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GaN-based micro-LED arrays on flexible substrates for optical cochlear implants

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Abstract
Currently available cochlear implants are based on electrical stimulation of the spiral ganglion neurons. Optical stimulation with arrays of micro-sized light-emitting diodes (\(\mu\)LEDs) promises to increase the number of distinguishable frequencies. Here, the development of a flexible GaN-based micro-LED array as an optical cochlear implant is reported for application in a mouse model. The fabrication of 15 \(\mu\)m thin and highly flexible devices is enabled by a laser-based layer transfer process of the GaN-LEDs from sapphire to a polyimide-on-silicon carrier wafer. The fabricated 50 \(\times\) 50 \(\mu\)m\(^2\) LEDs are contacted via conducting paths on both p- and n-sides of the LEDs. Up to three separate channels could be addressed. The probes, composed of a linear array of the said \(\mu\)LEDs bonded to the flexible polyimide substrate, are peeled off the carrier wafer and attached to flexible printed circuit boards. Probes with four \(\mu\)LEDs and a width of 230 \(\mu\)m are successfully implanted in the mouse cochlea both in vitro and in vivo. The LEDs emit 60 \(\mu\)W at 1 mA after peel-off, corresponding to a radiant emittance of 6 mW mm\(^{-2}\).

Keywords: cochlear implant, optogenetics, micro-LED array, flexible substrate

(Some figures may appear in colour only in the online journal)

1. Introduction
1.1. Towards optical cochlear implants

Auditory defects are the most common sensory deficits in humans. Electrical cochlear implants (CIs), which are the most widely spread neuroprostheses, allow profoundly hearing impaired and deaf patients to acoustically communicate with their environment [1]. The 30 000 spiral ganglion neurons (SGNs) are ordered tonotopically, corresponding to certain sound frequencies. CIs employ electrical sound coding and frequency-specific excitation along the tonotopic axis, which allows one to distinguish different frequency channels. Nevertheless, the wide spatial spread of current during electric stimulation limits the frequency resolution of hearing with CIs [2]. Optical stimulation may yield a higher spatial resolution in the excitation of the nerve cells along the tonotopic axis, as light can be focused on the excitation spot. Therefore, sound coding with an optical implant promises higher frequency and intensity resolution and a greater number of independent stimulation channels. This could enable a better speech understanding, particularly in background noise, and better appreciation of music. One possibility is the excitation of the SGNs with infrared light. This method is thought to involve an opothermal effect of the infrared light on the nerve cells [3]. Another optical approach is through optogenetics. Here, the SGNs are made light-sensitive by genetically introducing light-gated ion channels. As an example, channelrhodopsin-2
(ChR2) integrated into the cell membrane makes neurons sensitive in the blue spectral range with maximum sensitivity between 460 and 480 nm [4]. Specific advantages compared with infrared excitation are lower power consumption by a factor of 7–70 [5] and, therefore, a reduction in potential tissue damage, both of which critical for potential future use in clinical restoration of hearing.

Transgenic mouse models are suitable for studying the feasibility of optogenetic CIs. This work aims at the first demonstration of probes with integrated micro-sized light-emitting diodes (µLEDs), which are implantable in the mouse cochlea. The scala tympani of the snail-like shaped mouse cochlea is about 6 mm long. It is about 300–400 µm wide at its base near the round window, which is about 250–300 µm in diameter [6]. The probe shall be inserted into the scala tympani through the round window. Thus, the diameter of the probes should not exceed 300 µm to enable insertion.

1.2. State of the art in GaN-based LEDs

(AlGaN)N has proven to be the material of choice for high-brightness LEDs in the ultraviolet, blue and green spectral range [7]. Standard LED chips do have a dice-like geometry with an edge length and substrate thickness of 200 µm or more. Those devices are too large for implantation in the mouse cochlea. Wafer bonding and subsequent laser lift-off (LLO), which was applied in this work, is frequently used for removing the sapphire substrate [8, 9]. In high-brightness LEDs, the main advantage of the sapphire removal is the possibility of roughening the outcoupling semiconductor surface of the LED, which maximizes the light extraction efficiency.

Flexible arrays of µLEDs were first published for the (AlGaN)P material system, which covers the orange and red spectral range [11]. Recently, there have been reports on GaN-based LEDs on flexible substrates comprising a chip-level transfer of GaN-LED onto a flexible polymer substrate [12].

1.3. Probe design

The probe design in figure 1 is based on the wafer level transfer of GaN-µLEDs from the growth substrate sapphire to a flexible substrate wafer. This heterointegration provides a powerful tool for miniaturizing LED probes.

The design entirely relies on techniques that are well established in high-brightness LED production, especially wafer bonding and LLO. Au–In metal interdiffusion wafer bonding is carried out to join the two wafers. Subsequently, the sapphire wafer is removed using LLO. Finally, the probes are peeled off the carrier wafer after the completion of the wafer process.

We are aiming at multi-channel addressing of the µLEDs. This is achieved by employing separate levels for n- and p-contacting of the LEDs. Thus, we can achieve matrix-addressable one-dimensional LED arrays. In principle, \( N_n \times N_p \) LEDs can be addressed using multiplexing of different LED blocks, where \( N_n \) and \( N_p \) are the number of contact lines on the n- and p-sides, respectively. The current probes do exhibit two conducting paths on both p- and n-sides.

The highly integrated fabrication process avoids handling of individual LED chips. Thus, the present scheme can be easily scaled up in terms of numbers of LEDs per probe and allow the fabrication of a large number of probes in parallel.
Figure 2. Laser-assisted 3L process for structuring the GaN epitaxial layer into electrically and mechanically separated LEDs using mesa (L1), trench (L2) and burst (L3) laser fabrication techniques.

2. Probe fabrication

In the following, the implementation of the probe design is described in detail. LED structures were grown on 2 inch c-plane sapphire by metalorganic vapour phase epitaxy (MOVPE). The 5 µm-thick LED structure comprises a three GaInN quantum well (QW) active region on top of a Si-doped n-GaN layer, followed by an AlGaN : Mg electron blocking layer and a Mg-doped p-GaN layer. The wavelength can be adjusted between 400 and 465 nm by varying the indium content of the QW. Ni/Au-based p-GaN metallization was conducted via electron beam evaporation and was subsequently annealed to form a low-resistive ohmic contact.

2.1. Laser-assisted LED processing: 3L-process

A novel scheme for the processing of GaN-LEDs on sapphire using three laser fabrication steps (3L) was implemented. Laser fabrication gives the opportunity for rapid prototyping of any kind of LED structures and has recently been demonstrated for thin-film solar cells [13]. Two laser work stations with complementing capabilities were employed to process the LEDs. The first tool is an excimer laser work station. It is equipped with a KrF laser emitting 20 ns pulses at a wavelength of 248 nm. The laser beam is homogenized to obtain a flat top beam of 2 × 2 mm². The second work station is a picosecond laser micromaching tool comprising a frequency tripled Nd : YVO₄ laser emitting 10 ps pulses at a wavelength of 355 nm.

In the 3L process, LEDs were defined on sapphire. The electrical definition of the p–n junction area was carried out with the excimer laser work station in the laser mesa process L1. In the laser trench L2, the laser micromaching tool was used to fabricate 10 µm-wide trenches around each LED. L1 and L2 processes are described in detail in [14]. The laser burst L3 was employed to remove large areas of surplus GaN between the individual LEDs employing a projection mask. The GaN–sapphire interface was backside illuminated through the sapphire substrate and the GaN layer was removed with a single laser pulse. The process scheme of the 3L process is depicted in figure 2.

A silicon nitride layer was deposited to passivate the mesa and trench sidewalls. Subsequently, a titanium-based diffusion barrier and the indium bond metal were evaporated onto the gold-based p-GaN metal contact. The GaN-on-sapphire LED wafer before bonding is shown in figure 3(a).

2.2. Layer transfer process: wafer bonding, underfill and LLO

The GaN-on-sapphire LEDs fabricated in the 3L laser process were transferred to a corresponding PI(polyimide)-on-silicon substrate in a three-step process comprising wafer bonding, underfilling and LLO.

The PI-on-silicon wafer was fabricated on 4 inch wafer level comprising a 5 µm-thick PI layer, which was spin-coated using U-Varnish-S from UBE. 300 nm Pt is used as metal path. The metal lines were covered with a second polyimide layer, which was structured using dry etching. Finally, the bond pads of the µLEDs were thickened using Au electroplating (figure 3(b)).

Aligned wafer bonding was carried out using commercial wafer bonding equipment. The minimum bond temperature was found to be 140 °C, which is below the melting point of indium. Thus, spreading of the liquid indium phase can be avoided.

After wafer bonding, the gap between the LED wafer and the carrier wafer was filled with an epoxy (Epo-tek 301-2FL from Epoxy Technology). In a subsequent process step, the low-viscosity and solvent-free underfill material was thermally cured.

After curing the underfill, the sapphire wafer was removed using a LLO process. A two-step LLO was employed for lifting the sapphire wafer. In the first LLO step, all GaN-LEDs were illuminated through the polished sapphire wafer with a single 20 ns laser pulse at a fluence of 0.80 J cm⁻² with appropriate projection masks. The underfill mechanically stabilized the
Figure 4. 380 µm-wide probe comprising 3 × 5 LEDs with 150 × 150 µm² emission area each.

Figure 5. (a) Side view of the peeled-off probe. (b) Bending test of the delaminated probe.

GaN-LEDs during laser illumination. After delamination of all GaN-LEDs, the entire wafer was illuminated with single pulses at a fluence of 0.25 J cm⁻². This process delaminated the underfill material from sapphire. Afterwards, the sapphire wafer could be removed. Figure 3(c) shows a GaN-µLED after the transfer to the PI-on-silicon substrate wafer. The n-conducting metal paths and an insulating polymer top layer were added after sapphire removal to complete the wafer process.

2.3. Probe characterization

Probes were fabricated comprising 15 LEDs with 150 × 150 µm² emission area, which could be addressed in three separate channels. The width of the probe was 380 µm (figure 4).

The probes were cut out using laser micromachining and subsequently peeled off from the silicon carrier wafer. Bending tests were performed with a peeled-off array furled around a wire with a diameter of 1 mm (figure 5). The radius of curvature was chosen to equal that of the mouse cochlea. No cracks were observed after the bending test.

Probes were attached to a flexible printed circuit board (PCB) to connect the delicate probe with an electrical connector and to enable handling of the probe during in vivo animal experiments. At the end of the flexible PCB, a standard connector system was used. We mounted a linear probe with a width of 230 µm and four functional 50 × 50 µm² LEDs in one channel and measured the electro-optical characteristics (figure 6). The output power of the four LEDs was 60 µW at a wavelength of 405 nm and a current of 1 mA at a voltage of 3.5 V. The reverse bias leakage at −5 V was 1 µA. The optical power density was 6 mW cm⁻², which is above the published typical excitation density of 4 mW cm⁻² for cochlear optogenetics.

Figure 6. Implantation-ready device mounted in a flexible PCB including electro-optical characteristics, i.e. current–voltage and power–current curves.

The bendability of the device was tested using a quasi two-dimensional (2D) model of the rat cochlea (figure 7), which imitates the spiral-shaped canal of the scala tympani. The device with three functional LEDs in the middle of the probe was inserted through the first opening of the model, corresponding to the round window of the cochlea. Afterwards, the device was bent backwards and put through the second opening, not affecting the functionality of the LEDs. The probe, which follows the first half-turn of the model, showed a bending radius of 900 µm. Thus, the probe flexibility was compatible with the bending radius in the rat cochlea. Electrical failures due to bending were not observed. Nevertheless, sharp kinks are to be avoided, as they cause breaking of the metal conducting paths.

3. Application

Implantation tests with functional probes were carried out in the explanted mouse cochlea of a 5-week-old animal. Visualization was facilitated by immersing the cochlea in 2,2'-thiodiethanol as an index-matching fluid. The µLED-array on
the flexible substrate was encapsulated with silicone. Thus, the device thickness was increased to 170 $\mu$m. The width of the probe was 230 $\mu$m. The first two functional LEDs could be inserted through the round window. This corresponds to an implantation length of about 800 $\mu$m (figure 8(a)). The non-functional probe could finally be inserted following the scala tympani for three quarters of a full round trip without damaging the cochlea walls, corresponding to an implantation length of about 3 mm (figure 8(b)). Thus, the feasibility of insertion regarding the flexibility of the device could be clearly shown.

4. Summary and outlook

In this work, flexible $\mu$LED probes were realized for the potential use as CIs in a mouse model. The flexible linear LED arrays were fabricated combining two wafer level processes and the layer transfer of the LEDs from the sapphire growth substrate to a carrier wafer via metal wafer bonding and LLO. We fabricated devices with three individually addressable channels on wafer level. Implantable 230 $\mu$m-wide probes with four 50 $\times$ 50 $\mu$m$^2$ LEDs emitting 60 $\mu$W at 1 mA were attached to flexible PCBs. The fabricated probes were well suited for acute implantations regarding flexibility and handling.

Future work will address the increase in individual addressable LEDs by further miniaturizing the p- and n-side conducting paths. Testing the long-term biocompatibility of the devices will also be a challenging aspect of future progress.

For clinical application, the power efficiency and overall power budget should be comparable to state-of-the-art electrical CIs. The efficiency of the $\mu$LED probes can be enhanced by patterning the n-GaN outcoupling surface to increase the light extraction efficiency. Collecting and directing the light output of the $\mu$LEDs towards the SGNs will be an effective means to make more efficient use of the light generated, and thus to increase the overall power efficiency. This can be achieved by, e.g., narrowing the Lambertian emission profile of the LEDs using micro-optical lenses, diffractive optics or resonant-cavity LEDs.

The combination of light-based methods and electrophysiology provides the possibility to separate excitation and detection mechanisms. Nevertheless, the role of electrical artefacts due to capacitive coupling of the probes needs to be taken into account [15].

Future challenges for clinical translation may include studies of biosafety for optical stimulation as well as optimization of channelrhodopsins and virus-mediated gene transfer for the efficient and safe long-term expression of channelrhodopsin DNA in auditory neurons [5].

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