Time-of-Flight Based Endoscopy for NOTES Interventions: Challenges and Limitations

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Abstract

The emerging Time-of-Flight (ToF) technology is looking for its way to minimally invasive surgery. Detailed analysis of the clinical workflow and surgical challenges during Natural Orifice Translumenal Endoscopic Surgery (NOTES) interventions explicitly pointed out that finding the optimal access point to the peritoneal cavity as well as intraoperative collision control and navigation support are of utmost importance for the success of NOTES and have to be solved by methods of information science. ToF based endoscopy has the potential to address these issues successfully by measuring the time of flight of an actively emitted modulated reference signal in each pixel and thus providing distance information for each pixel. Various options for optimizing the overall signal transmission of the modulated reference signal and consequently the measuring uncertainty have been determined and are subject to current research. We furthermore describe how to correctly model the signal transmission properties of a ToF based endoscope and are thus able to draw valid conclusions on an achievable measurement uncertainty less than 2mm.

1 Introduction

The field of NOTES interventions [1] has again brought an issue into the focus of interest, which has already been investigated in the context of minimally invasive surgery for a long time: The acquisition of real-time three-dimensional information of the operation area via an endoscope optic which is an integral part of the operation room of the future [2]. Recent approaches mainly focused on the extraction of three-dimensional information from a calibrated endoscope optic and a sequence of acquired images from a rigid operation area [6, 7]. Inherently, these approaches are not real-time capable. Other approaches projected patterns or structured light into the operation area [5, 9]. Although, real-time constraints could be met by these techniques, the utilized modifications of the endoscope optic where too costly to make the approaches usable. The major contribution of our novel approach is utilization of a real-time capable sensor system with the ability to derive three-dimensional information of the operation area at constant lateral resolution. The utilized ToF sensor [8] computes the distance of the

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observed scene in every pixel by measuring the flight time of an actively emitted modulated optical light signal via the phase shift of the emitted and reflected signal. Such ToF sensors are also available as integrated parts of camera systems. Additionally, ToF sensors do not only provide distance information for every pixel, but also intensity information: As the intensity relates to the amplitude of the reflected modulated signal these intensity values are referred to as amplitude values.

By mounting a ToF sensor at the end of the endoscope optic and transmitting the obligate modulated optical light signal via the available optical fibers, the ToF measurement principle can be realized via an endoscope optic. The available data is the same as is available for ToF camera systems: Distance information, from which 3D coordinates can be easily computed, and amplitude values for every pixel.

Nevertheless, the two-dimensional color information provided by standard endoscope systems is of utmost importance for NOTES interventions and also for minimally invasive interventions in general. Thus, the proposed approach of only attaching a ToF sensor to the endoscope optic will hardly be accepted for use during NOTES interventions. This problem can be solved by additionally attaching an optical beam-divider, which projects the incoming optical signal onto a standard CCD sensor, which provides the color information, and onto a ToF sensor, which provides the 3D information. Having such a camera system calibrated using standard approaches [10] the 3D points can be virtually re-projected onto CCD sensor and a colored overlay of the 3D points can be provided. A scheme of this enhanced endoscope optic is given in Figure 1.

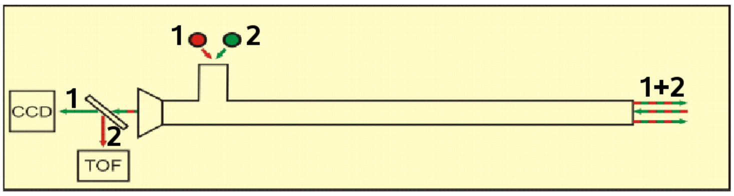

Figure 1: Scheme of the ToF based endoscope: Modulated light (1) and standard illumination (2) are simultaneously transferred in to the operation area (1+2). After being reflected the standard illumination light is projected onto a CCD sensor; the modulated light is projected onto a ToF sensor.

2 Modeling an appropriate ToF based endoscope

The measurement uncertainty of a ToF sensor can be modeled by [3]

$$\sigma = \frac{c}{2\sqrt{8}f} \cdot \frac{1}{M} \cdot \sqrt{B + \frac{M}{2}}, \tag{1}$$

where σ is the standard deviation of the range error, f is the modulation frequency, c is the speed of light, M is the number of photoelectrons generated by the modulated light signal and B is the number of photoelectrons generated by background light. Using $f = 100\ MHz$ and $M = 1000000$ the achievable measurement uncertainty σ was computed for varying values of B. Figure 2 shows the results. It can be observed that a measurement uncertainty of 2 mm can well be achieved even in the presence of as much background photoelectrons $B = 100000$ as signal photoelectrons $M = 100000$. This leads to two conclusions:

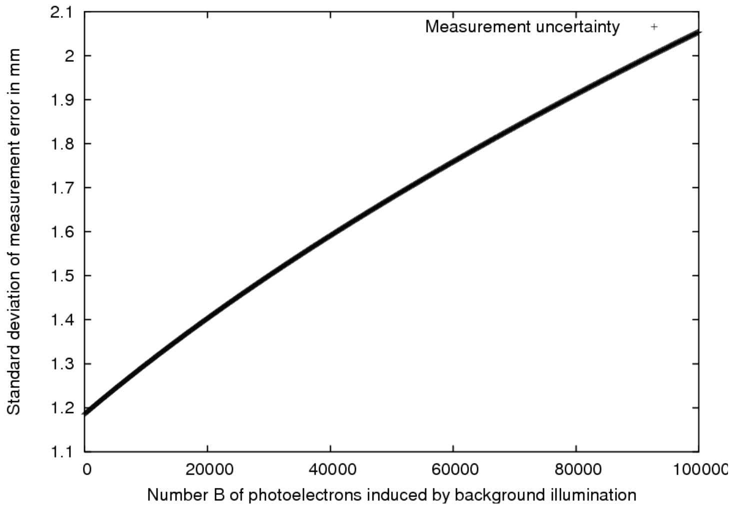

Figure 2: Measurement uncertainty σ in mm with respect to background illumination B.

1. The light source used for the ToF endoscope has to be modulated with 100 MHz.
2. The generated light signal must generate approximately 100000 photoelectrons on the ToF sensor

In the following we want to discuss the second question in detail.

3 Conclusions for the hardware setup and achievable measurement uncertainty

The modulated light signal is emitted with an optical power P. During its way towards the object and back to the ToF sensor it is attenuated like every electro-magnetic wave with respect to the traveled distance, which we call R (the distance of the observed object). Some amount of the light is absorbed by the object: This effect is modeled by the remission $\phi \in [0..1]$ of the object. The signal receiving unit (lens and ToF sensor) is modeled by the aperture D of the lens, the area A_S of the ToF sensor, the area A_P of one pixel on the ToF sensor, and the quantum efficiency $q(\lambda)$, which again relies on the utilized base wavelength λ. Finally, the factor $k \in [0..1]$ models attenuation effects introduced by utilized lenses and the parameter T models the integration time during which the measurement is accomplished. Briefly, this models how much of the incoming optical power is projected onto each pixel and how many photoelectrons N_e are generated in each pixel.
As $N_e = 100000$ have to be generated, conservative values are chosen for the other parameters of the model:

- $\phi = 0.8$, i.e. the observed object is not too absorbing; due to the fact that tissue in the operation area is relatively glossy, this assumption can be considered valid.

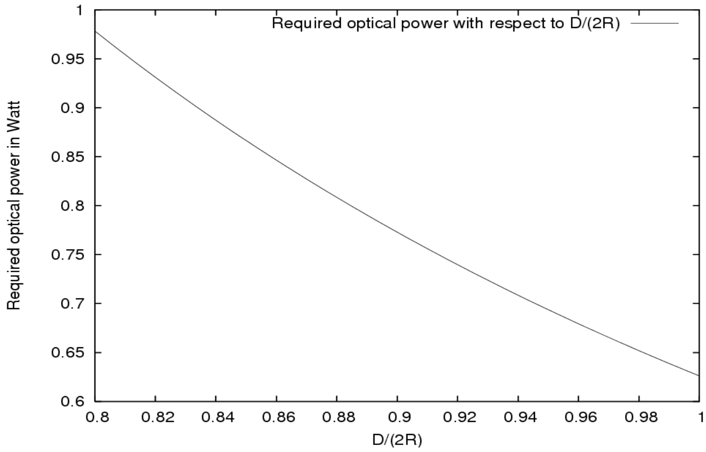

Figure 3: Required optical power of illumination source with respect to $\frac{D}{2R}$ in the range of $0.8...1.0$

- $k = 0.3$, i.e. the utilized lenses attenuate the incoming signal strongly; this is valid as the optical lense system of an endoscope optic consists of many lenses and not only one single lense.
- $\lambda = 850nm$, i.e. the near infrared range is used; this is a valid choice for the wavelength as higher wavelength are more likely to induce heat damages.
- $q(\lambda) = 0.9$, i.e. only a fraction of the incoming photons generate an electron.
- $T = 10ms$, i.e. the integration time is chosen small enough to enable real-time usage and take into account some time necessary for post-processing of the acquired data.
- The ratio $\frac{D}{2R}$ describes the ratio of lense aperture and object distance. This term is considered as a degree of freedom in the following discussion.

For the chosen parameter values, the necessary optical power P was computed with respect to the ratio $\frac{D}{2R}$, i.e. lense size vs. object distance. Thus, a value of $\frac{D}{2R} = 0.5$ means that the object is as far away as the size of the lens is. But this is only valid under the assumption that the object is a Lambert reflector and was practically validated with ToF cameras only for observable distances in the range of several meters. Both assumptions are likely to be not fulfilled for endoscopic ToF data acquisition. Thus, we concentrate on the range of $[0.8..1.0]$ for the ratio $\frac{D}{2R}$. The formula used for the computation of the necessary optical power is [3]:

$$P = \frac{N_e \cdot \frac{A_S}{A_P} \cdot h \cdot c}{\phi \cdot (\frac{D}{2R})^2 \cdot k \cdot q(\lambda) \cdot \lambda \cdot T}. \tag{2}$$

The results are shown in Figure 3. It can be observed that a measurement uncertainty of 2mm can be reached under valid assumptions with an optical illumination power of approx. 1W, which is modulated with 100MHz.

4 Conclusion and future research

The previous discussion has determined the physical requirements for achieving a measuring uncertainty which initially makes the ToF based endoscope feasible for NOTES interventions and minimally invasive procedures. Future research will have to focus in the appropriate design and implementation of the illumination unit and once this is successfully accomplished on appropriate processing routines to provide threedimensional intraoperative navigation assistance and collision control. [4]

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