Optical Wireless Communications for In-Body and Transdermal Biomedical Applications

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Abstract

This article discusses the fundamental architectures for optical-wireless-systems for biomedical-applications. After summarizing the main applications and reporting their requirements, we describe the characteristics of the transdermal and in-body optical channels (OCs) as well as the challenges that they impose in the design of communication systems. Specifically, we provide three possible architectures for transdermal communications, namely electro-optical monitoring, opto-electrical, and all-optical (AO), for neural stimulation, which are currently under investigation, whereas for in-body communications, we provide a nano-scale AO concept. For each architecture, we discuss the main operation principles, the technology enablers, and research directions for their development. Finally, we highlight the necessity of designing an information-theoretic framework for the analysis and design of the physical and medium access control layers, which takes into account the channels' characteristics.

Introduction

Medical implants (MIs) and nano-scale wireless networks (NWNs) have been advocated as an effective solution to numerous health issues. Typical MIs consist of an out-of-body (in-body) unit that captures the stimulus (bio-signal), converts it into an RF signal, and wirelessly transmits it into the in-body (out-of-body) unit, which stimulates (monitors) the corresponding nerve (bio-process). The main disadvantage of this approach is its incapability to support the high data rates required for neural-prosthesis applications under reasonable transmission power. Additionally, there is a lack of effectiveness and flexibility in coexisting with the huge amount of RF devices that are expected to conquer the beyond 5G wireless world. Finally, the use of RF prohibits the utilization of nano-scale biomedical devices.

In view of the advances in optoelectronics and optogenetics, very recent innovative designs have been reported, which, by employing optical-wireless communications (OWCs), deliver more compact, reliable, and energy-efficient solutions, while at the same time reducing the RF radiation concerns [1]. The main advantages provided by OWCs are the abundantly available bandwidth, no interference from other devices, higher achievable data rates, and increased safety. As a result, OWCs are expected to be used for both transdermal and in-body link establishment; hence, they will influence the fundamental technology trends in biomedical applications for the next 10 years and beyond. Due to the transdermal and in-body OC particularities, the compactness and energy limitations, as well as the directionality of the OWC links, the design and development of OWC-based biomedical applications needs to leverage breakthrough technological concepts. Indicative examples are the co-design of communication and stimulation units, the presentation of digital signal processing (DSP) schemes for the end-to-end link from the device (biological unit) to the biological unit (device), and the utilization of new electro-optical (EO), opto-electrical (OE), all-optical (AO), and nano-scale AO (NAO) interfaces. Likewise, new channel and noise models that take into account the peculiarities of the transdermal and in-body optical medium need to be developed. Building on these, a novel information-theoretic framework is required for the design of energy-efficient physical (PHY) and medium access control (MAC) schemes, as well as the development of simultaneous light information and power transfer (SLIPT) policies and resource allocation strategies [2].

Motivated by the above, the aim of this article is to present the vision and approach of delivering safe and high quality of experience (QoE) in MIs and NWNs, and identify the critical technology gaps and the appropriate enablers. In particular, after presenting the biomedical-applications, which can be OWC-enabled, together with their critical design requirements, we introduce three possible architectures for transdermal communications that are currently under investigation, whereas for in-body communications, we discuss the NAO concept. These architectures are expected to drastically increase the achievable data rate and significantly reduce the in-body devices' energy consumption. For each architecture, we report the main operation principles, technology enablers, and critical technology gaps. We emphasize the need to develop novel channel models that take into account the transdermal and in-body optical medium particularities as well as derive a corresponding theoretical framework for the performance assessment and design of the PHY and MAC schemes.

Beyond Classical Biomedical Applications

Optical-wireless MIs (OWMIs) and NWNs are envisioned to enable a vast variety of novel biomedical applications, such as smart drug delivery and tissue recovery, as well as to improve the QoE and safety of the conventional ones, not only by targeting

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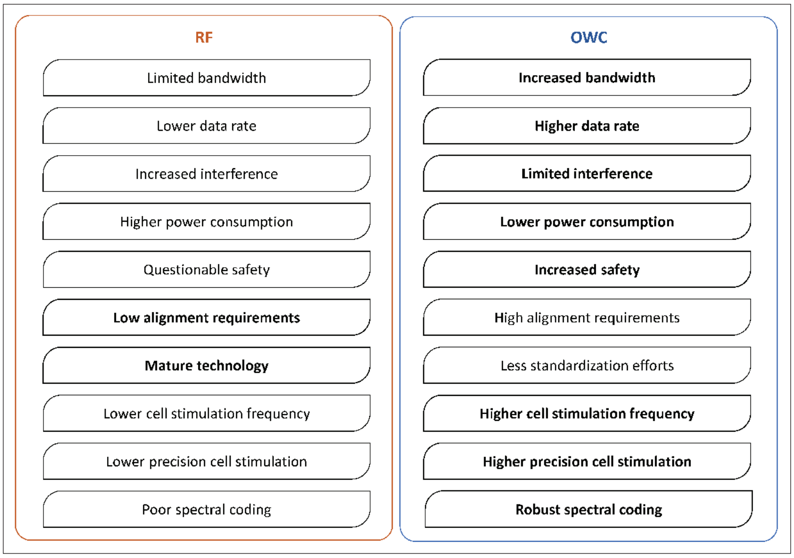

FIGURE 1. RF vs OWCs.

10–100 times higher data rates, but also by combining them with reliability and compact designs. Vital signal, pathogen, and allergen real-time continuous monitoring, detection of tissue and molecule abnormalities, as well as smart drug delivery are only some examples of several highly challenging applications. To support such scenarios, the research world turns to the adaptation of THz technologies, which, by leveraging graphene designs, are expected to utilize nano-scale transceivers. However, the safety of THz links is still questionable, and these technologies are still immature. A comparison between RF and OWCs is provided in Fig. 1. To break these barriers, we need to directly research toward mature OWC concepts, identify and categorize the possible applications, and design suitable architectures and systems [3]. In this direction, we present the main application categories based on the link nature and scenarios as well as their critical design parameters. These applications are depicted in Fig. 2.

Biomedical Applications Requiring the Establishment of Transdermal Links: The main representatives of this category are cochlear, retinal, cortical, and foot drop implants, gastric stimulators, wireless capsule endoscopes, insulin pumps, and implantable orthopedic devices. Maybe the most successful application is cochlear implants (CIs), which have restored partial hearing to more than 350,000 people, half of whom are pediatric users, who ultimately develop nearly normal language capability. Conventional CIs exploit near-field magnetic communication technologies and typically operate in low-RF frequencies, from 5 to 49 MHz, while their transmit power is on the order of tens of milliwatts. As a result, they cannot support high data rates (in the order of megabits per second) under reasonable transmit power constraints, which are required to achieve similar performance to the cochlea. Transdermal optical link (TOL) is an important building block to guarantee high-speed connectivity between the external and internal units with low energy consumption [4]. Although the need for higher data rates is detrimental for the performance of retinal implants, it plays an important role in the quality of CIs as well. One reason is the higher sound quality that can be achieved with higher data rates. A high definition recording (e.g., 24 b/192 kHz) requires a minimum data rate of 4.6 Mb/s for single-channel audio and 9.2 Mb/s for dual-channel audio, which is impossible with current state-of-the-art CIs. Furthermore, another issue that justifies the requirement of higher data rates is the fact that higher transmission frequency, which is entangled with the communication data rate, plays an important role in the precision of the stimulation of the cochlear neurons, which results in the information to be conveyed faster to the implanted part of the system and the stimulation of the targeted neurons to be performed in a more timely manner. In this scenario, apart from the high targeted data rates (10 Mb/s), the critical parameters are the latency, which should be on the order of 0.1 ms, uninterrupted connectivity (i.e., outage probability lower than 10^{-5}), and low bit error rate (BER $\leq 10^{-4}$). Additionally, these devices should support SLIPT in order to guarantee the internal device energy autonomy.

After the successful clinical validation of CIs, neural prostheses were exploited in the treatment of visual impairments by developing retinal implants (RIs). In this application, the implanted device is an epiretinal prosthesis that includes a receiver antenna, a DSP unit, and an electrode array attached to the retina surface. The external unit consists of a miniature video camera attached in glasses and a transmitter coil, connected via a small video processing-unit, which converts the captured video to electrical pulse. These pulses are transmitted wirelessly to the implanted device. Unfortunately, the visual acuity of this approach is quite low due to the limitations of the stimulation unit and the achievable data rate of the RF communication link. To break through this barrier, two solutions have been reported, namely, the use of optical stimulation approaches in order to reduce the pixel dimensions and OWCs to increase the data rate [1]. The critical parameters for this application are the pixel population, which should be higher than 60×60 pixels, the data rate (~ 100 Mb/s), and the latency (less than some tens of milliseconds).

Gastric stimulators are used in the treatment of gastric dysmotility disorders and obesity. They consist of two units, out-of-body and in-body. The former's main responsibility is to wirelessly transfer power to the latter, which consists of a received RF antenna, a charge pump, and a peripheral interface controller that generates the electrical pulse and is connected to the electrodes. The transmission medium is body tissue of 1 to 5 cm thickness. This is translated to about 4 dB penetration loss when RF-signals are employed. However, if optical signals are employed, this loss is further constrained. Except for the transmission range, another critical parameter for this application is the in-body unit compactness.

Biomedical implants are also used for bio-signal and bio-process monitoring, such as neural implants, insulin pumps, and wireless capsule endoscopes. Neural implants, pacemakers, and insulin pumps, in addtion to high data rates, demand high energy autonomy and compactness of the in-body unit. In other words, efficient power transfer taking into account the space constraints needs to be utilized. On the other hand, a wireless capsule endoscope is an ad hoc setup. Thus, its battery covers the energy demands, while its main challenge is to sustain continuous connectivity between the in-body and out-of-body units independent of the in-body unit orientation.

In-Body Biomedical Applications: Nanotechnology provides a new set of tools to control matter at the atomic and molecular scales, thus enabling the development of nano-scale biomedical applications for nano-sensing and actuation. Nano-devices utilized in such systems require the use of bio-compatible materials and communication techniques with limited electromagnetic radiation on the biological tissues [5]. Moreover, several breakthroughs in the field of nano-photonic devices enabled nano-scale PWCs. These applications include diagnostic and therapeutic techniques utilizing nano-devices, healthcare monitoring solutions that combine in-body nano-photonic bio-sensors, and non-intrusive brain-machine interfaces. Among others, nano-LEDs can be leveraged as energy-efficient compact signal sources for optical-wireless links, which meet both size and power requirements of nano-devices [6]. Similarly, sensitive nano-photodetectors (PDs) were developed and can be combined with nano-LEDs to form nano-sensors, which can be utilized in compact nano-transceivers to forge fast, short-distance wireless communication links. The aforementioned nano-transceivers, when paired with nano-antennas, can serve as nano-gateways to bridge the communication with out-of-body devices.

Despite the latest advances, numerous challenges still exist that need to be addressed. For example, in the case of wireless network on a chip, the main concern is to increase the capacity and data rate of the network, while dealing with a more predictable and controllable system model. Therefore, new challenges in terms of nano-photonic devices and communication protocols, such as energy efficiency, link capacity, and reliability, have to be addressed before NAO networks can be established as a realistic solution.

Candidate Architectures

For transdermal optical links: As illustrated in Fig. 3, three candidate architectures for transdermal OWCs have been identified, namely OE, EO, and AO. OE and AO are suitable for translating external stimuli, such as audio and video, into nerve stimulation signals. OE was introduced in [1] for next-generation CI designs. The main difference between EO/OE and AO is that the former's in-body units perform DSP. Thus, an energy autonomy demand for the internal device arises. To satisfy this, a SLIPT policy is utilized. In particular, the DC part of the received signal is used for energy harvesting, while the AC part conveys the neural stimulation message. A simple AC and DC separator (ADS) is employed at the internal device consisting of a capacitor that blocks the DC component of the received signal and forwards it to the energy harvesting branch. At the same time, the AC component of the signal at the output of the ADS is inserted to the DSP unit, which together with the STM translates the received message into the appropriate neural stimulation signal.

On the other hand, in AO, no SLIPT is required, since the stimulation signal is directly emitted by the out-of-body device. The in-body device consists of a fiber coupling unit, which is usually a MEM responsible for capturing the transmitted signal and forwarding it to the optical stimulator. This concept may be utilized by employing commercially available optics and optogenetics modules. In this

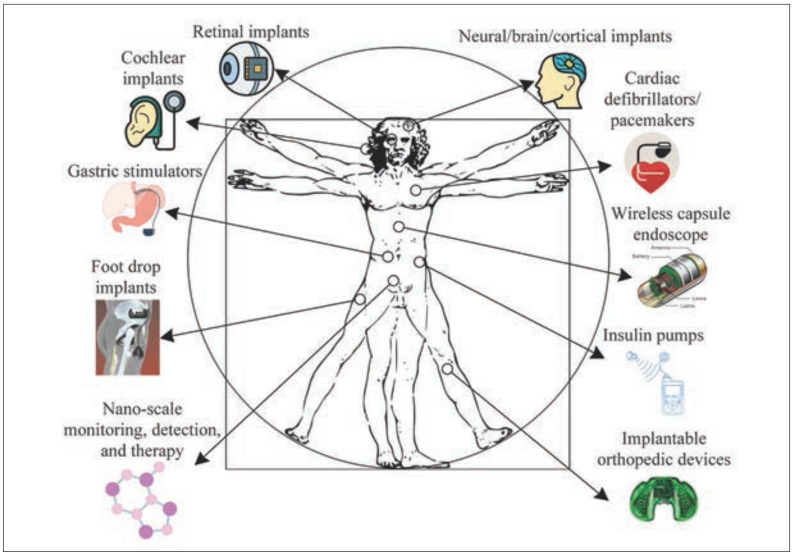

FIGURE 2. Biomedical-applications that can be supported by OWCs.

scenario, DSP for mitigating the impact of both the TOL impairments and the stimulation signal is required. The main challenge of AO is modeling the link from the external unit to the corresponding nerve. Another challenge is mapping the transmission signal into the stimulation one.

The EO architecture can be used for monitoring and sensing bio-signals; hence, it can be employed in neural implants and insulin pumps. The in-body device consists of a sensing/monitoring-unit, a DSP unit, and a light source that emits the modulated sensed message. To cover the energy requirements of the in-body device, an energy harvesting branch is utilized. The out-of-body device consists of a photodetector that captures the light signal and forwards it in a DSP unit, responsible for message detection. The detected signal is forwarded to the imaging device. Moreover, the out-of-body unit has a DC signal generator with a dimming controller that feeds a light source, responsible for energy transfer. The major challenge of EO is to present suitable dimming control strategies that guarantee the energy autonomy of the in-body device. Moreover, this concept allows the integration of the internal device front-end into a single module with superior compactness and energy efficiency.

For nano-scale systems: For the utilization of NAO networks, the two architectures depicted in Fig. 4 have been identified. A common nanonode structure is considered in both cases. Each nano-node is equipped with a nano-LED and a nano-PD that enable communication with others, as well as an energy-harvester that replenishes the energy stored in a nano-capacitor. This enables circulating nano-nodes to overcome the energy limitations and even have practically infinite lifetime. Depending on the application, nano-nodes can be equipped with nano-sensors and nano-actuators for monitoring and manipulation of biological processes, respectively.

The first architecture incorporates an intermediate nano-device, called a nano-gateway. This device is responsible for forwarding the data collected from the nano-nodes to the external interface. The link between them can be established on the optical or RF spectrum, depending on the application characteristics. For example, in close-range commu-

In-body and transdermal optical-wireless channel models are considered to be the main building blocks for developing the aforementioned architectures. In comparison with the out-of-body models, these models are required to accommodate a number of different characteristics and biological particularities.

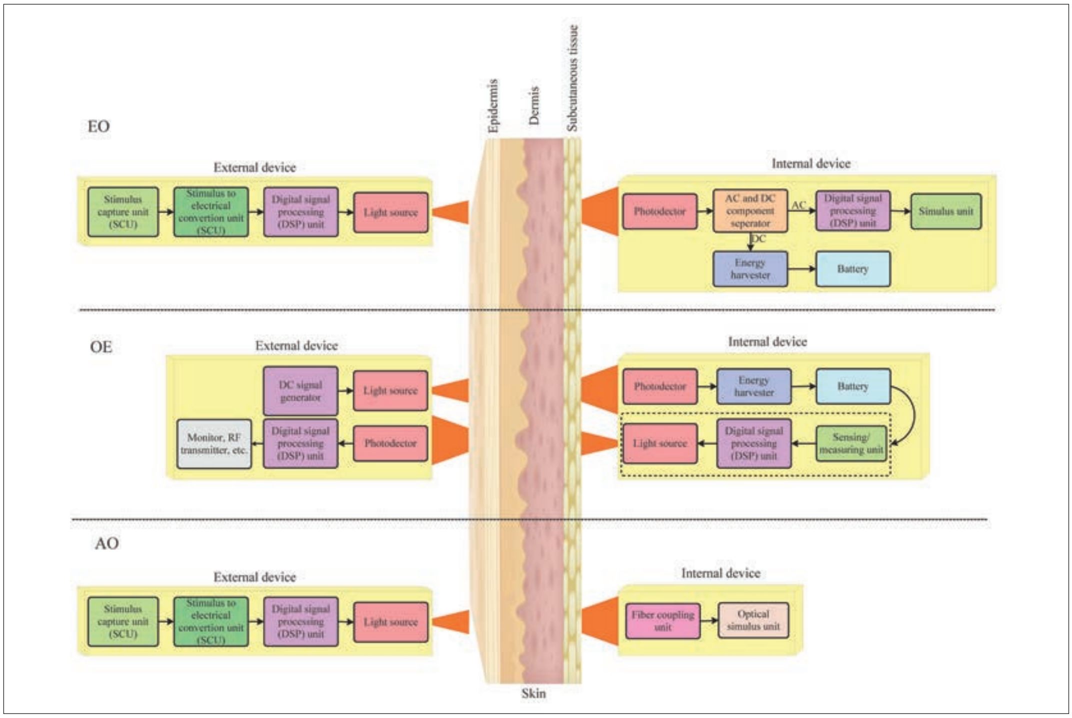

FIGURE 3. Candidate architectures for transdermal-applications.

nications, the optical link has been proven to outperform the corresponding RF, while for longer-range applications, the required optical beam intensity for achieving adequate signal quality can be destructive to the human tissue due to increased temperature.

The second architecture consists of the nano-nodes and the nano-router. The former move throughout the body with a pattern that varies with the application, while the latter collects the gathered information when the nano-nodes are in range [7]. For example, if we consider a scenario where the nano-nodes are injected into the bloodstream and the nano-router is placed on the outer surface of the skin, as the nano-nodes travel through the circulatory system, they communicate with the nano-router when they are positioned close enough.

The main difference between the two architectures is scalability. The first architecture can be multiplied throughout the human body, both in deep tissue and near the surface. By placing the nano-gateways with appropriate distance between them, the first architecture shown in Fig. 4 can be repeated multiple times. However, both architectures share the same concerns such as limited energy consumption, limited computational capabilities, and stochastic network topology.

Design Principles and Technology Enablers

In order for the candidate architectures to provide solutions to the OWC-based biomedical applications' challenges, rethinking of the fundamental design principles and adoption of signal transmission and neural stimulation, joint design approaches are necessary. In this direction, a generalized optical-wireless channel model is needed that accommodates the particularities of such systems and can support the extraction of their fundamental limits. Moreover, new neural stimulation approaches need to be examined in order to countermeasure the pixel population requirement. To cover the energy autonomy demand, SLIPT approaches are discussed. Finally, the development of PHY and MAC schemes capable of supporting the data rate, range, and latency limitations as well as neural stimulation functionalities becomes a necessity.

Optical-Wireless Channel

In-body and transdermal optical-wireless channel models are considered to be the main building blocks for developing the aforementioned architectures. In comparison with the out-of-body models, these models are required to accommodate a number of different characteristics and biological particularities. For instance, in TOLs, the dermis content in hemoglobin can significantly influence its light absorption in the blue and green-yellow regions. On the other hand, the dominant absorber in the ultraviolet and infrared regions is the epidermis melanin. Finally, chromophores, including bilirubin, carotene, lipids, cell nuclei, filamentous proteins, and so on, can cause further absorption in different wavelengths. Of note, in spite of the abundance in all tissues, due to the short communication distance, water is not a significant light absorber in these systems. This is one of the differences between transdermal/in-body and out-of-body optical channels.

The impact of the optical-wireless channel particularities with regard to path loss is presented in Fig. 5. This figure summarizes the main contributing elements of any tissue and is applicable to the communication paradigms under investigation. Moreover, the path loss that occurs when light of a specific wavelength travels through tissue is a function of the absorption coefficient of the tissue, which can be calculated based on the tissues' concentrations and absorption coefficients of water, blood, fat, and melanin [9]. In order to calculate the impact on the link's performance only due to the path loss, we assume perfect alignment, negligible noise, and perfectly aligned receivers with divergence angle and field of view such that they completely receive the transmitted beam, and the received signal is normalized to unity. From

Fig. 5, we observe that the attenuation due to the existence of blood plays a detrimental role in the performance of the communication link. If we consider only the blood, it is evident that a transmission window exists after 600 nm. On the contrary, water more intensively attenuates optical signals with higher wavelength, while the rest of the elements have a more consistent impact. As a result, the appropriate transmission wavelength for each use case has to be selected based on the composition of the intermediate tissue.

Apart from path loss, transdermal and in-body optical links experience wavelength-dependent particulate scattering. In particular, within the skin, the main source of scattering is filamentous proteins (e.g., keratin in epidermis and collagen in dermis). Note that since these particles are comparable to or larger than the transmission wavelength, scattering can be approximated as a Mie solution to Maxwell's equations. On the other hand, in in-body applications, tissues, such as membranes, striations in collagen fibrils, macromolecules, lysosomes, vesicles, mitochondria, and nuclei, are the main scatterers. Notice that membranes are usually lower than 1/10 of the wavelength, while all the other structures are comparable to the wavelength. Thus, scattering in tissues can be modeled as a mixture of Rayleigh and Mie processes.

Inhomogeneities in the body content in light-absorbing and scattering structures lead to a variation of the reflective index along the transmission path, which causes random fluctuations in both the amplitude and phase of the received signal (i.e., channel fading). To provide the theoretical framework for the performance assessment and design of optical biomedical implants, it is of high importance to deliver a stochastic model for the accommodation of this type of fading. In this direction, stochastic geometry approaches have been employed [10, references therein]. Finally, another source of received power uncertainty, which arises from the directional nature of the optical links, is transceiver misalignment. As reported in [4], this type of uncertainty can be modeled through a stochastic process and can significantly affect the system performance.

Another differentiation of in-body/transdermal optical communications in comparison to other out-of-body applications is the existence of neural noise. Specifically, optical approaches for tracking neural dynamics have modeled the physical bounds on the detection of neural spikes as photon counting statistics (shot noise). Besides the neural noise, other noise sources come from the receiver, like thermal, background, and dark current noises, and can be modeled as zero-mean Gaussian processes. Their power depends on the communication bandwidth, the detector's responsivity, the background optical power, and the intensity of the dark current, which is generated by the PD in the absence of background light. Depending on the architecture, different types of noise determine the fundamental bounds. For example, in AO, neural noise is expected to play the most detrimental role, while in OE and EO, its intensity tends to zero.

The above observations motivate the development of generalized deterministic path loss and statistical in-body particulate models for TOLs and in-body optical links that will aid in the appropriate transmission waveform design, the development

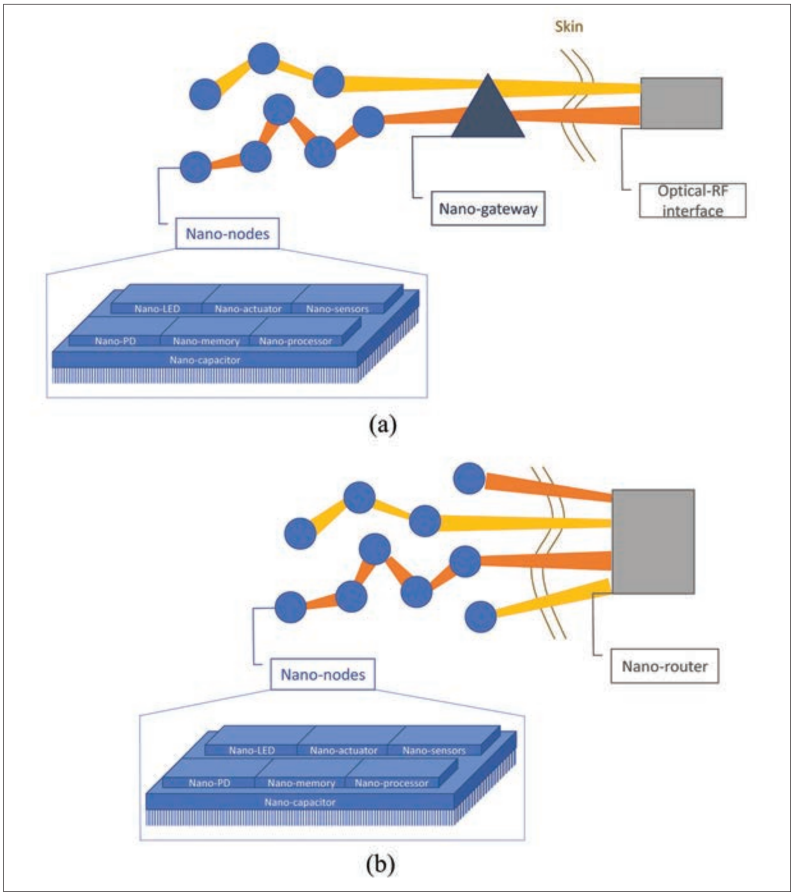

FIGURE 4. Candidate architectures for in-body optical applications.

of the PHY and MAC, as well as the utilization of the candidate architectures. These models should take into account the patient's particularities, such as skin color, the composition of different structures (collagen, melanin, etc.), the transmission wavelength, and the misalignment intensity.

Electrical and Optical Neural Stimulation

Based on the stimulation type, implants can be classified into electrical and optical. Electrical stimulation methods use an external electrical current to manipulate the membrane potential directly and have been extensively used for igniting action potentials in various types of human cells. Numerous medical implants, including CIs, pacemakers, and more, take advantage of these techniques. However, the performance of electrical stimulation is hindered by poor spectral coding and bandwidth scarcity of the RF spectrum. These limitations motivated the research community to turn its attention to the optical spectrum, which is capable of providing large amounts of unexploited bandwidth and higher safety for the human body [11]. As a result, optical stimulation has been greatly investigated over the past decade due to the various advantages it offers over the established electrical kind. Optogenetic techniques are based on the development of several genetically modified proteins, called opsins, which when illuminated generate a flow of ions through the cellular membrane that alters the membrane potential [8]. This implementation has been experimentally proven to control the cell dynamics with higher accuracy and frequency and greater spatial resolution.

Regarding their performance comparison, as indicated in Fig. 6, under the same power consumption and stimulation frequency, optical stimulation is proven to outperform the equivalent electrical one in terms of accuracy and power consumption, respectively. In this figure, the stimulation success rate is calculated as the percentage of the transmitted pulses that successfully generate action potential at the targeted neurons. It should be noted that the vertical dotted line, which corresponds to 15 Hz, marks the upper frequency limit for the electrical stimulation. It is evident that optical stimulation is capable of generating accurate action potentialwith higher frequency while maintaining similar power consumption. This observation is in agreement with several publications, which state that optogenetics offer more precise control over the excited neurons due to both the higher achievable frequency and the temporal accuracy provided by the variety of existing photosensitive opsins. In addition, optical cell stimulation outperforms the corresponding electrical in terms of the stimulation success rate. This is expected due to the fact that the optical spectrum offers more bandwidth. However, if we consider the fact that optical signals attenuate faster as they travel through the human body, thus offering lower geometric spread or, in other words, higher spatial resolution, we can deduct that optical stimulation is an overall more robust solution for cell stimulation.

The safety and stability of radiating the human body with light pulses with repetition rates of a few hundred Hertz need to be evaluated. Such stimulations at high intensities have been proven to cause phototoxicity or heating [12]. Pulses with optical irradiance up to 75 mW/mm^2 are considered safe for in vivo application, while tissue heating has been reported for amplitude of approximately 200 mW/mm^2, and phototoxic effects are absent up to 600 mW/mm^2. Furthermore, these limits were acquired for pulse width of 5 ms, while reliable excitation can be achieved at about 1 ms and possibly even shorter with future advances.

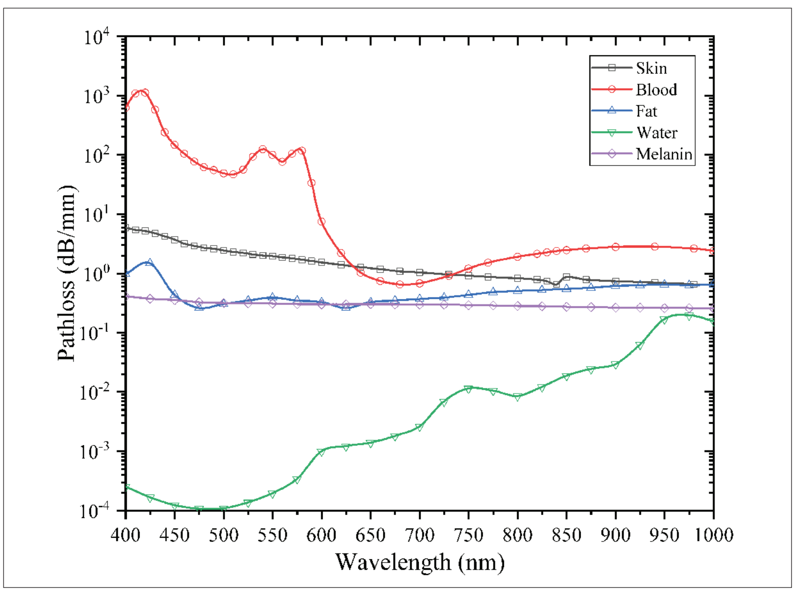

FIGURE 5. Optical-wireless-channel path-loss. Of note, the path loss was evaluated based on the model presented in [8].

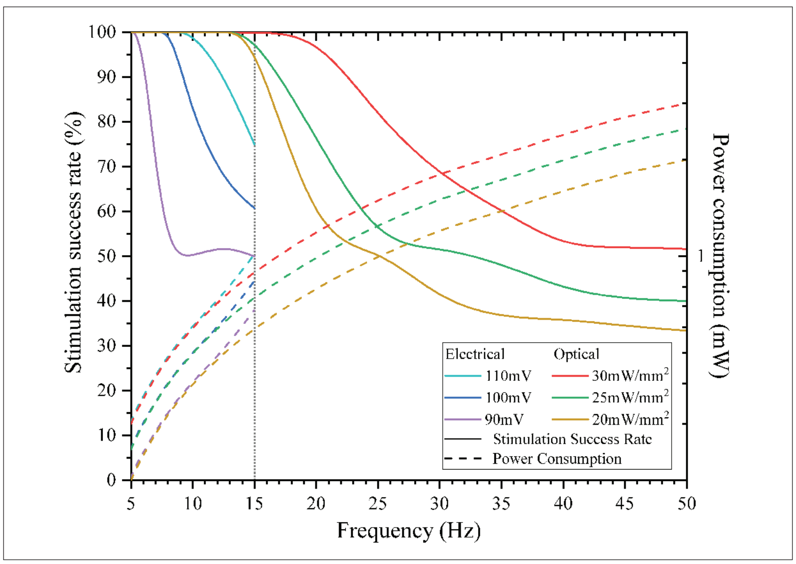

FIGURE 6. Optical vs. electrical stimulation performance.

Energy Transfer and SLIPT

A critical system design parameter in several in-body biomedical devices is their lifetime, which depends on the amount of available power. Replacing or removing and recharging the battery of such devices is impractical or even impossible. To counterbalance this, energy harvesting from the body using thermoelectric, piezoelectric, and electromagnetic generators attracted the eye of biomedical engineers [13]. However, these approaches have two inherent disadvantages:

- Low harvesting efficiency (~ 20 percent)
- Requiring the installation of extra modules that increase the implementation cost and size

As a consequence, SLIPT policies seem to be more appealing to biomedical-applications.

In biomedical devices with SLIPT capabilities, the PD, which is a common choice due to the high data rate that it supports, needs to be replaced with solar cells, which provide apparent advantages in terms of harvesting efficiency [14, 15]. Solar cell dimensions can be different, depending on the application for which they are used, from some nano-meters to tens of centimeters. They are basically photoelectric converters with zero bias and can be used to support both energy harvesting and information decoding. From the SLIPT utilization point of view, two fundamental policies can be used, namely time and signal splitting. In the former, the transmission period is divided into energy harvesting and information transfer, while in the latter, the DC and AC parts of the received photo-current are used for energy harvesting and information decoding, respectively.

Although SLIPT approaches and policies were extensively investigated in different types of OWC systems, they cannot be straightforwardly applied in biomedical systems because of the receiver particularities. Receivers designed for out-of-body OWC applications with SLIPT capabilities are equipped with solar panels. On the other hand, in biomedical applications, they are equipped with solar cells. As a consequence, SLIPT is identified as a promising open research direction, and further investigation on the suitability of each policy in each one of the aforementioned applications needs to be conducted.

PHY and MAC Layers

By exploiting the capabilities of the optical transceivers, we report the new challenges and opportunities at the PHY and MAC, including energy-efficient modulation and coding (MC), energy transfer, and SLIPT policies. Consistent with the demands of ultra-low power consumption, compact size, reliability, and ultra-low complexity, new MC schemes capable of accommodating the different types of noise and channel particularities need to be developed. Intensity modulation and direct detection with power adaptation is a possible approach to deal with a channel's stochastic behavior, whereas transmission and/or reception diversity through the exploitation of multiple light sources and PDs can be employed to deal with pointing errors while offering diversity through repetition coding schemes, which consequently can contribute to a significant error rate reduction.

Moreover, channel particularities and the transceiver limitations require the development of novel channel-codes for in-body/transdermal OWCs. The conventional capacity-approaching channel codes, which are designed to maximize the data rate for a given transmit power, demand additional transmission power and time for decoding, which may violate the application requirements. Thus, we need to characterize the error sources (i.e., both the noise and channel natures) and examine the trade-off between transmission power and decoding time in order to design channel codes. The use of low-complexity error prevention coding schemes might be a solution. However, the identification and design of the optimal coding scheme is still an open issue.

From the MAC perspective, the directional nature of OWCs combined with the high data rates increase the spatial synchronization requirements in nano-scale optical networks and call for novel MAC protocols that guarantee transceiver alignment and enable mobility management. In this direction, schemes that allow devices to periodically operate in sleeping, discovering, receiving, and transmitting modes may be the solution. During the discovering phase, each device sends synchronization signals in all directions in order to announce its availability as an RX or its intention to transmit a message to a specific node. Meanwhile, its neighbors are capable of discovering the aforementioned device. A distributed routing protocol then needs to be designed that allows the TX to communicate with the intended RX through intermediate nodes. The design of the optimal root should take into account not only the minimization of the network's energy consumption but also the power availability to each one of the intermediate nodes. Another functionality that needs to be utilized is fast optical tracking and steering. Further research is needed to address the development of appropriate field-of-view restrictions, effective steering, and adaptive control systems for nano-scale biomedical applications. Finally, note that due to the directional nature of both THz and OWC networks, several of the aforementioned challenges are the same. The main difference between OWC and THz networks resides in the different channels, each with distinct particularities, which generate new PHY and MAC layer challenges for OWC nano-scale networks. Therefore, solutions that have been applied in THz nano-scale communications need to be re-examined for possible adoption in OWC nano-scale networks.

The main difference between OWC and THz networks resides in the different channels, each with distinct particularities, which generate new PHY and MAC layer challenges for OWC nano-scale networks. Therefore, solutions that have been applied in THz nano-scale communications needs to be re-examined for possible adoption in OWC nano-scale networks.

Conclusions

In this article, we present the concept of optical-in-body and transdermal-optical communications. In particular, after presenting the possible biomedical applications and identifying their requirements, we describe candidate system architectures and features. Due to the fact that their designs are in an early stage, it is difficult to determine their final form. However, their enablers and technology modules are reported, namely channel modeling, the development of novel PHY, MAC, and SLIPT schemes, and the design of appropriate stimulation units. Finally, future research directions are highlighted. Note that this article is not just a review or tutorial, since it further aims at identifying the technology enablers that open the road to another promising application of OWCs.

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