Macrobending losses of glass fibers for optical cochlear stimulation

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Laser stimulation of the cochlea could be a potential alternative to conventional electric cochlear implants in treatment of sensorineural hearing loss. Light delivery by an optical fiber necessitates fiber bending inside the inner ear, which can lead to energy losses. In order to assess those effects, measurements of attenuation and breakage due to fiber bending were performed. Fibers with smaller diameters can withstand smaller bend radii. Energy losses could be reduced with smaller core diameters and higher NA. In conclusion, the results provide a basis for further studies regarding the optical stimulation of the cochlea.

1 Introduction

Direct electrical stimulation of the cochlea is the used method for treatment of sensorineural hearing loss [1]. Cochlear implants (CI) are inserted in the *scala tympani*, one of three helical-shaped ducts of the cochlea, to stimulate the auditory nerve electrically. For a higher spatial resolution and the corresponding improvement in frequency resolution, alternative stimulation methods are investigated. For optical stimulation with laser light it is discussed controversially in literature whether the excitation method is based on direct neuronal stimulation [2,3] or on the optoacoustic effect [4,5,6,7]. In a previous study, we have clearly shown that the basic mechanism is the optoacoustic effect [7]. Nevertheless, the laser-induced intracochlear stimulation requires means of light delivery, such as glass fibers, to stimulate different areas inside the cochlea. These optical fibers are placed inside the helical-shaped cochlea, so that photons can reach the inner ear structures. Here, we investigate the energy losses due to fiber bending and the radius of the fiber breakpoint.

2 Materials and Methods

An optical parametric oscillator (OPO, Ekspla, Vilnius, Lithuania, λ = 420 nm – 2300 nm, τ = 4 ns) with a fiber coupling unit was used to couple the light into an optical fiber (Thorlabs, NJ, USA). To quantify losses of multimode fibers, energies were measured in front of the fiber and at the fiber tip with an energy detector (Gentec-EO, Quebec, Canada). For the measurements, optical fibers with different core diameters (50 µm, 105 µm, 200 µm) and NAs (0.22, 0.39), different laser wavelengths (480 nm, 1070 nm, 1860 nm), bending radii (10 mm – 1.5 mm) as well as bending angles (0°-1080°) were used. The point of fiber breakage was determined experimentally. For different bending radii the bending angle was increased until the fiber broke. A human cochlea was segmented (Slicer) to calculate the relevant bending radii for a simulation of these experiments (Fig. 1). A computer simulation of the fiber bending experiments was performed using CAD software (SolidWorks) in combination with optical design software (ZEMAX, Fig. 2). Based on the dimensions of the segmentation and the bending experiments, the position inside the cochlea of the expected failure of the fiber due to breakage and energy losses was predicted.

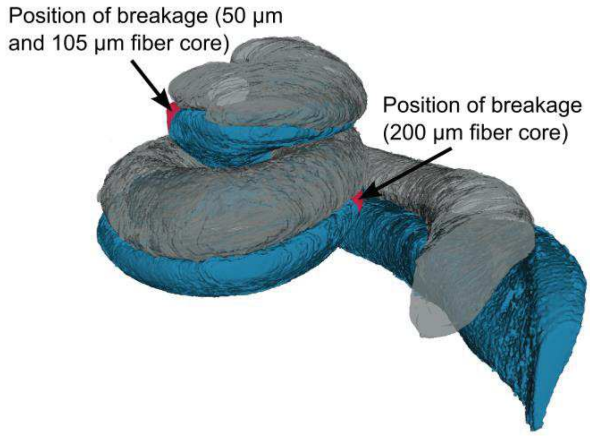

***Fig. 1** Segmented human cochlea (blue: Scala tympani, gray: Scala vestibuli and ductus cochlearis) with the position of the expected failure due to fiber breakage (red).*

3 Results

The energy loss increases with decreasing bending radii (Fig. 3). An optical fiber with a 105 µm core diameter (0.22 NA) breaks at a radius of 1.5 mm and an angle of 720° (Fig. 3a). The dependency of the energy attenuation on the fiber core diameter was analyzed with three different core diameters (50 µm, 105 µm, 200 µm, Fig. 3b).

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Smaller fiber core diameters have lower energy losses than larger cores (Fig. 3b). Three different wavelengths were used to analyze the attenuation dependence on the wavelength. No differences were detected for wavelengths of 480 nm, 1070 nm and 1860 nm each with bending radii of 8 mm and 2 mm (Fig. 3c). The fiber with 0.39 NA shows no losses for bending radii of 10 mm and 3 mm (Fig. 3d). The fiber with 0.22 NA shows high losses for both radii (Fig. 3d). The position of the experimentally determined failure due to the fiber breakage for a 50 µm as well as a 105 µm fiber is at an angle of 540° and for a 200 µm fiber at an angle of 360° (Fig. 1). The simulation showed an attenuation of 0.30 dB/mm for 105 µm fiber core, a length of 27 mm, and NA = 0.22. Experimentally, an attenuation of 0.31 dB/mm was measured for these parameters. The corresponding break radius was 1.5 mm.

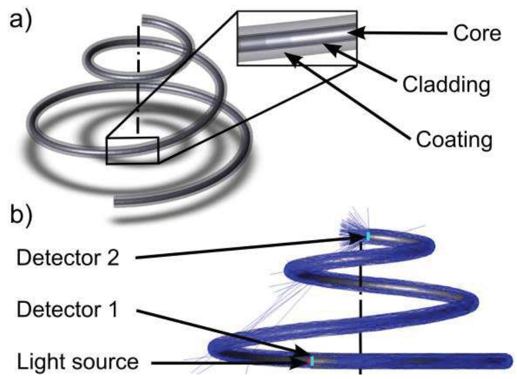

__Fig. 2__ a) CAD construction of a 105 µm core fiber with SolidWorks including core, cladding and coating. The bending radii are determined based on the segmented human cochlea (Fig. 1). b) The multimode fiber imported into the optical design software ZEMAX. The refractive index is assigned to core, cladding and coating and a light source (red) as well as detectors (turquoise) are added to calculate the energy losses due to fiber bending.

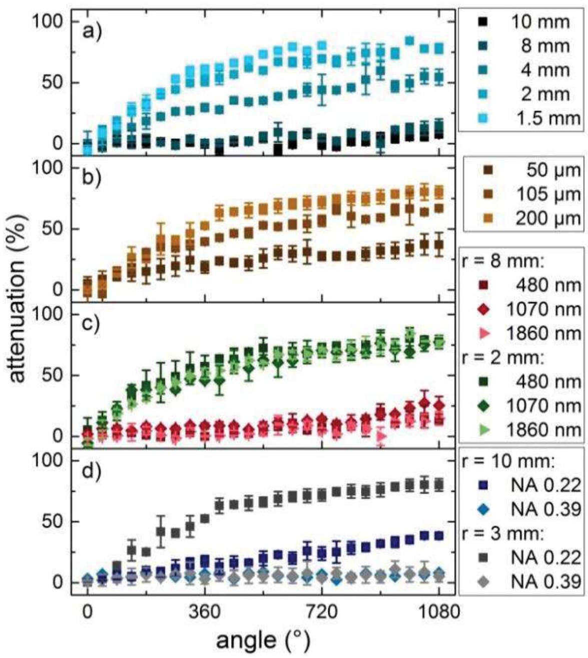

__Fig. 3__ Energy losses due to bending of a) a 105 µm fiber (λ = 1860 nm, 0.22 NA) using defined bending radii, b) different fiber core diameter using a bending radius of 3 mm (λ = 1860 nm, 0.22 NA), c) different wavelengths with bending radii of 8 mm and 2 mm (105 µm fiber core, 0.22 NA), and d) different NA using bending radii of 10 mm and 3 mm (200 µm fiber core, λ = 1860 nm).

4 Discussion

Due to bending, the angle of incidence falls below the critical angle of total internal reflection. The energy loss is caused by coupling light from the core into the cladding and coating. Energy losses could be reduced with smaller core diameters and higher NA. These findings were confirmed by the simulation. Fibers with smaller core diameters can withstand smaller bend radii. None of the used fibers could be inserted into the apex of the cochlea without breaking. The results of the simulation and the experiments match very well. The bending radii of the experiment were not entirely the same as for the simulation. Energy losses of smaller bending radii than 1.5 mm can only be simulated, but not measured, because of the fiber breakage.

In conclusion, the results provide a basis for further studies regarding the optical stimulation of the cochlea.

5 Acknowledgements

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