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Effects of Different Hardware and Measurement Parameters on Diffuse Correlation Spectroscopy

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Abstract: The precision of diffuse correlation spectroscopy (DCS) measurements was evaluated for varying instrumental and experimental parameters to provide recipes for making practical choices when designing devices.

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1. Introduction

Diffuse correlation spectroscopy (DCS) is an emerging near-infrared diffuse optical technology capable of non-invasively monitoring tissue hemodynamics. It allows to recover the microvascular blood flow of the tissue investigated by measuring the decay of the speckle intensity autocorrelation function due to light scattering by moving particles (i.e. red blood cells) [1]. Applications of diffuse correlation spectroscopy in clinical and pre-clinical environment nowadays spread among human brain, preterm infants neuromonitoring, tumors diagnosis and therapy monitoring, muscle function, spinal cord, bones, and intra-operative monitoring. All these different applications have different requirements as to the allowed compromises in precision, accuracy and temporal resolution [1]. These compromises are often directed by physiological parameters (e.g. the dynamics of the parameter of interest) or hardware limitations (e.g. cost).

In principle, high quality results can be obtained by increasing the detected photon count-rate [2], by increasing the intensity of the laser light injected in the tissue or increasing the measurement duration. These are strictly limited for many clinical scenarios for a variety of reasons. For example, low count-rates are common when measuring highly absorbing tissues such as the human thyroid, or when investigating deep tissues (since a probe with a large source detector fibers separation is needed). The source intensity is limited due to safety reasons and is further limited in some scenarios such as the measurements on preterm born infants due to their sensitive skin and underdeveloped eye reflexes. Long averaging times are afflicted by the instability of the tissue physiology and/or are limited due to the need to measure transient signals.

From an hardware point of view, another way to increase the precision of DCS acquisitions is by adding a number of independent detectors [3] probing different speckles in the very same region of the tissue. The improvement in the signal-to-noise by this method follows the common "squared-root of the number of independent measurements" trend which implies a rapid increase in the cost and complexity of the device.

As part of the LUCA project, which is an international, multi-disciplinary project (LUCA[1] - Laser and Ultrasound Co-Analyzer for thyroid nodules), we have investigated these parameters to design a system suitable for the highly absorbing thyroid tissues. We stuck to the "traditional" DCS approach and considered different aspects of the problem. The output is a study allowing a potential recipe for defining *a priori* device and experimental settings that optimize the precision of DCS measurements, balancing costs (that is, the number of the detectors necessary) and temporal resolution.

2. Experiments and discussion

Experiments on liquid phantoms have been performed by using a custom DCS system consisting of 16 single photon detectors (SPCM-AQ4C, Excelitas, Germany), a 16-channel hardware correlator (HemoPhotonics, Spain)

[1]http://www.luca-project.eu. Acknowledgments: This project has received funding from the European Union's Horizon 2020 research and innovation programme under grant agreement N. 688303.

and a 785 *nm* wavelength continuous wave single longitudinal mode laser. The liquid phantom consisted of a solution of water and Lipofundin20% (B.Braun Melsungen, Germany), with the concentration of Lipofundin20% chosen in order to obtain a reduced scattering coefficient (μ_s') of 5 cm^{-1} and water absorption [4]. Light has been injected in the phantom by a 400μm core fiber and the diffused reflectance was collected at a distance of 2.5 *cm* by a bundle of 16 single mode fibers.

We have acquired the intensity autocorrelation curves (g_2) every second, continuously for approximately 100 *s*, for all the acquisition channels, at different detection count-rates (form 5 to 150 kHz), obtained by attenuating the injected light intensity. The data have been analyzed by calculating, for every count-rate considered, the coefficient of variation of the fitted scatterer Brownian diffusion coefficient D_B, defined as $CV = \sigma_{D_B}/\overline{D}_B$ where σ_{D_B} is the standard deviation and $\overline{D}_B$ is the average value, considering different measurement duration times and different numbers of detection channels. The results obtained have been compared with the theoretical model of DCS noise developed by Zhou et al. [2], which states that the precision of the DCS acquisitions scales as $N^{-\alpha}$, with $\alpha = 1/2$ when N is the number of the detection channels considered or the measurement duration time, and $\alpha = 1$ when N is the intensity of the detected signal.

Examples of the results obtained are reported in figure 1, where we plotted the dependence of the CV on the instrumental and experimental parameters considered in loglog scale, to highlight the comparison with the theoretical model (red dashed line, only the slope is indicative for comparing with the experimental results). In panel (a) is reported explicitly the dependence of the CV on the number of detection channels, at different measurement duration times and at fixed count-rate (20 kHz), in panel (b) the dependence of the CV on the measurement duration time, at different count-rates and at fixed number of detection channels (8) and in panel (c) the dependence of the CV on the count-rate, at different number of detected channels, and at fixed measurement duration (1 *s*). The results retrieved qualitatively reproduce the power law dependence predicted by the theory, having measured exponents α in the range $0.8 \div 1$ for the dependence on the photon count-rate (theory prediction $\alpha = 1$), and in the range $0.25 \div 0.5$ when considering the dependence on the number of detection channels and measurement duration time (theory prediction $\alpha = 0.5$). The underestimation of α measured in some cases has to be attributed to not optimal experimental and environmental conditions, comparable to standard clinical measurement conditions.

In conclusion, this study represents a recipe for *a priori* hardware design (i. e. number of detection channels) and optimizing measurement settings such as the measurement duration time (i. e. the temporal resolution), for different conditions implying different detected photon count-rates. We will discuss different approaches to further improve these considerations.

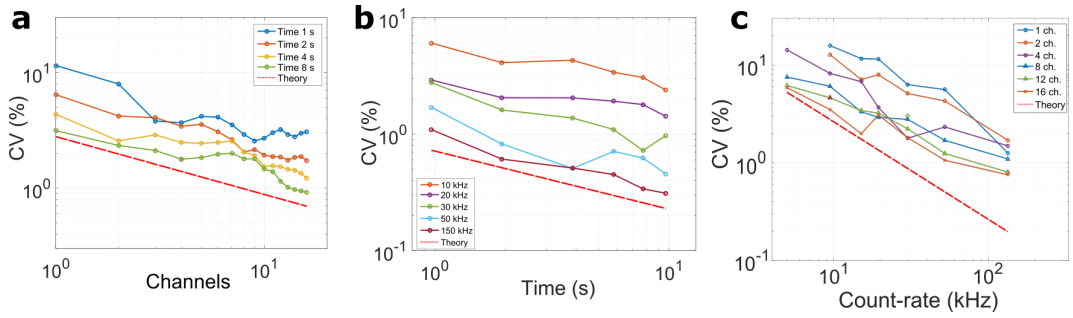

Fig. 1. Examples of calculated CV at (a) fixed count-rate (20 kHz) for varying number of detection channels and at different measurement duration time, (b) fixed number of detection channels (8) for varying measurement duration time and at different detected count-rate, and (c) fixed measurement duration time (1 *s*) for varying detected count-rate and with different number of detection channels. Red dashed lines represent the expectation from the theory (only the slope is indicative).

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