Distance optical sensor for quantitative endoscopy

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Abstract. We present a novel optical sensor able to measure the distance between the tip of an endoscopic probe and the anatomical object under examination. In medical endoscopy, knowledge of the real distance from the endoscope to the anatomical wall provides the actual dimensions and areas of the anatomical objects. Currently, endoscopic examination is limited to a direct and qualitative observation of anatomical cavities. The major obstacle to quantitative imaging is the inability to calibrate the acquired images because of the magnification system. However, the possibility of monitoring the actual size of anatomical objects is a powerful tool both in research and in clinical investigation. To solve this problem in a satisfactory way we study and realize an absolute distance sensor based on fiber optic low-coherence interferometry (FOLCI). Until now the sensor has been tested on pig trachea, simulating the real humidity and temperature (37°C) conditions. It showed high sensitivity, providing correct and repeatable distance measurements on biological samples even in case of very low reflected power (down to 2 to 3 nW), with an error lower than 0.1 mm.

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1 Introduction

In medical endoscopy, the knowledge of the distance from the endoscope tip to the surface of the anatomical cavity is crucial information toward obtaining the actual dimensions and areas of the anatomical object under investigation. On the other hand, every system aiming at solving such a problem ought to be easily integrable with the actual endoscopic probes. A variety of techniques have been developed to measure organs and lesions dimensions, such as imaging techniques [computed tomography (CT) or magnetic resonance imaging (MRI)] or ultrasounds. However, concerning the investigation of hollow organs (airways, intestine), endoscopy is the gold standard technique because of its capability of imaging lesions directly and at the same time of observing organ functionality by a dynamic view. It is a relatively cheap and routine procedure, that can be used for periodic evaluations. Moreover it can be performed directly by the specialist, in contrast with other imaging techniques that require waiting times.

Nevertheless, until now no system has been successfully demonstrated for the endoscopic quantitative measurement of anatomical dimensions. In fact, with current flexible endoscopes it is impossible to measure the actual size of the imaged anatomical objects, as the instrument requires the presence of a wide-angle lens aimed at enlarging the field of view, but at the same time it magnifies details, depending on their distance from the endoscope tip, which is unknown and can constantly change during the examination.

In clinical practice, empirical methods are used to obtain an approximate measure of lesion size, but they are inaccurate, hardly repeatable, and unfortunately, often invasive. Recently, many approaches to this problem have been proposed. If possible, an object of known size is put near the organ of interest for image calibration, but such a solution is not always applicable in clinical practice.[1,2] Other authors propose the projection of a pattern on the field of view, and the dimension of anatomical objects is computed by processing the pattern image,[3,4] but this process requires some computing time, and in this way, the clinician is unable to monitor in real time the quantitative data that can be useful during the examination as well.

Currently empirical methods are used to measure the unknown distance: either by invasive methods, such as the insertion of a marker in the endoscope tip,[5] or indirectly, i.e., by measuring a relative displacement between endoscope and anatomical wall.[6] Therefore, with the current techniques, quantitative dimensions are subjective estimates, and intraobserver and interobserver variations are likely to be high or, in some cases, undefined. On the contrary, our approach is based on the direct measurement of the absolute distance between the endoscope tip and the anatomical walls, eliminating the delay time necessary to extrapolate this information by image processing.

The goal has been to realize an optical sensor able to measure distances in a reproducible way, to be inserted in the operative channel present in most clinical endoscopes. For this reason, we employ optical fiber technology to implement our sensor, so that it can run parallel to the image acquisition fiber bundle. This is based on the optical technique of fiber optic low-coherence interferometry[7] (FOLCI), since it offers high precision in a range of typical distances (1 to 30 mm) in an endoscopic examination, and because of its good immunity to the optical noise connected with the unstable conditions in human body cavities.[8] A near-IR source is used because of the low absorption of the human tissues at these wavelengths. Thanks to these distance data, it is possible to obtain from the endoscopic images a variety of quantitative data that will help clinicians in diagnosis and therapy.

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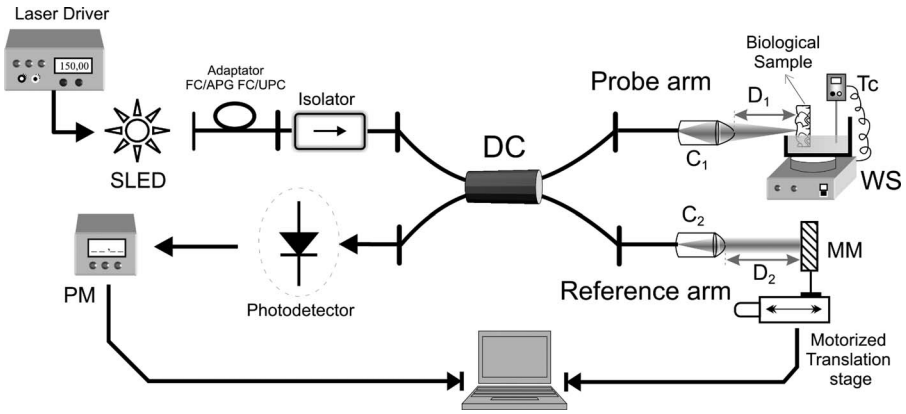

Fig. 1 Distance sensor experimental setup: SLED, broadband laser source; DC, directional coupler; C_1, focal lens; C_2, collimator lens; Tc, temperature control; WS, heating magnetic stirrer; MM, motorized mirror; and PM, power meter.

2 Experimental Method

Figure 1 shows a sketch of the experimental setup. Low-coherence interferometry is performed by means of a superluminescent laser diode (SLED) (λ_c=1550 nm, $\Delta\lambda$=60 nm, power=150 μW) with a coherence length (L_c) of about 40 μm. An optical signal is sent from the source to a fiber optic interferometer, represented by a 3-dB fiber optic directional coupler. One arm of the interferometer, to be inserted in the operative channel of the endoscope, is the probe arm, while the other arm, used as reference, is placed externally. Thanks to the source bandwidth, interference will occur in the detected signal only when the optical paths are the same in both interferometer arms, with an uncertainty[9] of about 20 μm, which is acceptable in our application. The reference arm optical path is scanned moving a reference mirror (MM) by an automatic stepping motor, while at the same time, the interference signal is detected and acquired by a data acquisition (DAQ) PC card. In this way, we obtain an interference pattern versus the motor position. If the fiber branches used in the two arms are of the same length, the real distance $D1$ of the biological sample is equal to the motor position $D2$ in correspondence of the maximum of the interference pattern. To perform time-repeated scans of the distance from the endoscope to the anatomical wall, the moving and acquiring process is completely automated using a PC system. With the aim of reducing the acquisition time, we are working at the software optimization to obtain a real-time measurement. The results reported were obtained with an acquisition time in the range of 0.5 to 1.5 s. Currently the only time limitation is due to our preliminary acquisition software, which is not yet optimized. We did not try to further reduce the acquisition time, since the goal of these first tests was to demonstrate the usefulness of the technique for endoscopic applications. However, there are no intrinsic limitations either in the technique or in the hardware used (step motor and acquisition board) to increasing the scanning speed by 30 to 50 times.

When dealing with an interference system, sufficient fringe visibility is necessary to be able to identify the maximum of the interference pattern. Unfortunately, working with biological samples, a problem we faced has been the very low reflection from the anatomical walls. Moreover, the surfaces of the tissue under examination often present high roughness, increasing losses by light scattering. To collect most of the signal reflected by the biological tissue and to facilitate the alignment procedure, different kinds of lenses at the probe arm tip were tested. A great improvement was provided by the use of a lens with focal length of 1.1 cm and a focal point size of about 50 to 80 μm. In this way, illuminating a very small spot of the biological sample, we can reduce the roughness problem. This solution enables collecting a good signal from the tissue and measuring the distance $D1$ in the range from a few millimeters to 2 to 2.5 cm, depending on the specific application. The scanning range of the translator is 26 mm, which is an appropriate interval to identify the correct distance value.

The broadband source guarantees high accuracy in the distance measurement since it is possible to get interference only if the difference $\Delta d = |D1 - D2| \leqslant Lc$. However, in the real conditions of an endoscopic examination the SNR of the interferogram may be very low, because of the misalignment of the lens with respect to the anatomical wall, or the light scattering by the biological surface, making it difficult for the acquisition software to detect the correct distance value. For this reason, we used a peak detection algorithm based on the Hilbert transform, that increases the sensitivity of the measurement. We compute the envelope of the intensity signal, according to[10]

$$E(t) = [i^2(t) + \hat{i}^2(t)]^{1/2}, \tag{1}$$

where $i(t)$ is the acquired intensity signal, and $\hat{i}(t)$ is the Hilbert transform of $i(t)$. The maximum value of the envelope $E(t)$ corresponds to the zero optical path difference position.

3 Results

We demonstrated the effectiveness of this sensor on biological tissues by performing distance measurements on pig trachea samples. LabVIEW software was used to acquire and process the experimental data. To simulate the real environmental conditions of the measurement, we heated the samples to 37°C by dipping them in warm water (Fig. 1). A typical behavior of the interference pattern obtained is reported in Fig. 2. We can observe a flat signal until we cross through the interference region, where immediately an interference pattern appears. The small asymmetry shown by the measured interferogram is due only to the imperfect matching of the

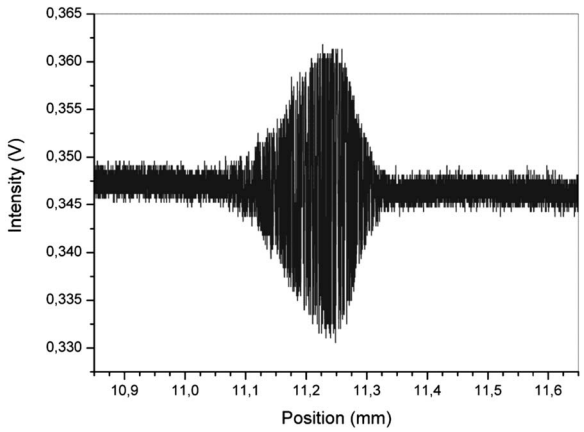

Fig. 2 Interference signal as a function of the motorized mirror position. After a calibration of the stepping motor, the center of the interference pattern corresponds to the distance of the probe arm tip from biological sample.

length of optical fibers used in the two arms. In fact, the laser emission of a broadband source is formed by continue "wave trains" of ultrashort duration (about 100 fs) that traveling along the fiber are strongly affected by dispersion. A few millimeters of fiber path difference are enough to evidence the pattern interference asymmetry.

Figure 3 shows the envelope obtained by the Hilbert transform. In correspondence of its maximum value, the software identifies the position of zero optical path difference, thus obtaining the absolute distance $D1$ of the biological sample. To verify the correctness of the measurements obtained with our distance sensor, we measured the real $D1$ distance using a caliper with 0.05-mm precision. We repeated every measurement more than 10 times, always obtaining the correct value within the experimental error. In the example reported in Fig. 3, we obtained a distance of 11.26 ± 0.02 mm with the optical sensor, and a distance of 11.20 ± 0.05 mm with the caliper, showing very good agreement between the two measures.

In this example, the power at the probe arm was about 20 nW, but we must often work with only few nanowatts of signal power. In this case, the fringe visibility of the interference pattern is very low, making it difficult to identify the interference maximum without any signal processing. Our peak detection algorithm is capable of identifying the interference maximum even in case of very low power at the probe arm, down to 2 to 3 nW, or in presence of "false maxima" outside the interference region.

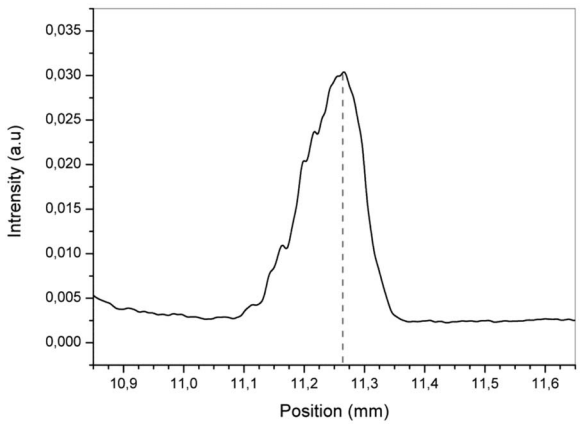

Fig. 3 Envelope obtained by Hilbert transforming the detected intensity signal as a function of the motorized mirror position.

4 Discussion and Conclusions

We realized and tested an absolute distance sensor, based on the optical FOLCI technique, to be used in parallel with a medical endoscope to quantify areas and dimensions in images. Tests performed on biological samples showed that the system is very sensitive, providing correct and repeatable distance values even in the case of very low SNRs. Thanks to the effectiveness of FOLCI for this kind of application and to our signal processing based on the Hilbert transform, we obtained acceptable results even in the case of very low collected signal by the probe arm, down to 2 to 3 nW. Further improvements can be made in enhancing the light collected by the probe arm, for instance by increasing the source power, according to the safety rules on the maximum laser radiation exposure of biological tissues, or by choosing the SLED central wavelength to maximize the reflectance of biological tissues.[11]

In conclusion, we provide a technique for measuring, in a precise and repeatable way, the distance between the endoscope tip and the anatomical object under investigation, to obtain quantitative data from endoscopic images. Our sensor consists in a technical solution that can be easily built in a traditional clinical endoscope, then maintaining all the advantages of such imaging technique. Moreover, it could enable faster and easier introduction of the device in the clinical field than a completely new imaging technique, avoiding time-consuming and expensive training.

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