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Functional near-infrared spectroscopy for neuroimaging in cochlear implant recipients

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ABSTRACT

Functional neuroimaging can provide insight into the neurobiological factors that contribute to the variations in individual hearing outcomes following cochlear implantation. To date, measuring neural activity within the auditory cortex of cochlear implant (CI) recipients has been challenging, primarily because the use of traditional neuroimaging techniques is limited in people with CIs. Functional near-infrared spectroscopy (fNIRS) is an emerging technology that offers benefits in this population because it is non-invasive, compatible with CI devices, and not subject to electrical artifacts. However, there are important considerations to be made when using fNIRS to maximize the signal to noise ratio and to best identify meaningful cortical responses. This review considers these issues, the current data, and future directions for using fNIRS as a clinical application in individuals with CIs.

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Contents

Abbreviations: BOLD, blood-oxygen level dependent; CI, cochlear implant; CW, continuous wave; EEG, electroencephalography; FD, frequency-domain; fMRI, functional magnetic resonance imaging; fNIRS, functional near-infrared spectroscopy; HbO, Oxygenated hemoglobin; HbR, Deoxygenated hemoglobin; MEG, magnetoencephalography; NIR, near-infrared; PET, positron emission tomography; TD, time-domain; SNR, signal-to-noise ratio; SRT, speech reception threshold

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1. Introduction

Cochlear implants (CI) have restored hearing to over 90,000 individuals in the United States in the past 30 years (FDA, 2015). Significant advances in speech processor design, signal processing and surgical techniques have resulted in progressively enhanced performance (Rubinstein, 2004; Roland et al., 2006; Srinivasan et al., 2013). As a result, cochlear implantation has become a highly successful prosthetic solution to replace the function of a sensory organ. Intervention with deaf children has been particularly successful: many children who would otherwise have been placed in schools for the deaf and taught sign language are now learning alongside mainstream peers in a regular classroom environment. The primary goal of cochlear implantation is now open-set auditory-only speech understanding in everyday listening environments. However, while the majority of implant recipients achieve this goal, many still perform poorly (Lazard et al., 2012; Miyamoto et al., 1994).

The factors that contribute to the wide variations in individual outcomes following cochlear implantation are diverse and not completely understood (Lazard et al., 2014; Peterson et al., 2010). Numerous reports have identified age of implantation as a strong predictor of better CI outcome (e.g., the younger, the better) (Kirk et al., 2002; Nikolopoulos et al., 1999; Robinshaw, 1995). Investigators have also demonstrated that children who communicate orally achieve better speech perception skills than children who use visual sign communication (Osberger and Fisher, 2000; Geers et al., 2003). Finally, family income predicted language outcomes in pediatric CI recipients (Holt and Svirsky, 2008). In order to more fully understand how such neurobiological, cognitive, and societal factors influence language outcomes post-implantation, it may be beneficial to examine the neural processing during the perception of auditory stimuli through a cochlear implant. Together with behavioral measures, neurophysiological indicators have the potential to guide post-implant programming in support of deaf patients' speech and language outcomes and, eventually, even predict results for an individual CI patient before implantation occurs.

Functional near-infrared spectroscopy has already been shown to be a reliable neuroimaging modality in both adult and pediatric populations (Fava et al., 2014a,b; Giraud et al., 2001; Quaresima et al., 2012; Wilcox et al., 2005). Generally, reviews of this literature have focused on the use of fNIRS in research on language development and language processing in healthy populations (Crosson et al., 2010; Elwell and C. E. Cooper, 2011; Gervain et al., 2011; Lloyd-Fox et al., 2010; Quaresima et al., 2012; Fava et al., 2011, 2014a,b; Wilcox et al., 2005). More recently, an emerging body of reviews addresses the imaging instrumentation and methodology, as well as approaches to statistical analysis of fNIRS data (Bandettini, 2009; Piper et al., 2014; Scholkmann et al., 2014; Tak and Ye, 2014). However, most relevant to CI research is the fact that fNIRS is compatible with these devices. This review explores applications and limitations of fNIRS in the CI population, comparing it with traditional neuroimaging methods. We summarize the existing literature on the use of fNIRS in adult and pediatric CI recipients, and conclude by outlining possible directions for future research and clinical applications using this promising imaging technique in the CI population.

2. Neuroimaging options in cochlear implant users

Because auditory perception occurs within and beyond the auditory cortex, neuroimaging has the potential to provide an additional clinical measure for assessing whether the electrical stimulation of the cochlea by the CI is reaching and stimulating auditory-specific cortical regions of the brain similar to normal-hearing subjects (Pasley et al., 2012; Steinschneider et al., 2014). Such information can supplement behavioral tests, which are often limited in young CI users (Choi and Oghalai, 2005; Katzenstein et al., 2009; Lin et al., 2010; Oghalai et al., 2009; Santa Maria and Oghalai, 2014; Williamson et al., 2009; Ying et al., 2013). However, there are inherent limitations in the use of all of the currently available neuroimaging modalities in CI recipients, as outlined below and summarized in Table 1.

Functional neuroimaging attempts to identify the brain systems responsible for different behaviors by comparing brain activity during contrasting states (Aine, 1995; Crosson et al., 2010). The logic is that neurons in different areas of the brain associated with specific cognitive processing tasks generate electrical signals when they are active. As a result of this activation, the metabolic needs of neurons change: increased oxygen demand results in increased cerebral blood flow and thus oxygen delivery to that area, with a consequent decrease in deoxygenated hemoglobin (HbR) (Babiloni et al., 2009). Certain neuroimaging modalities, such as EEG, measure this neural activation directly by recording the average electric field potential at different regions of the scalp. In contrast, metabolic neuroimaging methods, such as fMRI, PET, and fNIRS, are indirect, surrogate measures of neuronal activity (Castañeda-Villa

Table 1
Characteristics of the functional neuroimaging techniques currently available for research involving cochlear implant users. See explanations in text, Sections 2.1–2.3.

Technique	Spatial resolution	Temporal resolution	Cochlear implant compatibility	Flexibility in auditory stimuli paradigm	Potential for use in infants	Comments
fNIRS	+ + +	+ + +	Yes	Yes	Yes	
fMRI	+ + + + +	+ +	No*	No**	No	* Structural imaging possible ** Loud background noise
PET	+ + + +	+	Yes	No*	No	* Limited to block design paradigms
EEG	+	+ + + + +	Yes	No*	Yes	*Limited to sound bursts/clicks
MEG	+ +	+ + + + +	No*	No**	Yes	* Requires use of magnet-less implant and simultaneous radio frequency head shield ** Limited to sound bursts/clicks

fNIRS: functional near-infrared spectroscopy, fMRI: functional magnetic resonance imaging, PET: position emission tomography, EEG: electroencephalography, MEG: magnetoencephalography.

et al., 2012; Girouard, 2006; Levitin and Menon, 2005; Mc Laughlin et al., 2013).

Although functional neuroimaging technologies have the potential to provide insight into the cortical changes that take place in patients with cochlear implants, obtaining meaningful measurements of cortical responses in CI recipients has proven challenging. This is primarily because the traditional imaging methods have limitations when used in implanted patients, and so alternative neuroimaging strategies have been sought. In this context, functional near-infrared spectroscopy (fNIRS) has been a welcome addition to a limited choice of neuroimaging modalities suitable for use in CI recipients. Here we outline the primary techniques and assess their appropriateness for use in combination with CIs. Because it is important understand the benefits and downsides to each technique when selecting an imaging modality, we briefly review several commonly-used techniques including fMRI, PET, EEG, and MEG, before moving on to an in depth explanation of fNIRS.

2.1. Functional MRI

Functional MRI provides high spatial resolution and is often the neuroimaging technology of choice in unimplanted subjects. However, conventional CIs are incompatible with fMRI for several reasons. The primary reason is that CIs contain internal magnets and ferromagnetic components, including a coil used to transcutaneously relay data from the external processor to the surgically implanted components (Doucet et al., 2006; Gilley et al., 2008; Majdani et al., 2008). Such ferromagnetic implants exposed to electromagnetic fields or radiofrequency energy may heat, induce a current, or become dislocated (Azadarmaki et al., 2014; Portnoy and Mattucci, 1991; Teissl et al., 1999). Thus, the most important concern in using fMRI to study a subject with a CI is patient safety. Furthermore, the magnet and coil interact with the electromagnetic fields found in MRI scanners, producing interference that can disturb data transfer, and malfunction of the implant can occur due to demagnetization of the CI internal magnet via the imaging magnet (Majdani et al., 2008; Ponton et al., 2000). Finally, CIs produce considerable artifacts on the MR image, obscuring cortical regions proximal to the internal magnet (Majdani et al., 2009). Thus, these signal-void areas can compromise accurate diagnosis of certain medical conditions when used for medical imaging and make it nearly impossible to measure activity within the ipsilateral temporal lobe when used for functional imaging.

In response to these limitations, certain manufacturers have designed CIs with removable internal magnets. Unfortunately, large artifacts often remain on the MRI even after the internal magnet is removed (Risi et al., 2004). Other models of CI have MRI-conditional internal magnets that do not need to be removed prior to scanning. Regardless of the status of the internal magnet, the external processors for all CI devices are MRI unsafe (Azadarmaki et al., 2014) and the radiofrequency fields generated by the MRI interfere with the transcutaneous radiofrequency link between the external and internal coils (Lazeyras et al., 2002; Seghier et al., 2005). Auditory stimulation by the implant during imaging is therefore generally precluded, though anatomical images can be acquired for medical purposes (Baumgartner et al., 2001; Crane et al., 2010; Gubbels and McMenomey, 2006).

The limitations of using fMRI with the CI population extend beyond equipment incompatibility issues. MRI is subject to movement artifacts (Quaresima et al., 2012), requiring subjects to remain completely still and to avoid overt vocalizations while in the scanner. In infants, this translates into the need for restraints and even sedation and/or anesthesia. Sedatives and anesthetics, of course, alter brain activity and therefore change cortical responses to auditory stimuli (Marcar et al., 2006). Such circumstances considerably restrict the use of fMRI in this age group.

It is also important to consider that fMRI is a noisy imaging modality, which introduces a potential confounding effect as the background noise cannot be matched between deaf and hearing participants (Dewey and Hartley, 2015). Moreover, the acoustic noise associated with fMRI creates an intrusive testing environment for younger children and disturbs the presentation of auditory stimuli relevant to CI users (Gervain et al., 2011). Finally, the BOLD (Blood Oxygenation Level Dependent) signals obtained using fMRI relate to changes in HbR only and do not directly convey information about HbO.

2.2. PET scan

Nuclear functional imaging techniques such as PET scans have more frequently been used in studies involving CI users. Previous investigators employed PET scans to examine various auditory cognitive processes in the CI population (Limb et al., 2010; Naito et al., 2000; Wong et al., 1999), and several dedicated reports have even been published for reviewing the use of PET scans in language processing research on CI recipients (Aggarwal and Green, 2012; Giraud et al., 2001). Several factors account for the popularity of this neuroimaging modality for use with CIs among the scientific community. First, PET is fully compatible with CIs. It also has good spatial resolution and, as with MRI, it can image activity in deep, subcortical structures (Bandettini, 2009). Because PET is a relatively quiet imaging modality, it is suitable for studies involving auditory stimuli. Finally, it is tolerant to subtle subject movements thanks to rapid image acquisition times, a significant advantage over fMRI (Crosson et al., 2010).

The significant drawback of using this imaging modality is the exposure of the research subjects to radiation and the necessary limitation in the number of scans that this implies. The radioactive tracers or carrier substances need to be injected into the blood stream, which many subjects find aversive. For these reasons, PET is rarely used in research studies involving children. Though understandable, this is unfortunate because children are a demographically important age group within the CI population. The use of PET to study neuroplasticity post-implantation is also ethically challenging, as measuring such changes would require sequential longitudinal testing in the same subject (Giraud et al., 2001). Limited temporal resolution, or the accuracy on a temporal scale with which a neural event can be characterized (Crosson et al., 2010), is another shortcoming of PET. This is because PET's ability to resolve neural events is on the order of tens of seconds compared to only a few seconds for fMRI (Bandettini, 2009). Such limited temporal resolution requires averaging over long blocks of events; higher sampling rates are generally preferred in functional studies because they allow the use of event-related paradigms, which offer greater flexibility and more precision in experimental inquiry (Aine, 1995).

2.3. EEG and MEG

Unlike fMRI and PET, EEG and MEG directly measure the electrophysiological response of neural activation. The resulting advantage of this technique is an unrivaled temporal resolution in the sub-millisecond range (Babiloni et al., 2009), however at the expense of spatial resolution (Posner and Levitin, 1997). Studies have shown that auditory evoked potentials recorded in EEG provide a useful objective metric of performance in CI patients (Castañeda-Villa et al., 2012; Mc Laughlin et al., 2013). It is therefore not surprising that the EEG literature in CI users is abundant and, indeed, has greatly contributed to the understanding of auditory processing in this population (Sandmann et al., 2010; Zhang et al., 2010). In addition, the combination of the high temporal resolution and an excellent safety profile make EEG and MEG ideally suited for follow-up studies requiring several successive assessments, such as those investigating cortical plasticity following implantation (Doucet et al., 2006; Gilley et al., 2008). Finally, EEG is tolerant to subtle movements and can even be used with fully awake infants.

On the other hand, as mentioned, EEG and MEG offer relatively poor spatial resolution due to the inverse Poisson problem: the location of activity within a sphere is ambiguous when measuring from the surface of that sphere (Posner and Levitin, 1997). While the reconstruction of brain responses to specific cortical regions is possible (Ferree et al., 2001; Song et al., 2015), the accuracy of this localization remains inferior to other modalities such as fMRI or PET (Ponton et al., 2000). Data corruption by the electrical components of the implant is another major limiting factor for the use of EEG in combination with CIs. To minimize the electrical artifacts produced in EEG recordings, only short auditory stimuli such as tone bursts or clicks can be employed in CI studies, which significantly limits the flexibility of the experimental paradigm (Gilley et al., 2008). Despite the various techniques that have been described to filter this artifact, the interpretation of auditory evoked potentials in EEG remains challenging (Mc Laughlin et al., 2013; Sandmann et al., 2009). Additionally, MEG measures very weak magnetic fields that can only be recorded in magnetically shielded rooms equipped with detectors that are highly sensitive to minute changes in magnetic signals (Crosson et al., 2010). Similar to fMRI, MEG instrumentation interacts with the internal magnet of most CI models, precluding any useful recording. To successfully monitor neural activity in CI users using MEG, certain conditions must be fulfilled. This unique experimental setup is described by Pantev et al. (2006), who reported the only MEG study involving CI users. The basis for the methodological success of this study is twofold. First, the two participants enrolled were recipients of Clarion (Advanced Bionics, Valencia, CA) magnet-less implants – now withdrawn from the market. Second, a unique radio frequency shield was applied between the head of the patients and the MEG device, preventing interference from radio frequency signals transmitted by the CI. Such setups, however, are very rare and extremely costly.

3. fNIRS

Before fNIRS was adapted for use in people with CIs, PET was reported to be the only technique suitable for measuring brain responses in the CI population for all of the reasons outlined above (Giraud et al., 2001; Truy, 1999). Because the concepts, features, and instrumentation of fNIRS have been described in substantial detail in previous reports (Elwell and C. E. Cooper, 2011; Gervain et al., 2011; Lloyd-Fox et al., 2010; Quaresima et al., 2012), we will only briefly address them in this review. Here we focus primarily on the characteristics of fNIRS that are relevant to its use with the CI population.

3.1. General principles

fNIRS is an optical imaging technique: it uses near-infrared (NIR) light to detect changes in cerebral blood flow as a proxy for neural activation. When a beam of light is directed onto tissue, three factors can interfere with its undisturbed propagation (i.e. transmission) through it: reflection/refraction, absorption and scattering (Niemz, 2002). The contribution of reflection/refraction can essentially be ignored in opaque media such as the skull. The intensity of the transmitted light therefore depends on the amount of non-absorbed and non-scattered photons(Welch and van Gemert, 2011; Gervain et al., 2011). Biological tissues preferentially absorb light in the visible spectrum, while being relatively transparent to light in the NIR wavelengths (650–1000 nm) (Smith, 2011). As a result, NIR light can penetrate through superficial biological layers, enabling sampling of deeper tissue structures. For neuroimaging, this means that fNIRS can effectively probe the surface of an adult brain to a depth of up to 1.5 cm (Elwell and C. E. Cooper, 2011).

fNIRS is capable of measuring changes in cerebral blood flow because hemoglobin is the main pigmented molecule in human tissues that is present in clinically significant quantities to exhibit oxygenation-dependent absorption of light in the NIR spectrum (Delpy and Cope, 1997). In tissues, hemoglobin exists in an oxidized (oxygenated hemoglobin, HbO) and reduced (HbR) form, each characterized by a unique absorption spectrum. The aim of NIRS neuroimaging is to quantify the concentrations of these two hemoglobin chromophores in the tissues traversed by NIR light. This is possible using the Beer–Lambert Law, an equation that describes the light absorbance (A) at a given wavelength (λ) in a medium (Crosson et al., 2010):

$$A = -\log\left(\frac{I}{I_o}\right) = c \cdot \varepsilon_\lambda \cdot l$$

Shining light of an appropriate wavelength at a given intensity (incident light, I) on the head, and measuring the intensity of the light that leaves the tissues (transmitted light, I_o) allows for the calculation of the concentration of the medium, "c" (i.e. the concentration of HbR, HbO and total hemoglobin). This concept assumes that the molar extinction coefficient of the medium at that specific wavelenght (ε_λ) and the optical pathlength "l" in the tissues (the path the light travels between the source and the detector) are known.

The application of this physical principle forms the basis of fNIRS neuroimaging. Of course, other factors need to be considered. Light scattering caused by skin, hair and skull, also contributes to light attenuation in tissues, resulting in an unknown light loss that needs to be accounted for (Delpy and Cope, 1997). Furthermore, light does not travel through biological tissue in a straight line. The Beer–Lambert Law was therefore modified to take into account the scatter and the non-linear trajectory of light in tissues, referred to as the differential pathlength factor (Cope et al., 1988). These two factors cannot be measured directly using continuous-wave NIRS systems (see below), therefore only changes in HbO and HbR concentrations, as opposed to absolute values, can be obtained. A detailed description of the mathematical model underlying light absorption in scattering media can be found elsewhere (Gervain et al., 2011; Hoshi, 2003; Sassaroli and Fantini, 2004).

Practically speaking, fNIRS is performed on human subjects by placing a light source and a light detector adjacent to each other above the brain area to be measured. This source-detector pair is called a channel. A convex banana-shaped tissue region is sampled, corresponding to the light path through the tissue between the source and detector. The depth of penetration of the NIR light in brain tissue is approximately half of the source-detector distance. To reach a clinically relevant depth of cortical area, the source-detector distance should be 2–3 cm in infants and 3–5 cm in adults (Quaresima et al., 2012). The choice of the wavelength pair is also important, as it affects the quality of the fNIRS signals. Ideally, one wavelength should be sensitive to HbO; the other to HbR. This is possible because HbO and HbR demonstrate differential absorption in the NIR spectral range (except at the isosbestic point, where the extinction coefficients of these two chromophores are equal). Generally, wavelengths below the isosbestic point are used to measure HbR responses (below 760–770 nm), whereas longer wavelengths are more sensitive to HbO (up to 920 nm) (Boas et al., 2004). Theoretical models also revealed that the highest signal-to-noise ratios were obtained if one wavelength was below 720 nm, and the other higher than 730 nm (Uludag et al., 2004). The 690 nm and 830 nm pair is commonly reported in fNIRS literature, but a variety of other systems capitalizing on different wavelength contrasts are commercially available (Lloyd-Fox et al., 2010).

Three different fNIRS instrumentation techniques are currently available, and they vary in the type of illumination employed (Ferrari and Quaresima, 2012). The first modality, continuous wave (CW) light, is the most commonly used and the least costly. It is based on constant tissue illumination and simply measures changes in light attenuation as it passes through the head. This technique does not allow calculation of light scattering or optical path length in tissues and, as a result, can only determine relative changes in HbO, HbR and total hemoglobin concentrations (Scholkmann et al., 2014). However, relative values of hemodynamic parameters are usually sufficient in functional brain studies. The last two techniques, time-domain (TD) and frequency-domain (FD), are equivalent in that they both measure the time needed by light to travel through tissues (i.e. time of flight) to determine optical path length (Wolf et al., 2007). They differ in their approach to time of flight measurements, and in the resulting instrumentation that this implies. TD systems emit extremely short pulses of light into tissue, and directly measure the arrival times of the scattered photons that emerge (Torricelli et al., 2014). Such recordings require very sensitive photon-counting detectors. The time of flight multiplied by the speed of light in the tissue provides optical path length. In contrast, FD technique uses intensity-modulated light to illuminate the brain at very high frequencies, and measures both the attenuation and the phase delay of the emerging light (Wolf et al., 2007). Time of flight is then obtained by Fourier analysis of the phase delay, and can be used to calculate optical path length. The resulting advantage of TD and FD imaging is that knowledge of optical path length allows calculation of absolute values of HbO, HbR and total hemoglobin concentrations. On the other hand, such systems are associated with higher costs, bulky instrumentation, and slower acquisition times. The characteristics of the different fNIRS technique have been described in much greater detail in recent reviews (Wolf et al., 2007; Scholkmann et al., 2014; Torricelli et al., 2014).

3.2. Advantages, limitations and considerations for using fNIRS with CIs

Compared to other techniques, fNIRS has several clear advantages that encourage its use in CI research. One of its most appealing features is its full compatibility with CI devices. Owing to the optical nature of the technology, fNIRS data are not corrupted by the electronic or ferromagnetic components of the CI device during acquisition. PET is the only other neuroimaging modality that provides a matching level of compatibility. However, unlike PET, fNIRS does not require injection of tracer substances in the blood stream and does not expose individuals to radiation. The number of examinations is therefore not restricted, and repeat assessments through longitudinal studies can be performed. fNIRS is also ideally suited for research involving young infants. Measurements can be recorded without the need for sedation or restraints because it is robust to motion artifacts. In fact, recording during overt speech in even possible (Hull et al., 2009; Quaresima et al., 2012). This is of great significance for CI investigators, as a large field of CI research involves the pediatric population.

Good research tools are safe, but also practical. To carry NIR light, fNIRS uses optic fibers that are light, flexible, and therefore suitable for a range of head positions and postures. Some centers replaced the plastic optic fibers with glass optic fibers and have reported reduced weight of the optic bundles on the headgear (Lloyd-Fox et al., 2010). Furthermore, fNIRS requires only a compact measurement system. The setup typically consists of a mobile cart carrying a computer tower and monitor, an optical NIRS module and the optical fibers connected to that module. This increases portability and allows for measurements in non-intrusive environments and even in clinical settings. PET scans, on the other hand, can only be performed in a radiation-proof radiological suite and require the presence of a radiochemist and a cyclotron for the production of radioisotopes (Crosson et al., 2010). Advances in optical technology have even allowed the production of a wireless, completely wearable, multi-channel fNIRS system suitable for use in unrestrained settings (Piper et al., 2014). Cost is another important factor to consider when choosing a research instrument. fNIRS is among the most affordable neuroimaging modalities, after EEG. There are no disposables and minimal maintenance is required. In comparison, the instrumentation and maintenance fees associated with MRI, PET and MEG are on the order of millions of dollars (Bandettini, 2009).

The temporal resolution of fNIRS is the highest among the hemodynamic neuroimaging techniques, reaching up to 100 Hertz (Hz) with CW systems (Huppert et al., 2006). Although inferior to EEG and MEG by one order of magnitude, this fine temporal resolution allows the use of event-related paradigms and allows for nuanced examination of the temporal dynamics of cortical blood flow. The spatial resolution of optical topography is typically estimated at 1 cm (Ferrari and Quaresima, 2012), enabling the localization of brain responses to specific cortical regions with reasonable precision. The spatial resolution is dependent on the arrangement of source-detector fibers on the scalp. Increasing the density of channels, among other things, achieves finer sampling of the cortex (Minagawa-Kawai et al., 2008). At our institution, we transitioned from a four channel system to a 140 channel system, allowing us to generate topographic activation maps of the auditory

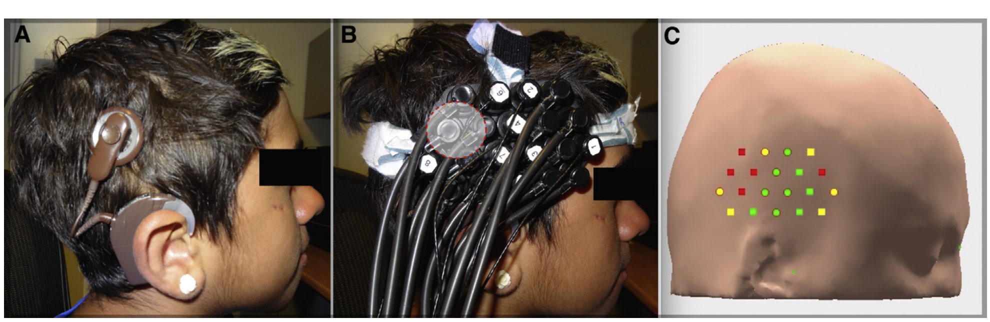

Fig. 1. fNIRS headset placement over a cochlear implant device. A) The location of the cochlear implant's external magnet and coil interferes with headset placement over the temporal area. B) The fNIRS headset is simply apposed over the magnet (shaded area). C) Diagrammatic representation depicting the quality of scalp contact of the optode array, obtained from custom analytic software using real-time fNIRS recordings. The optodes obstructed by the magnet postero-superiorly lose their scalp contact (red), while the remaining optodes are unaffected and can still be used (green). The status of scalp contact was indeterminate for certain optodes (yellow).

cortex (Pollonini et al., 2014; Sevy et al., 2010). It is even possible to generate three-dimensional images of the optical properties of the brain given a sufficient number of sources and detectors placed around the head (Minagawa-Kawai et al., 2008). This technique, called optical tomography, is costly and is usually restricted to young infants, as adults' larger heads usually result in too much light attenuation (Gibson et al., 2005). Another advantage of fNIRS is that it offers quantitative monitoring of HbO, HbR, and total hemoglobin, generating a more complete evaluation of the cortical hemodynamic response than the fMRI BOLD response which tracks HbR (Scholkmann et al., 2014). Lastly, the fNIRS hardware is silent, which makes it ideal for the presentation of accurate auditory stimuli in an acoustically-quiet environment, and artifact-free response measurement.

The major spatial limitation of NIRS is that it only probes a thin top layer of the cortex, up to 1.5 cm deep (Fukui et al., 2003). This is a considerable drawback for cognitive studies that aim to investigate deep regions such as the brainstem, basal ganglia, or amygdala (Minagawa-Kawai et al., 2008). However, a substantial amount of research can be done probing the upper layers of the auditory, visual, somatosensory or frontal cortices in CI research. Depth resolution is also highly dependent on the age of the subjects and varies somewhat from region to region even within a particular age group (Beauchamp et al., 2011). In adults, thicker scalp soft tissues and skulls significantly restrict NIR light penetration, impacting the accuracy of the recording. Deeper neural activity can be probed by increasing the source-detector distance, although at the cost of lower signal-to-noise ratio due to a reduction in the number of transmitted photons.

Good contact between the optodes and the skin of the scalp is also critical for a high signal-to-noise ratio (SNR) and a good quality recording. Hair is a nuisance in fNIRS recordings because (1) it interferes with this contact and (2) hair pigments significantly scatter and absorb NIR light and therefore attenuate the detected signal. In subjects with thick, dark hair, a researcher can spend a considerable amount of time trying to optimize the positions of the optodes to maximize the SNR. The use of gel can help to keep hair pushed out of the way. Nevertheless, the best recordings often come from subjects who are bald or have thin, blond hair — this makes fNIRS particularly suitable for work with infants.

Another drawback to fNIRS is the need to separate signals of cerebral origin from those of extra-cerebral tissues. For instance, blood volume changes in the scalp and within the muscles beneath the optical probes create noise in the fNIRS recordings and must be filtered during data analysis. Physiologic noise originating from heart rate and changes in respiratory effort may also be a source of confounding cerebral blood flow signals and must be accounted for during analysis (Gagnon et al., 2012). To remove the noise

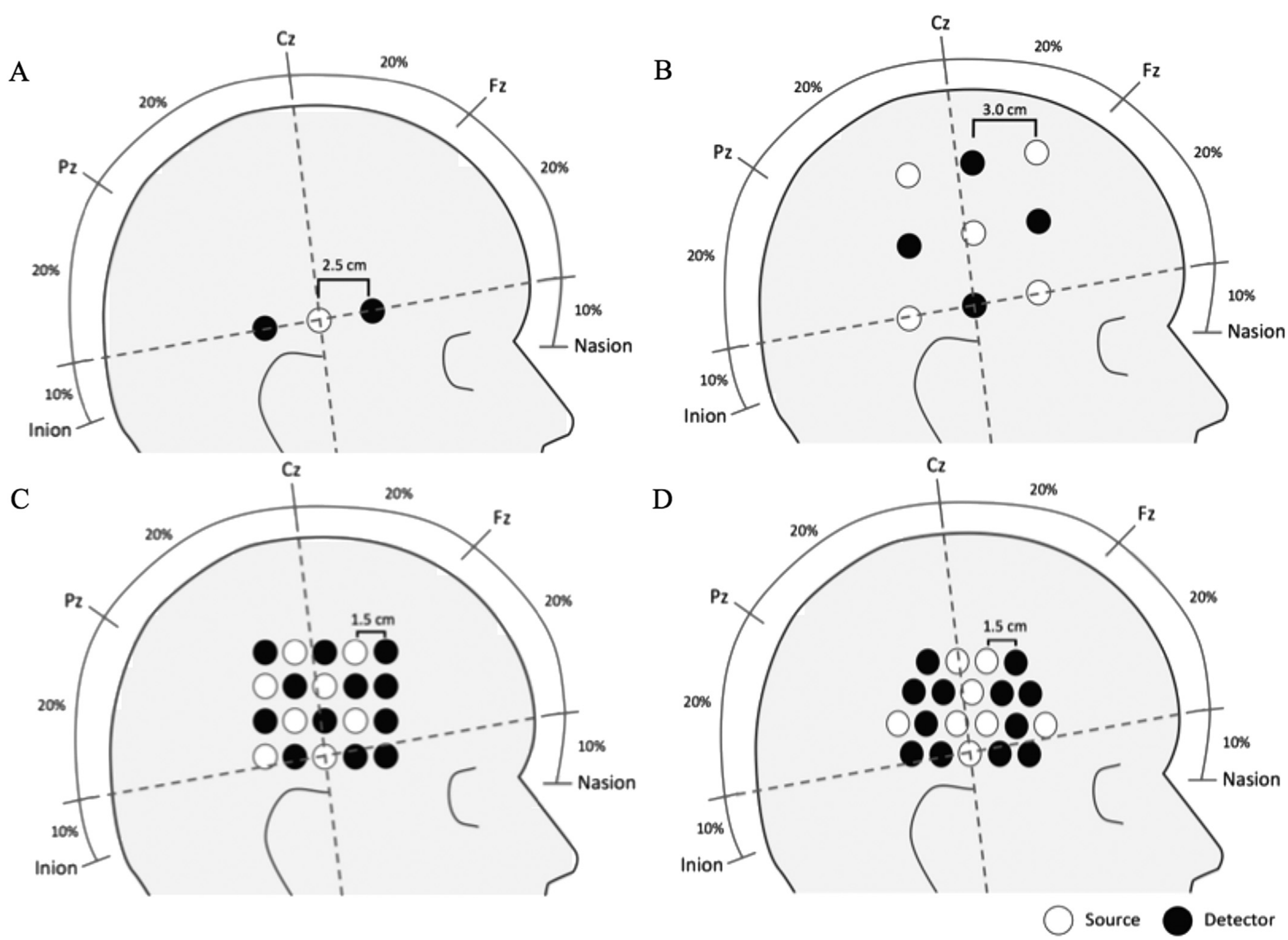

Fig. 2. Comparison between fNIRS probe layouts previously reported for CI use. A, Sevy et al. (2010); B, Dewey and Hartley (2015); C, Pollonini et al. (2014); D, Our new honeycomb-shaped headpiece. The optode arrangement in all headsets is based on the International 10/20 system: A is centered at the T3/T4 position; the optode located in the middle of the bottom horizontal line in B, C and D is aligned with the T3/T4 position.

component from the raw data, analytical strategies must be adopted. While some institutions use their own custom software, others turn to freely available software packages. However to date, there is a lack of a standard method for data analysis in fNIRS (Tak and Ye, 2014).

Similar to EEG, MEG and PET, the raw fNIRS data not provide an anatomic image upon which neural activity can be superimposed. Therefore, to localize brain activity to known anatomical locations, the optodes must be carefully positioned according to a standard for the recordings. The 10–20 (EEG) system is often used (Minagawa-Kawai et al., 2008). Once this is done, the optode layout is precisely aligned, and therefore the functional data obtained with fNIRS can be overlaid onto structural MRI images or anatomical atlases, if desired (Crosson et al., 2010).

Certain considerations must be taken into account when acquiring fNIRS data from CI users. Depending on the probe layout and the size of the headset, the external magnet of the CI device can interfere with headset placement over the temporal area. In such circumstances, we simply place the headset over the magnet (Fig. 1). While this obstructs the scalp contact of certain channels, the remaining channels can still be used. In our experience, however, the external magnet is generally posterior and inferior enough so as not to interfere with headset placement that permits the measurement of responses within the regions of interest, such as primary auditory cortex. Of course, care must be taken not to displace the magnet, as the implant would turn off. Gentle manipulation is also required when placing the headset in the crease between the pinna and the temporal skin to avoid repeated contact with the CI microphone and the resultant unpleasant noise for the CI user.

In an attempt to facilitate recording in the CI population, we designed a custom probe layout and headset at our institution. This arrangement features six light sources clustered in the center of the headpiece and an additional source anteriorly and posteriorly. Detectors are positioned in between (Fig. 2D). The center-to-center distance between adjacent optodes was 15 mm. Moving away from the checkerboard pattern described in our previous work (Fig. 2C; Pollonini et al., 2014), this new honeycomb-shaped design allows for a denser configuration of probes, while maintaining an equal number of channels. The result is a smaller and more convenient headpiece suitable for both adult and pediatric subjects, without compromising resolution. This dense multi-array headset allows spatial oversampling of a defined cortical area through adjacent channels that cross each other.

3.3. *What region(s) of the central nervous system should be studied?*

To understand the neural substrates involved in auditory processing through cochlear implants, it is necessary to observe activity within the brain when a sound stimulus is presented (Hall and Langers, 2014; Zhang et al., 2010). Ideally, one would track activity all the way from the level of the auditory nerve, through the ascending auditory pathways in the brainstem to the auditory and auditory-associated cortical regions. However, given its depth limitations, such whole-brain imaging is not possible with fNIRS. Because fNIRS is not a whole-brain technique, choices must be made about what portion of the cortex to record from in order to get the information most relevant to understanding auditory processing through a CI. A substantial body of fMRI data highlights the lateral temporal lobe and superior temporal gyrus (LTL/STG) as foundational to auditory processing at the cortical level.

Several studies have revealed preferential activity for the processing of acoustic parameters such as pitch, noise and spatiotemporal fluctuations in the LTL/STG (Hall and Plack, 2009; Humphries et al., 2010). Selective responses to species-specific vocalizations were demonstrated in the LTL/STG of humans and other mammals (Belin et al., 2002). In addition, studies using fMRI and implanted recording electrodes have shown localized responses within the left LTL/STG to phonemes, words, and phrases (DeWitt and Rauschecker, 2012). Of particular relevance to understanding hearing through a CI, Smalt et al. (Smalt et al., 2013). Demonstrated rapid neural adaptations in normal-hearing participants exposed to degraded sound, similar to what a CI user experiences.

While fNIRS does not provide whole-brain imaging, it can be used to dissociate music and language processing within constrained cortical regions such as the left and right LTL/STG thanks to stimulus specific processing differences across the cerebral hemispheres. Neuroimaging studies in normal-hearing subjects using PET and fMRI have previously shown that the left temporal lobe is primarily involved in speech and language processing, while the right temporal lobe preferentially responds to music (Hickok and Poeppel, 2007; Price, 2000; Belin et al., 2006). Furthermore, reports have demonstrated that secondary auditory areas in the right STG (surrounding Heschl's gyrus) are key to the processing of pitch information (Zatorre, 1998; Tramo et al., 2002). Temporal information, on the other hand, is preferentially processed by left-lateralized primary (core) auditory areas (Zatorre and Belin, 2001). Evidence also points toward a functional segregation between music and speech processing within the temporal lobes (Abrams et al., 2011; Levitin and Menon, 2003). Armony and colleagues not only revealed the existence of a region in the anterior STG (planum polare) that responds more strongly to music than voice, but their results also provide strong support for the presence of “music-preferring” neurons in this area (Armony et al., 2015). Moreover, several fMRI studies have demonstrated that the anterior portion of the STG is involved in higher-order music analyses such as extraction of melodic information (Rogalsky et al., 2011). Lesion studies have reinforced the idea that pitch and rhythm processing recruit separate neural subsystems within the auditory cortex: cortical damage can interfere with pitch discrimination without affecting rhythm performance, and vice-versa (Di Pietro et al., 2004; Ayotte et al., 2000). These and other findings indicate that the LTL/STG are the most clinically relevant regions of the cortex to focus on when imaging different classes of auditory perception in CI recipients using fNIRS.

3.4. *Data analysis techniques in multi-array fNIRS headsets*

A comprehensive review of analysis techniques available for use with fNIRS data is beyond the scope of this paper, and this topic has been extensively reviewed recently (Tak and Ye, 2014). Rather, in the following section we summarize current strategies to analyze recordings from dense multi-array headsets, as they are the most suitable for CI research. As with fMRI, signal pre-processing is initially performed to remove motion artifacts and physiologic noise. The first step requires identification of channels with good scalp contact. At our institution, we filter channels with excessive noise according to their scalp-coupling index (Pollonini et al., 2014). In brief, this technique relies on the fact that adequate scalp contact is characterized by a synchronous cardiac pulse signal recorded by both wavelengths of light emitted from a single probe. While a perfect correlation between each wavelength's cardiac signals is ideal (coefficient of 1), channels with an index threshold above 0.70 are reliable and can be retained.

The next step is motion artifact correction. Relative to hemodynamic-related changes, head movements will cause rapid changes, sharp spikes, and increases in the magnitude of the recorded signals (Tak and Ye, 2014). Previous reports have described the use of external accelerometers to estimate and correct baseline motion artifacts, but this requires additional instrumentation with its related cost and complexity (Virtanen et al., 2011). Many approaches to remove these artifacts without the need for motion sensors have also been described (Cui et al., 2010; Scholkmann et al., 2010). Our preferred technique consists of identifying start and stop times of motion artifacts by bandpass filtering each channel between 0.1 and 3.0 Hz to remove slow signal drift and by normalizing the intensity of the highest peak of the entire time course. We define peaks in the signal exceeding 20% of the maximum peak intensity as motion artifacts. These are then removed from the raw data by performing linear interpolation between the start and stop time points. Once motion artifacts are corrected, physiologic noise can be removed from the hemodynamic signal. This is usually accomplished by bandpass filtering between 0.016 and 0.25 Hz. The modified Beer–Lambert law is then used to calculate the relative concentrations of HbO and HbR for each channel and time point (see Section 3.1).

Once signal processing is complete, brain activation can be detected by performing inferential statistics on the fNIRS data. For each channel, all the trials of each stimulus first need to be averaged, a process called block-averaging (Scholkmann et al., 2014). The resulting block-averaged hemodynamic response is then compared to a predicted hemodynamic response. Predicted fNIRS responses can be modeled in a manner similar to the analysis of fMRI data (Cox, 1996). In such models, the HbO concentration rapidly rises after stimulus exposure, reaching a peak in a few seconds. The response then plateaus pending stimulus discontinuation, following which it slopes down until baseline HbO concentration is reached. Physiologically, this corresponds to an augmented blood supply required by the neuronal activation. Conversely, HbR concentration changes in a similar but opposite direction, decreasing during stimulus presentation. The quality of fit is determined by linear regression analysis of the measured and predicted responses, resulting in a T-statistic for each channel. Thus, each source-detector pair (channel) in the headset can be represented by a single number that describes the goodness of the fit. These T-statistics are then arranged in a spatial grid representing the position of the channel they derive from within the source-detector array. Multi-array fNIRS headsets provide spatial oversampling in the cortex since many channels cross each other at a given location. The resulting benefit is a reduction of noise in overlapping channels. A topographic (2 dimensional) activation map for each stimulus condition can then be generated by color-coding the T-statistic spatial grid. Alternatively, it is possible to project this colored T-statistic distribution map onto a standard brain image to create cortical activation maps that are easier to visualize and interpret.

4. Review of fNIRS neuroimaging studies in CI recipients

In 2013, fNIRS celebrated its 20th anniversary as a human neuroimaging modality. Jöbsis (1977) was the first to demonstrate the possibility of detecting changes of cortical oxygenation by transilluminating the cranium of anesthetized cats with NIR light (Jöbsis, 1977). However, it was not until 1993 that this emerging technology was first applied to human brains. That year, four research groups independently published the first single-site fNIRS human adult studies (Chance et al., 1993; Hoshi and Tamura, 1993; Kato et al., 1993; Villringer et al., 1993). fNIRS has since rapidly gained popularity among the neuroscience and clinical communities. If the number of annual publications reflects scientific enthusiasm, fNIRS has definitely emerged as one of the most popular research fields in the past 20 years: its publications have doubled every 3.5 years and have now reached over 200 per year (Boas et al., 2014). Despite this growing interest, the literature reporting the use of fNIRS in the CI population remains sparse. A comprehensive review across multiple databases of published articles mentioning fNIRS and cochlear implantation yielded five papers (Sevy et al., 2010; Pollonini et al., 2014; Dewey and Hartley, 2015; Lawler et al., 2015; Olds et al., 2015).

Sevy and colleagues report the first research application of fNIRS in CI users (Sevy et al., 2010). The authors used fNIRS to measure speech-evoked cortical responses within four subject cohorts: normal-hearing adults, normal-hearing children, deaf children who had over 4 months experience hearing through a cochlear implant, and deaf children who were tested on the day of initial CI activation. The speech stimuli consisted of digital recordings from children's stories in English. A four channel NIRS 2CE system (TechEn, Inc., Milford, MA) with 2 emitters mounted on a custom headframe was used to sample bilateral auditory cortices (Fig. 2A). The authors report successfully recording auditory cortical activity using this fNIRS setup in 100% of normal-hearing adults, 82% of normal-hearing children, 78% of deaf children who have used a CI for at least four months and 78% of deaf children on the day of CI initial activation. Interestingly, Sevy et al. had validated their NIRS experimental paradigm with fMRI in 3 normal-hearing adults. They showed that similar speech-evoked superior temporal gyrus responses were obtained with both fNIRS and fMRI. Such results were encouraging as they demonstrated that fNIRS was a feasible neuroimaging technique in CI users and that reliable hemodynamic cortical responses to speech could be recorded in these patients.

The same group later evaluated whether fNIRS was sensitive enough to detect differences in cortical activation evoked by different quality levels of speech in normal-hearing individuals (Pollonini et al., 2014). The investigators used a 140 channel fNIRS system (NIRScout, NIRx Medical Technologies LLC, Glen Head, NY) in a tight array to provide spatial oversampling, and permit averaging between channels to improve the SNR (Fig. 2C). By increasing the number of channels, the authors were able to generate topographic maps and measure the area of activation and center of mass. They also designed their own custom analytic software and developed novel data analysis techniques to filter channels with poor scalp contact or high SNR. The experimental paradigm consisted of four different stimuli: normal speech, channelized (vocoded) speech, scrambled speech and environmental noise (for previous use of these stimuli as cross-controls see, for example, Abrams et al., 2011; Humphries et al., 2001; Levitin et al., 2003). Their results revealed that speech intelligibility correlated with the pattern of auditory cortical activation measured with fNIRS: normal speech evoked the strongest responses, distorted speech produced less region-specific activation and environmental sounds evoked the least response. Again, the investigators validated their stimulus paradigm with fMRI on a single participant. Such results demonstrated that in normal-hearing individuals, fNIRS can detect differences in the response of the auditory cortex to variations in speech intelligibility. The conclusions of this study raise implications for the CI population. If fNIRS can provide an objective measure of whether a normal-hearing subject is hearing normal or distorted speech, then it has the potential to be used to assess how well speech information activates the brain in subjects hearing through a CI.

While Pollonini's study did not involve CI subjects, subjects hearing through a CI were studied with a similar technique (Olds et al., 2015). Olds' study used an experimental paradigm and fNIRS instrumentation comparable to that of Pollonini, but expanded the approach to participants with CI. Specifically, the authors aimed to better understand the variability in speech perception outcomes in CI using fNIRS. A NIRScout 1624 instrument (NIRx Medical Technologies, LLC, Glen Head, NY) with 140 channels was used to record the auditory cortical response of 32 post-lingually deaf adults hearing through a CI and 35 normal-hearing adults. Again, four auditory stimuli with varying degrees of speech intelligibility were employed: normal speech, channelized speech, scrambled speech and environmental noise. Speech reception thresholds (SRT), monosyllabic consonant-nucleus-consonant word (CNC Words) scores and AzBio sentence recognition scores were used as behavioral measures of speech perception. Results from this study demonstrated that the cortical activation pattern in implanted adults with good speech perception was similar to that of controls. In those two groups, less cortical activation was noted as the speech stimuli became less intelligible. In contrast, CI users with poor speech perception displayed large, indistinguishable cortical activations across all four stimuli. As the authors had hypothesized, the findings of this study demonstrated that activation patterns in the auditory cortex of CI recipients correlate with the quality of speech perception. Importantly, when the fNIRS measurements were repeated with the implant turned off, reduced cortical activations in all CI recipients were noted. This suggests that sound information is conveyed to the auditory cortex of CI users with poor speech perception, but that these subjects are unable to discriminate speech from the information that gets to the cortex.

To our knowledge, Lawler and colleagues are the only other research group actively using fNIRS neuroimaging in auditory processing studies in deaf individuals and CI recipients; to date, they have published two articles on that topic (Dewey and Hartley, 2015; Lawler et al., 2015). While this group's long-term aim is to examine cortical reorganization associated with deafness and cochlear implantation using fNIRS, none of these articles enrolled CI users thus far. The first report discusses maladaptive cross-modal plasticity in CI subjects and its role as a potential factor underlying poor performance following implantation (Lawler et al., 2015). Through this article, the authors describe their long-term research goals and introduce their plans for future fNIRS studies with deaf individuals and CI recipients. Later that year, Dewey and Hartley published a study on the use of fNIRS to detect visual and vibrotactile cross-modal plasticity changes in profoundly deaf but non-implanted individuals (Dewey and Hartley, 2015). Their setup consisted of a Hitachi ETG4000 (Hitachi Medical Corporation, Tokyo, Japan) optical topography system with 12 recording channels over each hemisphere (Fig. 2B). The authors reported that auditory deprivation is associated with cross-modal plasticity of visual inputs to auditory cortex. Practically speaking, such results highlight the ability of fNIRS to accurately record cortical changes associated with neural plasticity in profoundly deaf individuals. The application of these findings to the CI population is very promising, as they demonstrate the potential of fNIRS as an objective neuroimaging tool to detect and monitor cross-modal plasticity both prior to and following cochlear implantation.

5. Directions for future fNIRS application in CI users

5.1. Clinical applications

A promising future for fNIRS clinical applications includes the implementation of NIRS as a neuroimaging tool to guide post-implant programming in the service of improving deaf patients' speech and language outcomes. CIs need to be reprogrammed frequently to ensure they are accurately conveying the sound information within speech to the auditory nerve and, ultimately, to the auditory cortex. If the language areas of the brain are appropriately activated, then the child has the best chance of learning normal speech and language. Early identification of patients who do poorly is therefore critical, as prompt intervention can prevent delay in linguistic and psychosocial development (Robinshaw, 1995). Current cochlear implant assessment tools are limited and hard to administer in young infants, whose behavioral responses are difficult to elicit and are often not interpretable. An objective measure of how well speech information is processed within the cortex would provide an ideal tool for monitoring (and possibly predicting) language development in young CI users. Given that the number of imaging sessions is not restricted for fNIRS, repeat assessments through longitudinal studies can be performed to monitor rapid cortical modifications resulting from poor implant programming. In doing so, fNIRS studies may allow early identification of children on poor language development trajectories. If this can be achieved while the child is still within the critical time period when significant language development occurs (i.e. age 1–4 years), prompt intervention can be started. Ultimately, this type of early intervention could prevent delays in a child's psychosocial development, a process highly dependent on hearing (Yoshinaga-Itano et al., 1998). Using fNIRS to supplement our current clinical practice of CI programming and speech and language therapy is an exciting possibility.

5.2. Research applications

The opportunity for safe, repeated testing of CI recipients with fNIRS also provides investigators with the ability to explore the cortical changes associated with neural plasticity in this patient population. For instance, understanding the cortical reorganization that occurs following prolonged auditory deprivation in potential CI recipients may help predict their expected outcome post-implantation. This expectation is based on emerging evidence suggesting that cross-modal plasticity of visual inputs into a sensory-deprived auditory cortex may affect the ability of a CI recipient to process auditory information from their implant effectively (Sandmann et al., 2012). fNIRS may also provide insight into the cortical changes that take place in deaf patients following implantation. An example of such an application is the study of post-implantation training and its effects on brain plasticity. Pantev et al. examined the dynamics of auditory plasticity after implantation through MEG longitudinal imaging, suggesting that CI users would benefit the most from language training within the first 6 months after implantation (Pantev et al., 2006). As discussed, fNIRS is significantly easier to use in longitudinal studies compared to MEG. The opportunity to further explore cortical reorganization following hearing restoration has the potential to guide the design of post-implantation training strategies.

The neural basis for CI users' variable experience perceiving music is another interesting topic and one that merits further investigation. Despite advances in CI technology, music perception in CI recipients remains quite poor (Gfeller and Lansing, 1992). A growing body of psychophysical studies has better defined the limitations of music enjoyment and perception in CI users. For example, studies suggest that CI users perform poorly on pitch recognition tasks, whereas rhythmic perception remains relatively intact following implantation (W. B. Cooper et al., 2008; McDermott, 2004). Reports have also shown that appraisal ratings and overall listening time are significantly lower following implantation, with some CI users even describing music as “aversive” (Looi et al., 2012; Migirov et al., 2013). The challenges that CI users face in processing a complex auditory stimulus such as music can be explained by a number of technological, acoustical and biological constraints (Limb and Roy, 2014). While many of these have been addressed in the literature previously, the neural basis for poor music perception in CI users is under-investigated and poorly understood. This is at least in part due to inherent limitations on the use of most neuroimaging modalities with CI users, as outlined here. fNIRS is quiet and allows the use of event-related paradigms, thus offering greater flexibility in experimental inquiry. It is also relatively low cost, another factor that may have constrained examination of neural mechanisms underlying better or worse music perception in implant users in previous years. These and other features make fNIRS an ideal tool for evaluating music-evoked brain activation in CI recipients, as well as for examining the relationship between behavioral music performance and degree of auditory cortical activation in this patient population. Together, these inquiries would help achieve the long-term goal of higher-level music perception in CI recipients.

6. Conclusion

fNIRS is a safe, reliable neuroimaging technique that is compatible with CI devices. It offers many benefits over other approaches for examining cortical responses in CI recipients, although care must be taken in collecting and analyzing the data. While the existing literature on fNIRS neuroimaging in adult and pediatric CI users is currently limited, the future of this emerging technique is promising and numerous clinical and research applications remain to be explored.

Disclosure

Nothing to disclose. All authors have materially participated in the article preparation, and all have approved the final article.

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References

Abrams, D.A., Bhatara, A., Ryali, S., Balaban, E., Levitin, D.J., Menon, V., 2011. Decoding temporal structure in music and speech relies on shared brain resources but elicits different fine-scale spatial patterns. Cereb. Cortex 21, 1507–1518. http://dx.doi.org/10.1093/cercor/bhq198.

Aggarwal, R., Green, K.M.J., 2012. Cochlear implants and positron emission tomography. JLO 126, 1200–1203. http://dx.doi.org/10.1017/S0022215112002241.

Aine, C.J., 1995. A conceptual overview and critique of functional neuroimaging techniques in humans: I. MRI/FMRI and PET. Crit. Rev. Neurobiol. 9, 229–309.

Armony, J.L., Aubé, W., Angulo-Perkins, A., Peretz, I., Concha, L., 2015. Neuroscience letters. Neurosci. Lett. 593, 35–39. http://dx.doi.org/10.1016/j.neulet.2015.03.011.

Ayotte, J., Peretz, I., Rousseau, I., Bard, C., Bojanowski, M., 2000. Patterns of music agnosia associated with middle cerebral artery infarcts. Brain 123 (Pt 9), 1926–1938.

Azadarmaki, R., Tubbs, R., Chen, D.A., Shellock, F.G., 2014. MRI information for commonly used otologic implants: review and update. Otolaryngol. Head Neck Surg. 150, 512–519. http://dx.doi.org/10.1177/0194599813518306.

Babiloni, C., Pizzella, V., Gratta, C.D., Ferretti, A., Romani, G.L., 2009. Fundamentals of electroencefalography, magnetoencefalography, and functional magnetic resonance imaging. Int. Rev. Neurobiol. 86, 67–80. http://dx.doi.org/10.1016/S0074-7742(09)86005-4.

Bandettini, P.A., 2009. What's new in neuroimaging methods? Ann. N. Y. Acad. Sci. 1156, 260–293.

Baumgartner, W.D., Youssefzadeh, S., Hamzavi, J., Czerny, C., Gstoettner, W., 2001. Clinical application of magnetic resonance imaging in 30 cochlear implant patients. Otol. Neurotol. 22, 818–822.

Beauchamp, M.S., Beurlot, M.R., Fava, E., Nath, A.R., Parikh, N.A., Saad, Z.S., Bortfeld, H., Oghalai, J.S., 2011. The developmental trajectory of brain-scalp distance from birth through childhood: implications for functional neuroimaging. PLoS ONE 6, e24981. http://dx.doi.org/10.1371/journal.pone.0024981.

Belin, P., Zatorre, R.J., Ahad, P., 2002. Human temporal-lobe response to vocal sounds. Brain Res. Cogn. Brain Res. 13, 17–26.

Belin, P., Zilbovicius, M., Crozier, S., Thivard, L., Fontaine, A.A., Masure, M.-C., Samson, Y., 2006. Lateralization of speech and auditory temporal processing. J. Cogn. Neurosci. 10 (4), 536–540. http://doi.org/10.1162/089892998562834.

Boas, D.A., Dale, A.M., Franceschini, M.A., 2004. Diffuse optical imaging of brain activation: approaches to optimizing image sensitivity, resolution, and accuracy. Neuroimage 23, S275–S288. http://dx.doi.org/10.1016/j.neuroimage.2004.07.011.

Boas, D.A., Elwell, C.E., Ferrari, M., Taga, G., 2014. Twenty years of functional near-infrared spectroscopy: introduction for the special issue. NeuroImage 85, 1–5. http://dx.doi.org/10.1016/j.neuroimage.2013.11.033.

Castañeda-Villa, N., Cornejo, J.M., James, C.J., Maurits, N.M., 2012. Quantification of LLAEP interhemispheric symmetry by the intraclass correlation coefficient as a measure of cortical reorganization after cochlear implantation. Int. J. Pediatr. Otorhinolaryngol. 76, 1729–1736. http://dx.doi.org/10.1016/j.ijporl.2012.08.011.

Chance, B., Zhuang, Z., UnAh, C., Alter, C., Lipton, L., 1993. Cognition-activated low-frequency modulation of light absorption in human brain. Proc. Natl. Acad. Sci. U. S. A. 90, 3770–3774.

Choi, C.-H., Oghalai, J.S., 2005. Predicting the effect of post-implant cochlear fibrosis on residual hearing. Hear. Res. 205, 193–200. http://dx.doi.org/10.1016/j.heares.2005.03.018.

Cooper, W.B., Tobey, E., Loizou, P.C., 2008. Music perception by cochlear implant and normal hearing listeners as measured by the montreal battery for evaluation of Amusia. Ear Hear. 29, 618–626. http://dx.doi.org/10.1097/AUD.0b013e318174e787.

Cope, M., Delpy, D.T., Reynolds, E.O., Wray, S., Wyatt, J., van der Zee, P., 1988. Methods of quantitating cerebral near infrared spectroscopy data. Adv. Exp. Med. Biol. 222, 183e189.

Cox, R.W., 1996. AFNI: software for analysis and visualization of functional magnetic resonance neuroimages. Comput. Biomed. Res. 29, 162e173.

Crane, B.T., Gottschalk, B., Kraut, M., Aygun, N., Niparko, J.K., 2010. Magnetic resonance imaging at 1.5 T after cochlear implantation. Otol. Neurotol. 31, 1215–1220. http://dx.doi.org/10.1097/MAO.0b013e3181ec1d61.

Crosson, B., Ford, A., McGregor, K.M., Meinzer, M., Cheshkov, S., Li, X., Walker-Batson, D., Briggs, R.W., 2010. Functional imaging and related techniques: an introduction for rehabilitation researchers. J. Rehabil. Res. Dev. 47, vii–xxxiv.

Cui, X., Bray, S., Reiss, A.L., 2010. Functional near infrared spectroscopy (NIRS) signal improvement based on negative correlation between oxygenated and deoxygenated hemoglobin dynamics. NeuroImage 49, 3039–3046. http://dx.doi.org/10.1016/j.neuroimage.2009.11.050.

Delpy, D.T., Cope, M., 1997. Quantification in tissue near–infrared spectroscopy. Phil. Trans. R. Soc. Lond. 352, 649–659.

Dewey, R.S., Hartley, D.E.H., 2015. Cortical cross-modal plasticity following deafness measured using functional near-infrared spectroscopy. Hear. Res. 325, 55–63. http://dx.doi.org/10.1016/j.heares.2015.03.007.

DeWitt, I., Rauschecker, J.P., 2012. Phoneme and word recognition in the auditory ventral stream. Proc. Natl. Acad. Sci. 109, E505–E514. http://dx.doi.org/10.1073/pnas.1113427109.

Di Pietro, M., Laganaro, M., Leemann, B., Schnider, A., 2004. Receptive amusia: temporal auditory processing deficit in a professional musician following a left temporo-parietal lesion. Neuropsychologia 42 (7), 868–877.

Doucet, M.E., Bergeron, F., Lassonde, M., Ferron, P., Lepore, F., 2006. Cross-modal reorganization and speech perception in cochlear implant users. Brain 129, 3376–3383. http://dx.doi.org/10.1093/brain/awl264.

Elwell, C.E., Cooper, C.E., 2011. Making light work: illuminating the future of biomedical optics. Philosophical Transactions of the Royal Society A: Mathematical. Phys. Eng. Sci. 369, 4358–4379. http://dx.doi.org/10.1016/j.ejso.2010.10.006.

Fava, E., Hull, R., Bortfeld, H., 2011. Linking behavioral and neurophysiological indicators of perceptual tuning to language. Front. Psychol. 2, 174. http://dx.doi.org/10.3389/fpsyg.2011.00174.

Fava, E., Hull, R., Baumbauer, K., Bortfeld, H., 2014a. Hemodynamic responses to speech and music in preverbal infants. Child. Neuropsychol. 20, 430–448. http://dx.doi.org/10.1080/09297049.2013.803524.

Fava, E., Hull, R., Bortfeld, H., 2014b. Dissociating cortical activity during processing of native and non-native audiovisual speech from early to late infancy. Brain Sci. 4, 471–487. http://dx.doi.org/10.3390/brainsci4030471.

Ferrari, M., Quaresima, V., 2012. A brief review on the history of human functional near-infrared spectroscopy (fNIRS) development and fields of application. NeuroImage 63, 921–935. http://dx.doi.org/10.1016/j.neuroimage.2012.03.049.

Ferree, T.C., Clay, M.T., Tucker, D.M., 2001. The spatial resolution of scalp EEG. Neurocomputing 38, 1209–1216.

Food and Drug Administration, 2015. FDA Executive Summary, Prepared for the May 1, 2015 Meeting of the Ear, Nose, and Throat Devices Panel of the Medical Devices Advisory Committee. Accessible at. http://www.fda.gov/downloads/AdvisoryCommittees/CommitteesMeetingMaterials/MedicalDevices/MedicalDevicesAdvisoryCommittee/EarNoseandThroatDevicesPanel/UCM443996.pdf.

Fukui, Y., Ajichi, Y., Okada, E., 2003. Monte Carlo prediction of near-infrared light propagation in realistic adult and neonatal head models. Appl. Opt. 42, 2881–2887.

Gagnon, L., Cooper, R.J., Yücel, M.A., Perdue, K.L., Greve, D.N., Boas, D.A., 2012. Short separation channel location impacts the performance of short channel regression in NIRS. NeuroImage 59, 2518–2528. http://dx.doi.org/10.1016/j.neuroimage.2011.08.095.

Geers, A., Brenner, C., Davidson, L., 2003. Factors associated with development of speech perception skills in children implanted by age five. Ear Hear. 24 (Suppl. l), 24S–35S. http://doi.org/10.1097/01.AUD.0000051687.99218.0F.

Gervain, J., Mehler, J., Werker, J.F., Nelson, C.A., Csibra, G., Lloyd-Fox, S., Shukla, M., Aslin, R.N., 2011. Near-infrared spectroscopy: a report from the McDonnell infant methodology consortium. Dev. Cogn. Neurosci. 1, 22–46. http://dx.doi.org/10.1016/j.dcn.2010.07.004.

Gfeller, K., Lansing, C., 1992. Musical perception of cochlear implant users as measured by the primary measures of music audition: an item analysis. J. Music Ther. 29, 18–39.

Gibson, A.P., Hebden, J.C., Arridge, S.R., 2005. Recent advances in diffuse optical imaging. Phys. Med. Biol. 50, R1–R43.

Gilley, P.M., Sharma, A., Dorman, M.F., 2008. Cortical reorganization in children with cochlear implants. Brain Res. 1239 IS, 56–65.

Giraud, A.L., Truy, E., Frackowiak, R., 2001. Imaging plasticity in cochlear implant patients. Audiol. Neurootol 6, 381–393.

Girouard, H., 2006. Neurovascular coupling in the normal brain and in hypertension, stroke, and Alzheimer disease. J. Appl. Physiol. 100, 328–335. http://dx.doi.org/10.1152/japplphysiol.00966.2005.

Gubbels, S.P., McMenomey, S.O., 2006. Safety study of the Cochlear Nucleus 24 device with internal magnet in the 1.5 Tesla magnetic resonance imaging scanner. Laryngoscope 116, 865–871. http://dx.doi.org/10.1097/01.MLG.0000216807.03225.CE.

Hall, D.A., Langers, D.R.M., 2014. Special issue in Hearing Research: human auditory neuroimaging. Hear. Res. 307, 1–3. http://dx.doi.org/10.1016/j.heares.2013.09.002.

Hall, D.A., Plack, C.J., 2009. Pitch processing sites in the human auditory brain. Cereb. Cortex 19, 576–585. http://dx.doi.org/10.1093/cercor/bhn108.

Hickok, G., Poeppel, D., 2007. The cortical organization of speech processing. Nat. Rev. Neurosci. 8, 393–402. http://dx.doi.org/10.1038/nrn2113.

Holt, R.F., Svirsky, M.A., 2008. An exploratory look at pediatric cochlear implantation: is earliest always best? Ear Hear. 29, 492–511. http://dx.doi.org/10.1097/AUD.0b013e31816c409f.

Hoshi, Y., 2003. Functional near-infrared optical imaging: utility and limitations in human brain mapping. Psychophysiology 40 (4), 511–520. http://dx.doi.org/10.1111/1469-8986.00053.

Hoshi, Y., Tamura, M., 1993. Detection of dynamic changes in cerebral oxygenation coupled to neuronal function during mental work in man. Neurosci. Lett. 150, 5–8.

Hull, R., Bortfeld, H., Koons, S., 2009. Near-infrared spectroscopy and cortical responses to speech production. Open Neuroimaging J. 3, 26–30.

Humphries, C., Willard, K., Buchsbaum, B., Hickok, G., 2001. Role of anterior temporal cortex in auditory sentence comprehension: an fMRI study. Neuroreport 12 (8), 1749–1752.

Humphries, C., Liebenthal, E., Binder, J.R., 2010. Tonotopic organization of human auditory cortex. NeuroImage 50, 1202–1211. http://dx.doi.org/10.1016/j.neuroimage.2010.01.046.

Huppert, T.J., Hoge, R.D., Diamond, S.G., Franceschini, M.A., Boas, D.A., 2006. A temporal comparison of BOLD, ASL, and NIRS hemodynamic responses to motor stimuli in adult humans. NeuroImage 29, 368–382. http://dx.doi.org/10.1016/j.neuroimage.2005.08.065.

Jöbsis, F.F., 1977. Noninvasive, infrared monitoring of cerebral and myocardial oxygen sufficiency and circulatory parameters. Science 198, 1264–1267.

Kato, T., Kamei, A., Takashima, S., Ozaki, T., 1993. Human visual cortical function during photic stimulation monitoring by means of near-infrared spectroscopy. J. Cereb. Blood Flow. Metab. 13, 516–520. http://dx.doi.org/10.1038/jcbfm.1993.66.

Katzenstein, J.M., Oghalai, J.S., Tonini, R., Baker, D., Haymond, J., Caudle, S.E., 2009. Neurocognitive functioning of a child with partial trisomy 6 and monosomy 21. Neurocase 15, 97–100. http://dx.doi.org/10.1080/13554790802631910.

Kirk, K.I., Miyamoto, R.T., Lento, C.L., Ying, E., O'Neill, T., Fears, B., 2002. Effects of age at implantation in young children. Ann. Otol. Rhinol. Laryngol. Suppl. 189, 69–73.

Lawler, C.A., Wiggins, I.M., Dewey, R.S., Hartley, D.E.H., 2015. The use of functional near-infrared spectroscopy for measuring cortical reorganisation in cochlear implant users: a possible predictor of variable speech outcomes? Cochlear Implants Int. 16 (Suppl. 1), S30–S32. http://dx.doi.org/10.1179/1467010014Z.000000000230.

Lazard, D.S., Giraud, A.L., Gnansia, D., Meyer, B., Sterkers, O., 2012. Understanding the deafened brain: Implications for cochlear implant rehabilitation. European Annals of Otorhinolaryngology. Head Neck Dis. 129, 98–103.

Lazard, D.S., Innes-Brown, H., Barone, P., 2014. Adaptation of the communicative brain to post-lingual deafness. Evidence from functional imaging. Hear. Res. 307, 136–143. http://dx.doi.org/10.1016/j.heares.2013.08.006.

Lazeyras, F., Boëx, C., Sigrist, A., Seghier, M.L., Cosendai, G., 2002. Functional MRI of auditory cortex activated by multisite electrical stimulation of the cochlea. NeuroImage. http://dx.doi.org/10.1006/nimg.2002.1240.

Levitin, D.J., Menon, V., 2003. Musical structure is processed in “language” areas of the brain: a possible role for Brodmann Area 47 in temporal coherence. NeuroImage 20, 2142–2152.

Levitin, D.J., Menon, V., 2005. The neural locus of temporal structure and expectancies in music: evidence from functional neuroimaging at 3 Tesla. Music Percept. 22 (3), 563–575.

Levitin, D.J., Menon, V., Schmitt, J.E., Eliez, S., White, C.D., Glover, G.H., Reiss, A.L., 2003. Neural correlates of auditory perception in Williams syndrome: an fMRI study. Neuroimage 18 (1), 74–82.

Limb, C.J., Roy, A.T., 2014. Technological, biological, and acoustical constraints to music perception in cochlear implant users. Hear. Res. 308, 13–26. http://dx.doi.org/10.1016/j.heares.2013.04.009.

Limb, C.J., Molloy, A.T., Jiradejvong, P., Braun, A.R., 2010. Auditory cortical activity during cochlear implant-mediated perception of spoken language, melody, and rhythm. JARO 11, 133–143. http://dx.doi.org/10.1007/s10162-009-0184-9.

Lin, J.W., Mody, A., Tonini, R., Emery, C., Haymond, J., Vrabec, J.T., Oghalai, J.S., 2010. Characteristics of malfunctioning channels in pediatric cochlear implants. Laryngoscope 120, 399–404. http://dx.doi.org/10.1002/lary.20668.

Lloyd-Fox, S., Blasi, A., Elwell, C.E., 2010. Illuminating the developing brain: the past, present and future of functional near infrared spectroscopy. Neurosci. Biobehav. Rev. 34, 269–284. http://dx.doi.org/10.1016/j.neubiorev.2009.07.008.

Looi, V., Gfeller, K., Driscoll, V., 2012. Music appreciation and training for cochlear implant recipients: a review. Semin. Hear 33, 307–334. http://dx.doi.org/10.1055/s-0032-1329222.

Majdani, O., Leinung, M., Rau, T., Akbarian, A., Zimmerling, M., Lenarz, M., Lenarz, T., Labadie, R., 2008. Demagnetization of cochlear implants and temperature changes in 3.0T MRI environment. Otolaryngol. Head. Neck Surg. 139, 833–839. http://dx.doi.org/10.1016/j.otohns.2008.07.026.

Majdani, O., Rau, T.S., Gotz, F., Zimmerling, M., Lenarz, M., Lenarz, T., Labadie, R., Leinung, M., 2009. Artifacts caused by cochlear implants with non-removable magnets in 3T MRI: phantom and cadaveric studies. Eur. Arch. Otorhinolaryngol. 266, 1885–1890. http://dx.doi.org/10.1007/s00405-009-0994-8.

Marcar, V.L., Schwarz, U., Martin, E., Loenneker, T., 2006. How depth of anesthesia influences the blood oxygenation level-dependent signal from the visual cortex of children. AJNR Am. J. Neuroradiol. 27, 799–805.

Mc Laughlin, M., Lopez Valdes, A., Reilly, R.B., Zeng, F.-G., 2013. Cochlear implant artifact attenuation in late auditory evoked potentials: a single channel approach. Hear. Res. 302 IS, 84–95.

McDermott, H.J., 2004. Music perception with cochlear implants: a review. Trends Amplif. 8, 49–82. http://dx.doi.org/10.1177/108471380400800203.

Migirov, L., Kronenberg, J., Henkin, Y., 2013. Self-reported listening habits and enjoyment of music among adult cochlear implant recipients. Ann. Otol. Rhinol. Laryngol. 118, 350–355.

Minagawa-Kawai, Y., Mori, K., Hebden, J.C., Dupoux, E., 2008. Optical imaging of infants' neurocognitive development: recent advances and perspectives. Devel Neurobiol. 68, 712–728. http://dx.doi.org/10.1002/dneu.20618.

Miyamoto, R.T., Osberger, M.J., Todd, S.L., Robbins, A.M., Stroer, B.S., Zimmerman-Phillips, S., Carney, A.E., 1994. Variables affecting implant performance in children. Laryngoscope 104, 1120–1124. http://dx.doi.org/10.1288/00005537-199409000-00012.

Naito, Y., Tateya, I., Fujiki, N., Hirano, S., Ishizu, K., Nagahama, Y., Fukuyama, H., Kojima, H., 2000. Increased cortical activation during hearing of speech in cochlear implant users. Hear. Res. 143, 139–146.

Niemz, M.H., 2002. Laser Tissue Interactions: Fundamentals and Application, second ed. Springer, New York, ISBN 978-3-662-04717-0.

Nikolopoulos, T.P., O'Donoghue, G.M., Archbold, S., 1999. Age at implantation: its importance in pediatric cochlear implantation. Laryngoscope 109, 595–599. http://dx.doi.org/10.1097/00005537-199904000-00014.

Oghalai, J.S., Tonini, R., Rasmus, J., Emery, C., Manolidis, S., Vrabec, J.T., Haymond, J., 2009. Intra-operative monitoring of cochlear function during cochlear implantation. Cochlear Implants Int. 10, 1–18. http://dx.doi.org/10.1002/cii.372.

Olds, C., Pollonini, L., Abaya, H., Larky, J., Loy, M., Bortfeld, H., Beauchamp, M.S., Oghalai, J.S., December 2015. Cortical activation patterns correlate with speech understanding after cochlear implantation. Ear Hear. 1. http://dx.doi.org/10.1097/AUD.0000000000000258 [Epub ahead of print].

Osberger, M.J., Fisher, L., 2000. Preoperative predictors of postoperative implant performance in children. Ann. Otol. Rhinol. Laryngol. Suppl. 185, 44–46.

Pantev, C., Dinnesen, A., Ross, B., Wollbrink, A., Knief, A., 2006. Dynamics of auditory plasticity after cochlear implantation: a longitudinal study. Cereb. Cortex 16, 31–36. http://dx.doi.org/10.1093/cercor/bhi081.

Pasley, B.N., David, S.V., Mesgarani, N., Flinker, A., Shamma, S.A., Crone, N.E., Knight, R.T., Chang, E.F., 2012. Reconstructing speech from human auditory cortex. PLoS Biol. 10, e1001251. http://dx.doi.org/10.1371/journal.pbio.1001251.

Peterson, N.R., Pisoni, D.B., Miyamoto, R.T., 2010. Cochlear implants and spoken language processing abilities: review and assessment of the literature. Restor. Neurol. Neurosci. 28, 237–250. http://dx.doi.org/10.3233/RNN-2010-0535.

Piper, S.K., Krueger, A., Koch, S.P., Mehnert, J., Habermehl, C., Steinbrink, J., Obrig, H., Schmitz, C.H., 2014. A wearable multi-channel fNIRS system for brain imaging in freely moving subjects. NeuroImage 85, 64–71. http://dx.doi.org/10.1016/j.neuroimage.2013.06.062.

Pollonini, L., Olds, C., Abaya, H., Bortfeld, H., Beauchamp, M.S., Oghalai, J.S., 2014. Hearing research. Hear. Res. 309, 84–93. http://dx.doi.org/10.1016/j.heares.2013.11.007.

Ponton, C.W., Eggermont, J.J., Don, M., Waring, M.D., Kwong, B., Cunningham, J., Trautwein, P., 2000. Maturation of the mismatch negativity: effects of profound deafness and cochlear implant use. Audiol. Neurootol 5, 167–185.

Portnoy, W.M., Mattucci, K., 1991. Cochlear implants as a contraindication to magnetic resonance imaging. Ann. Otol. Rhinol. Laryngol. 100, 195–197.

Posner, M.I., Levitin, D.J., 1997. Imaging the future. In: Solso, R.L. (Ed.), The Science of the Mind: The 2lst Century. MIT Press, Cambridge, MA.

Price, C.J., 2000. The anatomy of language: contributions from functional neuroimaging. J. Anat. 197, 335–359.

Quaresima, V., Bisconti, S., Ferrari, M., 2012. A brief review on the use of functional near-infrared spectroscopy (fNIRS) for language imaging studies in human newborns and adults. Brain Lang. 121, 79–89. http://dx.doi.org/10.1016/j.bandl.2011.03.009.

Risi, F., Saldanha, A., Leigh, R., Gibson, P., 2004. Magnetic resonance imaging safety of Nucleus® 24 cochlear implants at 3.0 T. Int. Congr. Ser. 1273, 394–398. http://dx.doi.org/10.1016/j.ics.2004.07.033.

Robinshaw, H.M., 1995. Early intervention for hearing impairment: differences in the timing of communicative and linguistic development. Br. J. Audiol. 29, 315–334.

Rogalsky, C., Rong, F., Saberi, K., Hickok, G., 2011. Functional anatomy of language and music perception: temporal and structural factors investigated using functional magnetic resonance imaging. J. Neurosci. 31, 3843–3852. http://dx.doi.org/10.1523/JNEUROSCI.4515-10.2011.

Roland, J.T., Huang, T.C., Fishman, A.J., 2006. Cochlear implant electrode history, choices, and insertion techniques. In: Waltzman, S., Rol, J.T. (Eds.), Cochlear Implants, second ed. Thieme, New York, pp. 110–125.

Rubinstein, J.T., 2004. How cochlear implants encode speech. Curr. Opin. Otolaryngol. Head Neck Surg. 12 (5), 444–448.

Sandmann, P., Eichele, T., Buechler, M., Debener, S., Jancke, L., Dillier, N., Hugdahl, K., Meyer, M., 2009. Evaluation of evoked potentials to dyadic tones after cochlear implantation. Brain 132, 1967–1979. http://dx.doi.org/10.1093/brain/awp034.

Sandmann, P., Kegel, A., Eichele, T., Dillier, N., Lai, W., Bendixen, A., Debener, S., Jancke, L., Meyer, M., 2010. Clinical Neurophysiology. Clin. Neurophysiol. 121, 2070–2082. http://dx.doi.org/10.1016/j.clinph.2010.04.032.

Sandmann, P., Dillier, N., Eichele, T., Meyer, M., Kegel, A., Pascual-Marqui, R.D., Marcar, V.L., Jancke, L., Debener, S., 2012. Visual activation of auditory cortex reflects maladaptive plasticity in cochlear implant users. Brain 135, 555–568. http://dx.doi.org/10.1093/brain/awr329.

Santa Maria, P.L., Oghalai, J.S., 2014. When is the best timing for the second implant in pediatric bilateral cochlear implantation? Laryngoscope 124, 1511–1512. http://dx.doi.org/10.1002/lary.24465.

Sassaroli, A., Fantini, S., 2004. Comment on the modified Beer-Lambert law for scattering media. Phys. Med. Biol. 49 (14), N255–N257.

Scholkmann, F., Spichtig, S., Muehlemann, T., Wolf, M., 2010. How to detect and reduce movement artifacts in near-infrared imaging using moving standard deviation and spline interpolation. Physiol. Meas. 31, 649–662. http://dx.doi.org/10.1088/0967-3334/31/5/004.

Scholkmann, F., Kleiser, S., Metz, A.J., Zimmermann, R., Pavia, J.M., Wolf, U., Wolf, M., 2014. A review on continuous wave functional near-infrared spectroscopy and imaging instrumentation and methodology. NeuroImage 85, 6–27. http://dx.doi.org/10.1016/j.neuroimage.2013.05.004.

Seghier, M.L., Boëx, C., Lazeyras, F., Sigrist, A., Pelizzone, M., 2005. FMRI evidence for activation of multiple cortical regions in the primary auditory cortex of deaf subjects users of multichannel cochlear implants. Cereb. Cortex 15, 40–48. http://dx.doi.org/10.1093/cercor/bhh106.

Sevy, A.B.G., Bortfeld, H., Huppert, T.J., Beauchamp, M.S., Tonini, R.E., Oghalai, J.S., 2010. Neuroimaging with near-infrared spectroscopy demonstrates speech-evoked activity in the auditory cortex of deaf children following cochlear implantation. Hear. Res. 270, 39–47. http://dx.doi.org/10.1016/j.heares.2010.09.010.

Smalt, C.J., Gonzalez-Castillo, J., Talavage, T.M., Pisoni, D.B., Svirsky, M.A., 2013. Neural correlates of adaptation in freely-moving normal hearing subjects under cochlear implant acoustic simulations. NeuroImage 82, 500–509. http://dx.doi.org/10.1016/j.neuroimage.2013.06.001.

Smith, M., 2011. Shedding light on the adult brain: a review of the clinical applications of near-infrared spectroscopy. Philosophical Transactions of the Royal Society A: Mathematical. Phys. Eng. Sci. 369, 4452–4469. http://dx.doi.org/10.1007/978-1-4419-1241-1_41.

Song, J., Davey, C., Poulsen, C., Luu, P., Turovets, S., Anderson, E., Tucker, D., 2015. EEG source localization: sensor density and head surface coverage. J. Neurosci. Methods 256, 9–21.

Srinivasan, A.G., Padilla, M., Shannon, R.V., Landsberger, D.M., 2013. Improving speech perception in noise with current focusing in cochlear implant users. Hear Res. 299, 29–36. http://dx.doi.org/10.1016/j.heares.2013.02.004.

Steinschneider, M., Nourski, K.V., Rhone, A.E., Kawasaki, H., Oya, H., Howard, M.A., 2014. Differential activation of human core, non-core and auditory-related cortex during speech categorization tasks as revealed by intracranial recordings. Front. Neurosci. 8, 240. http://dx.doi.org/10.3389/fnins.2014.00240.

Tak, S., Ye, J.C., 2014. Statistical analysis of fNIRS data: a comprehensive review. NeuroImage 85, 72–91. http://dx.doi.org/10.1016/j.neuroimage.2013.06.016.

Teissl, C., Kremser, C., Hochmair, E.S., Hochmair-Desoyer, I.J., 1999. Magnetic resonance imaging and cochlear implants: compatibility and safety aspects. J. Magn. Reson Imaging 9, 26–38. http://dx.doi.org/10.1002/(SICI)1522-2586(199901)9:1<26::AID-JMRI4>3.0.CO;2-H.

Torricelli, A., Contini, D., Pifferi, A., Caffini, M., Re, R., Zucchelli, L., Spinelli, L., 2014. Time domain functional NIRS imaging for human brain mapping. NeuroImage 85 (Pt 1), 28–50. http://dx.doi.org/10.1016/j.neuroimage.2013.05.106.

Tramo, M.J., Shah, G.D., Braida, L.D., 2002. Functional role of auditory cortex in frequency processing and pitch perception. J. Neurophysiol. 87 (1), 122–139.

Truy, E., 1999. Neuro-functional imaging and profound deafness. Int. J. Pediatr. Otorhinolaryngology 47, 131–136.

Uludağ, K., Steinbrink, J., Villringer, A., Obrig, H., 2004. Separability and cross talk: optimizing dual wavelength combinations for near-infrared spectroscopy of the adult head. Neuroimage 22 (2), 583–589. http://dx.doi.org/10.1016/j.neuroimage.2004.02.023.

Villringer, A., Planck, J., Hock, C., Schleinkofer, L., Dirnagl, U., 1993. Near infrared spectroscopy (NIRS): a new tool to study hemodynamic changes during activation of brain function in human adults. Neurosci. Lett. 154, 101–104.

Virtanen, J., Noponen, T., Kotilahti, K., Virtanen, J., Ilmoniemi, R.J., 2011. Accelerometer-based method for correcting signal baseline changes caused by motion artifacts in medical near-infrared spectroscopy. J. Biomed. Opt. 16, 087005. http://dx.doi.org/10.1117/1.3606576.

Welch, A.J., van Gemert, M.J.C. (Eds.), 2011. Optical-Thermal Response of Laser-irradiated Tissue, second ed. Springer, New York, ISBN 978-90-481-8830-7.

Wilcox, T., Bortfeld, H., Woods, R., Wruck, E., Boas, D.A., 2005. Using near-infrared spectroscopy to assess neural activation during object processing in infants. J. Biomed. Opt. 10 (1), 011010–0110109.

Williamson, R.A., Pytynia, K., Oghalai, J.S., Vrabec, J.T., 2009. Auditory performance after cochlear implantation in late septuagenarians and octogenarians. Otol. Neurotol. 30, 916–920. http://dx.doi.org/10.1097/MAO.0b013e3181b4e594.

Wolf, M., Ferrari, M., Quaresima, V., 2007. Progress of near-infrared spectroscopy and topography for brain and muscle clinical applications. J. Biomed. Opt. 12 (6), 062104–062104–14. http://doi.org/10.1117/1.2804899.

Wong, D., Miyamoto, R.T., Pisoni, D.B., Sehgal, M., Hutchins, G.D., 1999. PET imaging of cochlear-implant and normal-hearing subjects listening to speech and nonspeech. Hear. Res. 132, 34–42.

Ying, Y.-L.M., Lin, J.W., Oghalai, J.S., Williamson, R.A., 2013. Cochlear implant electrode misplacement: incidence, evaluation, and management. Laryngoscope 123, 757–766. http://dx.doi.org/10.1002/lary.23665.

Yoshinaga-Itano, C., Sedey, A.L., Coulter, D.K., Mehl, A.L., 1998. Language of early- and later-identified children with hearing loss. Pediatrics 102, 1161–1171.

Zatorre, R., 1998. Functional specialization of human auditory cortex for musical processing. Brain 121 (10), 1817–1818. http://dx.doi.org/10.1093/brain/121.10.1817.

Zatorre, R.J., Belin, P., 2001. Spectral and temporal processing in human auditory cortex. Cereb. Cortex 11 (10), 946–953.

Zhang, F., Anderson, J., Samy, R., Houston, L., 2010. The adaptive pattern of the late auditory evoked potential elicited by repeated stimuli in cochlear implant users. Int. J. Audiol. 49, 277–285. http://dx.doi.org/10.3109/14992020903321759.