Published in final edited form as:

Conf Proc IEEE Eng Med Biol Soc. 2016 August ; 2016: 4475–4478. doi:10.1109/EMBC.2016.7591721.

A PDMS-Based Optical Waveguide for Transcutaneous Powering of Microelectrode Arrays

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Abstract

Implantable microelectrode arrays (MEAs) usually have on-site electronics that need to be powered, both in neural recording and stimulation applications. Interconnecting wires between implanted electrodes and the outside world constitute a major source of complications. Our solution to this tethering problem is to design a light waveguide that can collect the optical power transcutaneously and transmit it to the microelectrode array where it is to be converted to an electric current. A polydimethylsiloxane (PDMS)-based waveguide was fabricated and its attenuation was measured *in vitro* and found to be 0.36 dB/cm. The skin flap of the thenar web space in the hand was used to test the photon collection efficiency of the waveguide in diffuse light. The efficiency of the waveguide alone was $44 \pm 11\%$ (mean $\pm$ std), excluding the attenuation within the thenar skin, as measured in 13 subjects with different skin pigmentations. These preliminary results suggest that a PDMS waveguide may collect and deliver optical power with sufficient efficiencies to deep structures inside the body. Optical powering scheme can solve the tethering and breakage problems associated with metal wire connections.

I. INTRODUCTION

Neural electrodes have employed state-of-the-art technologies to produce the finest 2D and 3D constructs for selective neural stimulation and neural activity recording in the central nervous system (CNS) [1–4]. Thus far these electrode arrays are mostly tethered to the outside electronics via fine metal wires [5]. The brain continuously moving inside the skull as a viscoelastic structure causes two fundamental problems that are at a trade-off with each other [6]. First, the shear forces generated at the tissue/electrode interfaces increase with the thickness of the wires and the ever increasing number of channels and cause a chronic scar tissue formation around the electrode [7]. Second, the lifetime of the wires is severely limited by the continuous movement of the electrodes and wire breakage becomes a serious issue as they are made thinner to minimize the tethering forces [8]. Unfortunately, there are

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Research supported by NIH Foundation.

no electrically conductive, biologically inert materials that can provide a highly flexible and yet durable interconnection and thus alleviate both problems at the same time.

With wireless powering and signal telemetry, the electrode arrays can float inside the neural tissue and both the glial scar formation and the wire breakage problems are solved together. To this end, other forms of energy transmission can be considered to power the implants, such as the optical, electromagnetic, or acoustic methods. Optical form of energy has a particular advantage that it can be guided and transmitted through an optical fiber [9], unlike the latter two. A glass or plastic optical fiber perhaps is not a better choice than a metal wire with regards to its flexibility and durability inside a viscoelastic medium. However, highly flexible and optically transparent materials like the PDMS can be considered for power delivery to neural implants [10]. The PDMS is also highly durable under bending forces and biocompatible for chronic implants. Currently, the usage of PDMS waveguides in powering implantable electronics is non-existent in literature.

The amount of energy that can be delivered through a PDMS-based waveguide is limited only by the maximum permissible exposure (MPE) of light on the skin. This value is regulated by American National Standard (ANSI) Z136.1 and given approximately as 400mW/cm^2 near the red wavelengths [11] within the optical penetration window of live tissue. Since the MPE limits the amount of light power per unit area of the skin, what is needed to transmit more power is to increase the cross-sectional area of the waveguide. Then the trade-off becomes that of the one between the thickness of the optical waveguide and the maximum power that can be delivered, where thicker waveguides will obviously be less flexible. A significant relief to this equation comes from the fact that the Young's modulus for PDMS is about five orders of magnitude smaller than that of metal wires [12]. Therefore, optical waveguides even much thicker than the metal wires typically used for neural electrodes will be extremely flexible and capable of reducing the tethering forces by some orders of magnitude. In addition, optical communication has a much wider bandwidth than, for instance, electromagnetic method of signal telemetry.

In this brief study, we report the efficiency of an optical waveguide in collecting light energy transcutaneously in human subjects and the attenuation of light intensity along its length as an investigation of its feasibility for neural prosthetic applications.

II. METHODS

A. Participants

Healthy subjects with the age of >18 years were recruited following the guidelines approved by New Jersey Institute of Technology Institutional Review Board (IRB). Subjects with a history of malignancy, as well as people who had scar or wound on the hand skin were excluded. All subjects wore laser safety goggles during data collection. All participants reported their age, gender, weight and height. Body mass index (BMI) was calculated using the weight/height² formula.

Thirteen volunteers with an average age of 29.6 ± 7.1 (mean $\pm$ std.) met the inclusion criteria and enrolled in this study. In this group, the frequency of gender was Females=3 and

Males=10; the BMI was distributed as underweight=1, normal=7, and overweight=5; the distribution of skin color was pale white/white=3, cream white/moderate brown=7, and dark brown/dark=3. The skin colors of subjects were grouped into 3 categories following the Fitzpatrick scale [13]: I – Type 1 & 2 (pale white, white), II – Type 3 & 4 (cream white, moderate brown) and III – Type 5 & 6 (dark brown, darkest brown) based on visual observation.

B. PDMS-Based Optical Waveguide Fabrication

Dow Coming Sylgard 184 and OE-6550 silicon elastomers were used to fabricate two-layered core-cladding PDMS optical structure with different refractive indices for confinement of light. The core was fabricated with OE-6550 which had relatively higher refractive index (RI=1.55) than Sylgard 184 (RI=1.41), the cladding material.

The fabrication of PDMS core began by mixing a curing agent (OE-6550A) with PDMS liquid prepolymer (OE- 6550B) for about 15 minutes (ratio 1:1). After achieving a homogenous mixture, the solution was placed in a vacuum for 30 minutes to remove air bubbles that were trapped within the mixture. The PDMS solution was then withdrawn into a 10cc syringe. The mixture was injected into a 17cm Teflon tubing (0.8mm ID, McMaster) using a 19 gauge needle. Once the tubing was filled with OE-6550, it was baked at 60°C for 8 hours. Then the tubing was stripped from the cured core by making a cut at one end and pulling apart.

The cladding mixture was prepared following similar steps using Sylgard 184A/B liquid prepolymers (mixing ratio 10:1). The core prepared earlier was held by fine tip forceps and dipped into the cladding mixture, gently pulled up and then hung inside the oven to be cured vertically at 80°C for 2 hours. The cladding had a thickness of approximately $300\mu\text{m} \pm 50\mu\text{m}$. The attenuation measurements were made with the waveguides that turned out to have a total diameter of $1.1 \pm 0.05\text{mm}$.

C. Measurements of Optical Attenuation

In order to determine the attenuation loss per unit length of the fabricated cylindrical waveguides, the cutback method was employed. A 15-cm waveguide was coupled to a red laser diode (635nm, 4.9mW, Thorlabs) and the light exiting from the other end was collected with a photodiode (Silicon Photodiode, 10mm x 10mm active area, Thorlabs) for measurements of attenuation in straight positioning. Light intensity measurements were taken for waveguide lengths of 15cm to 5cm by progressively cutting off 5mm lengths of the waveguide. The voltage output from the photodiode array was converted to a light power value using the photodiode responsivity (0.72A/W) provided by the manufacturer. The total attenuation was determined using

$$A = -10 \log (P_{\text{out}}/P_{\text{in}}) \quad (1)$$

where A is the attenuation in decibels (dB), P_{in} is the power entering the waveguide. P_{out} is the output power measured by the photodiode.

D. Light Transmission Through the Skin

The skin flap (thenar web space) between the thumb and the index finger in both hands were cleaned using a 70% ethanol to remove residues that could affect light measurements. The setup consisted of a semiconductor laser module (650nm, 4.9mW, Flexpoint, Uberlingen, Germany) as a light source and a photodiode (FDS1010-CAL, 10mm x 10mm active area, Thorlabs, USA) as a detector. The diameter of the collimated light beam was set to Ø1.1mm by passing it through a pinhole and aligned with the Ø1.1mm waveguide on the other side of the hand (Fig. 1).

A commercial caliper was utilized to hold the light source and 5cm long waveguide in alignment (Fig. 1). The subject's hand was placed between the caliper jaws facing down and tightened gently to make a good skin contact while being careful not to compromise the blood circulation. The thickness of the skin was recorded as well as the light power (Pout) exiting on the ventral site of the thenar skin flap.

The local skin temperature was measured by a T-Type thermocouple sandwiched between the laser source and the skin, after allowing the temperature to stabilize for 5 seconds.

E. Data Analysis

Five measurements were taken from each hand first with the photodiode placed adjacent to the skin on the ventral side without the optical waveguide. These measurements were averaged as a representative amount of light transmitted through an average skin thickness for this particular hand (grey dot in Fig. 2). Multiple-point measurement was seen as necessary to eliminate outliers due to misplacement of the caliper across very thin or thick areas of the thenar skin. Five more measurements were made in the same hand with the waveguide inserted between the hand and the photodiode as shown in Fig. 1. A linear line was fit to the measurements and the transmission at the same skin thickness that was found as the average thickness without the waveguide was calculated by interpolation/extrapolation (yellow dot in Fig. 2). The ratio of light transmission at the same skin thickness with and without the waveguide (yellow dot / grey dot) was taken as the attenuation due to the waveguide, which in fact consisted of two parts. First, photons enter the optical waveguide in a very wide range of angles after passing through a scattering medium, such the human skin. Some of those photons cannot be contained in the waveguide because of the large angle of entry. Second, light is attenuated along the waveguide due to absorption. We could factor out the latter by measuring this value separately with a collimated laser beam (Section A in results). The net result was the light collection efficiency of the waveguide at the interface with the skin.

III. RESULTS

A. Attenuation in the PDMS Waveguide

The cylindrical optical waveguides we prepared had a good confinement of red light ($\lambda=635$ nm) and relatively low attenuation of approximately 0.36 ± 0.03 dB/cm (mean $\pm$ std, n=5, $R^2=0.96$) in straight positioning as measured separately (Fig. 3).

B. Transmission through the Hand Skin

The incident light intensity on the skin was $375\mu\text{W}$, as measured after the pinhole. The light intensity that was transmitted through the skin and received by a $\varnothing 1.1\text{mm}$ circular photodiode area ranged between $0.3\mu\text{W}$ to $1.0\mu\text{W}$ in different subjects while the skin thickness varied from 4.6mm to 2.6mm (squares in Fig. 4). This substantial attenuation of light intensity is mostly due to scattering of photons inside the tissue. The circles in Fig. 4 show the light intensities collected with the waveguide (also $\varnothing 1.1\text{mm}$), rather than the photodiode, after factoring out the attenuation along the waveguide that was shown in Fig. 3. Thus, the ratio of the circles over the squares in Fig. 4 in each hand gives the light collection efficiency of the waveguide in each case. The light collection efficiency averaged across all subjects (both hands) was found to be $44\pm 11\%$ (mean $\pm$ std), which was not significantly affected by the gender, the BMI, or skin pigmentation ($p>0.5$ for all groups). Overall, the 5cm long waveguides of this study had a total transmission efficiency of about 29% of the light that is transmitted to the other side of the hand skin (i.e. 44% of collection efficiency multiplied by 66%, which is the $0.36\times 5=1.8\text{dB}$ of attenuation along 5cm of its length).

C. Temperature Elevation of the Skin

The local hand temperature remained the same in all subjects during data collection due to low power illumination. The maximum change detected was 0.1°C , which is considered within the body maintaining temperature [14].

IV. DISCUSSION

The purpose of this study was to measure the light collection efficiency of PDMS-based optical waveguides across the human skin. We designed an experimental setup to determine how much of the laser energy could be collected by the waveguide and transmitted to a photodiode after having been scattered passing through the skin. The electrical power generated at the photodiode may then be utilized to power an implantable microelectrode array in order to avoid wire interconnections. There are some critical factors such as pigmentation and the thickness of various skin layers that directly effect the light absorption and scattering inside the tissue. The location on hands that we used for measurements consist of a layer of dermis between two layers of epidermis on each side. Because a small pinhole was used to collimate the beam, the incident power ($375\mu\text{W}$) was only a small fraction of what comes out of the laser source. The incident light beam was substantially attenuated also by the skin (300-500 times) at varying amounts depending on the thickness of the thenar skin in different subjects. The thenar skin flap was chosen only for convenience in this study. The skin thickness varies greatly around the body and the total transmitted light power will change depending on the site. The total power collected by the waveguide will also be a function of its diameter and the attenuation along its length. With the collection efficiencies and the longitudinal attenuations that we report here, one can estimate the total power for different fiber diameters and lengths. Fabrication techniques can be improved to achieve particularly blemish free PDMS surfaces and thus obtain better containment of light inside the waveguide. However, such imperfections will become more influential when the waveguide is bent, twisted, or stretched, and the attenuation measurements reported here

should be repeated under those conditions. But, the collection efficiency of the waveguide at the tissue-waveguide interface should be independent of the attenuation along its length.

V. CONCLUSION

Our results demonstrate that light energy could be collected transcutaneously and transmitted to a deeper location inside the neural tissue with high efficiencies using cylindrical PDMS waveguides. Considering the maximum allowed light exposures per unit area on the human skin, this efficiency implies that significant levels of optical power can be transmitted to an implant without using metal wires that frequently break and cause the implant to fail. A PDMS waveguide would generate much smaller tethering forces compared to metal wire interconnects. A trade-off exists between the size of the waveguide and the total power that can be delivered. A larger cross-sectional area of the waveguide would allow more optical power to be transmitted before reaching the exposure limits on the skin surface, however, generating larger tethering forces potentially on the implant.

Acknowledgment

This study was supported by the National Institute of Health Grant (NINDS, R01 NS0785).

REFERENCES

- [1]. Agorelius J, Tsanakalis F, Friberg A, Thorbergsson PT, Pettersson LM, and Schouenborg J, "An array of highly flexible electrodes with a tailored configuration locked by gelatin during implantation-initial evaluation in cortex cerebri of awake rats," *Front Neurosci*, vol. 9, p. 331, 2015. [PubMed: 26441505]
- [2]. Chen C, Zhang T, Zhang Q, Feng Z, Zhu C, Yu Y, et al., "Three-Dimensional BC/PEDOT Composite Nanofibers with High Performance for Electrode-Cell Interface," *ACS Appl Mater Interfaces*, vol. 7, pp. 28244–53, 12 30 2015. [PubMed: 26550840]
- [3]. Rector DM, Yao X, Harper RM, and George JS, "In Vivo Observations of Rapid Scattered Light Changes Associated with Neurophysiological Activity," in *In Vivo Optical Imaging of Brain Function*, Frostig RD, Ed., 2nd ed Boca Raton (FL), 2009.
- [4]. Ware T, Simon D, Hearon K, Liu C, Shah S, Reeder J, et al., "Three-Dimensional Flexible Electronics Enabled by Shape Memory Polymer Substrates for Responsive Neural Interfaces," *Macromol Mater Eng*, vol. 297, pp. 1193–1202, 12 1 2012. [PubMed: 25530708]
- [5]. Biran R, Martin DC, and Tresco PA, "The brain tissue response to implanted silicon microelectrode arrays is increased when the device is tethered to the skull," *J Biomed Mater Res A*, vol. 82, pp. 169–78, 7 2007. [PubMed: 17266019]
- [6]. Perge JA, Homer ML, Malik WQ, Cash S, Eskandar E, Friehs G, et al., "Intra-day signal instabilities affect decoding performance in an intracortical neural interface system," *J Neural Eng*, vol. 10, p. 036004, 6 2013. [PubMed: 23574741]
- [7]. Sridharan A, Rajan SD, and Muthuswamy J, "Long-term changes in the material properties of brain tissue at the implant-tissue interface," *J Neural Eng*, vol. 10, p. 066001, 12 2013. [PubMed: 24099854]
- [8]. Alex Mohit A, Samii A, Slimp JC, Grady MS, and Goodkin R, "Mechanical failure of the electrode wire in deep brain stimulation," *Parkinsonism Relat Disord*, vol. 10, pp. 153–6, 3 2004. [PubMed: 15036170]
- [9]. Abdo A, Sahin M, Freedman DS, Cevik E, Spuhler PS, and Unlu MS, "Floating light-activated microelectrical stimulators tested in the rat spinal cord," *J Neural Eng*, vol. 8, p. 056012, 10 2011. [PubMed: 21914931]
- [10]. Ersen A and Sahin M, "PDMS-based Optical Conduit for Powering Untethered Neural Microstimulators," in *University of Minnesota, Minneapolis, MN*, 2015, 4 16-17.

- [11]. Institute ANS, ANSI Z136.1- Safe Use of Lasers. Orlando, Florida: Laser Institute of America, 2014.
- [12]. Ratner BD, Biomaterials science : an introduction to materials in medicine, 3rd ed. Amsterdam ; Boston: Elsevier/Academic Press, 2013.
- [13]. Fitzpatrick TB, "The validity and practicality of sun-reactive skin types I through VI," Arch Dermatol, vol. 124, pp. 869–71, 6 1988. [PubMed: 3377516]
- [14]. Axelrod YK and Diringer MN, "Temperature management in acute neurologic disorders," Neurol Clin, vol. 26, pp. 585–603, xi, 5 2008. [PubMed: 18514828]

Figure 1.

Experimental setup for light collection through the hand skin. Inset: Representative image of right hand showing the point of measurement spotted in orange.

Figure 2.

Representative plot for determination of percent transmission in the left hand of subject 10 for a 5cm long waveguide. Grey dot: average value of 5 measurements without the waveguide, yellow dot: measurement with the waveguide at the same skin thickness calculated by extrapolation.

Figure 3.

Average attenuation of five cylindrical waveguides as a function of waveguide length. Error bars: one-standard deviation. Black dash-line: linear fit.

Figure 4.

Comparison plot to show the difference between the values of measurements with waveguide (circles) and hand alone without the waveguide (squares) as a function of thickness (52 data points from left and right hands of 13 subjects). Vertical dash lines connect the measurement pairs from the same hand.