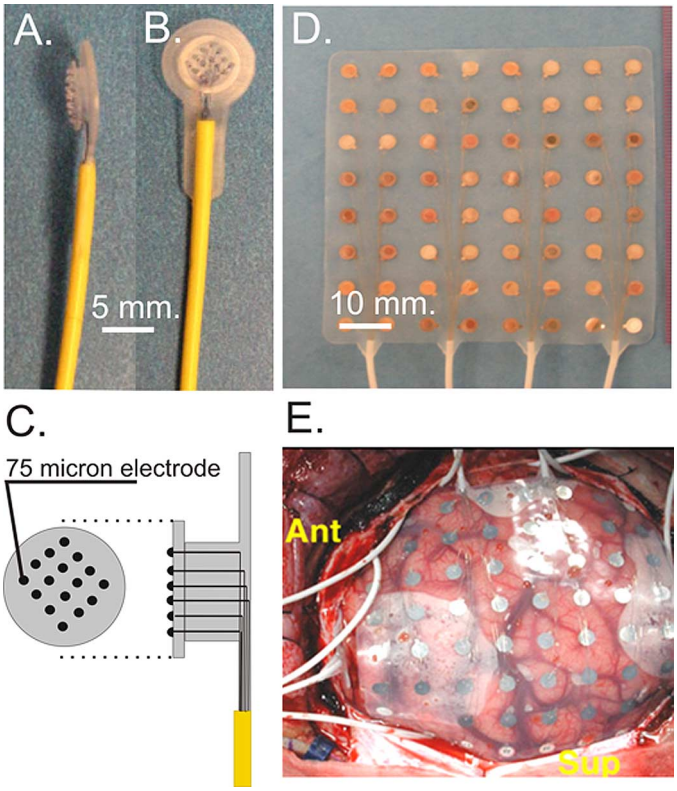

Fig. 2. Micro and macro ECoG arrays. A-B: Microgrid arrays. C: Schematic of microgrid array. D: Standard clinical macrogrid. E: Surgical placement of macrogrid.

roscience questions on cortical function and exploration of new ways for topographic localization of motor or language function [15]–[20]. These early efforts set the stage for more definitive evaluation of cortical changes associated with distinct frequency spectra that was first demonstrated using simple visuomotor tasks [21], [22]. With the growing interest in neuroprosthetics, these early clinical and neuroscientific studies gave way to fully multidisciplinary endeavors that combined experts from engineering, mathematics, and computer science to evaluate ECoG data as a potential BCI signal substrate in offline analyses [23]–[28]. In 2004, the first online ECoG BCI study by Leuthardt *et al.* [29] showed that ECoG can support accurate BCI operation with little user training. Additionally, this study also provided initial evidence that ECoG signals contain information about the direction of hand movements, which was one of the earliest demonstrations to show that specific details of motor function can be accurately inferred without measurements from individual neurons.

From both a performance and a clinical standpoint, ECoG appears to minimize some of the limitations that have hindered traditional noninvasive and invasive signal acquisition techniques. EEG is noninvasive and has supported important BCI applications, including two- and three-dimensional BCI control [14], [30]–[46]. However, the highest functioning EEG-based BCIs often require a substantial degree of user training and their performance is often not reliable. BCIs that are based on intracortical recordings of action potential firing rates or local field potentials are on the opposite end of the performance and clinical spectrum [47]–[56]. Though they can achieve a high level of multidimensional control, there still remain significant and unresolved questions regarding the long-term functional stability of intracortical electrodes, particularly for recording action potentials from individual neurons [57]–[59], although there are some encouraging reports of better reliability [60], [61]. This lack of signal durability has important clinical implications, because signal loss would require frequent and relatively unpredictable replacement of the implant, which would likely be unacceptable. In summary, despite encouraging evidence that current noninvasive and invasive BCI technologies can actually be useful to severely disabled individuals [41], [54], [62], these shortcomings and uncertainties remain substantial barriers to widespread clinical adoption and implementation in humans.

Compared to EEG, ECoG has a number of favorable characteristics: higher spatial resolution [i.e., 1.25 mm (subdural recordings [63], [64]) and 1.4 mm (epidural recordings [65]) versus several centimeters for EEG]; higher amplitude (i.e., 50–100 μV maximum versus 10–20 μV maximum for EEG); far less vulnerability to artifacts such as electromyographic (EMG) or electroocular (EOG) activity ([66] or [67], respectively); and broader bandwidth (i.e., 0–500 Hz [68] versus 0–40 Hz for EEG). With respect to the larger bandwidth of ECoG compared to EEG, it is important to note that this advantage may in part be related to the larger amplitude of ECoG. Because ECoG generally follows a $1/f$ drop-off in signal power [69], task-related brain signals may remain larger than the noise floor of the amplifier/digitizer, and thus be detectable, at higher frequencies than for EEG. This easy access in ECoG to amplitudes in higher frequencies appears to be important, as signals at higher frequencies have been shown to carry substantial information about cognitive, motor, and language tasks, and thus may provide critical information for BCI control that is not readily accessible with EEG. In addition to these advantages of signal and information quality, ECoG electrodes (which do not penetrate cortex) are likely to provide substantially greater long-term functional stability [70]–[74] than intracortical electrodes, which induce complex histological responses that may impair neuronal recordings [57], [75], [76]. In fact, a recent study by Chao *et al.* [77] showed in animals that the signal-to-noise ratio of ECoG signals, and the cortical representations of arm and joint movements that can be identified with ECoG, are stable over several months [78]. In addition, a recent study by Roland *et al.* [79] demonstrated that ECoG changes related to a simple motor task can be detected across a broad age group (11–59 yrs). This study also showed that the cortical area that exhibited gamma changes appeared to reduce with age, whereas the area that exhibited mu/beta changes appeared to expand.

Taken together, there is substantial theoretical and empirical evidence that ECoG could support a clinically and functionally reliable BCI with a high level of performance. Thus, it is reasonable to envision an ECoG-based implant that could substantially enhance the functional capability of a disabled patient by enabling their ability to modulate their environment, communicate, or control a prosthesis (see Fig. 3). Looking even further into the future, if the risk profile is low enough, it may also become reasonable to contemplate implants that augment capabilities in normal functioning adults.

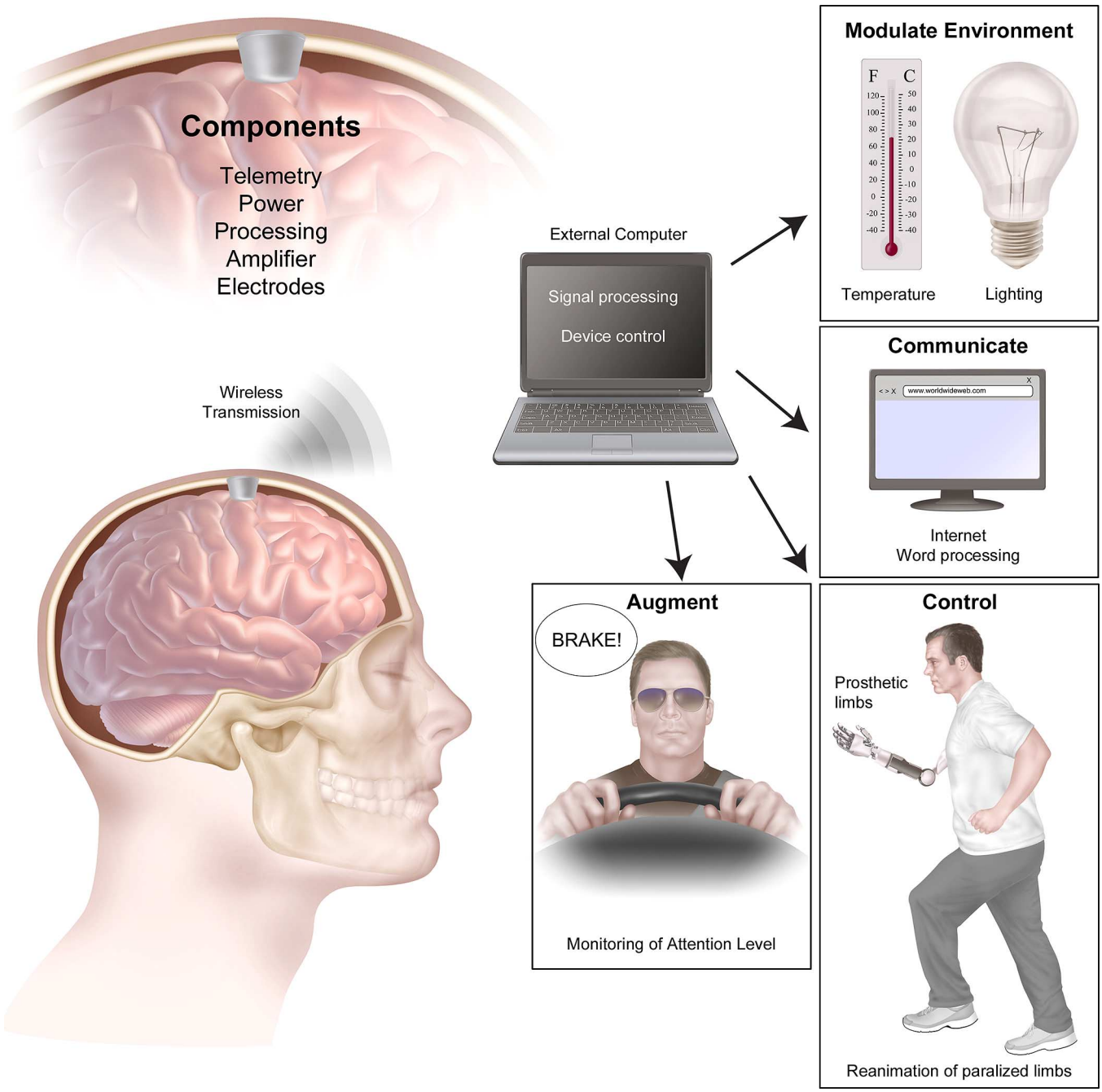

Fig. 3. Minimally invasive implant that integrates the electrodes, amplification, processing, power, and telemetry to wirelessly connect and control an external computer system. In disabled patients, this could have multiple output effector systems ranging from environmental communication and control to control of various robotic prosthetics. Alternatively, if the system has a low clinical risk profile, implants may eventually be utilized to augment capabilities in healthy individuals.

II. Electrophysiological Features Detected By ECoG

Because ECoG electrodes are beneath the skull and close to the cortical surface, they can detect brain activity very accurately. For this reason, many recent studies have begun to use ECoG to study the basic mechanisms of cortical processes. While it is difficult to make inferences about the nature of these processes using only ECoG measurements, specific ECoG features, in particular those extracted using frequency analysis, have been found to be related to important aspects of motor function, sensory perception, or cognition. The existing and emerging understanding resulting from the corresponding studies is summarized below.

When ECoG is analyzed using frequency analysis, the most relevant frequency bands in the lower frequency range (less than 40 Hz) include the mu and beta bands [80]. Different aspects of these bands have been extensively described in the classical EEG literature, as well as in numerous BCI studies in which they are used. Mu (8–12 Hz) and beta (18–26 Hz) rhythms are oscillations that are thought to represent post-synaptic potentials associated with thalamocortical modulation of motor cortex (e.g.,

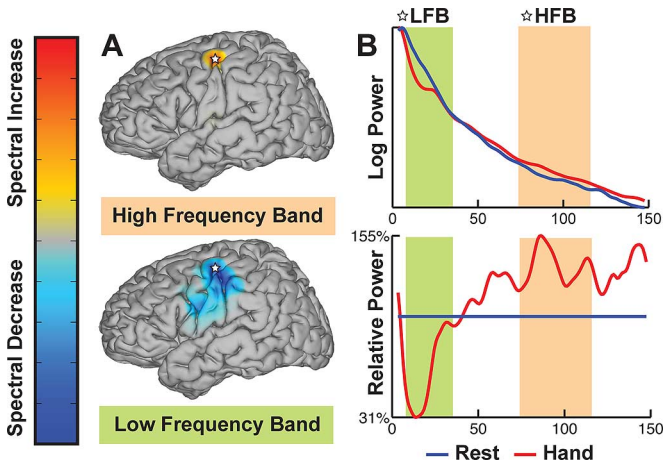

Fig. 4. Example of ECoG activity changes during the task of repetitively opening and closing of the hand and during rest. A: Signals in the mu/beta band (5–30 Hz) decrease with the task and are spatially less specific (i.e., they are broadly distributed topographically), whereas signals in the gamma band (i.e., 70–116 Hz as measured here) increase with the task and are spatially more specific (i.e., they are sharply focused topographically). B: Power spectrum on a logarithmic scale for the electrode marked with a star in the topographies illustrates the decrease in the mu/beta band (marked by the green bar) and increase in the gamma band (orange bar). From [115].

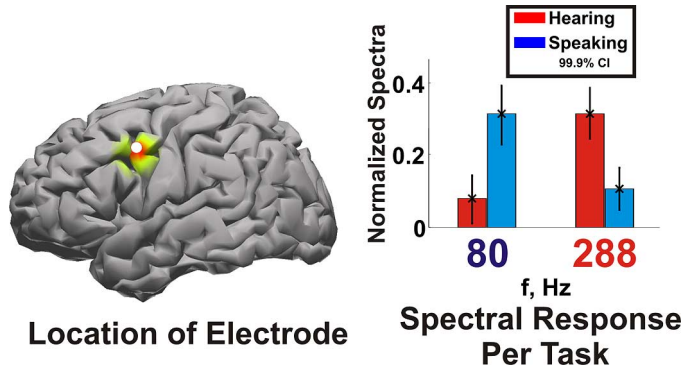

Fig. 5. Brain model on the left shows location of exemplar electrode found to have significant gamma power modulation during a word repetition task. Bars with 99.9 percent confidence intervals on the right illustrate the statistical significance of the difference in normalized power detected at this location between the two cognitive tasks at two distinct gamma subbands. Modified from [87].

[81]). Activity in each of these bands is focused narrowly spectrally and across relatively large areas along the central sulcus spatially. Motor actions, but also motor imagery, are usually associated with a decrease in spectral amplitude across the corresponding area of motor cortex (see Fig. 4, [21], and [82]). These changes in mu and beta activity have been found to reveal only limited information about localized cortical processing, such as that corresponding to different directions of hand movements [83].

EEG can readily detect activity up to approximately 40 Hz, which enables detection of mu and beta activity very well. Although recent advances in recording hardware and analysis methods suggest that the frequency range of EEG may be extended [84], [85], physiological and biophysical barriers will likely limit robust recording of activity in higher frequencies. In contrast to activity in the mu and beta band, which are narrowly focused spectrally, activity in the gamma band has been found from 40 Hz [86] to several hundred Hz [87]. Because ECoG electrodes are beneath the skull and in close proximity to the cortical surface, activity at these high frequencies can be more readily detected with ECoG than with EEG. This has important implications for BCI development, because gamma activity, unlike mu and beta activity, displays much more specific functional localization (e.g., see Fig. 4). Many ECoG studies [18], [20], [22], [28], [29], [63], [87]–[107] have demonstrated that spatially focused gamma activity correlates closely with specific aspects of motor, language, or cognitive function (e.g., see Fig. 6).

In summary, according to current understanding, oscillations in lower frequencies are thought to represent thalamic modulation, and activity in the gamma band represents cortical processing. While the precise mechanism for the generation of mu and beta rhythms is still somewhat debated, gamma activity has been shown to be strongly correlated to the firing rate of individual neurons [100], [108]–[110] and has also been closely linked to the blood-oxygen-level dependent (BOLD) signals detected by functional magnetic resonance imaging (fMRI) [94], [111]–[113]. It remains unclear to what extent its amplitude depends on the respective contributions of neuronal firing rates and synaptic potentials, and of their relative phases. In general, our understanding of the phenomenology of gamma activity continues to evolve. Early reports distinguished between a lower gamma band (35–45 Hz) and a higher gamma band (80–100 Hz) [88]. Gamma activity has also been posited to reflect the net result of asynchronous neuronal firing, which results in a more uniform broadband noise-like phenomenon that declines in amplitude as frequency rises [69], [98]. While this effect appears to be robust and ubiquitous across different functional areas, more recent evidence [87] provides new evidence for functionally separable frequency bands. In this study, Gaona *et al.* showed that during a word repetition task, gamma subbands distinguished stages of the task (for a given location) and differentiated cortical locations (for a given stage of task) (see Fig. 5 for an example). Additionally, these different gamma subbands, both on the macroscale and microscale, have been used for device control in humans [114]. In either case, ECoG's access to high frequency amplitudes provides an important although not yet fully explored advantage for basic neuroscientific investigations and for neuroprosthetic applications.

In addition to amplitude modulation in different frequency bands, ECoG signals also have functionally relevant characteristics in the time domain. Specifically, ECoG signals display discrete (i.e., evoked) and continuous time-domain features. Discrete time-domain ECoG features can be readily evoked by a sensory stimulus or motor action. Examples include visual evoked potentials (VEPs) [116] evoked by presentation of individual visual stimuli, steady-state visual evoked potentials (SSVEPs) [117] evoked by rapid serial visual presentation, P300 responses (e.g., [31], [118]) evoked by the presentation of an attended and rare stimulus, and motor evoked potentials (MEPs) induced by the onset of a movement [25]. The use of these signals for ECoG device control thus far has been fairly limited. There was early interest in the use of ECoG MEPs for

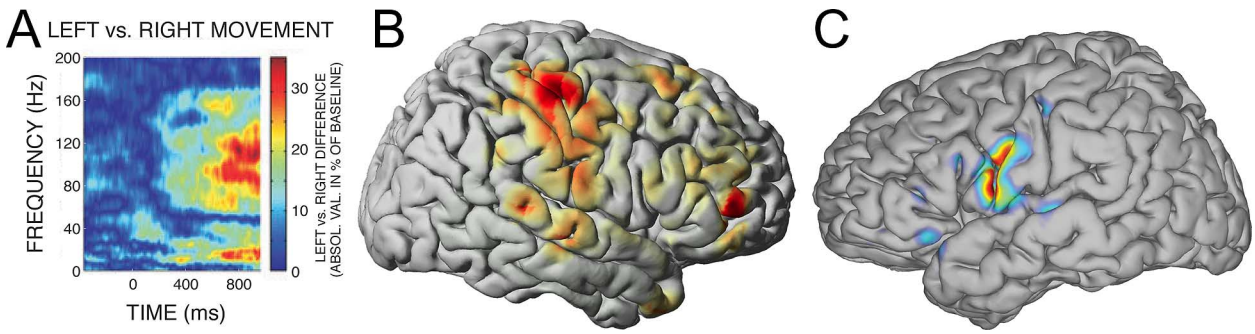

Fig. 6. Information in ECoG about the direction of hand movement and imagined vowels. A: ECoG in different frequency bands at one location in the contralateral hand area of motor cortex from one subject differentiates left and right movement directions. From [29]. B: Color-coded shading of average data from five subjects illustrates the information about hand movement direction provided by ECoG recorded over different cortical areas. Most of the information is captured over the hand representations of motor cortex. Modified from [90]. C: Color-coded shading of average data from six subjects illustrates the information about overt production of different vowels provided by ECoG recorded over different cortical areas. Most of the information is located over Wernicke's area. From [106].

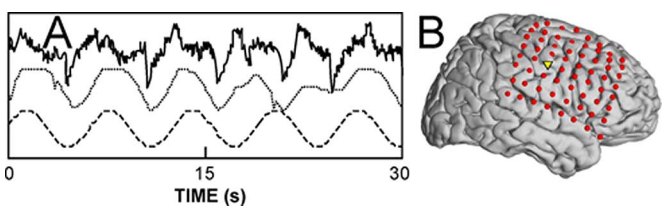

Fig. 7. Example ECoG time course during a tracking task. A: ECoG time course of one location [marked with a yellow triangle in (B)], as well as the horizontal positions of the cursor (dotted trace) and the moving target (dashed trace). B: Three-dimensional brain model with location of electrode grid. The time course of ECoG activity follows the time course of the cursor but not the target. Modified from [90].

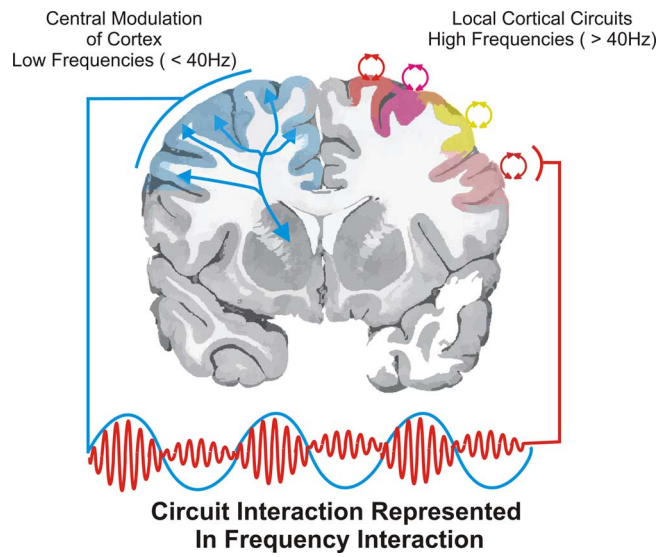

Fig. 8. Schematic of current and emerging understanding of ECoG signals and their physiological origin. The amplitude of mu/beta rhythm oscillations represent the level of thalamocortical interactions. Gamma activity represents the degree of local cortical processing. ECoG rhythm phase modulates local cortical processing.

BCI purposes [27], which to date has not resulted in real-time experiments. In the first study to use time-domain components in online experiments, a recent study [118] used VEPs and P300 evoked potentials to allow a subject to rapidly spell characters. Distinct from these discrete time-domain ECoG features, a continuous time-domain ECoG feature, referred to as the local motor potential (LMP), has recently been described [90], [97], [101]. The LMP shows close correlation with hand and finger kinematics (see Fig. 7). To date, the physiological origin of the LMP has not been defined, and the use of the LMP has also not been explored for real-time BCI application. Thus, further studies are required to define the LMP's potential to support neuroprosthetic applications. In addition to amplitude and time domain features, several recent studies have also begun to demonstrate that gamma activity is modulated by the phase of low-frequency brain rhythms (e.g., in the theta, mu, and beta ranges) [119]–[123]. These and other recent studies support a model in which: 1) local cortical processes can be detected using gamma activity; 2) function of a local cortical circuit (e.g., hand area of motor cortex) is facilitated or inhibited by the thalamus through thalamocortical oscillations whose neural correlates are represented in the lower theta/mu/beta rhythms [124]; and 3) coordination of cortical processing of different sites/networks are represented by the interactions of rhythm phase ([120], for review). See Fig. 8 for an illustration of this evolving understanding of cortical function and relevant ECoG signals. To what extent these coupling mechanisms could be useful in a BCI context remains unclear.

In summary, ECoG can detect several physiological processes and their interactions within and across sites. Gamma activity changes contain substantial information on cognitive, language, and motor function that can and has in part been used for BCI control. These activity changes will likely play an important role for future BCI development. The value of other physiological processes (e.g., the LMP, evoked potentials, low-frequency rhythms) and their interactions (e.g., cross-frequency coupling, phase-amplitude coupling), which also carry substantial information about cortical and in part subcortical processing, for neuroprosthetic application remains largely unexplored. In general, how the rich functional neurophysiologic representation detected in ECoG could or should be mapped to device control remains unclear at this point.

III. Current ECoG-Based BCIs

This section reviews the methods by which ECoG signals are commonly acquired, analyzed, and used for BCI applications, which to date mostly engaged human subjects. This is distinct

to development of many other biomedical techniques, which are usually studied first in animals before research begins in humans. This unusual situation occurred because ECoG-based research has emerged from investigations with people with intractable epilepsy. While important early demonstrations have been accomplished with this model, and will continue to contribute important understanding of human cortical physiology, further scientific research and technical development of ECoG-BCIs will greatly benefit from animal testing.

A. ECoG Signal Acquisition

ECoG recording devices are typically adjuncts to the clinical equipment in an epilepsy monitoring unit. The electrodes used for recording from the surface of the brain are typically standard FDA-approved electrodes that are used for seizure localization. After a craniotomy (i.e., a procedure to remove a large window of bone), an electrode array is placed that covers a wide area of cortex. The electrodes in such an array are usually 2.3 mm in diameter, spaced 1 cm apart, and are usually configured in a 8×8 grid configuration and embedded in a Silastic® base. As an alternative to a full craniotomy, small burr holes are made and 1×4, 1×6, or 1×8 electrode strips are placed bilaterally (this approach is typically used to lateralize the hemisphere of the seizure focus). More recently, smaller FDA-approved electrode arrays have been used in parallel with the clinical arrays for dedicated research use [64], [125]–[127]. These electrodes have ranged in size from 70 microns to 1.5 mm in diameter, with varied spacing of 1 to 4 mm. Beyond variance in sizing and spacing of electrodes, a new generation of microfabricated (i.e., micro-ECoG) electrode arrays that are embedded on thin films made of different biocompatible materials (e.g., polyimide, parylene, or silk) [128]–[130] is emerging. The choice of materials will determine the ability of the recording device to retain structural integrity and biocompatibility. In the long run, arrays that are thinner and more flexible will likely be able to accommodate less invasive surgical implantation and be better tolerated clinically. The microfabrication techniques used for these devices will also readily support the increase of the maximum number of electrodes.

Aside from the choice of materials, acquisition of ECoG signals requires attention to several important properties of the ECoG signal. Because ECoG amplitude attenuates rapidly as frequency rises [69], [122], i.e., from several hundred micro-Volt at low frequencies to several hundred nano-Volt at higher frequencies, effective ECoG recording requires high-fidelity amplifiers/digitizers with sufficient resolution in both time (i.e., adequate sampling rates) and amplitude (e.g., adequate voltage range and resolution). The sampling rate should be at least 1 kHz, the voltage range should be at least a few μV, and the resolution should be at least 16- or better 24-bit. In general, the noise floor of the amplification/digitization system needs to be lower than the amplitude of the signals that are to be detected (e.g., hundreds of nano-Volts for amplitudes in the high gamma band). Moreover, any analog low- or high-pass filtering at the amplification stage should accommodate the diverse physiological phenomena that are detected in ECoG. Ideally, there should be no high-pass filter (i.e., a dc amplifier) and the low-pass filter frequency needs to be less than half of the digitization rate to satisfy Shannon's sampling theorem. With regard to reference and grounding, the use of an intracranial reference and non-cortical (skull facing) grounding makes recordings less susceptible to noise compared to the use of scalp or cortical electrodes [87]. At present, most clinical (and even some research) ECoG amplification/digitization systems do not meet these stringent requirements, and thus may not be capable of acquiring ECoG signals with sufficient fidelity to capture all the information needed by a particular study.

Currently, since most ECoG-based BCI experiments are performed in the post-operative setting, there are numerous considerations that are quite different from elective EEG-based human studies and also from traditional well-controlled invasive animal studies. First and foremost, the patients who participate in these studies have had a major neurosurgical procedure. Their ability to cooperate will vary from day to day. Fluctuations in their capacity can be secondary to pain management, seizures and seizure recovery, and personal and social needs (e.g., needing to use the bathroom, family visitations, etc.). Thus, the research team is often "on call" to gather data when the patient is in an optimal state and willing to participate. It is important to emphasize that their participation is ultimately predicated on their generosity and patience during a very stressful time. That said, their involvement can often be a very positive experience. It often can aid in passing the time as the patient waits to have seizures. Additionally, since the patients' monitoring time is limited, and the research time within that monitoring period is further constrained by many factors, it can often be challenging to obtain an ongoing study that may require several days of training. In part due to these difficulties, relatively few online ECoG-based BCI studies have been published to date [29], [103], [114], [118], [126], [131]–[141]. Almost all of them, except [135], [136], [138], have used the highly flexible general-purpose BCI software platform BCI2000 [142], [143].

B. ECoG-Based BCI Protocol Design

The protocol of an ECoG-based BCI study usually has two parts. In the first part, the ECoG feature(s) to be used for BCI control are chosen. In the second part, the feature(s) are used for online BCI control of cursor movement or another output.

The first part of the typical protocol is to select those signal features that are to be used for BCI control, and is principally similar to other BCI studies using either EEG or single-unit recordings. For single-neuron recordings, this usually involves selection of particular neurons; for EEG or ECoG signals, this typically involves the selection of specific frequency bands and locations. Most commonly, the criterion for this selection is that the particular EEG/ECoG feature shows a clear difference between a particular action/imagery and rest. The majority of ECoG studies thus far have focused on associations with different types of motor or motor imagery tasks. More recently, several studies have used signals that were related to different non-motor tasks, such as auditory perception [132], speech production/imagery [114], or cognitive control [137]. A different possibility is to select features that correlate with specific parameters of an action or intention (e.g., velocity of hand

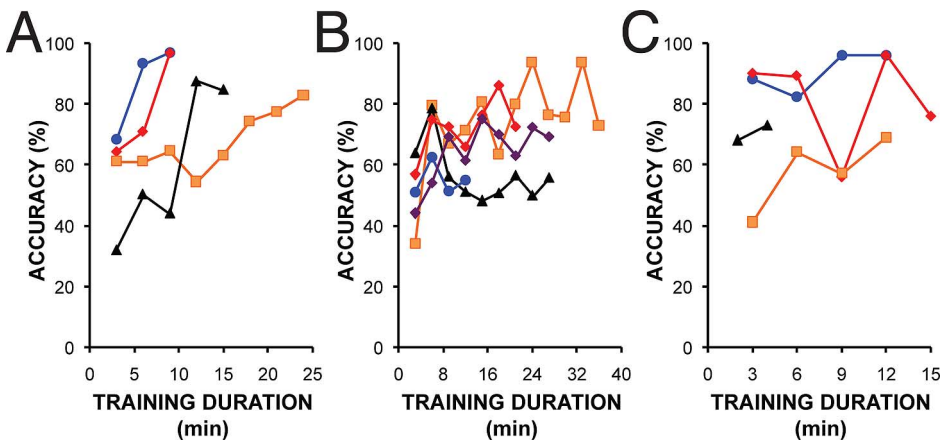

Fig. 9. A: Learning curves for ECoG control of vertical cursor movement using motor imagery to move up and rest to move down. (Accuracy in absence of control would be 50%.) Patient B (green trace) imagined opening and closing the right hand, Patients C (yellow trace) and D (red trace) imagined saying the word "move," and Patient D (blue trace) imagined protruding the tongue. Modified from [29]. B: Learning curves for ECoG control of two-dimensional cursor movement. (Accuracy in absence of control would be 25%.) Modified from [134]. C: Learning curves for ECoG control of one-dimensional cursor movement using phoneme articulation or imagery. (Accuracy in absence of control would be 50%.) Modified from [114].

movements, or distinct word articulations). This approach is encouraged by recent animal [77], [144] and human [29], [90], [92], [93], [95]–[98], [101], [145]–[147] studies that indicate that ECoG can give detailed information about the kinematic parameters of a concurrent movement. Of note, this capability of ECoG to reflect details of cortical function has been greatly underappreciated until recently [90], [97]. More recent studies have shown that different components of spoken or imagined words or their components are encoded in ECoG [106], [148] and can be used for BCI control [114]. This ready access to specific aspects of motor, cognitive, or language function possible in ECoG is likely due to the broad coverage and signal richness. At the same time, little is known about which brain functions (e.g., motor, sensory, speech, or attention), which task-related parameters (e.g., hand movement versus rest or hand velocity), which corresponding ECoG features, and what potential combination of all these factors will ultimately provide the best substrate for BCI control.

In the second part of the typical ECoG BCI protocol, the subject is trained to operate the BCI by using the chosen features to control movement of an output, i.e., in the existing studies usually a computer cursor. In most of the work to date, a linear combination of one or more of the chosen features controlled each dimension of movement. The parameters of the linear transformation [i.e., offset (intercept), gain (slope), and coefficients for the features (independent variables)] can be fixed throughout or set initially and then continually adapted on the basis of recent data to adjust for ongoing changes in the features (e.g., as done by [48] for single neuron-based control, or by [40], [46] for EEG-based control). While such feature adaptations are important for single neuron-based and EEG-based BCI movement control, they may turn out to be less important for ECoG-based movement. Single-neuron control may require adaptations to adjust for changes in the sample of neurons recorded by the microelectrodes, or in the tuning properties of individual neurons. EEG-based control usually requires adaptations to account for spontaneous changes in signals properties. In contrast, recent results [77], [126] suggest that cortical representations of function in ECoG are very stable and that spontaneous signal fluctuations are not prominent.

C. ECoG-Based BCI Studies

The initial studies of online ECoG-based BCI control using the methods described in the preceding paragraphs are promising [29], [103], [114], [118], [126], [131]–[140]. These studies have all been in humans, with the exception of [136], which used non-human primates.

Leuthardt *et al.* [29] was the first report of online ECoG-based BCI operation. In four subjects, it used different actual or imagined motor actions to chose the ECoG features to be used for online control of one-dimensional cursor movement to a target located at the bottom or top of a computer screen. Over brief training periods of only 3–24 min, and using features associated with different actual or imagined actions, the four subjects achieved online success rates of 74%–100% (with 50% expected by chance) (see Fig. 9(a) for learning curves). While the limited number of subjects and the limited number of study sessions do not permit quantitative comparisons of the performance (i.e., speed/accuracy) with that of EEG-based or single neuron-based BCIs, the acquisition of control [see Fig. 9(a)] appears to be faster than that typically associated with EEG-based BCIs. For example, Patient A imagined saying the word "move," and achieved close to 100% accuracy in less than ten minutes of training. Furthermore, offline analyses of data gathered from the same subjects while they were using a joystick to control two-dimensional cursor movement indicated that ECoG features at frequencies up to 180 Hz encode substantial information about both dimensions of movement [see Fig. 6(a)].

The online one-dimensional BCI control reported in this initial report was confirmed and extended by several other studies using similar experimental protocols. Wilson *et al.* [132] and Felton *et al.* [133] reported comparable control using closer electrode spacing (i.e., 5 mm) spacing and ECoG features associated with sensory (rather than movement) imagery. Van Steensel *et al.* [137] showed that ECoG recorded over left dorsolateral prefrontal cortex, an area involved in working

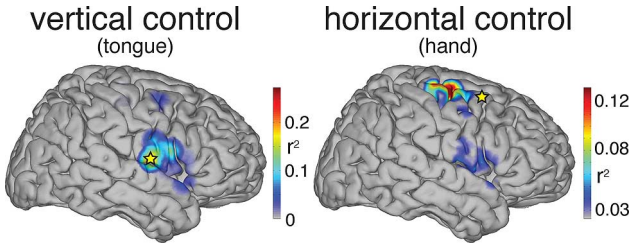

Fig. 10. Topographies of control of two-dimensional cursor movement for one subject, calculated for all locations and for the control signals provided by the ECoG features used online. These topographies show the color-coded correlation (as r^2 values) of the chosen ECoG features with vertical or horizontal movement, and thus indicate the level of task-related control of different cortical areas. This subject used imagined tongue movements for vertical control and imagined hand movements for horizontal control. Yellow stars indicate the locations used for control online. These figures suggest that selection of different locations could have yielded better online performance, in particular for horizontal control. This demonstrates that appropriate feature selection is important. Modified from [134].

memory, can also support rapid acquisition of movement control. In work with a single subject, Blakely *et al.* [126] found that an ECoG-based BCI with fixed parameters performed well over five days, which confirms the stability of ECoG features. Miller *et al.* [103] showed that motor imagery-based BCI control using locations in motor cortex can produce ECoG changes that exceed those produced by actual movements. Furthermore, in a study of potential importance for the development of practical long-term ECoG-based BCIs, Leuthardt *et al.* [131] found that an epidurally placed electrode could also support effective control (i.e., 100% accuracy using an electrode placed over premotor cortex). Finally, a recent study by Breshears *et al.* [141] showed that pediatric subjects were able to control a one-dimensional computer cursor with accuracies that were similar to those achieved by adult subjects.

Going beyond one-dimensional control, Schalk *et al.* in 2008 showed in the first and to date only two-dimensional ECoG BCI study that an ECoG-based BCI allowed five subjects to use imagined or actual motor actions to control a computer cursor in two dimensions [134]. Over a brief training period of 12–36 min [see Fig. 9(c)], each subject acquired substantial control of particular ECoG features recorded from several electrodes in a single array over one hemisphere (see Fig. 10). These features supported success rates of 53%–73% in a two-dimensional four-target center-out task in which chance accuracy was 25%. Acquisition of comparable levels of two-dimensional control using EEG typically requires substantially more training [40].

In a study of item selection (rather than movement control), Hinterberger *et al.* [135] showed that a motor imagery-based ECoG BCI could allow subjects to select characters. In this study, the subjects imagined one of two movements (e.g., moving a hand or the tongue). The BCI detected which of these two imageries the subject was attempting, and used the result in a multistep selection process to select a character. The best subject spelled one character in approximately 3 minutes. In a more recent item selection study, Brunner *et al.* [118] tested in one subject an ECoG-based matrix speller comparable to that developed for use with EEG [142], [149]. The subject achieved spelling rates (i.e., 17 characters/min (69 bits/min) sustained, 22 characters/min (113 bits/min) peak) several times higher than those reported for EEG (e.g., [150]–[155]). This ECoG-based matrix speller approach was recently further validated in two additional studies [139], [140].

In the first ECoG BCI study that used the language network, Leuthardt *et al.* [114] showed that ECoG allows for accurate discrimination of different overt and imagined phoneme articulations [68%–91% accuracy in a binary task, see Fig. 9(b)] with less than 15 minutes of training. Of importance to eventual clinical implementation of ECoG-based BCI systems, in one of the subjects, these results were achieved using recordings from a microarray consisting of 1 mm spaced microwires.

In the most recent online ECoG-based BCI study to date, Yanagisawa *et al.* [138] used ECoG to successfully decode the hand movements of a patient with chronic stroke and used the resulting commands to control a prosthetic hand in real time. With further improvements in accuracy and verification in more subjects, these results may eventually lead to accurate restoration of gripping function in people with stroke.

Rouse and Moran in 2009 [136] is to date the only online ECoG-based BCI study in monkeys. In this preliminary study with one monkey, ECoG features were used to control two different two-dimensional tasks: reaching and circle-drawing. For online control in either task, the authors used 65–100-Hz gamma activity in ECoG recorded from two arbitrarily selected epidural electrodes over the M1 area of motor cortex (i.e., primary motor cortex). The authors assigned gamma activity recorded from each of the two electrodes to horizontal or vertical cursor movement control, respectively. Over the course of five recording days, the subject was able to use gamma activity to achieve control of a cursor to successfully perform center-out reaching tasks as well as circle drawing tasks. This study also suggested the gamma frequency band that resulted in the best control.

In summary, the human and animal ECoG-based BCI studies to date show that ECoG recorded from different locations and in different experimental paradigms can support impressive BCI performance. ECoG might provide performance that exceeds that of EEG and requires substantially less training to acquire. This probable advantage may be due largely to the ability of ECoG to record high-frequency (i.e., gamma activity), which is minimal or entirely absent in EEG. Furthermore, ECoG may provide movement-related information comparable to that found in single-neuron activity [90] and could prove more stable for long-term use [77].

IV. Limitations

The studies reviewed thus far strongly support the value of ECoG BCI research and development. While these studies are encouraging, there are some important limitations to the current techniques that merit attention.

The most important limitation of current (mostly human) ECoG-based research efforts is related to the variability and constraints implied by the subject population. Current human ECoG-based research is almost exclusively limited to patients who have been temporarily implanted (usually for about one week) with an ECoG array in order to localize a seizure focus and essential cortical functions prior to surgical resection for treatment of their epilepsy. After implantation and post-operative recovery (about 1–2 days), patients are generally available

for research for only a few hours per day at best [due to appointments for imaging, clinical tests (in particular electrical cortical mapping of function), medication regimens (e.g., pain medication), and personal visits]. In addition, patients vary considerably in cognitive capability, level of interest in participating, and clinical status (e.g., seizures, pain, nausea, medications). Furthermore, the ECoG recordings are generally performed in a hospital room that has severe space constraints and may have considerable environmental noise [both auditory and electromagnetic (e.g., from electrical beds, pressurized stockings, or automated drug delivery systems)] that may be difficult or impossible to reduce. In order to be successful in these studies, personnel needs to be highly trained and efficient and should be ready to run experiments whenever the opportunity arises. Furthermore, experimental hardware and software needs to be streamlined, robust, and always available. Additionally, the researchers must have a sensitivity to the needs of the patients and a respect for their willingness to participate during the stressful clinical experience. A sensitivity to their needs, a respect for their willingness to participate, and hence particular social qualities of the experimenters, are paramount.

Another limitation is related to ECoG's lower spatial resolution compared to single-neuron recordings. ECoG's spatial resolution has been estimated to be around 1 mm (1.25 mm for subdural recordings [63] and 1.4 mm for epidural recordings [65]). The spatial resolution of single-neuron recordings is approximately one order of magnitude higher than that. Thus, it is unlikely that surface cortical recording can support the detection of the firing of individual neurons. On the other hand, ECoG's larger coverage and ability to detect different physiological processes [64], [114] appears to compensate for the lower spatial resolution [90], [97].

Ultimately, any ECoG BCI will always require an invasive procedure. Thus, there will always be some surgical risk associated with its implementation. For medical and perhaps for military applications of BCI technology, this issue may not substantially affect adoption and utilization. Furthermore, it is possible or even likely that invasive BCI procedures such as ECoG or single-neuron recordings will eventually become as safe as many other invasive medical procedures. Should the risk profile become low enough, such that it is comparable to other elective surgical procedures (e.g., cosmetic surgery), wider adoption for applications beyond medical restoration (e.g., gaming, attentional augmentation, brain-derived environmental control for normal individuals) may become a realistic possibility.

V. Important Questions and Areas for Further Research

Research and interest into ECoG-based BCIs has grown substantially over the past decade. The emerging science of surface-cortical physiology and its application for brain-derived control offers substantial promise for creating neuroprosthetic solutions for people with severe motor disabilities. Work so far, however, has barely begun to address additional important questions that need to be answered if that promise is to be fulfilled. Scientific and technical questions that still need to be addressed include: the best recording locations; the best ECoG features (e.g., broadband gamma versus gamma subband versus mu/beta versus LMP); the best recording method (subdural/epidural/skull screws); the best electrode diameter and density (i.e., inter-electrode distance); the best kinds of actual or imagined action (e.g., movements, sensations, speech, other cognitive functions); the maximum number of degrees of freedom that ECoG can support for online control or offline decoding (i.e., [134] demonstrated online two-dimensional BCI control in humans; [77] demonstrated offline decoding of seven degrees of freedom in monkeys); the best array designs for long-term biological impact and functional stability; and realization of wholly implantable systems. These optimizations would greatly benefit from conceptual frameworks that could be used to formally assess and optimize BCI performance given these different factors.

Just as for BCIs that use other signals, online studies are essential for developing optimal solutions. While many of these issues can be explored through offline analyses, online testing is necessary to establish the validity of their results. Thus far, human studies have largely been confined to relatively short-term studies in people implanted temporarily for other purposes. Fortunately, much of this work, particularly long-term studies, can take place in animals (mainly monkeys and rats). Indeed, for many of these issues, animal studies will be essential to justify and guide further human studies. Animal studies are likely to significantly contribute to the physiological bases of the different kinds of frequency-domain and time-domain features of ECoG (e.g., mu/beta, gamma, LMP, phase-amplitude coupling), which may enable informed choices for the configuration and implementation of ECoG-based BCI systems. Ultimately, however, chronic human trials will be essential to demonstrate the definitive clinical benefit of an ECoG-based platform.

The current generation of ECoG implants, in particular those implanted prior to surgical resections of a lesion (e.g., for epilepsy or tumors), are neither optimized nor even suitable for long-term BCI operation. The implant design (e.g., materials, electrode spacing) is generally determined purely by the clinical need to localize a seizure onset zone and functional regions of the brain. Suited only for relatively short-term use, these arrays are usually placed subdurally, cover areas up to 8×8 cm (thereby requiring a sizeable craniotomy), and have a percutaneous tethered connection to an external data acquisition system. This placement and the percutaneous connection increase the risk for infection and for epidural or subdural hematomas. In contrast, the design and clinical implementation of ECoG-based BCI systems suitable for long-term use would be quite small, wholly implantable, and may use arrays that cover relatively small areas of cortex. The work needed to develop the complete implants and establish their safety and effectiveness, first in animals and then in humans, has just begun.

Most components necessary for implementation of a chronic ECoG-based neuroprosthetic exist but have not yet received regulatory approval for human use. These components include ECoG implants that implement passive recording structures (see [129], [156], and Fig. 11(A) and (B)) or even active electronics on biocompatible substrates [130]. Such implants can often be used for recording and even stimulation [157].

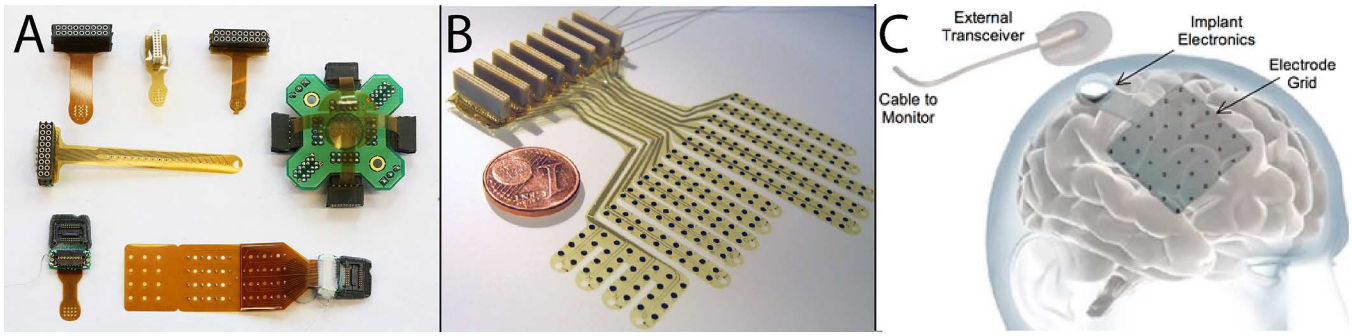

Fig. 11. Emerging generation of ECoG recording devices. A: Thin film-based ECoG devices and their connectors for recordings in different species. Picture courtesy of Dr. Justin Williams. B: ECoG recording device for high channel number recordings in a monkey. From [156]. C: Proposed ECoG grid with wireless interface. Courtesy of Ripple, Inc.

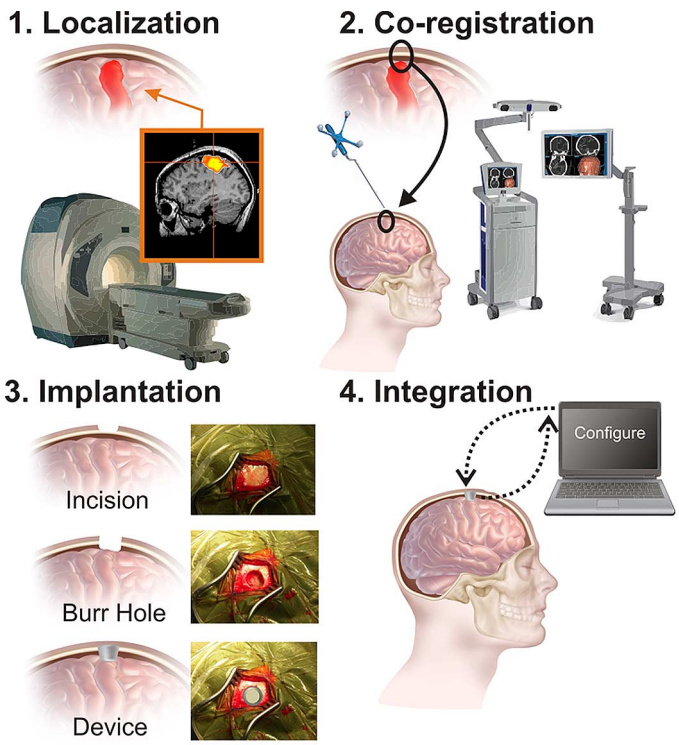

Fig. 12. Proposed clinical implementation of ECoG technology. Schematic illustration shows the multiple stages that will likely be required for the evaluation and application of an ECoG BCI in a human study. The first step will involve a preoperative localization of the functional region in question. The second step will require an anatomic coregistration of this functional region with the actual physical anatomy of the patient. The third step will involve the implantation of the device. The envisioned technology will be small and minimally invasive requiring only a small burr hole in the skull. See schematic diagrams on the left and actual surgical photographs on the right. Finally, after implantation, the device will have to configured to integrate with the user's various needs.

These implants could be connected to amplification/digitization devices [158] and/or wireless transmission units [159], [160] [Fig. 11(c)]. With the addition of a battery (which could be implanted in areas distant from the ECoG implant—in the chest, for instance), these components could be combined into a fully and permanently implantable system and validated in animal and subsequent human studies.

Though this still remains to be tested, implantable ECoG-based BCI systems may have some notable advantages over implantable intracortical (in particular single unit-based) BCI systems. First, because ECoG arrays can readily record from larger areas of cortex compared to microelectrode implants, ECoG-based systems may provide more practical comprehensive access to the cortical networks involved in various different cognitive operations. That said, minimizing an ECoG implant's invasiveness may require minimizing the size of the implant, thereby obviating this distinction. Second, the power demands of ECoG recordings are likely to be much more modest than those of single-neuron recording. This is an important consideration for wholly implantable systems. Action potential recording requires a digitization rate of $>$ 10 kHz per channel. With high numbers of channels, this requirement is difficult to satisfy in implanted low-power wireless transmission systems that do not generate undue heat. In contrast, ECoG requires a digitization of only 500–1000 Hz per channel, more than an order of magnitude less.[1] Furthermore, if an ECoG-BCI relies on gamma activity and an appropriate analog filter is employed, the digitization rate might be as low as 50 Hz per channel [161], more than two orders of magnitude less than action potential recording. In this case, a 1000-channel ECoG array would require a total digitization rate of only 50 kHz, the same rate required by only 3–5 microelectrodes. While the requisite clinically approved devices are still in development, fully implantable devices capable of this digitization rate could readily be implemented using current technologies. Thus, given the probable long-term stability of ECoG recordings [77], [78], appropriate modification or extension of current technologies could lead to wireless ECoG implants that could transmit ongoing brain activity from many thousands of cortical sites and could deliver robust signals over many years.

VI. Proposed Clinical Implementation of ECoG-Based BCIs

A chronically implanted ECoG-based BCI system would consist of either a subdural or epidural array that includes amplification/digitization/wireless electronics, is powered by a battery at a remote site (e.g., in the chest), and is permanently implanted through a small (e.g., 19-mm) burr hole in the skull. We envision that such ECoG-based systems would be implemented in a series of four steps that proceed from

[1]It is important to note that local field potential recordings from microelectrodes can also detect gamma activity and can also be satisfactorily recorded at sampling rates similar to ECoG.

functional localization to coregistration, to implantation, and to integration (see simplified illustration in Fig. 12).

The purpose of the first step, functional localization, is the identification of those cortical areas that represent the best substrate for BCI control and thus will identify the target location for subsequent grid implantation. The procedure for this localization may be similar to the first component of current BCI protocols (Section III-C) and may be realized using functional magnetic resonance imaging (fMRI).

The purpose of the second step, coregistration, is to relate the target location that was identified by the first step [which is defined in some coordinate system relevant to the imaging system used (e.g., Talairach coordinates)] to the physical position on the person's brain. This step can readily be achieved using conventional stereotactic navigation systems.

The purpose of the third step, implantation, is to place the ECoG sensing/transmission device over the identified location and to secure it to the skull. This procedure may also entail placement of a battery at a remote site and installation of related cabling.

The purpose of the fourth step, integration, is to configure the BCI system such that it properly identifies and detects relevant brain signals and relates them to the output function desired by the user.

VII. Summary

ECoG is generating strong and growing excitement for its potential to support basic neuroscientific investigations and powerful and clinically practical BCI systems. This interest is driven by several advantageous characteristics of ECoG recordings, as well as by the growing recognition of the limitations of existing noninvasive and invasive signal modalities. ECoG has greater amplitude, higher topographical resolution, and a much broader frequency range than scalp-recorded EEG, and is also less susceptible to artifacts. At the same time, ECoG is likely to have, and will likely continue to have, greater long-term stability than do intracortically recorded signals. Additionally, the technical requirements for ECoG-based systems are much lower than those for intracortical systems; thus, they should be more amenable to chronic implantation.

ECoG detects a number of physiological phenomena that are represented in different time- or frequency-domain components and their interactions. This includes activity in the mu and beta bands, which are related to general aspects of movements or cognition, and can also be detected in scalp-recorded EEG. Presumably more important for BCI function, it also detects gamma activity at higher frequencies, which show much greater functional and anatomical specificity than signals in the mu and beta bands, and can also not readily be detected by EEG. At the same time, the differential value of these and other ECoG phenomena (LMP, phase-amplitude coupling, etc.) for the BCI purpose has not yet been established.

To date, ECoG-based BCI studies have been limited almost exclusively to people that were temporarily implanted with ECoG recording arrays prior to surgery. Despite the many practical difficulties of such studies, the results are promising. They suggest that ECoG-based BCIs might provide control comparable or even superior to that reported for EEG-based BCIs. These results, combined with the likely practical and robustness advantages of ECoG, are encouraging further efforts to develop ECoG-based BCI systems. Scientific issues of particular importance include the determination of the best cortical systems (motor, sensory, language, attention, etc.), the best recording methods (epidural versus subdural, cortical location, and electrode spacing), the optimal features (mu, beta, gamma, LMP), and the most effective algorithm designs.

Ultimately, ECoG-based BCI systems suitable for chronic use must be wholly implantable and capable of performing reliably for many years. While such systems have not yet been developed, the individual components that would comprise them do exist or are under development. The extensive work needed to develop the complete systems and to validate them first in animals and then in humans has just begun. Its successful completion, combined with resolution of the other issues summarized above, could lead to ECoG-based BCI systems of great value to people with disabilities.

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