Feasibility Assessment of an Optically Powered Digital Retinal Prosthesis Architecture for Retinal Ganglion Cell Stimulation

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Abstract—Clinical trials previously demonstrated the notable capacity to elicit visual percepts in individuals with visual impairments caused by retinal diseases by electrically stimulating the remaining neurons on the retina. However, these implants restored very limited visual acuity and required transcutaneous cables traversing the eyeball, leading to reduced reliability and complex surgery with high postoperative infection risks. To overcome the limitations imposed by cables, a retinal implant architecture in which near-infrared illumination carries both power and data through the pupil to a digital stimulation controller is presented. A high efficiency multi-junction photovoltaic cell transduces the optical power to a CMOS stimulator capable of delivering flexible interleaved sequential stimulation through a diamond microelectrode array. To demonstrate the capacity to elicit a neural response with this approach while complying with the optical irradiance limit at the pupil, fluorescence imaging with a calcium indicator is used on a degenerate rat retina. The power delivered by the laser at the permissible irradiance of 4 mW/mm² at 850 nm is shown to be sufficient to both power the stimulator ASIC and elicit a response in retinal ganglion cells (RGCs), with the ability to generate of up to 35 000 pulses per second at the average stimulation threshold. This confirms the feasibility of generating a response in RGCs with an infrared-powered digital architecture capable of delivering complex sequential stimulation patterns at high repetition rates, albeit with some limitations.

Index Terms—Brain-machine interface, implantable electronics, neurostimulation, retinal prosthesis.

I. INTRODUCTION

AROUND 250 million people in the world are affected with moderate to severe vision impairment caused by uncorrected refractive errors, cataracts, glaucoma and degenerative retinal diseases [1]. Among them, retinal diseases such as age-related macular degeneration and retinitis pigmentosa are particularly difficult to treat due to the complex cellular organisation of this sensory membrane. The only currently approved treatment consists in functional neurostimulation to restore visual percepts by electrically stimulating the inner retinal neurons that survive the disease.

Existing clinically approved devices demonstrated the capacity to elicit visual percepts in patients by electrically stimulating the remaining neurons according to an image captured by a camera. They either use an external camera [2] (ARGUS II, Second Sight Inc., Sylmar, California, USA) or an internal photodiode array [3] (Alpha IMS, Retina Implant AG, Reutlingen, Germany) and replicate the image with stimulation pulses on an electrode array surgically affixed to the retina. Both implants receive power through an external inductive link connected to cables that traverse the eyeball. While they enable the generation of visual percepts with neurostimulation, the transcutaneous cables require intricate surgery involving possible complications such as conjunctival erosion, conjunctival dehiscence (reopening of the surgical incision), hypotony...
Fig. 1. Implant power and data delivery architecture. A near-infrared beam is sent through the pupil. A multi-junction photovoltaic cell captures the infrared light to power a CMOS stimulator ASIC and a photodiode recovers the data from the modulated laser beam. The ASIC delivers the stimulation through an ultrananocrystalline diamond substrate with conductive diamond electrodes.

(reduction of intraocular pressure) or endophthalmitis (infection) due to the permanent skin penetration [4], [5]. Moreover, the cables can lead to premature failing of the device. In the case of the alpha IMS prosthesis, the median lifetime of the cables was assessed at 1.2 years over 30 implanted first generation devices and at 7.0 years for the second generation [6].

To overcome the safety and reliability limitations induced by the transcutaneous cables, alternative designs use inductive power delivery with the receiving coil implanted within an artificial crystalline lens [7] or inside the eyeball [8], [9]. A notable design achieves power transmission efficiencies as high as 36% and data transfer rates of 20 Mb/s, even with a separation of 25 mm and a receiving coil diameter of 10 mm [7]. However, the coil size and separation requirements usually result in devices composed of interconnected components instead of a single integrated unit. This can complicate surgical procedures and potentially impact the device’s reliability.

Alternatively, wireless subretinal implants based on microphotodiode arrays were previously proposed with simpler packaging architectures [10], [11]. Optical power delivery using near-infrared light offers high power density, facilitating mechanical integration and reducing surgical constraints. In these systems, a camera mounted on a pair of glasses captures an image of the visual field and projects it on the retina at high intensity using an infrared projector. The photodiodes directly transduce the infrared image to stimulation pulses on electrodes to which they are individually coupled [12]. The duration and intensity of the projection determine the stimulation pulse widths and currents. While effective for indiscriminate bipolar cell stimulation, this architecture might be less suitable for potential epiretinal stimulation devices aimed at delivering cell-type specific stimulation [13]. In order to produce a stimulation pulse on a particular electrode, the eyeglasses would need to locate the implant with an accuracy finer than the electrode size at a high refresh rate [14], which can be highly challenging considering the rapid eye saccades and frequent eyeglasses displacements [15], [16]. Although the absence of digital electronics simplifies the design of the implant and minimizes the power consumption, it limits the possibility of dynamically varying some stimulation parameters such as the interphase gap and pulse polarity for selective cell-type stimulation [17].

It also limits the use of active charge balancing [18], [19] and the delivery of flexible multipolar stimulation patterns such as current steering [20], [21] and current focusing [22], which are demonstrated means of enhancing spatial resolution.

In order to provide wireless operation in a monolithic unit while retaining the flexibility of an implanted digital stimulation controller, we propose an implant architecture that A) receives both power and data through an optical link and B) decouples this link from the stimulation by embedding a digital controller capable of spatially confined stimulation strategies. To validate the feasibility of this power and data delivery method, a 288 electrode application-specific integrated circuit (ASIC) was designed in TSMC CMOS 65 nm LP [23] and packaged with a multijunction photovoltaic cell for power recovery. Calcium imaging fluorescence microscopy is used to validate that the device can elicit a response on retinal ganglion cells of rats affected by inherited retinal degeneration. Section II presents the implant architecture. Section III presents the materials and methods used to validate the retinal ganglion cells’ (RGCs) response. Section IV presents the stimulation results and Section V discusses the implications for future implant design.

II. IMPLANT ARCHITECTURE

The implant comprises multiple heterogeneous components to allow photovoltaic operation (Fig. 1). A high efficiency multi-junction photovoltaic cell recovers the optical power, and a photodiode, with a higher frequency response, receives
the data transmitted by modulating the infrared beam. A stimulator ASIC then decodes the stimulation data, and executes the stimulation pattern on a 288 diamond electrode array. An embedded analog-to-digital converter (ADC) characterizes the electrode properties and sends them back to a radiofrequency (RF) receiver mounted on a pair of smart glasses through a custom-designed RF transmitter. The photovoltaic cell, photodiode, RF transmitter and passive components are assembled on a printed circuit board interposer (Fig. 4), which is then mounted on the subassembly comprising the diamond array and the stimulator ASIC (Fig. 4). The next section details the rationale behind the design and the choice of each component.

A. Photovoltaic Cell

Since the retina is sensitive to temperature increases, the implant power supply is limited by the permissible optical power density that can enter the eye. Thermal damage can occur because of protein denaturation following light absorption in the retinal pigment epithelium. For an 850 nm beam entering the natural or dilated pupil, safety standards for ophthalmic devices dictate that the maximum permissible radiant power is limited to $6.93 \times 10^{-5} C_T C_E P^{-1}$ for chronic exposure at 850 nm, where the wavelength parameter $C_T = 2$ at 850 nm [10], [24], [25]. The pupil factor $P$ models its contraction and dilatation and is equal to one at 850 nm. For spot sizes larger than 1.7 mm in diameter, $C_E = 29.38 \, \text{W mm}^{-2}$. This results in a maximum permissible radiant power density of 4.06 mW mm$^{-2}$ that can enter the pupil.

Maximizing the power reaching the implant requires a high efficiency PV cell. Recent photovoltaic cells based on vertical epitaxial heterostructures achieve efficiencies up to 65 % for monochromatic sources [26]. By stacking multiple thin GaAs photovoltaic junctions with submicron absorption thicknesses, it is possible to achieve sufficient voltage for stimulation. The implant is designed around a $3 \times 3 \, \text{mm}^2$ photovoltaic cell, resulting in a maximum usable power of 36.5 mW, given the power density limit above. Since redesigning a cell with these custom dimensions requires costly developments, a commercial bare die optical power converter (Broadcom AFBR-POCXX4L) with dimensions of $1.7 \times 1.7 \, \text{mm}^2$ was instead chosen to demonstrate the proposed architecture.

Frequent blinking cause power interruptions, leading to imbalanced stimulation pulses in the case of an interruption mid-pulse. To address this, a 15 µF capacitor (C1 in Fig. 2) stabilizes the voltage output of the photovoltaic cell and acts as an energy reservoir to complete a stimulation pulse in the event of a power loss during. Additionally, a voltage threshold is set to ensure that sufficient energy is available before initiating a stimulation pulse. The photovoltaic cell connects to the ASIC (Fig. 2) through diode D1 (BAS116LP3-7, Diodes Incorporated) to prevent capacitor C1 from discharging into the photovoltaic cell when the laser does not reach the implant, and to prevent the PV cell maximum output of 4.4 V from exceeding the maximum supply voltage of the 65 nm technology.

B. Photodiode

The proposed receiving circuit is based on a transimpedance amplifier coupled to a comparator [23] that decodes the data from the photodiode (Albis PDCA04-100-GS). To prevent power variations during transmission and facilitate decoding, the glasses transmit the stimulation scheme using a DC-balanced Manchester code at 2 Mbits/s. The Manchester line code provides a transition in the middle of every bit interval, thus making bit clock recovery trivial at the receiver.

C. Stimulator ASIC

The stimulator ASIC is designed in 65 nm CMOS to allow integration of high-density digital circuits. Details about the ASIC are presented in a separate paper [23]. Its architecture (Fig. 2) includes 1) 288 electrode drivers, 2) a digital stimulation controller, 3) an optical data recovery circuit, 4) a power management module and 5) an electrode characterization circuit.

1) Electrode Driver: To prevent irreversible electrochemical reactions at the electrode-tissue interface, the electrode driver must provide charged-balanced biphasic pulses. However, in a typical CMOS current source and sink pair, the process variations will unbalance the cathodic and anodic currents. To prevent this, the ASIC uses a dynamic current copy architecture. It operates with a calibration phase where the current sink driver sets the current that flows through the current source driver. The current source driver then copies that current and stores the calibration, corresponding to the gate-source voltage of the transistor, on a capacitor [27], [28]. The electrode driver can provide pulse widths ranging from 10 µs to 700 ms in steps of 10 µs and with amplitudes from 50 µA to 255 µA in steps of 1 µA with a voltage range of up to ± 2.7 V.
2) Digital Stimulation Controller: One of the key requirements for the stimulator ASIC is to provide flexible stimulation patterns. Moreover, because the optical power delivery can be interrupted by an eye blink, the implant must also be able to restore stimulation quickly after power up. Some implantable ASICs require a configuration phase and a stimulation phase [28], and in the event of a power failure, this implies that the device must be reprogrammed before stimulation can resume. The digital stimulation controller operates in a stateless fashion, where each new frame fully configures the next stimulation pulses (phase durations, currents, and selection of active and return electrodes). Thus, as soon as the power is reestablished, the stimulation resumes its operation within 3 µs without the need for bidirectional communication [23].

3) Electrode Monitor: The characterization of electrode impedance enables adaptation of the stimulation to the available voltage dynamic range. To achieve this, any given electrode can be selected via a multiplexer for connection to a 8-bit ADC. Details about the electrode monitor circuit are presented in a separate paper [23].

4) Power, Data and Clock Recovery: The power recovery block linearly regulates the PV cell power to 3.0 V for the electrode driver and electrode monitor circuits and to 1.2 V for the digital circuits. Having two different voltages allows greater stimulation headroom while minimizing the power consumption of digital circuits. The clock recovery circuit generates the clock from the 37.4 MHz crystal, and divides it by 40 to provide a 935 kHz system clock. The data recovery circuit uses a transimpedance amplifier to recover the Manchester-encoded data from the photodiode, and oversamples it with the 37.4 MHz clock. Oversampling enables maximum energy transfer from the received bit and straightforward bit clock recovery (no phase-locked loop) to minimize power consumption.

D. Diamond Electrode Array and Package

The packaging and electrode design of retinal implants is critical to ensure reliability while immersed in a biological fluid environment. A possible failure mode is the electrode material degradation. Iridium Oxide and platinum electrodes are often used for retinal stimulation. However, platinum has a relatively low charge injection capacity and Iridium Oxide is deposited as a coating and can be subject to delamination [29]. Alternatively, ultrananocrystalline (UNCD) diamond can be made conductive with the co-deposition of a dopant (boron) or the inclusion of nitrogen during its production by chemical vapor deposition (CVD). This electrode material provides sufficient charge injection capacity for stimulation and while allowing non-conductive and conductive diamond to coexist in the creation of a monolithic package comprising both the enclosure and the electrodes [30], [31], [32].

Using this method, a 16 × 18 diamond electrode array was designed with 120 × 120 µm square electrodes separated by a pitch of 150 µm on which the stimulator ASIC will be assembled. The impedance of the electrodes ranges from 13 to 25 kOhm at a frequency of 1 kHz. Due to ongoing development in the ASIC-diamond substrate assembly process, the stimulator ASIC and components were instead mounted on a printed circuit board and linked via wires to a 5 × 5 electrode diamond array with identical pitch and electrode dimensions. The fabrication of the diamond array is presented in separate papers [31], [33].

E. Printed Circuit Board Interposer

In the complete implant, the photovoltaic cell, crystal oscillator, PV cell and RF transmitter will be assembled on a high density printed circuit board (Fig. 4). A copper trace surrounds the PCB and forms the RF antenna. This PCB will then be affixed to the diamond substrate (Fig. 3). For the feasibility study with calcium imaging experiment, the implant was physically separated from the diamond substrate and connected with a cable.

F. RF Transmitter and Antenna

Assessing electrode properties and monitoring their degradation over time is crucial for the long-term functionality of a prosthesis. A custom RF transmitter is incorporated into the circuit for this purpose. Due to power and area limitations, it is necessary to minimize the complexity of the implanted RF transmitter and antenna and relocate the complexity at the receiver side where there are less constraints. A typical oscillator-based transmitter requires multiple
Fig. 5. To validate the implant powering method using laser illumination, an apparatus was designed to facilitate calcium imaging where the implant components are assembled on a printed circuit board. A 35 mW, 850 nm laser powers the implant. A cable connects the implant to a 5 × 5 electrode array. A degenerate rat retina stained with a calcium indicator is placed on the electrode array with retinal ganglion cells facing up. The RGCs’s response is evaluated by measuring rapid fluorescence variations with a confocal microscope.

internal RF submodules and external components. To minimize the complexity, the transmitter operates from a simpler complementary cross-coupled LC oscillator architecture at 2.4 GHz. An internal on-chip capacitor and a loop PCB antenna inductor compose the LC resonant network. Since the resonant frequency changes with fabrication variations, the on-chip capacitor is digitally tunable to adjust the frequency. The transmission efficiency at higher frequencies of 2.4 GHz allows a good compromise between tissue losses and loop antenna efficiency, although the efficiency is expected to be significantly lower in a biological environment than in air [34].

The transmitter is implemented in 65 nm GP technology with a die size of 0.7 × 1.5 mm² and an active area of 30 × 60 µm². The power consumption varies from 0.2 mW to 0.5 mW during transmission depending on the selected transmission power.

III. MATERIALS AND METHODS

To validate the proposed infrared power and data delivery architecture, the neural response of degenerate rat retinas to electrical stimulation from a single electrode was measured with calcium imaging. Then, the implant power consumption budget is determined to evaluate the headroom for delivering complex stimulation patterns comprising multiple sequential pulses.

A. RGCs Response to Infrared-Powered Stimulation

The response of retinal ganglion cells under infrared-powered stimulation is evaluated by generating spatial threshold maps of retinal ganglion cells around a single electrode. A map is realized for short pulse widths of 100 µs and for longer pulses of 500 µs to replicate a typical configuration used by the first generation of retinal implants [35]. The next subsections details how the spatial threshold maps are realized.

1) Implant Test Bench: To deliver the stimulation pulses, the stimulator ASIC, photodiode, photovoltaic cell, crystal and passive components (C1, C2, C3, D1 from Fig. 2) are assembled on a printed circuit board (Fig. 5). Then, the electrode driver pads are connected with cables to a 5 × 5 electrode array assembled on a second printed circuit board. The power and data is sent to the implant using an 850 nm laser diode (L850P200, Thorlabs). The output power of the laser diode is adjusted by the laser driver (iC-NZ, iC-Haus Inc.) with a power meter to 35 mW. An ADRV9364-Z7020 System-on-Module controls the laser driver to encode the stimulation data with a binary amplitude shift keying (BASK) scheme.

2) Retina Preparation: Retina preparation is performed in accordance with the ethical protocol of the Animal Care and Ethics Committee of The University of Melbourne. Adult Royal College of Surgeons (RCS-p+) rats of either gender and older than 3 months are prepared. RCS rats have inherited retinal degeneration which causes their retina to lose most of its photoreceptors by 90 days after birth [36].

The retina is injected with a fluorescent indicator dye through the optic nerve for calcium imaging. The dye is Oregon Green 488 BAPTA-1 solution (OGB-1, Hexapotassium salt, Thermo Fisher Scientific, dissolved in deionised water). The retina preparation and calcium indicator loading is described in detail in a separate paper [33].

The retina is mounted on the diamond electrode array with the ganglion cell layer facing up and held with a steel harp fitted with Lycra threads (SHD-25GH, Warner Instruments). The diamond array is assembled on a printed circuit board which
constitutes the bottom of a 3D printed perfusion chamber. The chamber is perfused with a carbogenated Ames’ solution at a rate of 3-8 mL min\(^{-1}\) held between 35°C and 37°C. The electrode array is kept around 2.5 mm away from the optic nerve.

Although the implant is designed to be placed epiretinally, the electrode array is placed subretinally in this demonstration to facilitate the experiment with calcium imaging. For maximum light transmission to an upright microscope, the retinal ganglion cells need to face the top of the microscope. Thus, the electrode array is placed on the bottom face (subretinally) in order to avoid obstructing the line of sight of the microscope.

3) Calcium Imaging: The retina preparation is imaged with a confocal microscope (Olympus FluoView FV1200) with a 10× and a 20× lens, for a field of view of either 318 × 318 \(\mu\)m\(^2\) or 633 × 633 \(\mu\)m\(^2\). The calcium dye is excited with a 473 nm source, and images are captured at a rate of 7.8 Hz.

4) Electrical Stimulation: The electrical stimulation is delivered by the ASIC and consists of charge balanced, biphasic current-controlled pulses. The pulses are delivered with an anodic-first polarity, with phase durations of 100 \(\mu\)s and 500 \(\mu\)s with a 10 \(\mu\)s interphase gap. The dynamic current copy architecture of the stimulation drivers requires a calibration phase prior to the stimulation whose duration is set to 30 \(\mu\)s. The stimulation protocol is detailed in Fig. 6. An Ag|AgCl wire acts as the return electrode and is placed in the perfusion chamber, 2 cm away from the stimulating electrodes.

5) Data Analysis: Electrical responses are evaluated by identifying rapid temporal changes in the fluorescence image. To achieve this, the response is evaluated by filtering the signal of each pixel with a temporal high-pass filter (with coefficients \([2.1, -1, -2]\)), and then detecting activation by setting a threshold. The detected response in 50% of the cases. The data analysis is presented in detail in a separate paper [33].

B. Implant Power Budget

The peak stimulation capabilities of the ASIC are determined based on the power budget. This budget is derived by deducting the optical losses in the eye, the power dissipation of the PV cell at maximum load, and the standby power consumption of the implant from the allowable input power of 35 mW. The following section outlines the methods used to assess the stimulation capabilities and power budget.

1) Photovoltaic Cell Characterization: The photovoltaic cell is characterized by tracing the current-voltage and power-voltage curves under illumination with a 35 mW laser beam collimated on the photosensitive surface. The curves are traced with a Keithley 4200A source measurement unit (SMU). The efficiency is extracted at the maximum power point. Although the PV cell achieves peak power efficiency only when the ASIC’s load current aligns with the PV cell’s maximum power point, the power budget and maximum pulse rate results represent the device’s peak performance at this point. Below this load current, the ASIC still functions, albeit with lower PV cell efficiency.

2) Available Stimulation Power: The available stimulation power is derived from the implant power budget by subtracting the losses associated with the ocular medium, the photovoltaic cell and the implant standby power consumption from the 35 mW power source. The standby power usage is determined using a 10 \(\Omega\) shunt resistor in series with the photovoltaic cell. The current measured is then multiplied by the voltage of the PV cell at its maximum power point. This measurement represents the portion of power allocated to ASIC’s ancillary functions during peak load conditions. As the voltage is recorded at the PV cell, the results account for the losses caused by the internal linear regulators.

3) Maximum Repetition Rate: The maximum stimulation repetition rate is a key metric indicative of the capacity of the implant to eventually mimic neural code on a spike-by-spike basis [37]. This rate is constrained by the available power. To determine the maximum stimulation rate, the average stimulation thresholds are evaluated with calcium imaging for pulse widths of 100 \(\mu\)s, 150 \(\mu\)s, 250 \(\mu\)s, 500 \(\mu\)s with three different pieces of retinal tissue.

Then, the power consumption for a single electrode is evaluated by measuring the current and voltage at the PV cell’s output while delivering continuous square biphasic pulses in a physiological saline solution at the average stimulation threshold. The repetition rate during continuous stimulation is given by the equation:

\[
\frac{1}{f_{\text{continuous}}} = \frac{1}{2 \cdot t_{\text{pulse}} + t_{\text{interphase}} + t_{\text{calibration}}}
\]

where \(f_{\text{continuous}}\) is the stimulation repetition rate, \(t_{\text{pulse}}\) is the pulse width, \(t_{\text{interphase}}\) is the interphase gap (10 \(\mu\)s), and \(t_{\text{calibration}}\) is the calibration interval (30 \(\mu\)s).

Then, the maximum pulse rate for the array is estimated by calculating the number of electrodes (\(N\)) that can operate...
Fig. 7. Voltage waveform of a 1 ms stimulation pulse at 100 µA in a physiological saline solution with an oscilloscope.

Fig. 8. Spatial threshold map of retinal ganglion cells in a degenerate RCS rat retina for 100 µs biphasic charge-balanced stimulation pulses.

Fig. 9. Spatial threshold map of retinal ganglion cells in a degenerate RCS rat retina for 500 µs biphasic charge-balanced stimulation pulses.

Fig. 10. Photovoltaic cell characterization at 850 nm with a 35 mW beam collimated within the sensitive area. The efficiency peaks at 59.4 % at a voltage of 3.9 V.

simultaneously within the available power budget. The maximum repetition rate for each pulse width is then obtained by multiplying N by the continuous pulse rate.

IV. RESULTS

A. RGC Response to Infrared-Powered Stimulation

Firstly, the functionality of the device is verified by measuring the voltage waveform of a stimulation pulse with an oscilloscope (Fig. 7).

Figures 8 and 9 present the RGC spatial threshold maps from 100 µs and 500 µs pulses with the implant being powered by a 35 mW laser. In the threshold map, each circle represents one RGC, with the color indicating the threshold current. The RGCs that couldn’t be activated with the maximum available current are shown as open circles. The blue square indicates the electrode position. As reported previously, 100 µs pulses lead to a more confined activation pattern. Using 500 µs pulses, the larger activation spread is most likely due to the unintended stimulation of the axon bundles passing the electrode and network-mediated stimulation via bipolar cells [33], [38].

B. Implant Power Budget

1) Photovoltaic Cell Characterization: To evaluate the power budget of the implant, the photovoltaic cell is first characterized. Fig. 10 presents its current-voltage and power-voltage curves with a 35 mW laser. At peak power, the cell outputs 3.9 V with an efficiency of 59.4 %.

TABLE I

| Description                                      | Power (mW) |
|--------------------------------------------------|------------|
| Laser                                            | 35.0       |
| Eye optical losses (20 % of 35 mW at 850 nm) [39]| -7.0       |
| PV cell power dissipation (59.4 % of 28.0 mW)     | -11.4      |
| Implant standby power consumption                | -3.5       |
| Maximum available stimulation power              | 13.1       |

2) Available Stimulation Power: The implant power budget following the photovoltaic cell characterization is presented in Table I. With a maximum radiant power density of 4.06 mW mm$^{-2}$, a maximum of 36.5 mW can enter the eye for a 9 mm$^2$ photovoltaic cell, assuming uniform light distribution. The laser power entering the eye is set slightly below 35 mW. Because of the light absorption of the ocular medium, 20 % of the light is absorbed (7.0 mW is dissipated), so that 28.0 mW reaches the photovoltaic cell [39]. The latter then converts the beam reaching its surface with an efficiency of 59.4 % (11.4 mW is dissipated). The ASIC consumes 3.5 mW of standby power consumption, which leaves 13.1 mW of power for delivering stimulation pulses. The RF transmitter is excluded from the power budget as it is only utilized during calibration phases, not during typical operation.

3) Maximum Repetition Rate: During stimulation, the power consumption depends on the current amplitude required to trigger action potentials, which varies according to many factors, including electrode-neuron distance, electrode size, neuron physiology and pulse characteristics. For the current
Fig. 11. Power consumption of a single electrode at the average stimulation threshold for different pulse widths. The thresholds were averaged over three different retinas. The ASIC standby power consumption is excluded.

Fig. 12. Maximum pulse rate that can be distributed on the electrode array based on the available power.

For comparison, the Argus II device operates with electrode repetition rates between 3 and 60 Hz and uses 450 µs pulse widths across 60 electrodes [35]. In the proposed photovoltaic system, up to 14,000 pulses per second can be distributed across the array with a 450 µs pulse width. Dividing this by the 288 electrodes gives a maximum repetition rate of approximately 50 Hz per electrode, which provides a limited operational margin. By reducing the pulse width to 100 µs, the system can achieve up to 37,000 pulses per second, increasing the per-electrode repetition rate to about 128 Hz. This higher rate offers a more substantial operational margin, while staying within the typical range of repetition rate used by other epiretinal systems.

The 35 mW power source was chosen based on a photovoltaic cell with photosensitive dimensions of at least $3 \times 3 \text{ mm}^2$ in order to comply with the maximal permissible irradiance of $4 \text{ mW mm}^{-2}$ at 850 nm. However, the prototype was realized with a commercially available $1.7 \times 1.7 \text{ mm}^2$ from Broadcom to reduce its development costs, as opposed to developing a cell with custom dimensions. This difference in optical power density may affect the efficiency of the PV cell compared to the measurements presented in this paper. Designing the implant with the appropriate photovoltaic cell dimensions is critical to achieve permissible irradiance levels.

Additionally, the envisioned power delivery mechanism aims to provide a wide, uniform beam within tolerated irradiance limits. However, achieving perfect uniformity is impractical. This non-uniformity must be factored into the power budget of the final device, as the recovered power is maximized only when the implant is precisely aligned with the optical beam.

Although the proposed design can meet theoretical permissible irradiance levels, extensive safety evaluations would still be required to confirm the safety of the optical power delivery device and obtain regulatory approval. Furthermore, the presented power budget represents an absolute maximum and inherently includes sources of uncertainty. When designing a device and determining its maximum operating conditions, it is essential to incorporate a safety factor tailored to the specific design parameters.

B. Optimal Pulse Width Considering the Photovoltaic Cell Constraints

In order to achieve wireless operation, a retinal prosthesis must use a stimulation strategy that optimizes the power consumption. Although the instantaneous power consumption is higher for shorter pulse widths (Fig. 11), the energy per pulse is lower (calculated by the multiplication of the power by the pulse duration). This is caused by the lower charge thresholds required to elicit a response with shorter pulse widths [40]. This effect is expected to plateau with pulse widths significantly below the cell chronaxie, at pulse widths around tens of microseconds [41].

Practically, other factors limit the delivery of really short pulses. Shorter pulses require larger currents to deliver comparable amounts of charge. However, stimulators have a limited maximum current, especially in the case of wirelessly powered...
devices where high peak currents require a large energy reservoir. Additionally, the compliance voltage of the stimulator limits the pulse widths. At shorter pulse widths, the higher currents induce larger access voltages caused by the resistive component of the electrode-electrolyte impedance. With the proposed implant, the ±2.7 V compliance limit prevented reliable elicitation of a response with pulses below 100 µs.

The reported experiments have demonstrated effective stimulation with a single electrode and with a power significantly below the available power from the photovoltaic cell. This leaves headroom for activating multiple electrodes simultaneously. Fig. 12 presents the expected maximum repetition rate achievable given the experimental conditions. A higher repetition rate allows more accurate neural code reproduction in stimulation strategies based on a rapid sequence of electrical stimuli from a given dictionary of possibilities [37]. Rather than activating all electrodes simultaneously, as is common with subretinal implants, the implant optimally utilizes its power by sequentially activating electrodes with short, high-current pulses.

C. Experiment Limitations

1) Electrode Location: In this experiment, the electrode array is located subretinally instead of epiretinally to preserve the line-of-sight between the RGCs and the confocal microscope objective lens. The electrodes are consequently separated from the RGCs by the thickness of the retina, which varies between 100 to 200 µm. Such placement may evoke spikes in RGCs either through direct stimulation or via a network-mediated response involving bipolar cells. A similar study with identical electrode dimensions, placement, and material used synaptic blockers to determine the activation source. The study found that subretinal stimulation with very short stimuli (pulse durations ≤ 0.1 ms) directly stimulated RGCs, while longer stimuli (pulse durations ≥ 15 ms) indirectly stimulated RGCs through the retinal network [33]. These results were corroborated by observing consistent threshold charge densities at short pulse durations, both with and without the use of synaptic blockers.

Epiretinal stimulation often results in unintended axon bundle activation, leading to oblong rather than focal percepts [42], [43], [44]. While subretinal stimulation can circumvent this issue with appropriate parameter settings, some studies have explored methods to prevent axon bundle stimulation during epiretinal applications [45], [46]. These methods involve using small electrodes closely positioned to the neurons and incorporating recording capabilities to detect electrodes poorly positioned relative to axon bundles.

Thresholds in this experiment are in the order of 100 µC cm⁻² for 100 µs pulses. In contrast, study [43] reported similar thresholds for longer 100 ms pulses, but significantly lower thresholds (< 5 µC cm⁻²) for the same 100 µs pulse duration. This discrepancy likely results from the electrode-to-retina distance, as the electrodes in [43] were mounted on the inner retinal surface rather than the subretinal positioning in this study.

2) Electrode Size: Additionally, the 120 µm electrodes used in this experiment are relatively large compared to other experiments with electrode sizes as small as 5 µm [47], [48]. Using smaller electrodes is critical to attain the spatial resolution required for single-cell stimulation [49]. Alternatively, transparent indium tin oxide electrodes could be placed epi-retinally without obstructing the line of sight [43], but would not exactly reproduce the behavior of the diamond electrode array. With smaller 10 µm electrodes placed epi-retinally, stimulating with biphasic electrical pulses of 0.05-0.1 ms can result in thresholds in the order of 1 µA [46], [48]. This requires close proximity of the electrodes to the ganglion cells, which is achievable in in-vitro experiments, but can be highly challenging in a clinical context [50], [51]. Lower thresholds would reduce the power consumption and allow higher stimulation repetition rates.

3) Return Electrode: The experiments were done with a distant Ag-AgCl return electrode, contrasting with the local return electrodes intended for use in the implanted device version. Despite potential variations, a similar experiment cited in [33] did not report significant effects attributable to the type or placement of the return electrode.

4) Nature of the Experiments: The feasibility of the proposed system was demonstrated through ex vivo experiments using a different hardware configuration than the final envisioned system. Integrating the system requires mounting the implant components directly on the electrode diamond array, instead of using a separate interposer. Due to the non-standard nature of the diamond substrate, a custom assembly process must be developed to accommodate the relatively small interconnect pitch of 120 µm. Furthermore, adjustments to the experimental setup would be required to enable simultaneous imaging of retinal ganglion cells with the confocal microscope and powering the implant, now directly attached to the photovoltaic cell.

Additionally, these experiments were to demonstrate the feasibility of the proposed system for evoking responses in the retina and measuring population-based thresholds. Therefore, a detailed analysis of individual cellular responses was not performed. However, such an analysis was previously reported in [33], which comprehensively investigated the impact of stimulation parameters on individual RGC responses and their spatial resolution.

D. Comparison With Other Photovoltaic Architectures

Other implant architectures previously demonstrated the capability to elicit responses using 0.2 to 10 mW mm⁻² of irradiance at 905 nm [10]. These architectures target bipolar cells with subretinal electrodes. In contrast to these analog epiretinal devices that use direct coupling between photodiodes and electrode drivers, the current study introduces a digital epiretinal architecture targeting retinal ganglion cells, powered and controlled by a digital infrared link.

The NR600 photovoltaic system also demonstrates digital photovoltaic stimulation. However, it differs by its internal camera and stimulation strategy, where electrodes are activated in parallel with preconfigured patterns, reportedly targeting bipolar cells [52], [53]. In contrast, this paper combines an external camera with a faster data link, allowing sequential electrode activation without a predetermined pattern to target
retinal ganglion cells. Although multiple challenges remain in achieving precise single cell activation using this architecture, sequential stimulation is crucial for future epiretinal implants aiming to reproduce the retinal code on a spike-by-spike basis.

Different neural types respond very differently to electrical stimulation. Bipolar cells respond preferentially to longer pulse widths with low currents (around 25 ms) and retinal ganglion cells respond preferentially to shorter pulse widths with higher currents (around 0.1 ms) [33, 43, 54]. This leads to very different requirements in terms of stimulation strategies. When targeting bipolar cells, the longer pulse widths impose parallel stimulation strategies where most electrodes are activated simultaneously to achieve a reasonable refresh rate. Architectures based on photodiode arrays are well tailored to this approach, as each photodiode transduces the energy to the electrode to which they are coupled.

For retinal ganglion cells, shorter pulse widths of around 0.1 ms allow for multiple time windows in which to deliver stimulation pulses within the image integration time of the brain [55]. Thus, electrodes could be stimulated sequentially, one at a time or in small groups. In terms of power delivery, this corresponds to concentrating the available power to the few simultaneously active electrodes. The photovoltaic cell approach proposed in this paper has the capacity of concentrating the total incident optical power on the active electrode, thus allowing shorter pulses at higher currents, as deemed preferable for RCG stimulation.

VI. Conclusion

We presented an implant architecture based on an optical power and data link capable of eliciting a response in retinal ganglion cells while retaining the flexibility of a digital stimulation controller. The limited permissible radiant power entering the eye is sufficient to power the digital stimulation ASIC, ancillary circuits and deliver stimulation pulses that elicit a response in retinal ganglion cells. The proposed solution promises higher safety and reliability due to the possibility of encapsulating the device in a hermetic package without wires protruding of the implant and through the eyeball. With the goal of achieving meaningful visual acuity gains, next generations of epiretinal prostheses will need to deliver stimulation pulses that reproduce the neural code at a spatial resolution of cellular scale. Towards that goal, one of the next major challenges will be the realisation of a closed-loop device capable of wirelessly stimulating and recording with high electrode density.

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Conflicts of Interest

Steven Prawer was a shareholder in iBIONICS, a company developing a diamond based retinal implant. Steven Prawer and David John Garrett are shareholders and directors of Carbon Cybernetics, a company developing brain-machine neural interfaces. Arman Ahnood is a co-founder and shareholder in BrainConnect Pty Ltd., an Australian startup developing physiological and neurophysiological and interventional solutions for a range of neurological disorders.

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