Power budget considerations for in vivo continuous glucose monitoring using absorption spectroscopy

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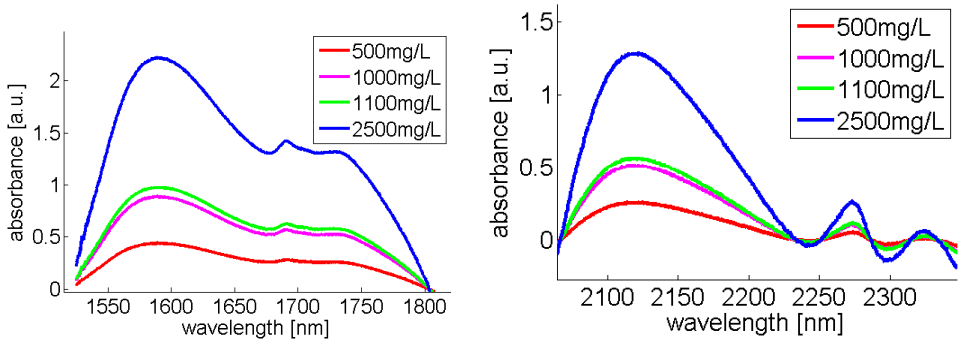
Diabetes is a serious health condition that can be treated more efficiently when the blood glucose level is monitored continuously. We study an implantable photonic chip that determines the glucose concentration by means of near infrared spectroscopy in the first overtone and combination band. In human tissue the glucose absorption spectrum is a weak variation on the water spectrum and light scattering is prevalent. Both conditions necessitate a high signal-to-noise ratio measurement. Power budget calculations of a post-dispersive integrated system elucidate how the low étendue of single mode optical waveguides in combination with strong water interference fail to realize such a high SNR. Unless adaptive optics is employed to compensate for wave front aberrations due to light scattering, laser spectroscopy is the preferred alternative.

Introduction

Diabetes mellitus is an auto-immune disease that deteriorates the insulin production of the pancreas. An elevation of the blood glucose level is a direct consequence (hyperglycaemia). The world health organization estimates that the number of diabetes patients worldwide accounts to 220 million people [1]. As diabetes is a chronic disease the cost for health care is enormous. These medical care costs can be reduced substantially by continuous blood glucose level control. The ideal implementation is a closed-loop control system with a glucose sensor and insulin pump to keep the blood glucose level in the healthy range of 60-110 mg/dL. Huge investments are made in research towards this goal. For clinical usefulness the glucose concentration needs to be detected with an accuracy of 100 mg/L. Apart from the sensor's accuracy, the sensor lifetime is a key parameter. Optical methods offer the largest chance on a long sensor lifetime as there is no need for reagentia whose effect diminishes over time. Still, methods using reagentia have a high specificity to glucose. Assisted by multivariate calibration strategies, optical methods like near-infrared and Raman spectroscopy also offer good specificity to glucose thanks to the uniqueness of the absorption and Raman spectrum. In the SBO-IWT GlucoSens project we aim at developing an implantable glucose sensor based on near-infrared absorption spectroscopy.

Glucose absorption features

The near-infrared (NIR) glucose absorption spectrum can be divided into two large regions. The first region is called the first overtone band (1550-1800 nm) and the second range is the combination band (2050-2400 nm). Figure 1 shows the different spectra.



(a) First overtone band

(b) Combination band

Figure 1: First overtone band(a) and combination band(b) absorption spectrum of glucose. The blue curve shows the glucose absorption during a hyperglycaemic state and the red curve during a hypoglycaemic state (overdose of insulin).

In the human body the glucose absorption is 3 orders of magnitude smaller than the water absorption. It implies that in a spectroscopic measurement the glucose spectrum appears as a weak variation on the water spectrum. The strong water absorption also limits the path length of the measurement to 5 mm and 1 mm in the first overtone and combination band respectively [2]. Apart from water absorption, the in vivo detection of glucose is hindered by the scattering nature of tissue. The resulting distribution of photon trajectories and according path lengths make it difficult to relate the absorption spectrum to the glucose concentration.

Layout of the implant:

Our device will consist of an on-chip spectrometer embedded in a biocompatible material like PMMA or PDMS with a total size of about 1 cm^2 x 0.3 mm. The spectrometer is integrated on the silicon-on-insulator (SOI) platform. This platform is CMOS-compatible allowing for future cheap mass fabrication. An electronic chip, 3D integrated with the photonic SOI chip, will drive the source and detectors. An artist impression of such a device is shown in figure 2.

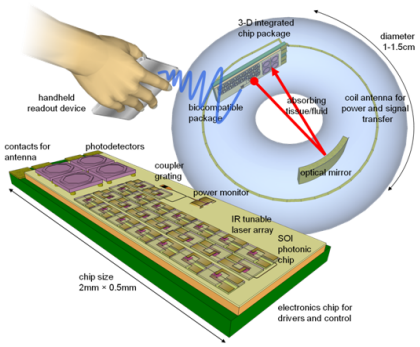


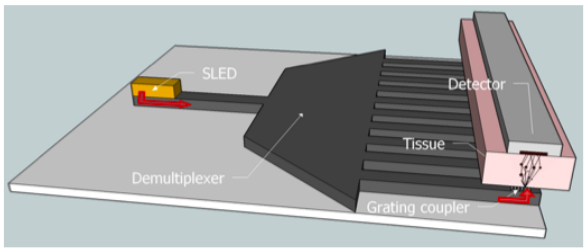
Figure 2: Possible device layout

The device can work in two distinctive ways: either a post-dispersive approach is implemented in which broadband light is coupled to the tissue and afterwards the transmitted

light is demultiplexed by a planar concave grating (PCG) [3] or in the pre-dispersive set-up we couple different wavelengths sequentially to the tissue and measure the transmission of 1 wavelength at a time. In the latter case we can use a broadband source which we demultiplex or we can use an array of laser with each a fixed emission wavelength.

Power budget calculation:

The choice between the different device layouts depends largely on the power budget. We compare the pre and post-dispersive set-up in the case we have a broadband light source like a superluminescent light emitting diode (SLED).



(a)

Loss factor	F.O. band	C. band
dT	38,5 dB	44,1 dB
GC_s	3 dB	3 dB
GC_b	6 dB	6 dB
L_{pcg}	3 dB	3 dB
L_{SLED}	6 dB	6 dB
L_{medium}	31,4 dB	13,6 dB

(b)

Figure 3: (a) Pre-dispersive layout with SLED and (b) a table showing the different loss factors for a realistic implementation.

For the pre-dispersive set-up as shown in figure 3 we can calculate the necessary total input power P_{in} [dBm] of the SLED using the following formula:

$$P_{in} > 10\log(3NEP) + dT + L_{medium} + GC_s + 10\log(N_{chan}) + L_{pcg} + L_{SLED} \qquad (1)$$

in which dT is the change in optical transmission due to 100 mg/L of glucose, L_{medium} is the loss in the medium due to absorption and scattering, GC_s is the loss of the grating coupler for narrowband light, N_{chan} is the number of channels, L_{pcg} is the loss of the demultiplexer, L_{SLED} is the loss for coupling the SLED light to a waveguide. The detector sensitivity is characterized by the noise equivalent power (NEP) which is defined as the input power P_s for which the signal-to-noise ratio (SNR) equals 1. A general expression for the SNR of a photodiode is [4]:

$$\frac{S}{N} = \frac{\eta P_s^2}{2h\nu B\left[P_s + \frac{h\nu i_D}{\eta e} + 2\left(\frac{h\nu}{e}\right)\left(\frac{kT_N}{e}\right)\left(\frac{1}{\eta R_{load}}\right)\right]} \qquad (2)$$

in which h is Planck's constant, k is Boltzmann's constant, ν is the light frequency, $B = \frac{1}{2t}$ is the measurement bandwidth with t the the integration time, T_N is the effective noise temperature, η is the quantum efficiency of the detector, e is the elemenatary charge, i_D is the dark current and R_{load} is the load resistance.

Similarly we can write down the needed power for a post-dispersive set-up:

$$P_{in} > 10\log(3NEP) + dT + L_{medium} + F + GC_b + 10\log(N_{chan}) + L_{pcg} + L_{SLED} \qquad (3)$$

in which GC_b is the loss in the grating coupler for broadband light and F is a factor representing the fraction of the transmitted light that couples to the fundamental mode of the input waveguide. This fraction is always smaller than 1 when scattering alters the wavefront. The inherent low étendue (characterized by F) of a post-dispersive system for highly scattering media like tissue suggest that the pre-dispersive set-up should be preferred. We estimate F to be minimally 20 dB for a tissue-like environment, but further measurements are necessary to set a definite value. For non-scattering media however, a special situation exist in which the post-dispersive set-up with a broadband light source can compensate for the lower étendue by using very small integrated detectors (20 um x 20 um). When the detector becomes dark-current-limited ($R_{load} > 10M\Omega$) decreasing the detector area results in a decrease of the NEP and the post-dispersive set-up can become beneficial compared to the pre-dispersive set-up with a large area detector.
When we calculate the needed SLED power with realistic values of $R_{load} = 1M\Omega$, F=20 dB and the loss values as found in figure 3, we obtain 13.58 mW for the pre-dispersive set-up and 461 mW for the post-dispersive set-up in the first overtone band. This shows that the necessary SNR cannot be realized using an integrated SLED. Instead of using a broadband light source we can implement a laser spectroscopy system to boost the signal-to-noise ratio. We envisage a laser array heterogeneously integrated on SOI with 30 lasers spaced 3 nm apart. The efficient light generation in lasers in combination with the ease of direct modulation offer clear advantages. Using the formula $P_{in} > 10\log(3NEP) + dT + L_{medium} + GC_s$ we obtain an input power of 60 μW per laser for the first overtone band. Small microdisk lasers with a diameter of 7.5 um heterogeneously integrated on SOI can emit up to 120 μW [5] showing that this input power poses no problems. Furthermore the power of the individual lasers can be adjusted in accordance with the water absorption.

Conclusions:

The stringent power budget requirements for a glucose detection scheme using absorption spectroscopy show that a broadband source like a SLED is not suitable for integration of a glucose sensor on SOI. An alternative measurement set-up using a heterogeneously integrated laser array is proposed.

Acknowledgements:

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