Chapter

Hearing Restoration through Optical Wireless Cochlear Implants

Stylianos E. Trevlakis,
Alexandros-Apostolos A. Boulogeorgos and
George K. Karagiannidis

Abstract

In this chapter, we present two novel optical wireless-based cochlear implant architectures: (i) optical wireless cochlear implant (OWCI) and (ii) all-optical cochlear implant (AOCI). Both the architectures aim to decisively improve the reliability and energy efficiency of hearing restoration devices. To provide design and development guidelines, we document their main components, discuss the particularities of the transdermal optical channel, and provide the analytical framework for their accurate modeling. Building upon this framework, we extract closed-form formulas that quantify the communication, the stimulation, and the overall performance. An overall comparison of OWCI and AOCI, as well as conventional cochlear implants, accompanied by future research directions summarizes this chapter. Our findings reveal that both the OWCI and the AOCI outperform conventional cochlear implant approaches; thus, they are identified as promising architectures for the next generation of cochlear implants.

Keywords: all-optical cochlear implants, biomedical applications, cell stimulation, neural stimulation, optical wireless cochlear implants, optical wireless communications, optogenetics

1. Introduction

The healthy ear functions much like a receiver (Rx) of acoustic signals, which can be described as time-varying pressure waves in a specific frequency range (20–20,000 Hz). These signals propagate toward the cochlear, which analyzes them based on their spectral content. Specifically, each pressure wave traveling inside the cochlea not only actuates inner and outer hair cells at different locations along its length based on the frequency components of the wave, but also determines the intensity of the perceived sound according to the amplitude of the wave [1]. The various spiking characteristics of the spiral ganglion neurons, such as spike rate, number, and location, encode the amplitude and frequency of the sound.

The most common sensory defect is hearing loss, which plagues more than 466 million people around the world and is mostly caused by cochlear abnormalities [2]. When unaddressed, hearing loss can negatively impact the quality of life in various ways, such as social isolation, limited education, and unemployment, which are estimated to cost 980 billion dollars annually. To counterbalance this, substantial research effort has been directed toward neuron regenerative techniques, such as pharmacological, gene, as well as cell therapies [3, 4]. Unfortunately, none of the aforementioned approaches is considered to be close to clinical use. Therefore, the most successful hearing restoration approaches to this day are based on cochlear implants (CIs). Of note, CIs can be used in almost all forms of hearing loss.

Conventional CIs are comprised of two parts: one external and one implanted. The former houses a sound receiver and the processor, while the latter contains the stimulation unit. Specifically, the captured sound signal is decomposed to its major frequency components that are assigned to the corresponding channels of the stimulation unit. Each channel delivers the electrical stimulation signal to the spiral ganglion neurons that match the frequency content of the decomposed electrically encoded sound signal. However, due to the relatively high electrical conductivity of the cochlea, the applied electrical stimulation spreads to nearby spiral ganglion neurons, thus stimulating wider spectral windows than the appropriate one. In conjunction with their low-dynamic range [5], conventional CIs offer limited spectral and intensity sound encoding, which is proven to be detrimental for their hearing restoration capabilities [6].

In this chapter, we introduce the major advances that paved the way for the revolution of CIs and the realization of hearing restoration. Initially, we investigate the current state of the art of hearing restoration through CIs. Next, an in-depth analysis of most promising techniques of light-based hearing restoration is presented. Finally, we offer design guidelines as well as future directions for the next generation of CIs.

2. Background

To aid the reader in understanding the requirements of hearing restoration, we provide some background that covers the CIs' evolution since their conceptualization as well as the current research progresses toward the next generation of CIs (**Figure 1**).

2.1 Evolution of CIs

The concept of hearing restoration through the electrical stimulation of the auditory nerve was conceived by André Djourno and Charles Eyriés in 1957. In their attempt to restore the functionality of the facial nerve through electrical signal applied via a wire, the deaf patient experienced auditory sensations [7]. Based on these findings, multiple attempts were made around the world to develop the first CI with William House performing the first implantation in 1972 [8, 9]. Moreover, the first cochlear implant manufacturing company was founded in 1982 under the name MedEl Corporation, closely followed by Cochlear Limited in 1984, and Advanced Bionics in 1996.

Since their creation, CI companies have iteratively updated their architecture designs, hardware, and optimizing stimulation techniques. The first generation of CIs was released in the early 1980s and included Nucleus 22 and Comfort CI, combined analog signal processing strategies with a multichannel stimulation unit that housed 22 and 4 channels, respectively. These designs were followed by the initial model of Advanced Bionics called Clarion in 1996 that was encased in a ceramic case, contained

Figure 1.
The evolution of CIs from 1982, when the first CI manufacturing company was founded, until the current state-of-the-art research that validated the feasibility of optogenetics-enabled optical CIs.

eight channels, and used rechargeable batteries. The second generation included Clarion II, Nucleus 24 Contour, and Combi 40+. These were introduced in the market with 24 electrodes and new sound processors with novel features such as precurved electrode arrays, backward compatibility, frequency modulation capabilities, dual electrodes, and behind-the-ear external components. However, in the early 2000s, completely redesigned highly customizable CI models, namely, Freedom, Pulsar, and HiRes90k, were developed. Their modularity and customization options were the distinguishing factors for these new models that were available in straight or precurved, standard, medium, condensed, and split electrode array architectures, based on the individual particularities of the cochlea of each patient. Moreover, these electrodes were encased in flexible plastic and housed a plastic tip that enabled nontraumatic implantation. In the era after 2010, the latest iterations of CIs have been focused toward higher fidelity sound that enhances the perception of music through state-of-the-art sound processing, wireless control, and software-enabled programability, as well as waterproof designs.

2.2 What the next-generation CI should be?

The utilization of light-based communications and stimulation has been proposed as a promising alternative for electrical hearing restoration techniques. The superior communication performance of optical wireless communications in trandermal applications revealed the benefits that can be achieved by utilizing light for the communication between the external and implanted components of CIs [10, 11]. Moreover, optogenetics was initially reported by Izzo et al. [12] and has been proven achieve more efficient coding of the spectral information of sound due to its higher temporal confinement compared to the electronic stimulation techniques [13–15]. Although optical stimulation has great potential, it exhibits increased energy requirements for achieving the actuation of spiral ganglion neurons, and thus, future research is necessary for developing more energy-efficient techniques [16]. Finally, the combination of optogenetics and optical wireless communications offers great promise for the realization of an all-optical CI architecture capable of achieving unprecedented performance [17].

3. Current research progress

Two main research directions remain to be investigated. First, transdermal communication plays an important role in propagating the sound information captured by the microphone of the external component toward the one implanted. Conventional CIs are based on magnetic coupling, a near-field technique that uses low radio frequencies (RFs) in the range of 5–50 MHz for communication [18, 19]. The required power of conventional CIs lies around some decades milliwatts. Although this technology has been successfully applied in the majority of CIs, it suffers from low data rates, which constrain the performance of artificial hearing aids in their attempt to simulate high-quality normal hearing [20–22]. In addition, the aforementioned spectral window is used by numerous applications, which generate a great amount of interference that diminishes the quality of communication [23–25]. On the other hand, the optical activation of the auditory nerve via optogenetics has been experimentally verified, but the propagation of the spiral ganglion neuron potential through the auditory pathway toward the brain and its successful perception have yet to be demonstrated [26]. Moreover, the superiority of optical over electrical cell stimulation must be validated in order to justify the research effort toward the all-optical cochlear implant (AOCI) [17]. Recently, multiple experiments have progressed these goals by implanting novel tiny optical fibers in animals models of human sensorineural hearing loss [27–29].

3.1 Communications

To overcome the aforementioned CI restrictions, researchers have investigated the viability of transdermal wireless networks that operate in nonstandard frequencies. Owing to increased bandwidth, surprisingly high tolerance to external interference, and partial skin transparency at near-infrared wavelengths, optical wireless communications have been applied to transdermal channels instead of the traditional RF-based techniques [30, 31]. In the past decade, numerous contributions have experimentally verified the practicality of transdermal optical links [32–36]. Abita established a transdermal optical link from the inside toward the outside component of a medial system achieving high-data-rate communications [32], while Ackermann et al. investigated the design principles and tradeoffs that are entangled to optical-based CIs [33, 37]. Moreover, Liu introduced a high-data-rate transdermal optical link for implantable biomedical systems with high energy efficiency under the assumption of deterministic misalignment [24]. Similarly, the interactions between data rate, transmission power, receiver characteristics, and tissue thickness as well as their impact on the system's performance were evaluated for transdermal optical links applied in neural signal extraction scenarios [38]. In addition, the same authors validated the proposed system by conducting *in vivo* experiments that achieved 2×10^{-7} bit error rate (BER) and 100-Mbps data rate under stochastic misalignment, but with relatively high power consumption in the order of 2 mW [36]. On the contrary, a novel retroflective architecture was presented for transdermal optical links [34], while Liu proposed a bidirectional transdermal optical link [35].

Building upon the aforementioned contributions, the development of optical-based CIs needs to leverage breakthrough technologies while taking into consideration the particularities of the transdermal and in-body optical channels, the space and energy design limitations, as well as the directionality of the optical links. Moreover, a novel information-theoretic framework is required for the design of energy-efficient physical and medium access schemes, as well as the development of simultaneous light information and power transfer policies and resource allocation strategies. Motivated by the above, recent research effort has been devoted toward delivering safety and high quality of experience in CIs and identifying the critical technology gaps and the appropriate enablers.

3.2 Neural stimulation

After communicating the information from the external environment toward the implanted component of the CI, the techniques of neural stimulation must be applied in order to excite the cochlear spiral ganglion neurons, which, in turn, will generate the desire action potential that will propagate along the acoustic nerve toward the brain. Over the years, various methods of neural stimulation have been developed. These can be categorized based on the nature of the applied stimulus as acoustic, thermal, magnetic, chemical, optical, and electrical, with the last ones being the most recognized [39]. Specifically, electrical neural stimulation is the most common technique and has been used in a wide gamut of biomedical applications [40–42]. Electrical neural stimulation applies an electrical stimulus (voltage, current [40], or charge [42]) on the target nerves that manipulates their membrane potential so that it exceeds a certain threshold and, therefore, generates or inhibits action potentials. Specifically, deep brain and cardiac muscle stimulation techniques that use voltage control mechanisms have been investigated with regard to power consumption [41, 43], while current-controlled electrical neural stimulation in CIs is characterized by power waste in the tissue that leads to limited longevity and tissue damage [40]. Voltage-controlled electrical neural stimulation is proven to be more power efficient and less complex, but with very limited stimulus tuning options that result in faster degradation of the electrode contacts. The opposite is valid for current-controlled electrical neural stimulation that can apply fine-tuned charge to the electrodes but exhibits lower power efficiency. Finally, charge control mechanisms for electrical neural stimulation have been applied on the peripheral neural system [42] and offer a middle ground between stimulus control and power consumption. Despite the control mechanism, the determining factors of electrical neural stimulation techniques include human safety, energy efficiency, stimulation waveform, and spatial resolution. The latter significantly affects the stimulation accuracy and is correlated with the distance from the targeted neurons as well as the size of the electrode, which is limited by maximum permissible charge per tissue surface and the electrode's manufacturing process. In addition, the unique characteristics of different types of neurons greatly affect their response to stimulations with variable waveform properties, such as amplitude, width, and frequency. To this end, a great amount of research effort has been devoted toward optimizing the waveform for the stimulus [44–47]. Finally, throughout the optimization procedure of electrical neural stimulation techniques, safety for humans must be ensured.

The solution to the several limitations of electrical neural stimulation was introduced almost two decades ago in the form of optical neural stimulation that uses light for the actuation and control of neurons. Specifically, light-gated ion channels found in proteins, termed opsins, have been proven to mediate light-driven action potentials in mammalian neurons by manipulating the polarization of their membrane and, therefore, suppressing or exciting them. Optical neural stimulation is highly dependent on the type of the utilized opsin, which incentivized research toward experimentally verifying its performance in terms of precision, accuracy, frequency, and scalability [48–51]. Optical neural stimulation was successfully applied in the motor control system of rodents [48], while the causal relationship of frequency-based optical neural stimulation and behavior state transitions was verified [49]. The increased specificity of exciting neurons was illustrated through efficiently mapping the spatial distribution of synaptic inputs [50]. Moreover, a high-precision optical neural stimulation technique for inhibiting neurons with temporal fidelity was developed [51]. The performance of this technique was evaluated based on novel key performance indicators such as light sensitivity. The aforementioned works illustrate that the development of opsins offering stable performance over multiple stimulations is accompanied by long desensitization periods and short channel-off durations. To this end, research was intensified toward developing opsins with different kinetic features and wavelength sensitivity for monitoring and controlling biological processes in subcellular and cellular levels [52, 53]. A major breakthrough was achieved with the application of channelrhodopsin 2 (ChR2) in mammalian neurons that enabled accurate stimulation with light pulses [54]. Since its development, ChR2 has been heavily investigated, and multiple variants have been introduced with applications in cardiology [55–58] and neuroscience [59, 60]. The performance of these variants greatly outperforms electrical neural stimulation in terms of stimulation pulse intensity and frequency (up to 200 Hz), as well as the ability to trigger large current action potentials with higher fidelity [61, 62].

4. Light-based hearing restoration

Based on the literature review presented in the previous section, the main bottlenecks of CIs are low accuracy and low precision of nerve stimulation methods, bandwidth scarcity and constraint capacity of RF communication techniques, and high energy consumption of both. To this end, we present two architectures capable of mitigating the effect of these limitations and even eliminating them [11, 17].

4.1 Optical wireless cochlear implant

The utilization of optical wireless communications in order to develop CI transdermal optical links has been recently investigated [11], where the authors proposed a novel system architecture, termed optical wireless cochlear implant (OWCI), that improves the power and spectral efficiency as well as the reliability of the transdermal optical link. Moreover, in the same contribution [11], the capabilities and feasibility of the OWCI are evaluated and design guidelines are provided. The main comparison points between OWCIs and conventional CIs are illustrated in **Table 1**. In addition, the presented advances in the communications of CIs are in line with optical neural stimulation advances on the acoustic nerve [21, 63–67].

The unique technical contributions of the OWCI entail the establishment of a novel system model for transdermal optical links that incorporates the various design variables such as the stochastic misalignment between the receiver (RX) and the transmitter (TX), the scale of the optical components, the skin thickness, and the

DOI: http://dx.doi.org/10.5772/intechopen.104622

OWCIs	Conventional CIs
Increased data rate	Low data rate
Abundant bandwidth	Limited bandwidth
High power efficiency	Low power efficiency
Safer for the human body	Questionable safety
Mature technology with promise of	Mature technology
higher performance in the same scale	with compact designs
Low solar and ambient light interference	Very high electronic interference
Stringent alignment requirements	**Susceptible misalignment**
Multiple design guidelines	**Mature standardization**
(IrDA, EU COST 1101, IEC LSS, IEEE Std 1073.3.2-2000, etc.)	**(IEEE Std 1073.3.2-2000, IEEE 802.15.4, etc.)**

Table 1.
OWCIs versus conventional CIs (bold fonts demonstrate the advantages).

transmission power. The external component of the OWCI is comprised of a microphone, the TX, and a digital signal processing (DSP) unit, while the implanted one contains the RX as well as a stimulation and a DSP unit. The external DSP unit is responsible for digitizing and compressing the sound signal from the microphone into coded signals, which are then forwarded from the TX to the RX over the transdermal optical link. In the implanted component, the DSP and stimulation units transform the received signal into a series of electrical pulses that will stimulate the auditory nerve (**Figure 2**). Based on this system model, the performance of the OWCI was evaluated with regard to the SNR, channel capacity, outage probability, and spectral efficiency. The results not only validated the feasibility of the proposed architecture and provided meaningful insights that can be used as design guidelines, but also revealed the superior effectiveness and reliability of the OWCI compared to the conventional CI.

In the aforementioned architecture, the transmitted signal, x, is conveyed over the wireless channel, h, with additive noise n. Thus, the received signal can be written as [68–70].

$$y_1 = Rhx + n \tag{1}$$

with η denoting the quantum efficiency of the photodiode, R the RX's responsivity, v the frequency of the photons, q the electron charge, and p the Planck's constant. It is highlighted that the channel coefficient can be expressed as $h = h_l \; h_p$, where h_l represents the deterministic channel coefficient caused by propagation loss, while h_p denotes the collected power fraction due to the geometric spread from the origin of the detector and is caused from the TX-RX misalignment (**Figure 3**).

The CI channel's deterministic term can be expressed as in ([71], Eq. (10.1))

$$h_l = \exp\left(-(\mu_\alpha(\lambda) + \mu_s(\lambda))\delta\right) \tag{2}$$

where λ is the transmission wavelength, δ is the skin thickness, $\mu_\alpha(\lambda)$ is the skin attenuation coefficient, and $\mu_s(\lambda)$ is the skin scattering coefficient, which can be acquired from a plethora of experimental results [72–76]. In this analysis, the term skin refers to the biological structure that consists from the stratum corneum, the

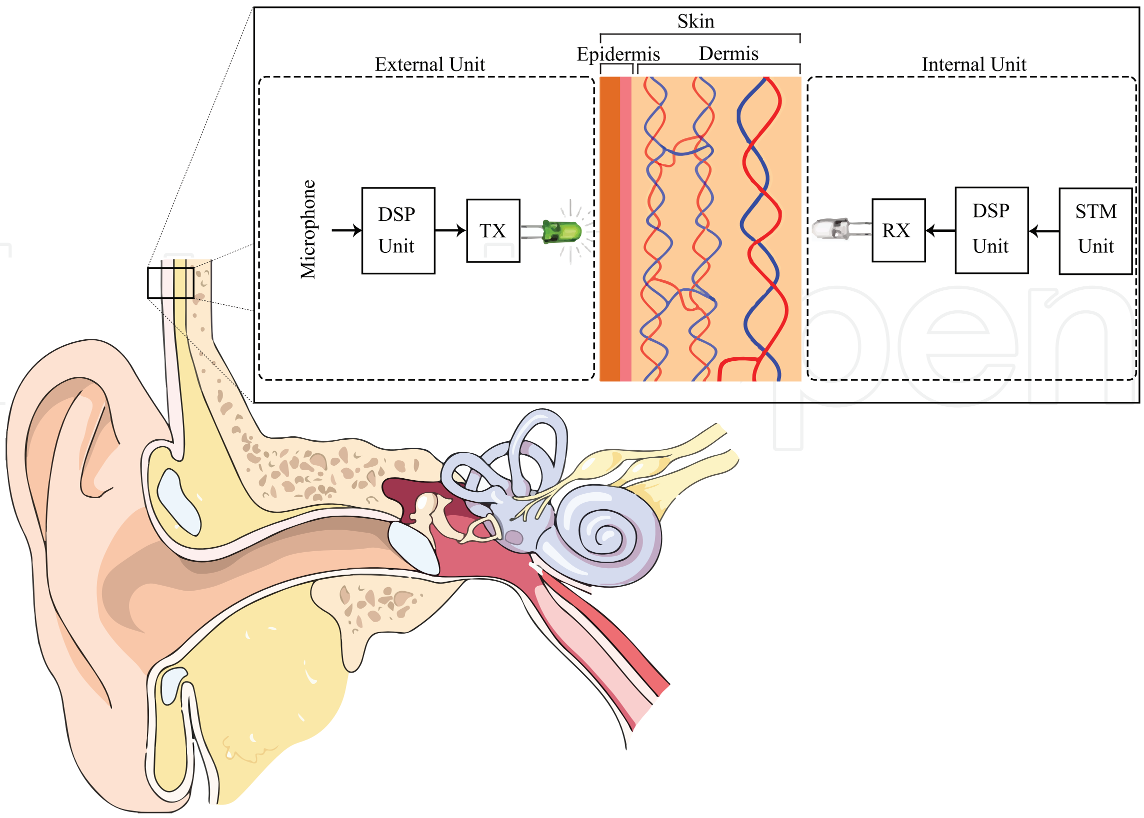

Figure 2.
Diagrammatic illustration of the architecture of the OWCI. The OWCI captures the sound information via the microphone located outside of the human body. Afterward, it utilizes optical wireless communications to transfer it toward the receiver fixed on the cranial bone. Finally, the implanted unit stimulates the acoustic nerve by delivering the appropriate signals via the stimulation electrode.

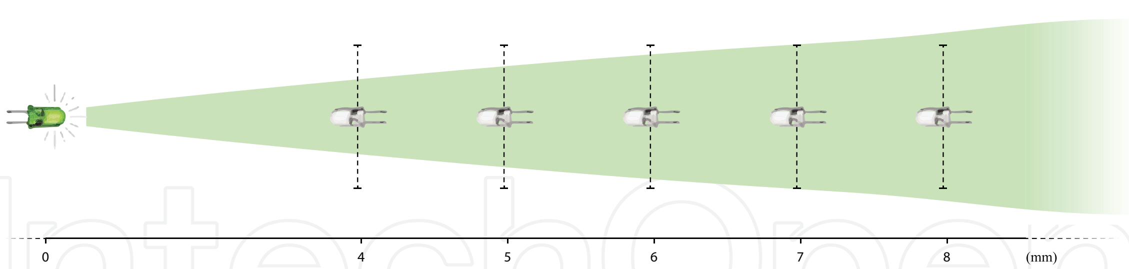

Figure 3.
The effect of misalignment with regard to skin thickness. As the RX (photodiode) moves away from the TX under the same severity of misalignment, its distance from the perfect alignment conditions becomes enclosed in the TX's beam width. This phenomenon creates an equilibrium between the optimal TX-RX distance and the TX beam width under fixed misalignment conditions.

epidermis, and the dermis [71], while both the RX and the TX touch with the inner (adipose) and outer (epidermal) side of the skin [24], and thus, the TX-RX distance is regarded equivalent to skin thickness.

The misalignment between the TX and the RX can be modeled as the stochastic channel coefficient, which expresses the collected power due to geometric spread with radial displacement r from the origin of the detector and can be written as

$$h_p \approx A \exp\left(-\frac{2r^2}{w_e^2}\right), \tag{3}$$

which is based on the assumption that at distance d from the TX, the circular aperture of the transmitted beam has a radius of R and the spatial intensity on the plane of the RX is w_d. In addition, w_e represents the equivalent beam waist radius and A expresses the collected power under perfect alignment. This approximation has been utilized in various previous works for modeling stochastic pointing errors [77, 78].

Based on this model, if we assume independently and identically Gaussian distributed horizontal and vertical displacement, it has been proven that r follows a Rayleigh distribution [79]. As a result, the probability density function (PDF) of the stochastic term of the channel coefficient can be written as

$$f_{h_p}(x) = \frac{\gamma}{A^{\gamma}} x^{\gamma - 1}, \qquad 0 \le x \le A, \tag{4}$$

where

$$A = (\mathrm{erf}(\beta))^2, \quad \beta = \frac{\sqrt{\pi} R}{\sqrt{2} w_d}, \quad \gamma = \frac{w_e^2}{4\sigma^2}, \quad w_e^2 = w_d^2 \frac{\sqrt{\pi}\,\mathrm{erf}(\beta)}{2\beta \exp\left(-\beta^2\right)}, \tag{5}$$

while σ^2 denotes the variance of the misalignment.

4.2 All-optical cochlear implant

The CI implementations presented so far counterbalance either the RF scarcity that plagues the communications part of the system or the nerve stimulation limitations. To this end, the AOCI has been proposed as an architecture that converts the audio captured from the microphone into a light signal inside the external component for propagation to the cochlea [17] (**Figure 4**). This way, the AOCI counterbalances the aforementioned challenges and, at the same time, eliminates the need for an energy-consuming DSP unit in the implanted component. The AOCI not only builds upon the fruitful characteristics of the OWCI but also proposes breakthrough alterations such as the fact that it consists of only passive components, and thus, the implanted component has no power demands, which eliminates the requirement of complex power transfer policies and boosts energy efficiency. Furthermore, the AOCI utilizes optical neural stimulation, which is characterized by higher fidelity than electrical neural stimulation due to the lower spread of optical signals in human tissues. The technical advancements include the introduction of the AOCI architecture, its main building blocks, and the end-to-end system model. The AOCI takes into account channel, building block, and biological particularities [17]. Moreover, a novel tractable expression is derived for the instantaneous coupling efficiency in scenarios with misalignment fading. The feasibility of the proposed architecture is proven through the theoretical framework, which also evaluates its performance with regard to the power efficiency, the photon flux, and a plethora of design parameters that greatly influence the success or failure of the system.

Much like OWCI, the architecture of AOCI consists of the implanted and the external component, with the former located on the skull and the latter on the external surface of the skin. The external component captures the acoustic signal with a microphone, performs the necessary DSP, and converts it into the appropriate optical signal capable of generating the desired action potentials on the targeted spiral ganglion neurons. This signal is transmitted from the TX, which is a laser source, to the implanted component, where the guiding lens, the microelectromechanical device,

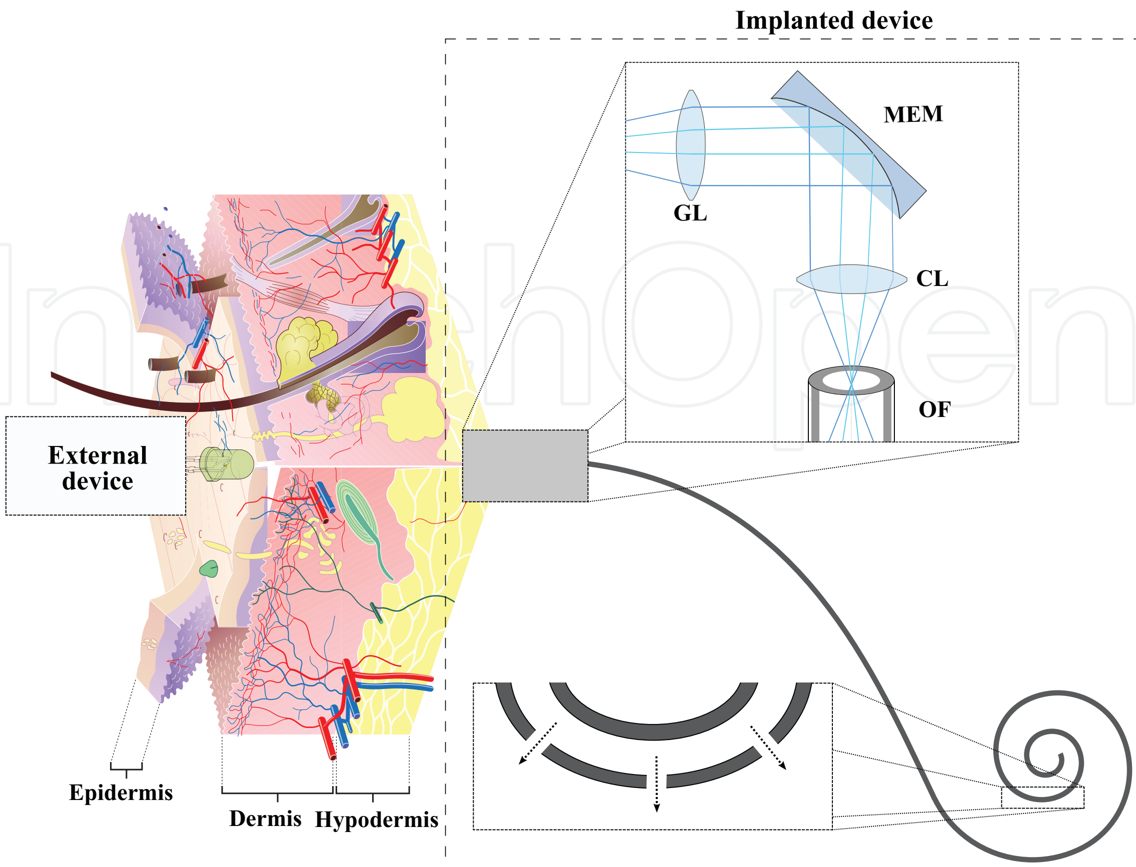

Figure 4.
Illustration of the architecture of the AOCI. The all-optical nature of the AOCI resides in the combined utilization of optical wireless communications and optogenetics for stimulating the auditory nerve. Initially, the auditory neurons are sensitized to optical radiation with optogenetic techniques. Next, the sound captured from the external microphone is converted into an optical signal capable of stimulating the light-sensitive nerves, which is then forwarded to the cochlea.

the coupling lens, and the optical fiber ensure its delivery to the appropriate place in the cochlea. Specifically, the guiding lens guides the light toward the microelectromechanical device to maximize the power of the received optical signal. Afterward, the microelectromechanical acts as a mirror that mitigates the misalignment to a degree by steering the light beam to the center of the coupling lens in order to be coupled into the optical fiber. Finally, the latter delivers the light into specific points along the cochlea based on their spectral content.

4.2.1 Microelectromechanical device

Microelectromechanical devices have been the subject of much hype during the past decade due to their adaptability as well as low cost, low weight, and small size [80–82]. In the case of the AOCI, the microelectromechanical device is required in order to account for the individuality of each patient. In particular, the AOCI is required to adapt to the particularities of the patient, such as different skin thickness and color or slightly varied orientation of biological tissues, in order to ensure uninterrupted hearing restoration. Moreover, imperfections during the implantation process can cause slight variations to the final placement of the implant. To this end, the microelectromechanical device provides an externally operated light control system by enabling the steering of the optical beam toward the coupling lens. Finally, the microelectromechanical device adjusts its optical properties and, thus, steers the beam after receiving the appropriate electrical charge that can be applied during implantation, while in normal operation, the need for adjustment is eliminated, and therefore, the microelectromechanical device operates passively [83, 84].

The signal received by the guiding lens presented in (1) is forwarded to the microelectromechanical device, which introduces a collimation gain [85].

$$G_c = \frac{1}{\sqrt{(1 - d_{in}/f)^2 + z_0^2/f^2}}. \tag{6}$$

Therefore, the updated received signal at the output of the microelectromechanical device can be expressed as

$$y_2 = G_c h_l h_p x + n. \tag{7}$$

4.2.2 Coupling lens

The coupling lens receives the optical beam from the microelectromechanical device and focuses it in the center of the optical fiber. The fact that incident light on the end of the optical fiber that arrives at a greater angle than the acceptable angle of the optical fiber is not coupled highlights the detrimental impact it plays on the maximum achievable coupling efficiency of the system. Moreover, the coupling efficiency is also affected by the dimensions of the coupling lens and the diameter of the optical fiber with its maximum value being in the order of 80% [86].

The coupling lens captures the optical signal that is reflected by the microelectromechanical device and couples it into the optical fiber. The signal that successfully enters the optical fiber can be written as

$$y_3 = \eta G_c h_l h_p x + n \tag{8}$$

with the coupling efficiency given by

$$\eta = \left(\frac{3.83\sqrt{2} D \omega_0}{1.22 \lambda F} \exp\left(-\frac{r^2}{\omega_0^2} \right) \Psi_2 \left(1; 2, 1; -\frac{3.83^2 D^2 \omega_0^2}{1.22^2 \lambda^2 F^2}, \frac{r^2}{\omega_0^2} \right) \right)^2. \tag{9}$$

In Eq. (9), ω_0, F, D, and ρ denote the optical fiber mode field radius, the focal length, the focusing lens diameter, and the radial distance on the focal plane, respectively, while it becomes obvious that the achievable coupling efficiency is dependent on the optical fiber's mode field radius, the coupling lens's focal length and diameter, as well as the intensity of misalignment and the transmission wavelength.

4.2.3 Optical fiber

The optical fiber of the AOCI takes the place of the electrode array of the conventional CI. The incident optical signal must be delivered to specific locations alongside the cochlea in order to generate action potentials at the targeted spiral ganglion neurons that are responsible for the appropriate sound frequency. To achieve the required performance, the optical fiber proposed in the AOCI architecture propagates the optical signal through its single-mode core with a Gaussian beam profile in the output [87, 88]. Furthermore, despite the fact that state-of-the-art conventional CIs can be equipped with a maximum of 20 electrodes, due to the limited spatial resolution of electrical neural stimulation, the sound perceived by the patient has the fidelity of eight functional electrodes [40]. In addition, to achieve speech and music perception under suboptimal noise constraints, CI must house approximately 32 electrodes, which is also the goal of the AOCI [89, 90]. Therefore, tilted fiber Bragg gratings (FBGs) were introduced in the AOCI architecture that enable light delivery in various locations alongside the optical fiber [91, 92]. These FBGs are located in the core of the optical fiber, along the propagation direction, with a periodic variation of the refractive index. These components have low insertion loss, low complexity structures, and high wavelength selectivity. Specifically, tilted FBGs allow a limited number of wavelengths to penetrate them by filtering the incident optical signal based on its spectral content and, at the time, redirecting it based to their angle [93, 94].

When the optical signal travels through the optical fiber, it attenuates due to the curvature of the optical fiber and the existence of FBGs, and therefore, the emitted signal can be expressed as

$$y_4 = k\eta G_c h_l h_p x + n, \tag{10}$$

where k denotes the propagation efficiency, which is limited to $0.14\,\text{dB}/90^{\circ}$ by the strong optical confinement of microfiber, even for increased bending radius or index values [95]. In addition, k incorporates the signal attenuation due to the existence of FBGs, which has been proven to be in the order of 10% [96].

5. The road ahead

From the presented analysis, it is evident that, despite their extensive applications, electrical neural stimulation techniques suffer from insufficient coding of spectral information, low power efficiency, low stimulation precision, accuracy, and frequency, as well as questionable safety. To this end, promising optical neural stimulation methods that surpass these limitations have been proposed. In an effort to establish these methods, the scientific community has pushed toward proving their feasibility as well as theoretically modeling and augmenting them. The state of the art of optical neural stimulation techniques offers great promise toward realizing next-generation biomedical systems.

One of the main offerings of optical neural stimulation is the outstanding stimulation precision it offers compared to electrical neural stimulation. In more detail, the increased precision can be translated into higher customization of the produced neural activity in two respects. First, the increased stimulation frequency that comes with optical neural stimulation leads to higher accuracy of excitation due to the fact that action potentials are delivered faster to the target spiral ganglion neurons and, therefore, to the brain, thus limiting the time between sound acquisition and perception. Second, optical neural stimulation depends on the optical particularities of light sensitive opsins with each one being expressed in a specific type of cell. Therefore, this offers another layer of light selectivity that can be leveraged by optical neural stimulation techniques [97]. The combination of these two aspects equips optical neural stimulation with the necessary tools to achieve unprecedented performance not only in the field of hearing restoration but also in other biomedical application such as retinal implants that would utilize this advantage to provide higher perceived image fidelity.

Another aspect that boosts the performance of optical neural stimulation is the exceptional spectral coding of the information carried by the optical signal. On the contrary to electrical neural stimulation techniques that are characterized by wide current spread from the electrode contacts, optical radiation attenuates with a greater rate when it propagates inside human tissue, and therefore, the applied optical stimulations are more spatially confined than electrical ones. The importance of this phenomenon is highlighted even more by the fact that human sound perception requires at least 32 stimulation channels in order to recognize music or sound in noisy environments [89, 90]. As a result, the superior spectral coding of optical neural stimulation enables support for stimulation units that can house significantly more channels.

Contrary to previous detrimental improvements offered by optical neural stimulation methods, their performance in terms of power efficiency is comparable to the one of electrical neural stimulation. In more detail, optimization is required for optical neural stimulation stimulation policies in order to achieve similar power consumption as electrical neural stimulation [58]. Therefore, the optimization of optical neural stimulation techniques in terms of their power demands is one of the key requirements for their successful application in future biomedical applications. Similarly, the safety and ethical concerns of optical neural stimulation pose another controversial aspect. On the one hand, the optical power that is required for the reliable activation of light-sensitive spiral ganglion neurons is below the limits defined in various standardization protocols [98]; on the other hand, the modification of the targeted spiral ganglion neurons in order to acquire light sensitivity poses ethical concerns.

From a purely biological perspective, action potentials generated from electrical stimulation signals resemble the morphology and waveform of the membrane potential. As a result, these electrical signals are superimposed on each other and become almost indistinguishable, which hinders hearing restoration [58]. However, owing to its core functionality, optical neural stimulation triggers action potentials that differ significantly from membrane potential based on the stimulation protocol and the type of the excited cell. Specifically, not only the waveform of the generated action potential is affected by the amplitude and the duration of the stimulation, but also the instant release of ions when opsins are illuminated, which causes the membrane to react immediately. In addition, each opsin-cell-type combination is characterized by a distinct morphology of transmembrane potential and in conjunction with the wide variety of opsins available; they ensure the generation of a distinct action potential.

Finally, from an engineering point of view, the plethora of opsins that have been developed can highly impact the performance of optical neural stimulation biomedical applications. All future research in this field should take into careful consideration the selection of the applied opsin, as suboptimal ones may result in low stimulation precision and reliability, which, in turn, can determine whether the application is successful or not. The most important design choices include the compatibility with the target cell type, the amplitude and morphology of the resulting action potential, and the nature and the direction of the released ions.

6. Conclusions

In this chapter, we have provided a vision for hearing restoration from an engineering point of view that could serve as a guide in the research and development of the next-generation CIs. We suggest that the future of digital hearing restoration lies in the optical spectrum, both in terms of communication and stimulation techniques.

We envisioned and explained potential architectures that enable the utilization of optical technologies in CIs. Finally, we introduced key features and performance indicators that could decide their success or failure.

Conflict of interest

The authors declare no conflict of interest.

Nomenclature

AOCI	all-optical cochlear implant
BER	bit error rate
ChR2	channelrhodopsin 2
CI	cochlear implant
DSP	digital signal processing
FBG	fiber Bragg grating
LED	light-emitting diode
MPE	maximum permissible exposure
OWCI	optical wireless cochlear implant
RF	radio frequency
SNR	signal-to-noise ratio

Author details

Stylianos E. Trevlakis[1], Alexandros-Apostolos A. Boulogeorgos[2]* and George K. Karagiannidis[1]

1 Department of Electrical and Computer Engineering, Aristotle University of Thessaloniki, Thessaloniki, Greece

2 Department of Digital Systems, University of Piraeus, Piraeus, Greece

*Address all correspondence to: al.boulogeorgos@ieee.org

References

[1] Moser T, Grabner CP, Schmitz F. Sensory processing at ribbon synapses in the retina and the cochlea. Physiological Reviews. 2020;**100**(1):103-144

[2] Deafness and Hearing Loss. Available from: https://www.who.int/news-room/fact-sheets/detail/deafness-and-hearing-loss

[3] Samarajeewa A, Jacques BE, Dabdoub A. Therapeutic potential of Wnt and Notch signaling and epigenetic regulation in mammalian sensory hair cell regeneration. Molecular Therapy. 2019;**27**(5):904-911

[4] Roccio M, Senn P, Heller S. Novel insights into inner ear development and regeneration for targeted hearing loss therapies. Hearing Restoration. 2020; **397**:107859

[5] Miller CA, Abbas PJ, Robinson BK, Nourski KV, Zhang F, Jeng FC. Electrical excitation of the acoustically sensitive auditory nerve: Single-fiber responses to electric pulse trains. Journal of the Association for Research in Otolaryngology. 2006;**7**(3):195-210

[6] Caldwell MT, Jiam NT, Limb CJ. Assessment and improvement of sound quality in cochlear implant users. Laryngoscope Investigative Otolaryngology. 2017;**2**(3):119-124

[7] Djourno A, Eyries C. Auditory prosthesis by means of a distant electrical stimulation of the sensory nerve with the use of an indwelt coiling. La Presse Médicale. 1957;**65**:1417

[8] House WF, Urban J. Long term results of electrode implantation and electronic stimulation of the cochlea in man. The Annals of Otology, Rhinology, and Laryngology. 1973;**82**(4):504-517

[9] Burian K. Letter: Significance of cochlear nerve electric stimulation in totally deaf patients. Laryngologie, Rhinologie, Otologie. 1975;**54**:530-531

[10] Trevlakis SE, Boulogeorgos AAA, Karagiannidis GK. On the impact of misalignment fading in transdermal optical wireless communications. In: 2018 7th International Conference on Modern Circuits and Systems Technologies (MOCAST). Thessaloniki, Greece: IEEE; 2018

[11] Trevlakis SE, Boulogeorgos AAA, Sofotasios PC, Muhaidat S, Karagiannidis GK. Optical wireless cochlear implants. Biomedical Optics Express. 2019;**10**(2):707

[12] Izzo AD, Walsh JT, Jansen ED, Bendett M, Webb J, Ralph H, et al. Optical parameter variability in laser nerve stimulation: A study of pulse duration, repetition rate, and wavelength. IEEE Transactions on Biomedical Engineering. 2007;**54**(6): 1108-1114

[13] Moser T. Optogenetic stimulation of the auditory pathway for research and future prosthetics. Current Opinion in Neurobiology. 2015;**34**:29-36

[14] Jeschke M, Moser T. Considering optogenetic stimulation for cochlear implants. Hearing Research. 2015;**322**: 224-234

[15] Trevlakis SE, Boulogeorgos AAA, Chatzidiamantis ND, Karagiannidis GK, Lei X. Electrical vs optical cell stimulation: A communication perspective. IEEE Access. 2020;**8**: 192259-192269

[16] Baumhoff P, Kallweit N, Kral A. Intracochlear near infrared stimulation:

Feasibility of optoacoustic stimulation in vivo. Hearing Research. 2019;**371**: 40-52

[17] Trevlakis SE, Boulogeorgos AAA, Chatzidiamantis ND, Karagiannidis GK. All-optical Cochlear implants. IEEE Transactions on Molecular, Biological and Multi-Scale Communications. 2020; **6**(1):13-24

[18] Zeng FG, Rebscher S, Harrison W, Sun X, Feng H. Cochlear implants: System design, integration, and evaluation. IEEE Reviews in Biomedical Engineering. 2008;**1**:115-142

[19] Agarwal K, Jegadeesan R, Guo YX, Thakor NV. Wireless power transfer strategies for implantable bioelectronics. IEEE Reviews in Biomedical Engineering. 2017;**10**:136-161

[20] Kim HJ, Hirayama H, Kim S, Han KJ, Zhang R, Choi JW. Review of near-field wireless power and communication for biomedical applications. IEEE Access. 2017;**5**:21264-21285

[21] Thompson AC, Wade SA, Pawsey NC, Stoddart PR. Infrared neural stimulation: Influence of stimulation site spacing and repetition rates on heating. IEEE Transactions on Biomedical Engineering. 2013;**60**(12):3534-3541

[22] Ko WH. Early history and challenges of implantable electronics. Journal on Emerging Technologies in Computing Systems. 2012;**8**(2):1-9

[23] Islam MN, Yuce MR. Review of medical implant communication system (MICS) band and network. ICT Express. 2016;**2**(4):188-194

[24] Liu T, Bihr U, Anis SM, Ortmanns M. Optical transcutaneous link for low power, high data rate telemetry. In: Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC). San Diego, CA, USA: IEEE; 2012. pp. 3535-3538

[25] Pinski SL, Trohman RG. Interference in implanted cardiac devices, Part I. Pacing and Clinical Electrophysiology. 2002;**25**(9):1367-1381

[26] Dieter A, Keppeler D, Moser T. Towards the optical cochlear implant: Optogenetic approaches for hearing restoration. Molecular Medicine. 2020;**12** (4):e11618

[27] Wrobel C, Dieter A, Huet A, Keppeler D, Duque-Afonso CJ, Vogl C, et al. Optogenetic stimulation of cochlear neurons activates the auditory pathway and restores auditory-driven behavior in deaf adult gerbils. Science Translational Medicine. 2018;**10**(449): eaao0540

[28] Keppeler D, Merino RM, de la Morena DL, Bali B, Huet AT, Gehrt A, et al. Ultrafast optogenetic stimulation of the auditory pathway by targeting-optimized Chronos. The EMBO Journal. 2018;**37**(24):e99649

[29] Mager T, de la Morena DL, Senn V, Schlotte J, Feldbauer K, Wrobel C, et al. High frequency neural spiking and auditory signaling by ultrafast red-shifted optogenetics. Nature Communications. 2018;**9**(1):1750

[30] Ghassemlooy Z, Alves LN, Zvanovec S, Khalighi MA. Visible Light Communications: Theory and Applications. Boca Raton: CRC Press; 2017

[31] Chowdhury MZ, Hossan MT, Islam A, Jang YM. A comparative survey of optical wireless technologies: Architectures and applications. IEEE Access. 2018;**6**:9819-9840

[32] Abita JL, Schneider W. Transdermal optical communications. Johns Hopkins APL Technical Digest. 2004;**25**(3):261

[33] Ackermann DM Jr, Smith B, Wang XF, Kilgore KL, Peckham PH. Designing the optical interface of a transcutaneous optical telemetry link. IEEE Transactions on Biomedical Engineering. 2008;**55**(4):1365-1373

[34] Gil Y, Rotter N, Arnon S. Feasibility of retroreflective transdermal optical wireless communication. Applied Optics. 2002;**51**(18):4232-4239

[35] Liu T, Anders J, Ortmanns M. Bidirectional optical transcutaneous telemetric link for brain machine interface. Electronics Letters. 2015; **51**(24):1969-1971

[36] Duncan K, Etienne-Cummings R. Selecting a safe power level for an indoor implanted UWB wireless biotelemetry link. In: IEEE Biomedical Circuits and Systems Conference (BioCAS). Rotterdam, Netherlands: IEEE; 2013. pp. 230-233

[37] Ackermann DM, Smith B, Kilgore KL, Peckham PH. Design of a high speed transcutaneous optical telemetry link. In: Engineering in Medicine and Biology Society, 2006. EMBS'06. 28th Annual International Conference of the IEEE. New York, NY, USA: IEEE; 2006. pp. 2932-2935

[38] Liu T, Anders J, Ortmanns M. System level model for transcutaneous optical telemetric link. In: IEEE International Symposium on Circuits and Systems (ISCAS). Beijing, China: IEEE; 2013. pp. 865-868

[39] Luan S, Williams I, Nikolic K, Constandinou TG. Neuromodulation: Present and emerging methods. Frontiers in Neuroengineering. 2014;7:27

[40] Srinivasan AG, Landsberger DM, Shannon RV. Current focusing sharpens local peaks of excitation in cochlear implant stimulation. Hearing Research. 2010;**270**(1–2):89-100

[41] Hardesty DE, Sackeim HA. Deep brain stimulation in movement and psychiatric disorders. Biological Psychiatry. 2007;**61**(7):831-835

[42] Rosellini WM, Yoo PB, Engineer N, Armstrong S, Weiner RL, Burress C, et al. A voltage-controlled capacitive discharge method for electrical activation of peripheral nerves. Neuromodulation: Technology at the Neural Interface. 2011;**14**(6):493-500

[43] Wong LSY, Hossain S, Ta A, Edvinsson J, Rivas DH, Naas H. A very low-power CMOS mixed-signal IC for implantable pacemaker applications. IEEE Journal of Solid-State Circuits. 2004;**39**(12):2446-2456

[44] Foutz TJ Jr, DMA, Kilgore KL, McIntyre CC. Energy efficient neural stimulation: Coupling circuit design and membrane biophysics. PLoS One. 2012; **7**(12):e51901

[45] Sahin M, Tie Y. Non-rectangular waveforms for neural stimulation with practical electrodes. Journal of Neural Engineering. 2007;**4**(3):227-233

[46] Albensi BC, Oliver DR, Toupin J, Odero G. Electrical stimulation protocols for hippocampal synaptic plasticity and neuronal hyper-excitability: Are they effective or relevant? Experimental Neurology. 2007;**204**(1):1-13

[47] Yip M, Bowers P, Noel V, Chandrakasan A, Stankovic KM. Energy-efficient waveform for electrical stimulation of the cochlear nerve. Scientific Reports. 2017;**7**(1):13582

[48] Aravanis AM, Wang LP, Zhang F, Meltzer LA, Mogri MZ, Schneider MB, et al. An optical neural interface: in vivocontrol of rodent motor cortex with integrated fiberoptic and optogenetic technology. Journal of Neural Engineering. 2007;**4**(3):S143-S156

[49] Adamantidis AR, Zhang F, Aravanis AM, Deisseroth K, de Lecea L. Neural substrates of awakening probed with optogenetic control of hypocretin neurons. Nature. 2007;**450**(7168):420-424

[50] Petreanu L, Mao T, Sternson SM, Svoboda K. The subcellular organization of neocortical excitatory connections. Nature. 2009;**457**(7233):1142-1145

[51] Lin JY, Sann SB, Zhou K, Nabavi S, Proulx CD, Malinow R, et al. Optogenetic inhibition of synaptic release with chromophore-assisted light inactivation (CALI). Neuron. 2013;**79**(2):241-253

[52] Renault R, Sukenik N, Descroix S, Malaquin L, Viovy JL, Peyrin JM, et al. Combining microfluidics, optogenetics and calcium imaging to study neuronal communication in vitro. PLoS One. 2015; **10**(4):e0120680

[53] Schmid F, Wachsmuth L, Schwalm M, Prouvot PH, Jubal ER, Fois C, et al. Assessing sensory versus optogenetic network activation by combining (o)fMRI with optical Ca2+ recordings. Journal of Cerebral Blood Flow & Metabolism. 2016; **36**(11):1885-1900

[54] Nagel G, Szellas T, Huhn W, Kateriya S, Adeishvili N, Berthold P, et al. Channelrhodopsin-2, a directly light-gated cation-selective membrane channel. Proceedings of the National Academy of Sciences. 2003;**100**(24): 13940-13945

[55] Entcheva E. Cardiac optogenetics. American Journal of Physiology-Heart and Circulatory Physiology. 2013; **304**(9):H1179-H1191

[56] Jiang C, Li HT, Zhou YM, Wang X, Wang L, Liu ZQ. Cardiac optogenetics: A novel approach to cardiovascular disease therapy. EP Europace. 2017;**20**: 1741-1749

[57] O'Shea C, Holmes AP, Winter J, Correia J, Ou X, Dong R, et al. Cardiac optogenetics and optical mapping—Overcoming spectral congestion in all-optical cardiac electrophysiology. Frontiers in Physiology. 2019;**10**:182

[58] Williams JC, Entcheva E. Optogenetic versus electrical stimulation of human cardiomyocytes: Modeling insights. Biophysical Journal. 2015; **108**(8):1934-1945

[59] Boyden E. A history of optogenetics: The development of tools for controlling brain circuits with light. F1000 Biology Reports. 2011;**3**:11

[60] Yizhar O, Fenno LE, Davidson TJ, Mogri M, Deisseroth K. Optogenetics in neural systems. Neuron. 2011;**71**(1):9-34

[61] Gunaydin LA, Yizhar O, Berndt A, Sohal VS, Deisseroth K, Hegemann P. Ultrafast optogenetic control. Nature Neuroscience. 2010;**13**(3):387-392

[62] Berndt A, Schoenenberger P, Mattis J, Tye KM, Deisseroth K, Hegemann P, et al. High-efficiency channelrhodopsins for fast neuronal stimulation at low light levels. Proceedings of the National Academy of Sciences. 2011;**108**(18):7595-7600

[63] Goßler C, Bierbrauer C, Moser R, Kunzer M, Holc K, Pletschen W, et al. GaN-based micro-LED arrays on flexible substrates for optical cochlear implants. Journal of Physics D: Applied Physics. 2014;**47**(20):205401. Available from:

http://stacks.iop.org/0022-3727/47/i=20/a=205401

[64] Kallweit N, Baumhoff P, Krueger A, Tinne N, Kral A, Ripken T, et al. Optoacoustic effect is responsible for laser-induced cochlear responses. Scientific Reports. 2016;**6**(28141):1-10

[65] Schultz M, Baumhoff P, Kallweit N, Sato M, Krüger A, Ripken T, et al. Optical stimulation of the hearing and deaf cochlea under thermal and stress confinement condition. In: Optical Techniques in Neurosurgery, Neurophotonics, and Optogenetics. International Society for Optics and Photonics. San Francisco, California, United States: SPIE; 2014. p. 892816

[66] Richter CP, Tan X. Photons and neurons. Hearing Research. 2014;**311** (Supplement C):72-88. Annual Reviews

[67] Duke AR, Cayce JM, Malphrus JD, Konrad P, Mahadevan-Jansen A, Jansen ED. Combined optical and electrical stimulation of neural tissue in vivo. Journal of Biomedical Optics. 2009;**14**(6):060501-060501

[68] Li J, Uysal M. Optical wireless communications: System model, capacity and coding. In: IEEE 58th Vehicular Technology Conference. VTC 2003-Fall (IEEE Cat. No.03CH37484). Vol. 1. Orlando, FL, USA: IEEE; 2003. pp. 168-172

[69] Zedini E, Alouini MS. Multihop relaying over IM/DD FSO systems with pointing errors. Journal of Lightwave Technology. 2015;**33**(23):5007-5015

[70] Popoola WO, Ghassemlooy Z. BPSK subcarrier intensity modulated free-space optical communications in atmospheric turbulence. Journal of Lightwave Technology. 2009;**27**(8): 967-973

[71] Faria M, Alves LN, de Brito André PS. 10. In: Transdermal Optical Communications. Vol. 1. Boca Raton: CRC Press; 2017. pp. 309-336.

[72] Bashkatov AN, Genina EA, Tuchin VV. Optical properties of skin, subcutaneous, and muscle tissues: A review. Journal of Innovative Optical Health Sciences. 2011;**4**(01):9-38

[73] Simpson CR, Kohl M, Essenpreis M, Cope M. Near-infrared optical properties of ex-vivo human skin and subcutaneous tissues measured using the Monte Carlo inversion technique. Physics in Medicine and Biology. 1998;**43**(9):2465

[74] Du Y, Hu X, Cariveau M, Ma X, Kalmus G, Lu J. Optical properties of porcine skin dermis between 900 nm and 1500 nm. Physics in Medicine and Biology. 2001;**46**(1):167

[75] Troy TL, Thennadil SN. Optical properties of human skin in the near infrared wavelength range of 1000 to 2200 nm. Journal of Biomedical Optics. 2001;**6**(2):167-176

[76] Bashkatov A, Genina E, Kochubey V, Tuchin V. Optical properties of human skin, subcutaneous and mucous tissues in the wavelength range from 400 to 2000 nm. Journal of Physics D: Applied Physics. 2005;**38**(15):2543

[77] Farid AA, Hranilovic S. Outage capacity optimization for free-space optical links with pointing errors. Journal of Lightwave Technology. 2007;**25**(7): 1702-1710

[78] Sandalidis HG, Tsiftsis TA, Karagiannidis GK, Uysal M. BER performance of FSO links over strong atmospheric turbulence channels with pointing errors. IEEE Communications Letters. 2008;**12**(1): 44-46

[79] Arnon S. Effects of atmospheric turbulence and building sway on optical wireless-communication systems. Optics Letters. 2003;**28**(2):129-131

[80] Zou Y, Zhang W, Tian F, Chau FS, Zhou G. Development of miniature tunable multi-element Alvarez lenses. IEEE Journal of Selected Topics in Quantum Electronics. 2015;**21**(4): 100-107

[81] Hwang K, Seo YH, Jeong KH. Microscanners for optical endomicroscopic applications. Micro and Nano Systems Letters. 2017;**5**(1): 2213-9621

[82] Zhou G, Lee C, editors. Optical MEMS, Nanophotonics, and their Applications. Boca Raton: CRC Press; 2017

[83] Khoshnoud F, de Silva CW. Recent advances in MEMS sensor technology— Biomedical applications. IEEE Instrumentation & Measurement Magazine. 2012;**15**(1):8-14

[84] Sezen AS, Sivaramakrishnan S, Hur S, Rajamani R, Robbins W, Nelson BJ. Passive wireless MEMS microphones for biomedical applications. Journal of Biomechanical Engineering. 2005;**127**(6):1030-1034

[85] Sabry Y, Khalil D, Saadany B, Bourouina T. In-plane optical beam collimation using a three-dimensional curved MEMS mirror. Micromachines. 2017;**8**(5):134

[86] Nguyen H, Arnob MMP, Becker AT, Wolfe JC, Hogan MK, Horner PJ, et al. Fabrication of multipoint side-firing optical fiber by laser micro-ablation. Optics Letters. 2017;**42**(9):1808-1811

[87] Wells JD, Xing A, Bendett MP, Keller MD, Norton BJ, Owen JM, et al. Laser-based nerve stimulators for, EG, hearing restoration in cochlear prostheses and method. Google Patents; 2014. US Patent 8.792.978

[88] Marcuse D. Gaussian approximation of the fundamental modes of graded-index fibers. Journal of the Optical Society of America. 1978;**68**(1):103

[89] Friesen LM, Shannon RV, Baskent D, Wang X. Speech recognition in noise as a function of the number of spectral channels: Comparison of acoustic hearing and cochlear implants. The Journal of the Acoustical Society of America. 2001;**110**(2):1150-1163

[90] Kong YY, Cruz R, Jones JA, Zeng FG. Music perception with temporal cues in acoustic and electric hearing. Ear and Hearing. 2004;**25**(2): 173-185

[91] Zhou K, Zhang L, Chen X, Bennion I. Optic sensors of high refractive-index responsivity and low thermal cross sensitivity that use fiber Bragg gratings of $>80^{\circ}$ tilted structures. Optics Letters. 2006;**31**(9):1193-1195

[92] Cotillard R, Laffont G, Ferdinand P. Regeneration of tilted fiber Bragg gratings. In: 23rd International Conference on Optical Fibre Sensors. Santander, Spain: SPIE; 2014. p. 91572S

[93] Bharathan G, Hudson DD, Woodward RI, Jackson SD, Fuerbach A. In-fiber polarizer based on a 45-degree tilted fluoride fiber Bragg grating for mid-infrared fiber laser technology. OSA Continuum. 2018;**1**(1):56-63

[94] Mou C, Zhou K, Zhang L, Bennion I. Characterization of 45°-tilted fiber grating and its polarization function in fiber ring laser. Journal of the Optical Society of America B: Optical Physics. 2009;**26**(10):1905

[95] Yu H, Wang S, Fu J, Qiu M, Li Y, Gu F, et al. Modeling bending losses of optical nanofibers or nanowires. Applied Optics. 2009;**48**(22):4365-4369

[96] Wang G, Wang C, Yan Z, Zhang L. Highly efficient spectrally encoded imaging using a 45° tilted fiber grating. Optics Letters. 2016;**41**(11):2398

[97] Cardin JA, Carlén M, Meletis K, Knoblich U, Zhang F, Deisseroth K, et al. Driving fast-spiking cells induces gamma rhythm and controls sensory responses. Nature. 2009;**459**(7247): 663-667

[98] International Commission on Non-Ionizing Radiation Protection. ICNIRP guidelines on limits of exposure to laser radiation of wavelengths between 180 nm and 1,000 μm. Health Physics. 2013; **105**(3):271-295