Miniaturized magnetic-driven scanning probe for endoscopic optical coherence tomography

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Abstract: We designed and implemented a magnetic-driven scanning (MDS) probe for endoscopic optical coherence tomography (OCT). The probe uses an externally-driven tiny magnet in the distal end to achieve unobstructed 360-degree circumferential scanning at the side of the probe. The design simplifies the scanning part inside the probe and thus allows for easy miniaturization and cost reduction. We made a prototype probe with an outer diameter of 1.4 mm and demonstrated its capability by acquiring OCT images of ex vivo trachea and artery samples from a pigeon. We used a spectrometer-based Fourier-domain OCT system and the system sensitivity with our prototype probe was measured to be 91 dB with an illumination power of 850 μW and A-scan exposure time of 1 ms. The axial and lateral resolutions of the system are 6.5 μm and 8.1 μm, respectively.

References and links

1. Introduction

The last two decades have witnessed rapid development of the high-resolution biomedical imaging modality, optical coherence tomography (OCT) [1, 2], and associated endoscopic probes [3, 4]. Various endoscopic OCT probe designs, including both side-imaging [5–12] and forward-imaging probes [13, 14], were proposed to extend the application of OCT in imaging internal organs. The side-imaging OCT probes have been widely used in gastrointestinal, respiratory, and intravascular imaging. The designs of side-imaging probes can be categorized into two patterns: proximal rotation [5, 6] and distal rotation [7–12]. Proximal rotation is implemented by rotating the whole probe with actuation located at the proximal end. There is no scanning part at the distal end, and thus the probe can be made very small, e.g., with an outer diameter of 0.31 mm [6]. However, the proximal-rotation probes usually required a rotary joint to separate the fixed and rotated part of probe which will introduce some loss and thus impair the system sensitivity. Furthermore, the rotation of the whole probe impose additional requirement on the catheter and make it unsuitable for some specific applications, e.g., in winding internal channels [5]. Distal-rotation probes can avoid the abovementioned disadvantages by placing the scanning part in the distal end of the probe, however, with the cost of more complicated design for the distal-end scanning. In recent years, most popular distal-rotation probes are implemented by placing a micromotor [7–9] or micro-electromechanical system (MEMS) [10, 11] attached with a micromirror at the distal end of the probe. In these cases, the probe size will be determined by the state-of-the-art of miniaturization of micromotor or MEMS devices. Furthermore, one usual problem in the micromotor-based distal-rotation probes is that the electric wire used for driving the motor will block the light at some scanning angle and introduce wire shadows in the OCT image [14].

In this paper, we propose a distal-rotation side-imaging OCT probe based on magnetic-driven scanning (MDS). The probe exploits a tiny cylindrical magnet, which is radially magnetized, attaching to a reflection micromirror. For OCT scanning, we use magnetic field generated by a pair of larger magnets which are attached to a motor externally to drive the rotation of the tiny magnet inside the probe. As detailed in later sections, our MDS probe can be made very small and low cost because of its simple structure with external actuations. Furthermore, the MDS probe allows unobstructed 360-degree circumferential scanning because there are no connecting wires in the scanning part.

In the rest of the paper, we will describe the details of the MDS probe design and our OCT system in Section 2, then show the OCT imaging results of trachea and artery from a pigeon in Section 3, and finally discuss and summarize our work in Section 4.

2. Design of the MDS-OCT probe

Figure 1(a) shows a photograph of the prototype MDS probe compared to a ruler and a Chinese coin. The enlarged view of the probe tip is shown in Fig. 1(b). The schematic of the distal end of the probe is shown in Fig. 1(c). All the scanning structure is enclosed in a glass tube with an outer diameter of 1.4 mm and an inner diameter of 1.05 mm. A glass ferrule with 8-degree face cut couples the light signal into the probe from a single-mode fiber. A gradient-index (GRIN) lens (ComFiber Communications Technology) with 8-degree face cut is placed...
in front of the glass ferrule to focus the light beam. Both the glass ferrule and the GRIN lens have diameters of 1 mm. The length and corresponding pitch of the GRIN lens are 2.56 mm and 0.24, respectively. The focused light beam is then reflected by a reflection micromirror into the side of the probe. The micromirror is cylindrical-wedge shape with diameter of 0.9 mm. The micromirror is attached to a rotor, which consists of two identical tiny NdFeB magnets (AIC Magnetics) that are radially magnetized and connected by a stainless steel wire. Both the diameter and thickness of the tiny magnets are 0.9 mm. The stainless steel wire is enclosed in a short piece of PTFE (Polytetrafluoroethylene) plastic tube glued on the inner wall of the glass tube in order to prevent axial movement of the rotor. The tip of the probe is sealed by epoxy for protection purpose.

Fig. 1. (a) A photograph of the prototype MDS probe comparing to a ruler and a coin. The outer diameter of the probe is 1.4 mm; (b) enlarged view of the probe tip; (c) The schematic of distal end of the MDS probe.

Figure 2 shows the schematic our OCT system using the MDS probe as the sample arm. The OCT system is a traditional spectrometer-based Fourier-domain system [2]. The light source is a superluminescent diode (InPhenix SLD IPSDD0804C, 4.5 mW) with center wavelength of 843 nm and 3-dB bandwidth of 48 nm. The spectrometer is built with a collimator, a grating (Wastch Photonics, 1200 lines/mm), an achromatic lens and a line-scan
CCD (E2V SM2CL2014, 2048 pixels with pixel size of 14 x 14 μm). During the signal processing, we apply the dispersion compensation algorithm [15] to alleviate the dispersion mismatch between the sample and the reference arm.

To achieve circumferential scanning of the mirror, external rotating magnetic field is applied to drive the magnets inside the probe. Here we exploit a device showed in Fig. 2. The device consists of a direct-current motor and a pair of arms attached with large NdFeB magnets in rectangular shape with size of 5 × 4 × 0.5 cm. Both magnets are magnetized in the direction perpendicular to their largest face. The magnetic pole of one magnet is facing the opposite magnetic pole of the other magnet. The two magnets are separated by 9 cm and form an angle θ so that both magnets are directly facing the rotor magnets inside the probe. During the imaging experiments, the MDS probe is placed in between the large external magnets and inserted into the sample. When the motor drives the large magnets pair to rotate, the rotor inside the probe will rotate synchronously because of the magnetic force.

As we can see from the MDS probe structure, the MDS probe has two major advantages compared to other types of OCT probe: (1) The probe rotor consists of only two tiny magnets, a stainless steel wire, and a PTFE tube. All the components are low cost and can be made very small easily. (2) The exit OCT light from the probe will be unobstructed during 360-degree circumferential scanning because there are no connecting wire in the scanning part which is common in OCT probes using micromotors.

One important consideration of the MDS probe is the required distance between the external magnets and the probe tip, which is determined by the magnetic force. In the following we will estimate the magnetic torque exerted on the rotor inside the probe by the two external magnets. Figure 3 shows the schematic of the interaction between the two external magnets and the two rotor magnets. Assuming all the magnets are uniformly magnetized with the same magnetization $M$, and the distance between the large and small magnets $z$ is much larger than the sizes of the external magnets, then the magnitude of magnetic induction generated by the two external magnets can be written as [16, 17]

$$B_i = B_z = \frac{\mu_0 MV}{2\pi z^3}$$  \hspace{1cm} (1)

where $\mu_0$ is the magnetic permeability in vacuum and $V$ is the volume of the magnet. According to the abovementioned arrangement of magnetic poles, the directions of $B_1$ and $B_2$ are shown in Fig. 3, thus the total B field exerted on the rotor magnet is

$$B = 2 \cdot \frac{\mu_0 MV}{2\pi z^3} \cdot \cos \frac{\theta}{2} = \frac{\mu_0 MV}{\pi z^3} \cos \frac{\theta}{2}$$  \hspace{1cm} (2)

with direction shown in the figure, where $\theta$ is the angle between the two large magnets. The torque exerted on the two rotor magnets can be written as $N = m \times B$, where $m$ is the total magnetic moment of the rotor magnets and its magnitude can be calculated as $m = VrM$, where $V_r$ is the total volume of the two rotor magnets and $M$ is the magnetization. Here we consider the two rotor magnets together since they are very close to each other compared to $z$. The maximum torque $N_{max}$ will be achieved when $m$ is perpendicular to $B$ and can be calculated as

$$N_{\text{max}} = \frac{\mu_0 M^3 V r}{\pi z^3} \cos \frac{\theta}{2}$$  \hspace{1cm} (3)

According to the geometric relations indicated in Fig. 3, Eq. (3) can be written as

$$N_{\text{max}} = \frac{\mu_0 M^3 V r}{\pi} \frac{a}{(D^2 + a^2)^{3/2}}$$  \hspace{1cm} (4)
where $2a$ is the distance between the centers of the two large magnets, and $D$ is the effective working distance measured from the midpoint between the two large magnets.

To roughly estimate the maximum permissible $D$, we assume that a torque of 1 $\mu$N·m is enough to drive the rotor with 0.9 mm diameter according to typical numbers in micromotors with similar sizes [18, 19]. According to Eq. (4) and using the practical parameters $\mu_0 M = 1.6$ T [20] (here we assume saturated magnetization), $V = 5 \times 4 \times 0.5 = 10$ cm$^3$, $V_r = \pi \times 0.45^2 \times 0.9 \times 2 = 1.1$ mm$^3$, and $a = 4.5$ cm, then the torque of 1 $\mu$N·m corresponds to $D = 12.6$ cm. In the experiment, the rotor can rotate smoothly when $D < 8$ cm. We can expect a larger permissible $D$ if the internal frictions can be further reduced with better fabrication of the probe.

![Diagram](image)

**Fig. 3.** The schematic of the interaction between the two external magnets and the two rotor magnets.

### 3. Experimental results

Before the imaging experiments, we first calibrate the performance of OCT system with our prototype MDS probe. The probe has a working distance of 0.7 mm measured from the outer wall of the probe. The system sensitivity is measured to be 91 dB with an illumination power of 850 $\mu$W and CCD exposure time of 1 ms. This is smaller than the theoretically predicted OCT system sensitivity of 122 dB mainly because of the loss in the prototype probe. The axial resolution is measured to be 6.5 $\mu$m, which is consistent with the theoretical value according to the SLD spectrum. The lateral resolution is 8.1 $\mu$m, calculated by paraxial Gaussian beam propagation. Here we notice that the glass tube of the probe will introduce additional geometrical aberration of the exit beam.

We then used our MDS-OCT probe system to scan the *ex vivo* trachea and aorta sample removed from a pigeon. During the scanning, we captured the B-scan image using 930 A-scans with a rotation scanning speed at 2 rpm, and formed the circumferential cross-sectional image. Here the effective working distance $D$ is set as 6 cm. Figures 4(a) and 4(b) shows the imaging results of the trachea and aorta, respectively. In Fig. 4(a), the respiratory epithelium layer, the mucosal stroma, and the cartilage that has lower reflectivity can be clearly identified. In Fig. 4(b), we can discern the tunica intima, tunica media as well as tunica adventia of the pigeon’s aorta. The wall of glass tube of the probe can also be seen in the figures. Actually during the scanning, we observed minor deviation (< 100 $\mu$m) of the rotation axis because of imperfect probe construction, as shown in Fig. 4(c), which is the uncorrected image corresponding to Fig. 4(b). These deviations were calibrated and removed in the final image by knowing that the glass tube of the probe is circular.
4. Discussion and conclusion

To apply the MDS probe for \textit{in vivo} imaging, we need to consider several factors, including the dimension of external magnets pair, distance between the external and internal magnets, and the rotation scanning speed. For imaging small animals such as pigeon, mouse and rat, it will be relatively easy to use a large magnets pair that can enclose the whole small animal while keeping small enough distance between the external and internal magnets to ensure sufficient magnetic torque. For large animals and human, the MDS probe is more suitable for imaging locations that is close to the body surface, for example, less than 8 cm depth into the surface if using magnets similar as we used in the experiment. For deeper locations, stronger magnetic field will be needed which might require electromagnets or even superconducting magnets. Notice that simply increase the number of external magnets can also increase the magnetic field intensity. This is equivalent to increase the magnet volume $V$ in Eqs. (1)-(4).

The rotation scanning speed and uniformity depend on the rotation of the external magnets and the friction experienced by the magnets rotor. Besides the mechanical issue, the rotation speed will be ultimately limited by the acquisition speed of the OCT system.

In summary, we have demonstrated a miniaturized magnetic-driven scanning (MDS) side-imaging probe with spectrometer based Fourier domain OCT system. The probe has 1.4 mm outer diameter with working distance 0.7 mm, which is suitable for endoscopic scanning. In principle, it’s easy to further reduce the probe size with the same design. The resolution of the MDS-OCT probe system is $6.5 \mu m$ (axial) $\times 8.1 \mu m$ (lateral). We successfully obtained OCT images that clearly present the biological structures from \textit{ex vivo} tissues of the pigeon. The probe can be potentially used in endoscopic applications where the requirement for probe size is critical.

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