A novel active MR-Probe using a miniaturized optical link
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ABSTRACT <200WORDS

Applying an active intravascular MR-catheter device that allows signal transmission from the catheter tip requires special means to avoid RF-induced heating. This paper presents a novel, all-optical active MR-probe to use with real-time MRI in minimally invasive interventions for catheter guidance and intravascular imaging. An optical link transmits the received MR-signals from the catheter tip to the MR-receiver with inherently RF-safe optical fibers. Furthermore, power is supplied optically to the transmitter as well. The complete integration into a small tube of 6 F size was realized using chip components for the optical modulator and a miniaturized optical bench fabricated from silicon substrates with 3D self-aligning structures for fiber integration.

In MRI phantom measurements projection based tip tracking and high resolution imaging were successfully performed with the optical link. Images were obtained in a homogeneous phantom liquid and first pictures were acquired from inside a kiwi that demonstrates the potential of the MR-safe optical link. It is demonstrated that the novel optical link exhibits an SNR comparable to a direct electrical link.

Keywords: TBD.
INTRODUCTION

Currently, most catheter interventions are performed under X-ray fluoroscopy which causes ionizing radiation that is harmful for the patient and the staff. MRI is a promising alternative imaging modality that avoids ionizing radiation and provides a superior soft tissue contrast compared to other imaging techniques. Standard catheters with their extended electrical conducting materials, however, can cause resonance effects that may lead to serious tissue heating which has so far prevented the clinical use with MRI. Purely polymer-based catheters on the other hand are not visible in MRI images. Therefore a tracking mechanism is needed to localize the catheter during the intervention. To localize the position of a catheter inside an MRI environment there are passive and active tracking methods. Passive tracking methods rely on signal amplification or cancellation of catheter regions in images, whereas active tracking methods collect the MR-signals using a resonant circuit at the catheter tip and transmit the received MR-signals to the scanner. This method allows an active guidance of the catheter with real-time tracking during the intervention and was first presented in [1]. Furthermore, the signals can be used to obtain high resolution and high signal-to-noise-ratio (SNR) images of the surrounding tissue near the micro receiver coils in relatively short time [2]. The feasibility of intravascular imaging using a micro catheter coil was also shown in [3]-[8] by using solenoid, opposed solenoid or loop coils. However, extended electrical wires for signal transmission during a catheter intervention inside an MRI could lead to serious tissue damage [9]-[11]. The maximum temperature rise allowed by a catheter inside a patient is 4°C [12].
In [14] and [15] a concept of a RF-safe signal transmission technique for catheter applications has been presented. There the RF-safety is achieved by segmenting the electrical line by miniature resonant transformers. It is the most relevant directly analogous method to the optical link, since it also provides general purpose MR-signal transmission.

A previous work [13] examined and demonstrated the feasibility of a novel miniaturized optical link that can be integrated into a catheter tip. Any extended electrical wires are avoided and the system is therefore inherently RF-safe. The system was examined in MR and transmission experiments. Projection-based tip tracking was achieved, and high-resolution images were obtained in a phantom. The basic principle was shown, but the obtained images suffered from a 20 dB lower SNR compared to a direct electrical transmission.

The purpose of this paper was a significant improvement of the SNR and a further miniaturization of the optical catheter link compared to the former work [13]. With the new system projection based tip tracking was performed and high-resolution and high-SNR micro images were obtained in a phantom.

**METHODS**

*Concept of the MR-probe*

The presented system uses an optical signal transmission to overcome the RF-safety issue for active interventional catheter devices. Signals received at the catheter tip are converted into an optical signal and are then transmitted to the MR-receiver. The tip set-up is powered optically to provide the active system with energy. Figure 1 shows the concept of the MR-probe. The system uses two optical fibers. The first
fiber is connected to the optical modulator that transmits the signal to an optical receiver. The optical receiver converts the optical back into an electrical signal and transmits it to the MRI scanner. The second fiber is used for the optical power supply that is needed to supply the MR-probe.

The electrical circuit of the MR-probe at the catheter tip as shown in Fig. 2 can be separated into three parts: the resonant receiver circuit, the optical modulator and the optical power supply. MR-signals out of the vicinity of the MR-probe are received by the resonant receiver circuit (L, C₁). The resonant circuit is passively decoupled by a crossed pair of diodes (D₁/₂). The received signal is fed directly into the transistor gate (T) of the optical modulator that modulates the current through the laser diode (LD). The transistor bias point is set with the resistor (Rₛ). Furthermore, the circuit contains an optical power supply via a second fiber. The power supply uses three photovoltaic elements (PPC) with three elements connected in series, that provide the optical modulator with the appropriate voltage current characteristic.

**Design for optimal SNR**

The signal transmission of MR-signals via an optical link requires a good SNR to achieve high quality images of the vicinity near the MR-probe. Therefore the transmission system is examined to determine the effects that limit the SNR.

Figure 3 depicts the electrical equivalent circuit for RF-signals. The resonant receiver circuit amplifies the induced voltage $U_{ind}$ from the coils by the Q-factor of the resonant circuit. This signal $U_{GS}$ is fed to the gate of the transistor. The input voltage $U_{GS}$ multiplied by the transconductance $g_m$ is current $i_D$ which modulates the laser. $R_{laser}$ in the equivalent circuit is the differential resistance of the laser diode. The modulation depth $m$ of the optical signal is calculated from the modulated laser current $i_D$, the threshold current $i_{th}$ and the bias current $i_0$ [16]:
To estimate the maximum SNR of an optical link the modulation depth \( m \), the relative intensity noise \( RIN \) of the laser and the receiver bandwidth \( \Delta f \) is used [17], [18]:

\[
SNR_{\text{max}} = \frac{m^2}{RIN \cdot \Delta f} = \frac{(g_m \cdot Q \cdot U_{\text{ind}})^2}{(I_0 - I_{th})^2 \cdot RIN \cdot \Delta f}
\]

Eq. (2)

The signal and receiver bandwidth depends on the applied imaging protocol of the MRI. The \( RIN \) and the threshold current \( I_{th} \) of the laser are fixed values for the VCSEL used in this work. Furthermore, a minimum level for the bias current \( I_0 \) is needed to operate the laser in the linear range. The parameters to maximize the SNR are therefore a high \( Q \)-factor of the receiver circuit and a high transconductance \( g_m \) of the transistor.

**MR-receiver coils**

The MR-receiver coils used for the presented investigations are dedicated micro Helmholtz-coils developed for an optimal catheter integration. The coils are fabricated on a flexible polyimide foil and wrapped around 5 F catheter tips as shown in Fig. 4. The coil tracks are electroplated using an UV-LIGA process that allows thick copper layers, here 50 \( \mu \)m thick tracks for the coil windings. The electroplating increases the \( Q \)-factor of the coils [19] that is essential for a high SNR as given by Eq. (2). Furthermore, the coils are optimized for imaging and tracking applications for small 5-F catheter.

**Optical modulator and optical power supply**

The components for the optical modulator have to be chosen carefully to meet the requirements of the system performance of a low energy consumption and a high
SNR as shown in Eq. (2). Additionally the components have to be small enough to fit into the catheter tube of a 5 F catheter.

Concerning the power requirements of the optical modulator, it is important to choose a laser diode with a low threshold current $I_{th}$ to keep the circuit current $I_0$ low and therefore minimize the required power of the system. As the optical source for the optical modulator a small laser diode die (VCSEL ULM850-10-TT-N0101U, Ulm-Photonics, 250 x 250 x 150 µm³) is used. It has a low threshold current of 0.55 mA and the die is small enough to be integrated into the catheter tube. The optical signal is coupled into an optical fiber (GIF 625, 125 µm, Thorlabs, USA NJ). In this work two different transistors are examined and compared for integration into the optical modulator. The JFET (2N4393, Central Semiconductor USA, 533 x 457 µm²) had already been used previously [13]. This transistor has a gate-source capacitance $C_{GS}$ of about 20 pF and a transconductance $g_m$ of 10 mA/V. The source resistor $R_S$ is 900 Ω. The second transistor is a pHEMT (TGF2022-06, 566x530x100 µm³, TriQuint Semi-conductor, USA) with a low gate-source capacitance $C_{GS}$ of 1.5 pF and a transconductance $g_m$ of 225 mA/V. The optimized source resistance $R_S$ is 700 Ω. The modulator circuits with these devices require a bias current $I_0$ of 1 mA and a voltage of about 3 V that has to be supplied by the optical power supply.

A GaAs photo voltaic power converter PPC (PPC3FM, 500x500x150 µm³, JDSU, USA CA) is used to supply the electrical power. Its light sensitive area is segmented into three diodes connected in series. The PPC generates an open circuit voltage of 3.3 V as required by the optical modulator. The necessary current of 1 mA is generated by an optical power of 13 mW at a wavelength of 850 nm. This optical power is supplied by an external laser (ADL-85501TL, Thorlabs, USA NJ) through an optical fiber (GIF 625, 125 µm, Thorlabs, USA NJ). To achieve a maximum efficiency
it is essential to illuminate each segment of the photovoltaic power supply equally. Any misalignment will cause a drop in efficiency. Therefore a dedicated reflector is developed that deflects the light from the fiber to the light sensitive area of the power supply. Figure 5 shows a schematic of a self-alignment structure on a micro optical bench for the optical reflector system. The fiber is aligned in a V-groove which together with the reflector is fabricated within a silicon substrate using an anisotropic KOH wet chemical etch. To ease the alignment of the reflector to the optical power converter, a self-alignment structure is incorporated into the base substrate, into which the reflector can be easily placed. The substrate provides the conductive leads to connect the electronic and 3D structures in silicon to mechanically support the alignment of the discrete components. The structures are etched with anisotropic deep reactive ion (DRIE) and KOH silicon etching processes.

In Fig. 6.a) the flexible micro coil with the capacitors of the resonant circuit is depicted. The setup of the laser diode and the optical modulator on the micro optical bench is shown in Fig. 6.b). Figure 6.c) and 6.d) show the optical power supply with the self-alignment structures and the reflector. The system is set up on a PCB to do the alignment and electrical testing and characterization. After the system is tested, it is integrated into a catheter tube (Fig. 7). To protect the mechanically sensitive system an additional tube is pushed across. The catheter is then connected by optical fiber plugs.

**Optical receiver**

To convert the optical catheter signals back into an electrical signal an optical receiver is used (HFBR-2416, Agilent Technologies). The receiver uses a pin-photodiode with a low noise transimpedance amplifier. The output impedance of 30 Ω
of the receiver is transformed to the 50 Ω input of the MR-preamplifier by an inductive and capacitive matching network.

**Electrical Characterization**

The resonance of the receiver circuit was measured using weak inductive coupling of the micro coil to a pick-up coil connected to a network analyzer. The Q-factor was determined from the bandwidth B and the resonance frequency from the resonance chart.

To characterize the gain and the dynamic range the optical link was examined with a spectrum analyzer. The micro coil was coupled to a transmit coil connected to the RF-output of the analyzer while the analyzer sends a signal at the Larmor frequency of 63.87 MHz. The optical receiver transmits the received signal back to the analyzer. To measure the noise level the RF-output of the spectrum analyzer was turned off, so that the noise level could be measured without the signal.

**MRI Measurements**

MR imaging and tracking experiments were performed with a phantom inside an MRI scanner (Achieva, Philips Medical Systems, Best, Netherlands). The catheter tip with the MR-probe is immersed into a bowl filled with standard phantom liquid (aqueous solution of 0.08 % CuSO₄ and 0.2 % NaCl). Additionally micro images are taken from within a kiwi to show the imaging capability of the micro optical link.

Tracking experiments were performed using fast projection tip tracking, where the volume of interest is excited. Then the frequency encoded projections were obtained for the x, y, and z-direction on a dedicated catheter channel. From the signal peaks the special position of the tip was calculated automatically by the MR-system [20].
This position is then marked in the overview images acquired by an additional surface C2 coil.

First the MR micro imaging experiments are performed with the micro coils inside the phantom liquid using a direct electrical connection with a capacitive matching network. The coils are matched to the $50 \, \Omega$ electrical transmission line, and a crossed pair of diodes is applied for passive decoupling. This measurement is used as benchmark of the optical link, since the feasibility of directly transmitted MR-signals from a catheter coil was presented in [3]-[8]. This measurement is then repeated with the MR-catheter using the optical link. The catheter system with the optical link and the direct electrical transmission is performed with the same MRI 3D-FFE sequence:

- TR = 16 ms
- TE = 7,6 ms
- Flip angle=15°
- FOV = 64 mm2
- Signal bandwidth = 95,7 Hz/px * 128 px = 12,25 kHz
- Resolution = 0,5 mm
- Acquisition time = 36,8 s (5 slices)
- Slice thickness = 2 mm

The signal to noise ratio SNR is then calculated in the micro imaging experiments for the maximum signal intensity $S$ in the coil center and the mean noise signal $\langle N \rangle$ of the zero signal region in the corners of the image [21]:
\[ SNR = \sqrt{\frac{\pi}{2}} \cdot \frac{S}{\langle N \rangle} \]  

Eq. (3)

**RESULTS**

**Electrical characterization**

The fabricated micro coils show a Q-factor of about 30 in a parallel resonant circuit. When the coils are matched to a 50 \( \Omega \) transmission line the Q-factor decreases to 15 due to the matching network and the 50 \( \Omega \) load. The Q-factor reduces to 11.3 when the parallel resonant circuit is connected to the optical modulator with the JFET due to the high load of the 20 pF gate-source capacitance of the transistor. When the receiver circuit is connected to the optical modulator in this version with the pHEMT the Q-factor only decreases to 28. This high Q-factor is achieved due to the small load caused by the pHEMT with its low gate source capacitance of 1.5 pF.

Figure 8 shows the output power of the optical receiver over the induced voltage into the receiver coil of the MR-probes inside the catheter, measured with help of a spectrum analyzer. The pHEMT system shows a 21 dB higher gain compared to the JFET system. This is caused by the higher Q-factor and the higher transconductance as discussed earlier. Both systems show a dynamic range of 67 dB above the noise level to the 1 dB compression point where the nonlinear region begins. The noise level is –86 dB(1mW) at a receiver bandwidth of 12 kHz according to the MR-imaging experiments, which is approximately the theoretical value resulting from the relative intensity noise of the laser (–89 dB (1mW) for a bandwidth of 12 kHz). The additional noise in the measurement will originate in the optical receiver.
MRI Measurements

Figure 9.a) shows the 3D-tracking projection peaks received on the catheter channel with the optical link from the catheter tip. Figures 9.b-d) depict different slices obtained with the surface C2-coil. Furthermore, the images show the position of the MR-probe and the imaging slices containing the MR-probe. The spatial position is calculated automatically from the projection peaks in the frequency domain by the MR-system. Here fast projection-based tip tracking was performed successfully during real-time MR-imaging.

The result of the MR-imaging experiment is shown in Fig. 10.a), where the MR-probe is immersed into the phantom liquid. This image depicts the field distribution of the micro MR-coil. In this measurement the optical link with the JFET-system has a SNR loss of -17 dB compared to a direct electrical transmission line. The novel optical link with the pHEMT shows a SNR gain of +8 dB compared to the direct electrical method. This SNR gain of +25 dB compared to the JFET system is caused by the higher Q-factor and the higher transconductance of the pHEMT-system, as already demonstrated in the electrical characterization. Figure 10.b) depicts the MR-micro image obtained from inside a kiwi to show the imaging capabilities of this first fully optical high SNR MR-probe prototype with the pHEMT-system. This image reveals some kiwi structures besides the coil structure.

DISCUSSION

The experiments and measurements demonstrate a significant improvement of this fully optical system as compared to earlier work [13]. The increased signal amplification at the catheter tip results in a much improved SNR. This SNR improvement of 25 dB is verified by experiments in the MRI images for the pHEMT
against the formerly used JFET. The fact that the SNR of the pHEMT-version is larger than that of a direct electrical link may be explained by a flip angle amplification due to an insufficient decoupling by the crossed diodes in conjunction with the small receiver coils. It can, however, be concluded that the novel system exhibits an overall performance in imaging quality equal to a direct electrical link.

A further miniaturization and a first catheter integration was also achieved. The fragile MR-probe still has to be protected by an additional rigid tube that slightly increases the overall diameter of the catheter at the tip. This has to be eliminated in future by further miniaturization of the probe.

Previously, a transformer cable has been described for RF-safe MR-signal transmission in catheters [13]. The transformers limit currents associated with RF-heating, but introduce an SNR loss of about 10dB for devices used in 1.5T systems. In comparison, the optical transmission system provides a higher SNR and has the advantage to be inherently safe, because elongated conductors are avoided completely. However, advantages of the transformer concept are that it can be applied by simple replacement of the standard cable of an active device and that the transformer cable has a very low profile and does not limit the mechanical flexibility of devices. These considerations suggest to use the transformer concept for RF-safe active tracking due to its comparably low effort and to exploit the SNR advantage of the optical concept primarily in intravascular imaging applications.
CONCLUSION

This paper demonstrates the feasibility of MR-signal transmission of the optical catheter link. The complete integration into a small tube of 6 F size was realized using a miniaturized optical bench fabricated from silicon substrates with 3D self-aligning structures for fiber integration. Since the system avoids any elongated conductors it is inherently RF-safe. In MRI phantom measurements projection based tip tracking and high resolution imaging was demonstrated with the optical link. These MR-Experiments in phantom liquids and inside a kiwi showed that the novel optical link exhibits an SNR comparable to a direct electrical link and is therefore quite sufficient for intravascular tracking and high resolution imaging.

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Figure 1: Concept of micro MR-probe: The probe is placed inside a phantom in an MRI scanner. Two fibers are applied in the catheter. One fiber is used to carry the signal information from the optical modulator and the other fiber for the optical power supply. Outside the catheter the optical signal is transformed into an electrical signal and is transmitted to the MRI scanner.

Figure 2: Electrical circuit of MR-probe: The MR-signal is received by the resonant circuit (L, C1). The signal is directly fed into the gate of the transistor T of the optical modulator. The signal is then transmitted by the laser diode LD to the MRI. The power is supplied by a photovoltaic power converter (PPC).
Figure 3: Equivalent electrical circuit of the optical modulator. The modulated laser current $i_D$ is calculated by the input voltage $U_{GS}$ and the transconductance $g_m$ of the transistor. $R_{laser}$ is the differential resistance of the laser diode.

Figure 4: a) Flexible planar polyimide foil with coil structures; b) Via to the electrical lead on the backside; c) Micro-Helmholtz-coil wrapped around a 5-F catheter.

Figure 5: a) Concept of the MR-probe with self-aligning structures on the micro optical bench and the reflector; b) Reflector inside self-aligning structure.
Figure 6: a) Flexible planar micro coil and capacitors for the resonant receiver circuit; b) Optical modulator on micro optical bench; c) Photovoltaic power converter (PPC) in self-aligning structure; d) Reflector on top of optical power supply.

Figure 7: MR-probe inside the catheter tip and optical fiber connectors
Figure 8: Output power of the optical receiver vs. induced voltage for the JFET and pHEMT optical modulator. The noise level at a receiver bandwidth RBW of 12 kHz according to the MR-imaging experiments is –86 dBm.

![Graph showing output power vs. induced voltage](image)

Figure 9: a) Tracking projection in the frequency domain; b-d) Overview image obtained with a C2 surface coil and the tracking position of the MR-probe obtained by the catheter channel.

![Tracking projection and overview images](image)

Figure 10: a) MR-micro image from phantom liquid transmitted by the optical link; b) MR-micro image from inside a kiwi transmitted by the optical link.

![MR-micro images](image)
REFERENCES


