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A wafer-level miniaturized Michelson interferometer on glass substrate for optical coherence tomography applications☆

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ABSTRACT

The wafer-level fabrication of a Michelson interferometer using optical MEMS technologies is presented. The intended application is in Optical Coherence Tomography (OCT). The micro fabrication involves two steps: the 45° saw dicing of glass substrate and the subsequent deposition of a dielectric multilayer and metallic layers to obtain a beam splitter and waveguide micro mirrors, respectively. The Michelson interferometer was designed for use in the near-infrared range of 800–900 nm. A 50/50 non-polarized beam splitter was obtained with only four layers (using titanium dioxide and silicon dioxide). The micro mirrors for the required spectral range were fabricated by sputtering of chromium and gold layers. The dicing cuts, which were performed with a custom-made 45° dicing blade, resulted in smooth slopes. The surface's roughness is 19.76 nm at setting and can be reduced to approximately 50% with a soft additional dicing cut. The height of the 45° surfaces was approximately 400 μm, which is in accordance with the design. The micro Michelson interferometer can be easily integrated with other optical components into a complete OCT miniaturized system.

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1. Introduction

The intrinsic advantage of non-invasive or minimally invasive techniques, such as reduced burden on the patient, has made this a general trend in medical technology of high potential. The objective of minimum invasiveness has significantly affected the design of medical tools, but has also become a key issue in biomedical imaging [1,2]. Whenever possible, an attempt is made to eliminate and substitute invasive methods by others that provide the same results with a reduced negative impact on the object being examined. Optical methods play an important role in modern medical imaging. Optical Coherence Tomography (OCT) is a recently introduced non-invasive optical real-time imaging technology for high-resolution and cross-sectional tomographic imaging [2,3]. The basic OCT concept was firstly presented by Huang et al. in 1991 [4]. The first in vivo OCT image was presented by Fercher et al., in 1993 [5]. From the first applications in ophthalmology [6], OCT has also penetrated into other medical fields, such as dermatology [7], endoscopy [8] and cardiology [9]. The operating principle is based on the interference between a signal from an object under investigation and a local reference signal. The most-generic OCT concept is referred to as Low Coherence Interferometry (LCI), and makes use of near infrared (NIR) light to analyse biomedical samples [1,2,7,10,11]. LCI measures the echo time delay and the intensity of the backscattered light, by correlating it with the light that travels a known reference path [3,7]. The conventionally used configuration of LCI employs a Michelson interferometer for achieving the interference: a NIR light source emits a light wave to a beam splitter (BS) witch splits this wave in half. The backscattered light from the sample arm interferes with the reflected light from the reference mirror in the BS, and the interference is finally detected at the interferometer output [3]. There are two approaches for implementing LCI in OCT: time-domain OCT (TD-OCT) and frequency domain OCT (FD-OCT). The latter one is more attractive, because there is no need of a moving reference mirror, which is an essential component in TD-OCT for depth scanning [2].

Current OCT systems available in clinical practice operate in FD-OCT. These systems are mainly used in ophthalmology. The use of OCT in biomedical applications is presently limited by its high cost and large instrument size. The integration of several complex optical devices as miniaturized components on a single microchip is a significant challenge in OCT development [12,13]. MEMS technologies offer a huge potential to achieve such a goal [14,15]. In this paper the wafer-level fabrication of a micro BS and micro mirrors is presented for use in the Michelson interferometer within an OCT configuration. The basic approach was already presented in

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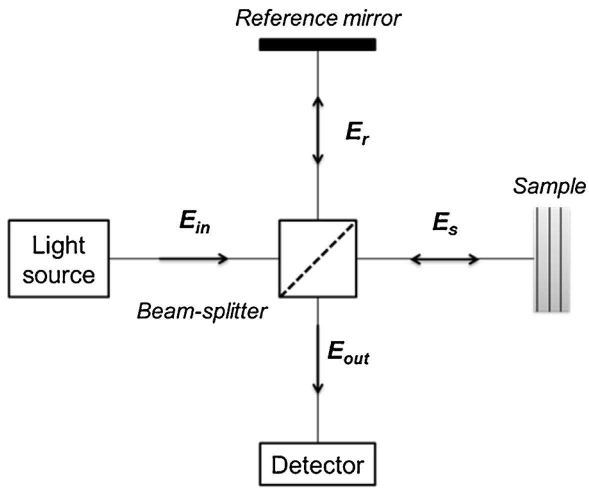

Fig. 1. Basic OCT configuration: Michelson interferometer with representation of variables E_{in}, E_{out}, E_r and E_s, which represent the optical fields in the input, output, reference and sample arms, respectively.

literature for application in the visible part of the spectrum [16]. However, the proposed prototype is designed for a wavelength range in the NIR region of electromagnetic spectrum, typically used in OCT imaging. The fabrication process is based on saw dicing of glass substrate and thin film deposition of a dielectric and metallic layers. The saw dicing process is well characterized for ensuring optimum conditions for the subsequent thin-film deposition. To the best of our knowledge this work represents the first step towards OCT miniaturization using optical MEMS technologies on a glass substrate. The proposed wafer-level Michelson interferometer can be easily integrated with a spectrometer in a complete spectral domain OCT system on a chip, representing the actual trend of scientific research in OCT miniaturization. Additionally, other optical components, such as a moving micro mirror in the sample arm to perform the sample scanning, can be added in the prototype, being an innovative potential of the current configuration.

2. OCT overview

2.1. Theoretical concepts

OCT is an interferometric technique based on the interference between two light waves: one from a reference mirror and another from the sample being examined [1]. This concept is represented in Fig. 1, with the basic Michelson interferometer used for OCT imaging. There are two approaches for implementing LCI for OCT imaging: TD-OCT and FD-OCT. FD methods can be implemented in two ways: (*i*) spectrometer based (SD-OCT) or (*ii*) by using a tunable laser or a swept-source (SS-OCT) [10].

2.1.1. TD-OCT

The first OCT generation operated in the time domain, where the interference pattern is obtained by the displacement of a reference mirror [1,2]. A TD-OCT setup is equipped by an optical source and a Michelson interferometer, which contains a BS and a moving reference mirror [7,9,10]. The sample wave, which travels from the BS to the sample and back to the BS, traverses the sample arm. Similarly, the reference wave traverses a reference path from BS to the reference mirror and back to the BS. The optical path difference (OPD) can be defined as the difference between sample and reference path lengths (OPD = |sample path length − reference path length|). TD-OCT operation principle is based on partial coherence interferometry: the photo-detector only senses variations in the interference when the OPD is less than the coherence length of the broadband source [7,10]. The TD-OCT interference pattern in each axial scan is given by the sum of two terms, assuming a lossless 50:50 split ratio BS.

$$I(\Delta Z) = \Gamma_0 + \Re\left\{\Gamma(\Delta Z)\right\} \tag{1}$$

The output of an TD-OCT system (Eq. (1)) is a function of reference path length (ΔZ) by integrating it over the source spectrum [1]. The term Γ_0 denotes only the contribution from self-interference, while the term $\Gamma(\Delta Z)$ includes the contribution from the cross interference:

$$\Gamma_0 = \frac{1}{4}\int_{-\infty}^{+\infty} S(w)\cdot(|H(w)|^2 + 1)dw \tag{2}$$

$$\Gamma(\Delta Z) = \frac{1}{2}\int_{-\infty}^{+\infty} H(w)S(w)\cos\left\{\phi(\Delta Z)\right\} dw \tag{3}$$

$S(w) = |s(w)|^2$is the intensity spectrum of the source and the function *H(w)* represents the overall reflection from all structures distributed in the *z* direction of the sample. $\phi(\Delta Z)$is the accumulated phase by translating the reference mirror with a geometric distance ΔZ [1].

2.1.2. SD-OCT

A SD-OCT setup makes use of a spectrometer in the detection system. As in TD-OCT configuration, a broadband source with short temporal coherence length is used in Michelson interferometer input. However, the depth information is obtained by spectral density measurement, using a spectrometer. The interference beam is dispersed by a diffraction grating and all the individual wavelength components are detected by an array detector [1,10,17]. The operation of a SD-OCT system is based on output demodulation of the optical spectrum. The axial depth scan is performed without a mechanical displacement of the reference mirror, as in TD-OCT configuration. Basically, the backscattered light from different depths in the sample overlaps with the reflected light from the reference mirror and results in a modulated spectrum. In order to reconstruct the structure of the measured object as a function of depth (i.e. the depth profile) in the direction of the light path, it is necessary to apply an inverse Fourier transformation to the modulated spectrum [2]. Since the reference mirror is at a fixed position, $\Delta Z = 0$ and a perfect 50:50 BS is assumed, the detected frequency spectrum is obtained as follows:

$$I(w) = \frac{1}{4}S(w)\left\{H(w) + 1\right\}^2 \tag{4}$$

The main advantage of SD-OCT, as compared to TD-OCT, is that the depth profile (*A-scan*) is measured from a single spectrum, without the mechanical movement of the reference mirror. This allows a faster acquisition rate, using a line scan array [2,9,10].

2.1.3. SS-OCT

In a SS-OCT configuration a narrowband source, such as a tunable laser, is used, instead of a broadband source. The tunable laser scans the entire operating spectrum during a specific time interval [10]. As in SD-OCT, no moving parts are required for depth profile scanning. This configuration requires rapidly tunable, narrow line-width lasers in combination with high-speed data-acquisition systems. The detection is made by single-point detectors, which allows high-speed detection: the signal is recorded by the detector when the laser is scanning over a specific wavelength. The recorded data is equivalent to the recorded data in a spectrometer. Consequently, as in the SD-OCT approach, an inverse Fourier transform provides the depth information [1,17].

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