System For Stimulation of the Spiral Ganglion Using Infrared Light

Final Assignment

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Abstract

Cochlear implants have historically used electrical stimulation of the auditory nerve. The system which follows makes the case that infrared light is a superior stimulus. Advances are shown in software configurability, density of stimulus, and delivery of light through novel optics. This system would make a solid platform for future study of infrared light as a replacement for electrical cochlear stimulation.

Introduction

In the United States there are about 4000 new cases of sudden deafness each year and approximately 2-3 children per 1000 are born deaf or hard-of-hearing.[1] Cochlear implants aim to restore hearing function to this patient population, specifically those whose hearing loss is sensorineural. Current commercial cochlear implants allow for the comprehension of speech in quiet environments, however they are severely deficient in noisy environments and for complex stimuli such as music.

Traditional implants excite the auditory nerve fibers of the cochlea using electric impulses delivered by an array of electrodes. These electrical implants are generally limited to about 20 channels due to overlapping electric fields from neighboring electrodes and performance generally starts to plateau at less than 20 channels. However, simulation studies have shown increasing levels of performance with larger numbers of electrodes which suggests that the performance plateau is specific to electrical implants.[2,3] Additionally, once placed, the position of each electrode cannot be adjusted. Electrical stimulation poses a number of problems the most prominent being the large number of sensory aberrations and poor targetability. Still by drawing on the technologies developed for these implants they can be easily adapted to conform to INS. Recent studies have shown that infrared light stimulation of the Crista ampullaris in the toadfish, auditory nerve in gerbils, and cats[4-6] can evoke electrical responses suggesting that an Infrared (IR) based cochlear implant could be feasible.

In light of these new developments we are proposing a complete system for IR based cochlear stimulation (Figure S1, supplemental). By integrating Digital Micromirror Device (DMD) technology and microscale monolithic flexible waveguides we can provide many more points of stimulation at a much finer resolution.

Biology

Cochlear Function

The Cochlea is an extremely complex and finely tuned hydro-mechanical sensory organ which serves to transduce the mechanical action of audio waves on hairs into electrical signals to be sent to the brain. For the purposes of successful Infrared Nerve Stimulation (INS) it is necessary to understand a few components of this intricate piece of anatomy. First, we must be able to build and maneuver such a device into a position that allows for ideal nerve stimulation. In order to best transmit auditory information with infrared light it is necessary to stimulate the spiral ganglia. To accomplish this, the cochlear implant must be inserted into the Scala-tympani by means of the circular window. From there the infrared light can be pointed directly at the Spiral Ganglia (Figure 1).

Cochlear Cross Section

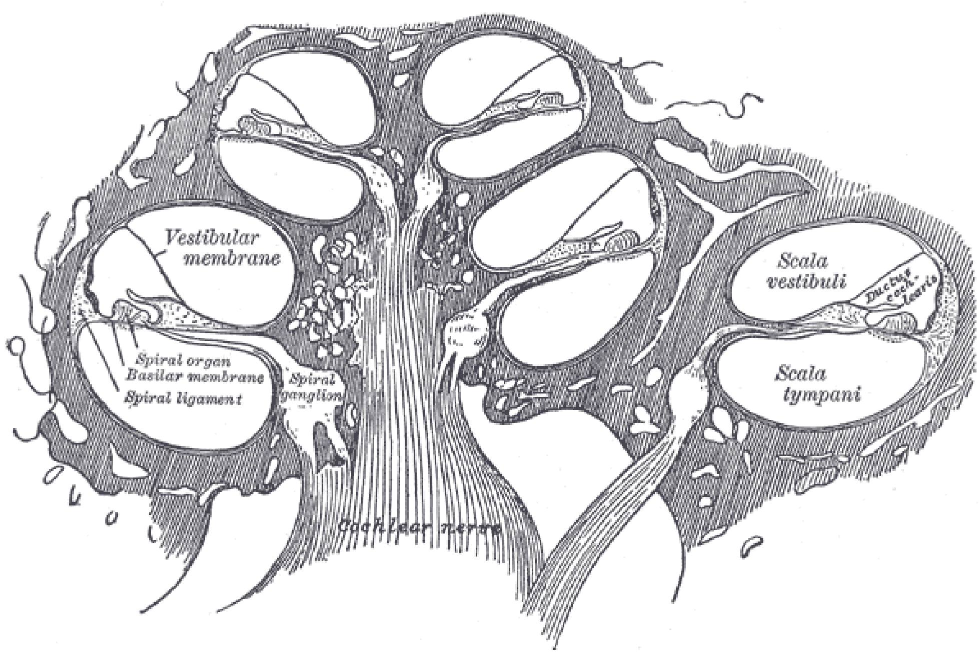

Figure 1 - ***from the 20th U.S. edition of*** **Gray's Anatomy of the Human Body**

The perception of pitch within the cochlea is determined primarily by two factors: spatial and temporal. For frequencies up to 5 kHz temporal phase locking plays a partial role in encoding pitch. As such, accurately representing these frequencies requires a device to be capable of stimulating the spiral ganglion nerves with a periodicity matching the apex of the wave for a given frequency up to about 4-5 kHz (Figure 2a). For frequencies up to and exceeding that range the spatial location of stimulation plays a much larger role. This is excellent for the prospects of encoding higher frequencies since it becomes unnecessary to work with high frequency signals. Instead, within the high frequency regime, a device merely needs to be able to spatially map stimulation areas (Figure 2b). Since spatial mapping plays a role at lower frequencies also those frequencies will require both spatial and temporal mapping.[7,8]

Phase Locking vs Spatial Mapping

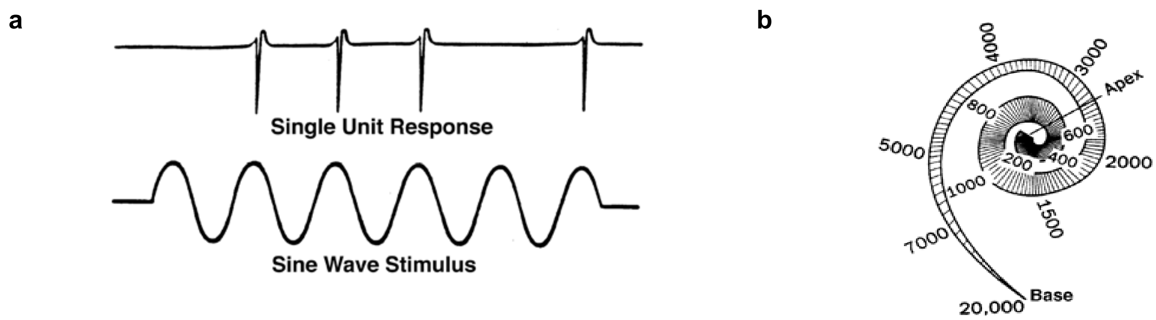

Figure 2: a. Temporal Frequency Phase-locking, b. Spatial Frequency Mapping

Lastly, while the biological mechanism behind INS remains a mystery, it is generally agreed that a change in the membrane potential induced by the photothermal effect is the driving factor. Which ion channels allow for INS is still up for debate but two possibilities are Ca^{2+} and Na^{+}.[6] The safety of chronic INS is also a concern. However, studies have confirmed that below a certain temperature threshold chronic damage is not evident.[4-6]

Electronics

To recreate the human perception of sound, electronics and optics are employed and conversions from analog to digital and back again must occur. The beginning of the flow is a microphone capturing the external sound to be translated into cochlear stimulation. The sound must be passed through an analog-to-digital converter followed by some digital signal processing. That data is then converted to light through a DMD mirror array which steers an IR laser to activate waveguides. At the end of the flow is the cochlea, stimulated by IR light to recreate the frequency originally presented to the microphone. (Figure 3)

From Sound to Light

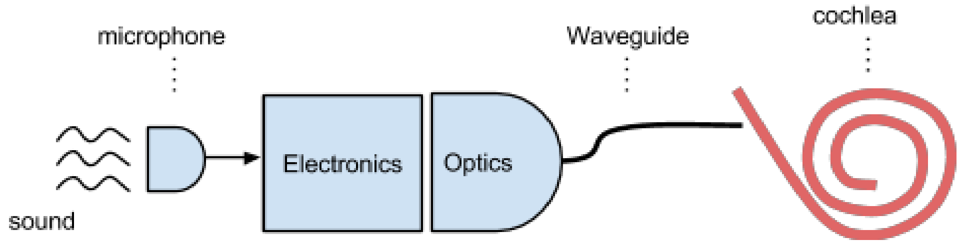

Figure 3

Audio signal input

The initial electronics are analog since they interface with the physical world. The microphone of the cochlear implant is important in that it should be tuned toward the frequencies of human speech (20 Hz to 4000 Hz) which are considered the most important area of the audible range. The microphone is often situated behind the ear and commonly in the same housing as further speech processing electronics. This analog audio signal is then passed to the signal processing portion of the system.

Signal processing

The digital electronics enter once the audio from the microphone is converted to digital data (Figure S2). An analog to digital converter (ADC) is a functional block integrated in many DSP chips such as the Blackfin ADSP-BF506F from Analog Devices where the converter offers 12-bits of precision.[9] After the external sound has been converted into digital data, a fast fourier transform (FFT) calculation must be performed on it to determine its frequency components. This mathematically intensive calculation is done in software running on the DSP chip which is streamlined for high speed math such as this. To conserve as much power as possible, the DSP will be tuned to use the slowest clock possible to handle the bandwidth of frequency stimulation. Additionally it will operate on the minimal voltage threshold and take advantage of all sleep modes designed into it.

The FFT derived frequencies are then compared to a table which was created shortly after the patient received their implant. (Figure S3) This is a calibration table designed to take a frequency range and map it to a specific fiber / cochlear location. Since the resolution of the DMD array is much greater than the fiber array, intensity of light can be controlled by the number of mirrors directing light on a given fiber and thus stimulation at that endpoint. For example, if 50% of the mirrors which are directed toward a fiber are directly facing it then the intensity will be 50% of the maximum. Thus intensity is tunable to a fine degree given the 100 to 1 ratio of DMDs to fibers.

Many advances have been made in audio processor for cochlear implants such as Automatic

Gain Control to tune the Input Dynamic Range[10] given the listening situation. Even greater dynamic mapping is offered through this system. Post-surgical calibration can be performed with traditional cochlear implants but this software element allows for adaptive enhancements in the system over time. A patient could feasibly have an application to fine-tune their hearing experience without the need for a doctor's visit. Similar studies have attempted to close the learning loop at the most optimal time, that is, during normal use.[11]

Signal to DMD

After the signal leaves the DSP chip it is passed along to a DMD controller. The controller will take the mirror location information from the DSP device and write to the SRAM elements which exist in each of the DMD mirrors. The mirror will not move until a clock pulse from an analog control chip (such as the DLP® DLPA200 DMD Micromirror Driver) arrives. At this clock edge the SRAM element value is sampled and the movement actuated at the DMDs torsion hinge. This approach only consumes power during transition and not when stable. The control device (DLPC200) will be implemented using a Field Programmable Gate Array (FPGA) device so that design upgrades can be made without having to replace any parts.

The signal which is sent to the DMD mirrors will be encoded in digital on-off pulses. (Figure S6) These will be exhibited as a data stream of ones and zeros. This data will be mechanically converted into light of the same modulation using the optics to be described in the following section.

Optics

Light Source

IR lasers are ideal due to their targetability and coherent nature. For the time being it is necessary to use an external laser source since implantable laser technology developed for INS is still too large and inefficient to be seriously considered.[12] Using an external laser has several benefits. First, with an external source, power is not an issue; also, replacing said laser would not require a surgical procedure. However, in order to use an external source it would be necessary for an optical fiber to penetrate the epidermis which could lead to infection. The risk should not exceed that of a typical surface piercing.

INS is best done with IR wavelengths with water absorption curves matching the ideal depth stimulation. In the case of the Spiral Ganglion, located in the cochlea, wavelengths of approximately 1470 or 1850 nm have been shown to work well.[12] While the majority of studies done on INS have used low frequency pulsed infrared lasers it has been demonstrated that a continuous wave (CW) laser can be just as if not more effective.[13] In the case of our system design a continuous laser is ideal for several reasons. Primarily it allows for signal modulation to be done exclusively with the DMD array so that a large variety of pulse widths and intensities can be created simultaneously. Also, it lowers the cost and complexity of the electronics necessary.

Recent studies have demonstrated that temperature is the primary factor in INS. The amount of stimulation is a result of the $\frac{dT}{dt}$ (change in temperature) above a threshold. The same stimulation done with a pulsed laser can be achieved with a continuous 7.5 Watt diode laser. Fortunately, it has been shown that 450 mW/cm^2 is sufficient if the nerve tissues are already warmed as they would be during chronic stimulation. Therefore, for our purposes we can use a IR diode laser with an output of one Watt if temperature regulation is in place.[13,16]

Optical Multiplexer and Signal Modulator

The combination of optics with the DMD array to allow for path selection is done with a simple integrated design (Figure S7). This portion of the system is a self-contained allowing for other components to be interchanged as needed. The package contains three connection points: one electrical, and two optical. The electrical connection goes to the DMD Control chip. The first optical connection is the input for the fiber optic laser source. At this junction a beam expander comprised of a concave lens followed by a convex lens expands and collimates the light to a diameter matching the width of the DMD array.

The digital micromirrors themselves are capable of pointing to two different locations, each 12 degrees off of the perpendicular. Typically one location is the active one and the other is the passive or dead space. In this design, one location will be the origin of the waveguides and the other will be a “dead space” from which the signal will not be transmitted. (Figure S4) This dead space is simply an area to point the laser when it shouldn’t be sent to the waveguide.

An individual DMD cell (Figure S5) is 10 μm square. The array is capable of holding 10,000 mirrors per square mm. Below the proprietary mirror cell is a CMOS SRAM logic bit which pulls the cantilevered mirror support post in one direction on a torsion hinge when storing a high voltage value (logic 1) and pulls the torsion hinge in the other direction when storing a low voltage value (logic 0).

The original design was intended for visible light only, but advances in mirror coating have allowed the usage of infrared light as well. The efficiency of the system is best for visible light with a reflection percentage of approximately 95%. However, with infrared light the efficiency of the mirror drops as low as 50% forcing the laser to use more power than if the light were visible. A solution to this was found in coating the mirror face with gold. The reflectivity of gold for IR light is as high as 98%.[17] For these purposes we would start with the DLP® 0.55 XGA Series 450 DMD (modified with gold coating), DLPC200® Digital Controller, and DLP® DLPA200 DMD Micromirror Driver[18].

The final connection belongs to the optical output which consists of an interface of many optical waveguides. In order to collect the light from the DMD mirrors and focus it into individual waveguides a square microlens array is employed. The square microlenses allow 98% of the incident light to be collected. With the addition of antireflection coatings the total attenuation caused by the μlens-waveguide interface can be kept below 3%.[19] For this system there is a ratio of 1 μLens to 100 mirrors. Since the mirrors are 10 μm^2 each μLens must be 100 μm^2. This allows for intensity to be controlled in increments of one percent. Given a diameter of 5mm for the collimated light, this system would contain from 1800 to 1900 potential stimulation points.

Part	**Typical Power Consumption**	**Size**
Blackfin ADSP-BF506F	~ 50 mw	12mm x 12mm
DLP 5500 - .55 XGA DMD	Max 2 mW/cm2	14mm x 14mm
DLP C200 - Controller	Data Dependent	29mm x 29mm
DLP A200 - Analog Driver	Data Dependent	12mm x 12mm

Waveguide

To direct the light from the lens array to the many stimulation sites a flexible waveguide can be fabricated. By creating a single monolithic waveguide through soft lithography we hope to simplify the construction of the device. Additionally, this approach can be scaled to a greater density of stimulation points as MEMS fabrication advances. We aim to create waveguides with a cross section of 50μm x 50μm at a spacing of 50μm. The total number of waveguides layers depends on the flexibility of the unit, but theoretically we could create a waveguide channel for each μLens.

There are a variety of substances which can be used for a waveguide core. However, if the final waveguide must be flexible as in this application, the waveguide components are limited to polymers materials. Some materials which are flexible and transparent to IR/NIR wavelengths are listed in the table below. For the proposed waveguide a core of SU-8 will be cladded in PDMS. The primary benefit to using SU-8 is its high transmissibility in the IR range (~95%, Figure S8) and in addition to being flexible, PDMS is a very biocompatible material.

Polymer Material	**Conventional Use**	**Refractive index**
SU-8	UV-cured photoresist	1.58[20]
UV15	UV-cured photoresist	1.50[20]
PDMS	stamp resin	1.45[21]

Topas 5013	optical polymer	1.53[22]

To fabricate the waveguide a modified microtransfer molding (μTM) procedure will be used. First a silicon master will be etched in two steps on the (100) plane. A buried oxide layer is first created at a depth of 50 μm. Next the waveguide ends are defined by masking off a section in a zigzag pattern to accommodate the different lengths of each waveguide. With an etching solution of TMAH + Triton, the (110) plane acts as a 45 degree etch stop to achieve a very smooth surfaces.[15] Next a new mask and resist are added to define the lengths of the waveguides and the interguide silicon is removed using the RIE process. To facilitate a denser packing of waveguides multiple waveguides of different lengths ranges will be stacked. A single silicon master is created for each of these ranges.

Fabrication of μTM master

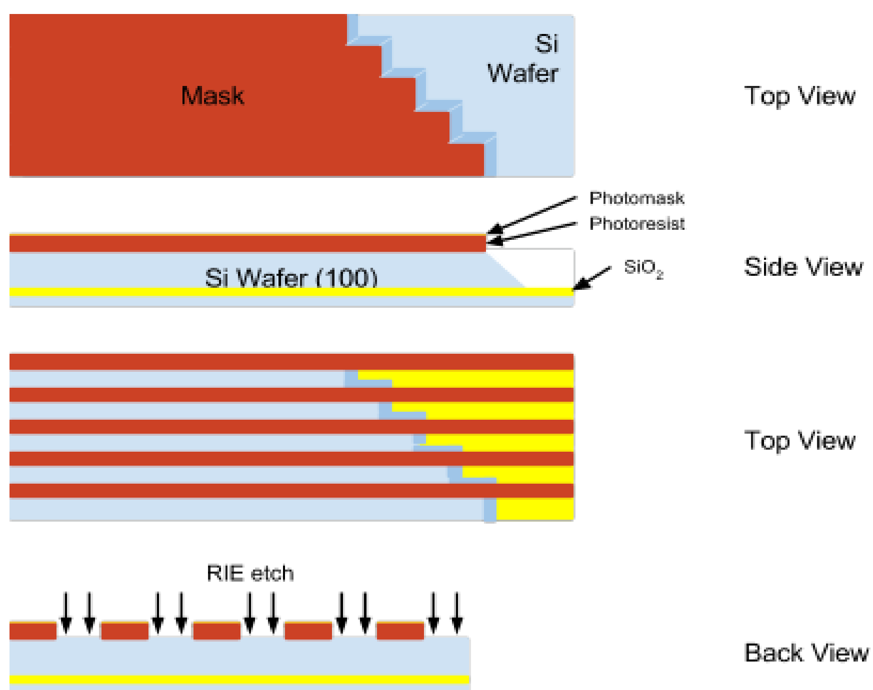

Figure 4

Next we create the PDMS mold from the master. However, unlike μMT, the mold will become a part of the device where it will function as the cladding of the waveguide. After the PDMS is cured it is peeled off the master and a photoresist-mask is created for the channels. A thin layer of gold is deposited using CVD into each channel. Finally, the waveguide channels can be filled with SU-8 and then physically scraped level. Once the SU-8 is cured with UV light the waveguide layer is complete (no transfer step). Each layer of the wave can then be ionized and fused together. Thus the final waveguide is mostly cladded in PDMS with small sections interfacing with air.

Fabrication of PDMS cladding and SU-8 Core

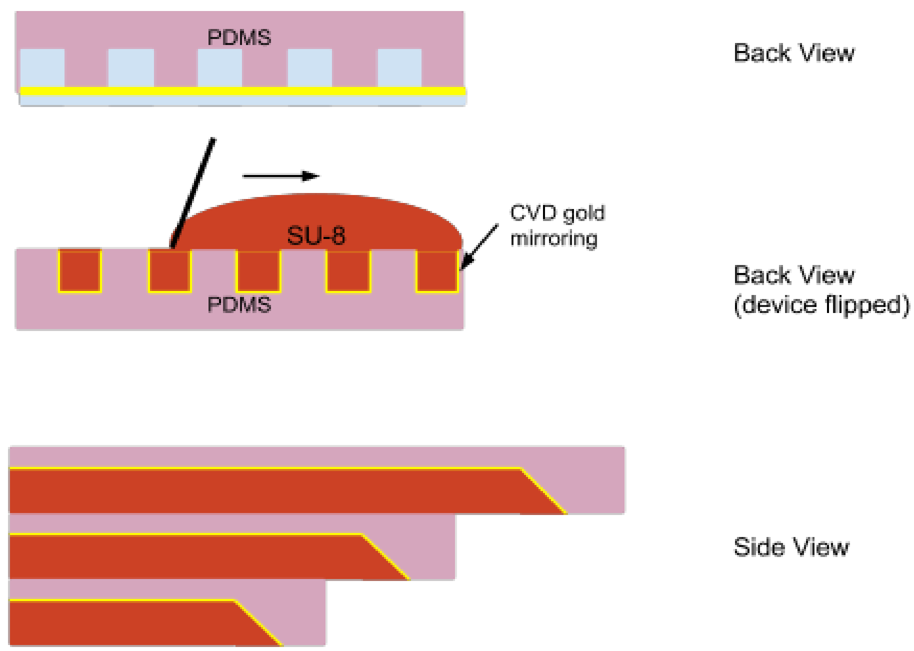

Figure 5

Attenuation

Below are estimates on attenuation drawn from data sheets and experimental work. We chose worst case values to provide plenty of room for error.

System Component	Estimated Attenuation
Beam Expander	<1%
μLens-Waveguide interface	<3%
Gold Coated DMD	<5%
Waveguide (no mirroring)	0.2dB/cm[26] - 3dB/cm[20] loss
- 6cm (0.2dB/cm)	25%
- 10cm (0.2dB/cm)	37%
Estimated Total Attenuation	<46%
Laser Power Needed	1 Watt

Cost Analysis

Although cochlear implants are expensive, especially including the surgical procedure, a number of studies on the price-utility benefits have concluded that the expenditures are more than recouped. One study shows that over the first 12 years for a child with partial hearing loss $78,000 was saved on special-needs programs; for a child with severe to complete hearing loss the amount saved was even greater.[23]

In the system we propose material costs are far below what might be spent on research and development. The total material cost is potentially less than $5000 as a prototype without applying savings for large scale production. In contrast, the price of the current procedure for the implantation, rehabilitation, and device runs $45,000 to $125,000 in the United States.[24,25] This cost savings was accomplished through off the shelf parts, the choice of a DPSS laser a DMD array and the use of PDMS in the waveguide. The largest cost was the silicon template, which could not be avoided.

Future Considerations

The currently proposed cochlear implant just scratches the surface of what is possible. The greater number of stimulation channels as compared to electrical stimulation allows for better accuracy for listening comprehension and a more enjoyable music listening experience. In order to determine the point of diminishing returns for so many possible stimulation channels, we could perform both animal and human trials. Animal trials could consist of auditory cortex imaging studies or behavioral studies. Human trials would involve accuracy listening tests in high noise environments. Due to the redundant number of light guides, we can further tune the prosthetic by selecting for the most sensitive locations in the subject.

If the cochlear implant electronics are given a communication interface (low power Bluetooth would be ideal) a data feedback loop would open up a world of possibilities. A patient could install an application on some smart portable device and configure their implant to have higher resolution during lectures, better response in the musical range during concerts and even turn it down when ambient noise is oppressive. The application could also sample ambient noise through the microphone in the smart device and the cochlear system and ask the subject to give positive or negative responses through the touch screen depending on their personal experience of the sound. From a technical perspective low power modes could be enabled or disabled as needed. The implanting doctor could change the patient’s mapping of frequency to fiber and intensity as time passed and could do so with a much greater level of control than in current designs. Finally, an artificially intelligent algorithm could one day use pseudo-random configurations and rate the patient’s happiness to “stumble upon” even better settings.

Additional considerations should be given to refinements at the light emitting end of the device. For example, by using photonics crystals the number of channels could be increased while still utilizing the same number of mirrors and waveguides. By designing wavelength specific routes, we can selectively stimulate different regions on the cochlea based on the fine variations in the laser source waveform. In addition, due to the largely variable water absorption curve over certain IR frequency ranges, a system might be employed to targeting different depths by changing wavelengths. Another possibility is to exploit the coherent waveform of the IR light source to spatially select certain regions through constructive and destructive interference.

Additional improvements could be made in how we handle the “dead-zone” light. One option would be to simply place a CMOS or CCD image sensor in the dead space plane. In doing so it would be possible to monitor the intensity levels for each simulation point as well as any issues with the mirror. A second options could be to utilize fibers in the deadspace to double the number of simulation points to choose from. By implementing a micro scale shutter we can replace the dead space with more stimulation points and dynamically choose which point is not used based on stimulation quality.

Closing the loop between stimulation of the nerve and would be another area of future consideration. In this way, the system could become more efficient delivering just the right amount of IR stimulus needed.

As VCSELs improve it may prove more practical to switch to a system of many pulsed sources instead of one. Using this type of light source it may be possible to do away with the majority of waveguides and optics by creating a device with one VCSEL for each simulation point in-vivo.

Conclusion

Infrared Nerve Stimulation has a great deal of potential for cochlear and other nerve interface applications. However, it is clear that the current state of technology and understanding behind INS stimulation requires significant improvement before a truly superior cochlear implant can be achieved. The system proposed will provide a powerful prototype for teasing out the precise parameters best suited to the task. The wealth of tunability in software, quantity of stimulation points and the original waveguide solution are all novel additions to the field. The large range of frequency, intensity, and spatial selectivity makes it an ideal first step towards realising substantial improvements in implant technology for the deaf and hearing impaired. Furthermore, it is possible that our technology can be adapted to a variety of other biological applications where precise nerve stimulation is needed such as the retina or even the brain.

Citations

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Similar Patents

- Lockheed Martin
    - OPTICAL-STIMULATION COCHLEAR IMPLANT WITH ELECTRODE(S) AT THE APICAL END FOR ELECTRICAL STIMULATION OF APICAL SPIRAL GANGLION CELLS OF THE COCHLEA http://www.faqs.org/patents/app/20130023967#ixzz2kgyxS17O
    - Cochlear implant and method enabling enhanced music perception US 20130023962 A1
- Richter
    - Apparatus and methods for optical stimulation of the auditory nerve http://www.google.fm/patents/US7833257 issued 2010

Supplemental Figures

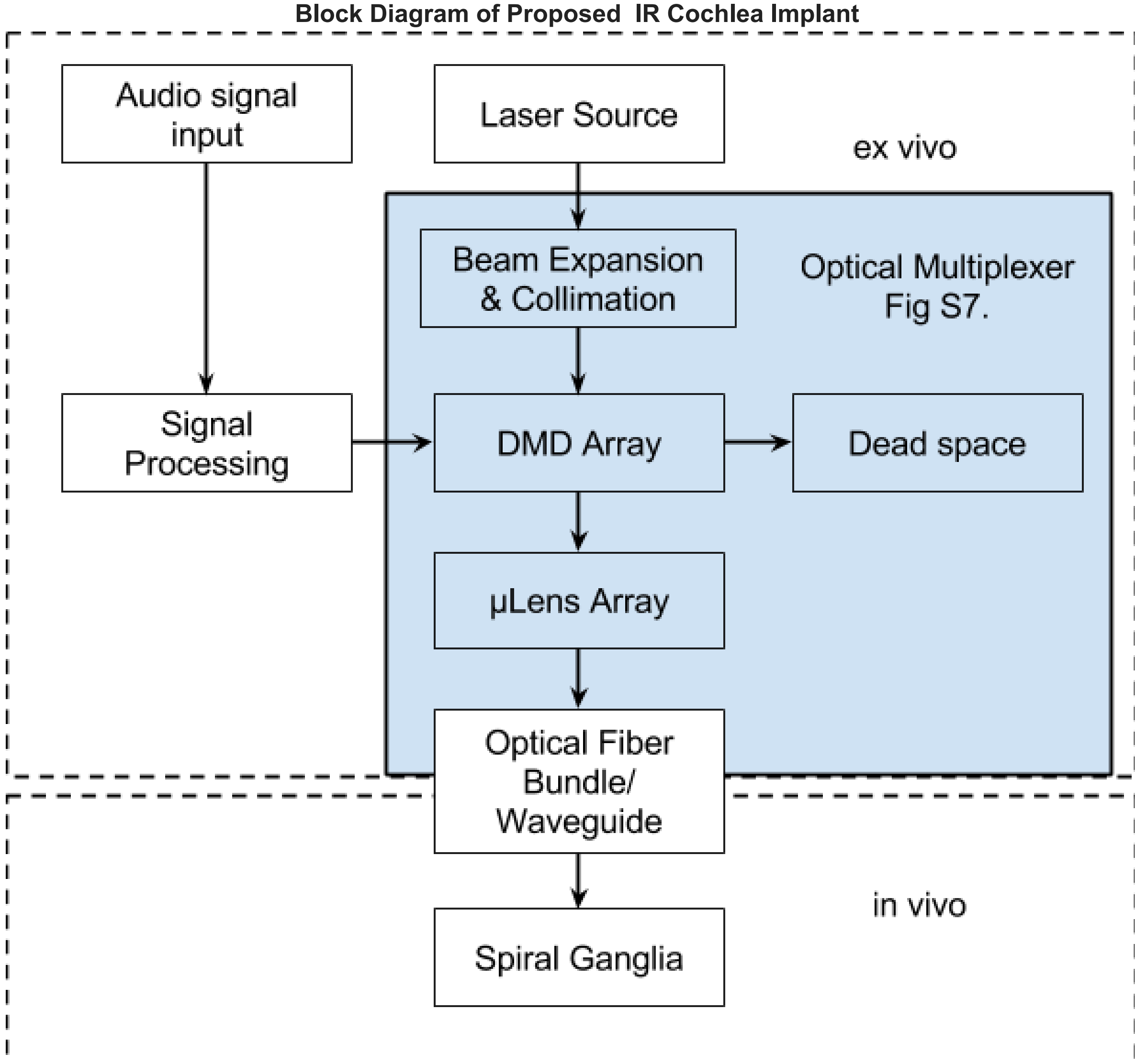

Figure S1

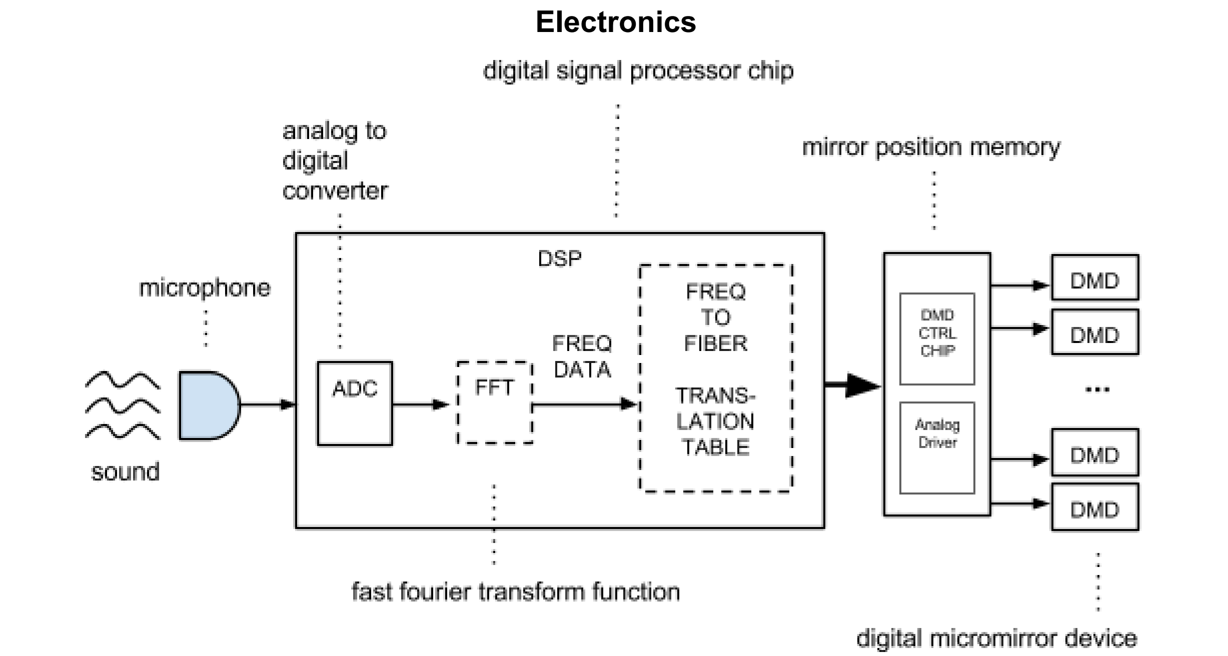

Figure S2

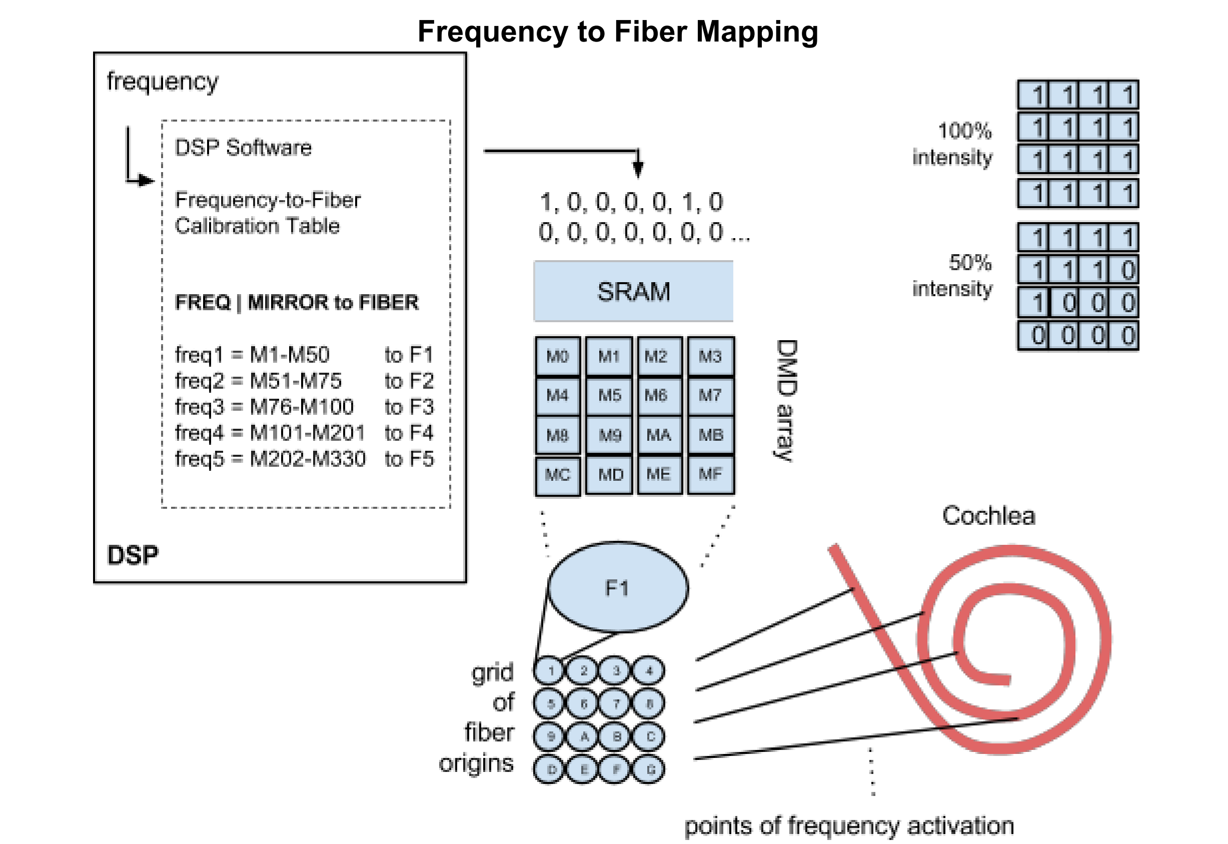

Figure S3

DMD Mirrors Detailed

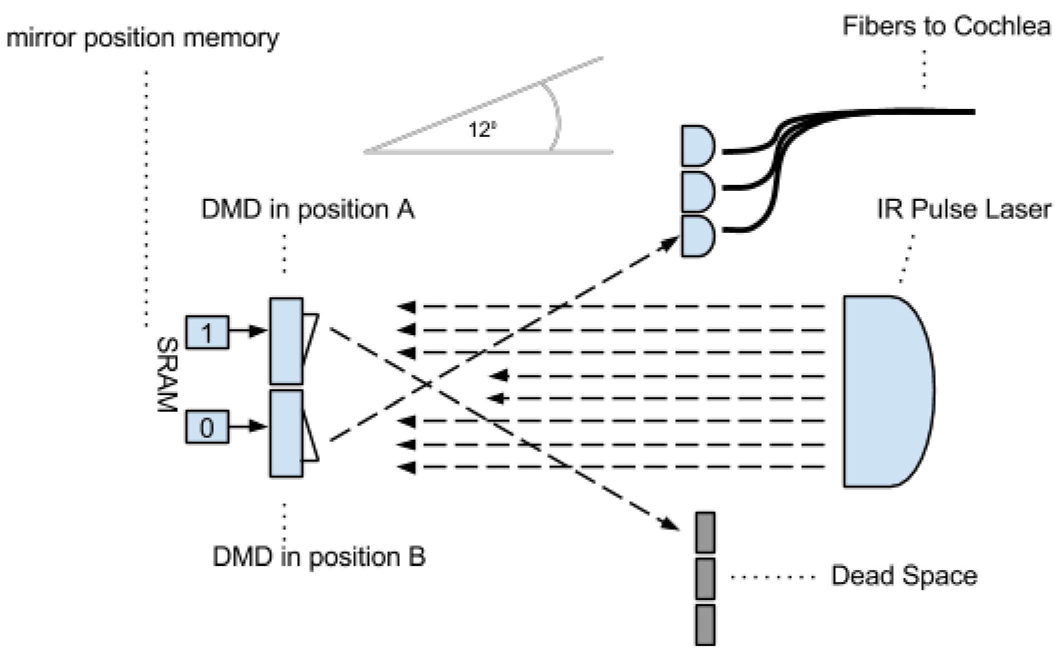

Figure S4

DMD Mirrors Expanded

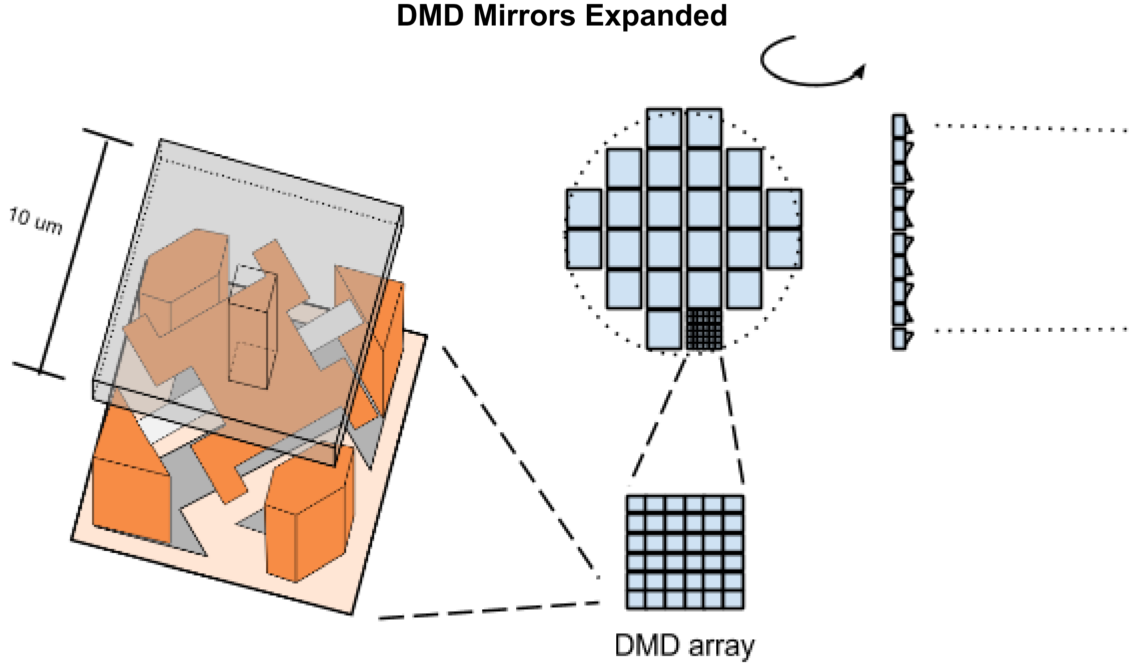

Figure S5

From Data to Light Pattern

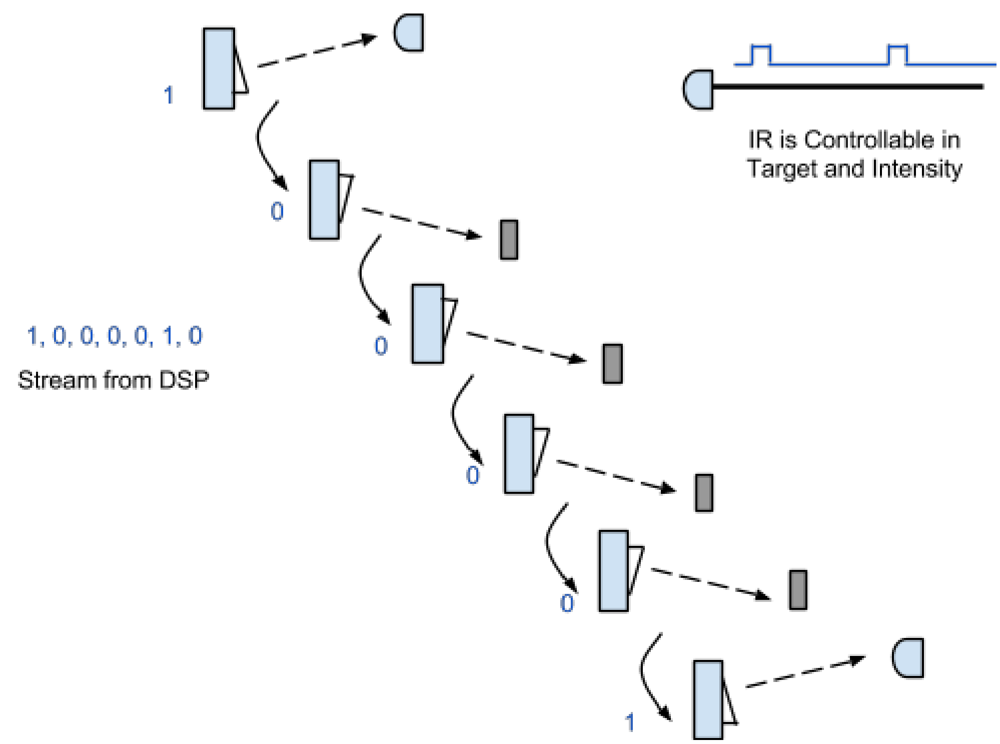

Figure S6

Optical Multiplexer Scale Design

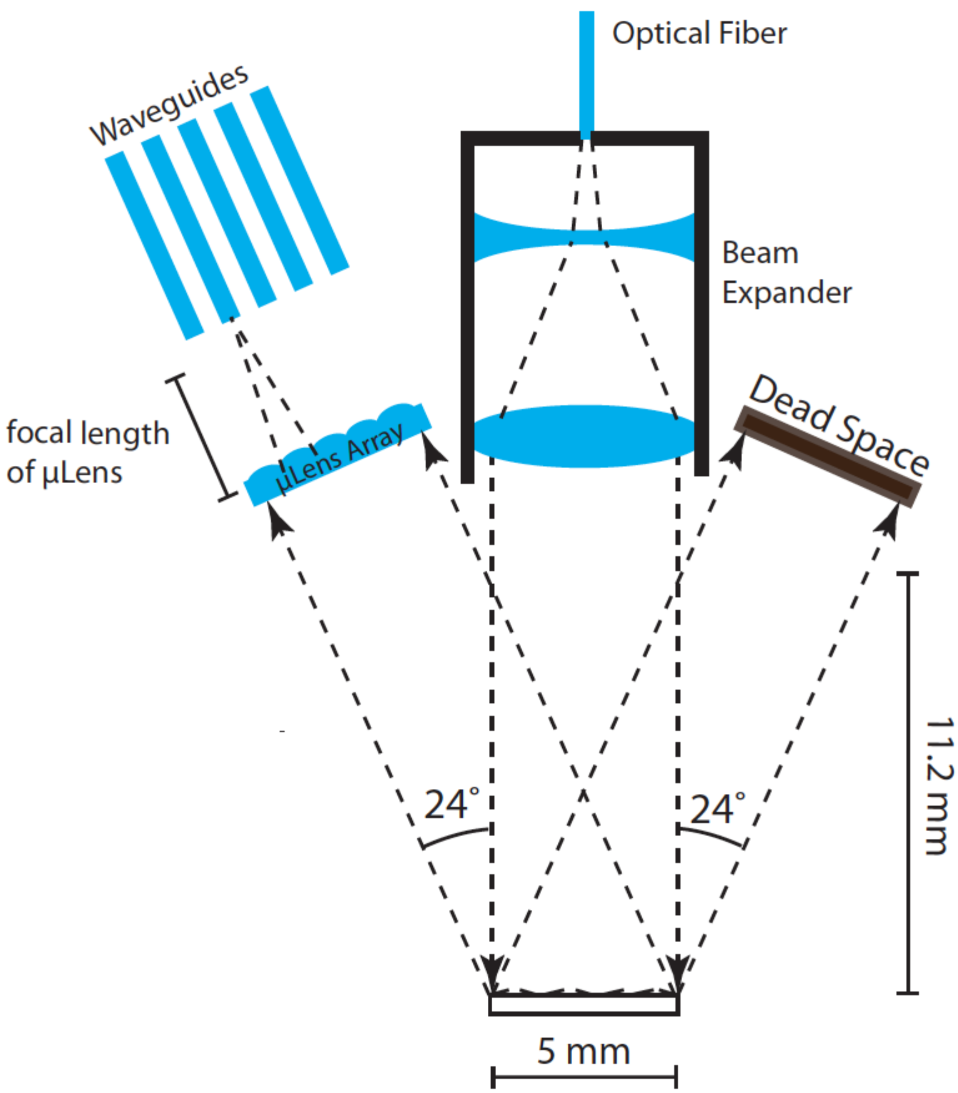

Figure S7

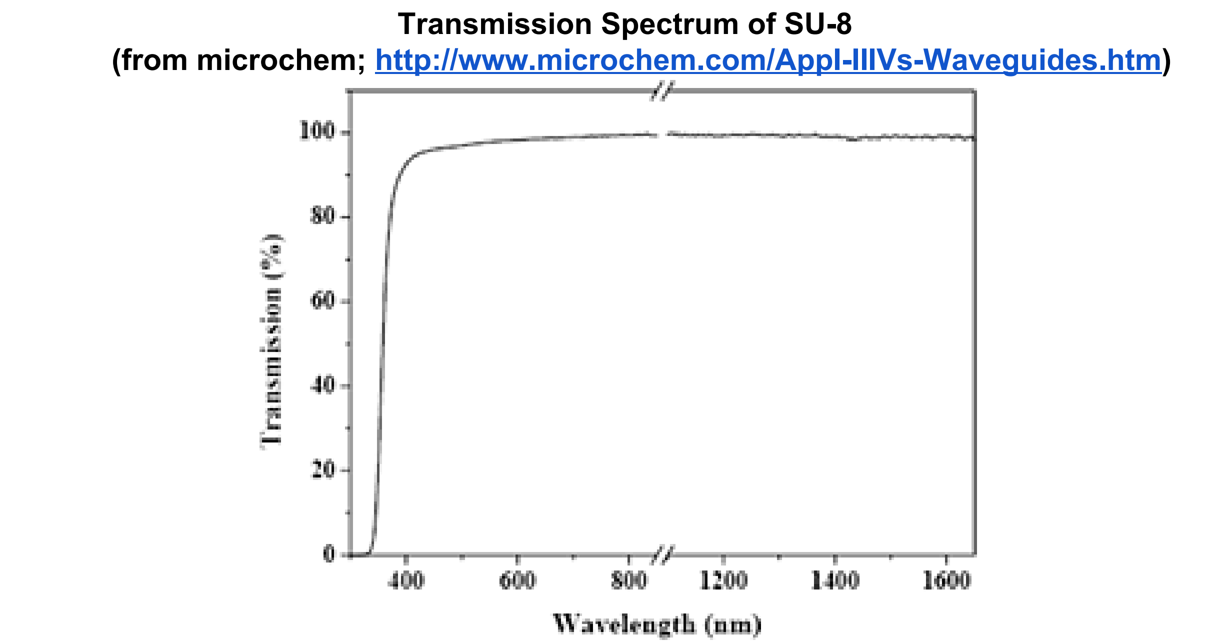

Figure S8

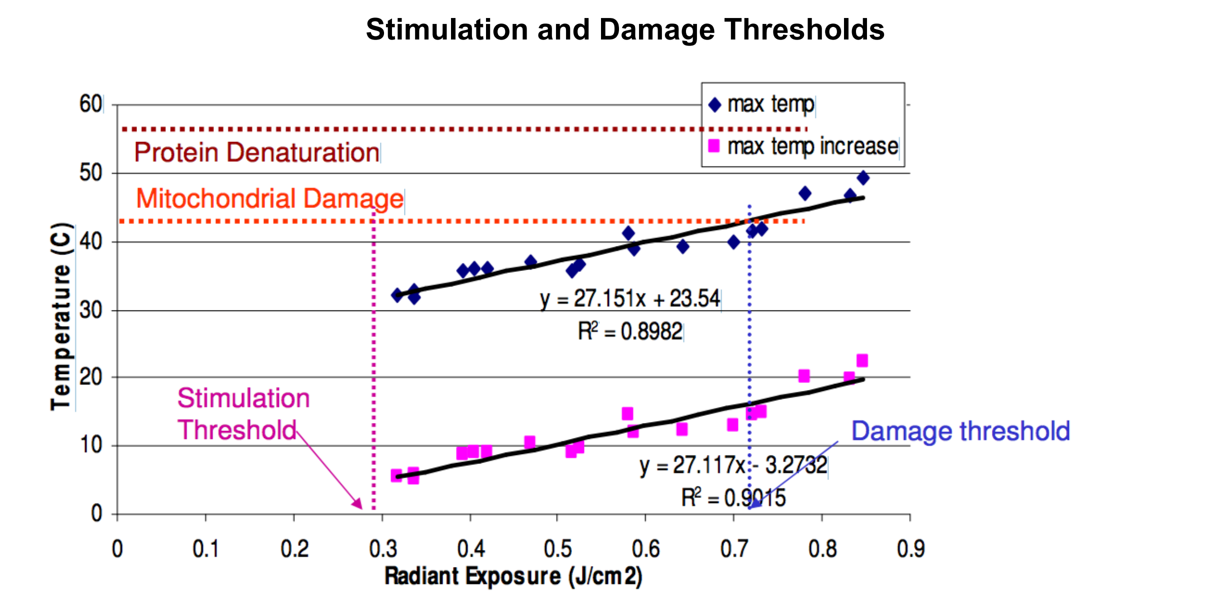

Figure S9