A Lensless Self-mixing Blood-Flow Sensor

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Abstract—We describe a novel layout for the measurement of the extracorporeal blood flow, performed by means of a laser diode placed directly in front of the blood vessel, without the need for any optical element. As the readout, we use the self-mixing interference in the laser diode due to light scattered by the red cells, whose Doppler shifted components comes from the intrinsic divergence of the laser diode. First we outline the sensor working principle, and then show that simulations match the measurements very well.

I. INTRODUCTION

Blood flow monitoring is a critical issue for checking working condition of extracorporeal blood circulator systems. The technique, most frequently employed in commercial instruments, is based on ultrasonic measurements [1]. Though the possibility of realizing an optical sensor for measuring blood flow is well-known in literature [2-6], no commercial tools are available, due to different reasons including cost. This work describes a novel optical approach for blood flow measurement, an evolution of the system presented in Ref.[7]. The working principle is based on a self-mixing interferometer [8], realized with low-cost components and managed by a dedicated data processing system. The novelty is represented by the total absence of optical components, such as lenses, and therefore no need for alignment. The Doppler signal is induced by the natural divergence of the laser beam. In order to better understand the system reliability and the sensitivity to mechanical positioning, the system has been numerically simulated and compared with some experimental results.

II. INTERFEROMETER SETUP

Back-reflection of light into the cavity of a laser diode (LD) induces amplitude and frequency modulation of the oscillating field [6]. Self-mixing interferometry (SMI) [8] takes advantage of this feedback effect in a single mode laser, in a simple optical configuration of measurement. The induced amplitude modulation depends on the phase $\phi = 2ks$ of the back-reflected optical field, and hence on the remote target position. The power emitted by the LD subject to optical feedback can be written as:

$$P(\phi) = P_0 [1 + m \cdot F(\phi)],$$

where $P_0$ is the power emitted by the unperturbed laser, $m$ is a modulation index and $F(\phi)$ is a periodic function of the interferometric phase $\phi = 2ks$ of period $2\pi$, where $k = 2\pi/\lambda$ (being $\lambda$ the emission wavelength), and $s$ is the distance between the laser diode and the remote target. The modulation index $m$ and the shape of the function $F(\phi)$ both depend on the so-called feedback parameter $C$ see Ref.[6,8] for more details.

The amplitude modulation signal is typically detected by the monitor photodiode, built-in the laser package, but it can be also detected by an external photodiode, at any position in front of the laser beam. The back reflection of a diffusing target induces a modulation index $m$ around $10^{-3}$ for a collimated beam. With such a reflection level, the self-mixing signal allows measuring displacement [9,10], absolute distances [11,12] or vibrations [13].

In order to measure a blood flow inside a duct, made of a polycarbonate, we propose the very simple optical setup as shown in Fig. 1. In this configuration, working without any collecting optics, the amplitude of optical back-reflection is particularly small, and therefore the system works in the so-called very-weak back-injection regime, where the signal distortion is almost negligible [7].

![Figure 1. Layout of the optical instrument](image)

The main novelty of the proposed setup is the absence of any external optical component: the laser diode is simply placed very close to the tube. The Doppler signal is achieved through the natural divergence of the laser diode, about 20 deg. When the liquid is moving at a speed $v$, the back-diffused light exhibits different contributions of Doppler shift, depending on the angle $\alpha$ between the laser beam and the speed direction. Each contribution is equal to

$$f_{\text{Doppler}} = \frac{2}{\lambda} \cdot v \cdot \cos \alpha \cdot$$

(2)
These Doppler contributions are summed up by the self-mixing interferometer and measured as an amplitude modulation of the laser power directly by the monitor photodiode. For this particular configuration, the resulting signal is not single tone, but a distributed spectrum resulting from the integral over different frequency contributions due to the local particle speed and angle $\alpha$.

**III. NUMERICAL SIMULATION**

In order to better describe the system behavior and estimate the mechanical tolerances sensitivity, a numerical simulator has been implemented. The mathematical model follows the one described in [14], but the approximation of paraxial beam is no more applicable in our case, because the Doppler signal is induced just by the laser beam divergence. Fig. 2 shows the coordinate system and the approximation of parabolic velocity distribution inside the duct.

![System schematization and coordinate axes.](image)

The model allows us to calculate the signal read by the monitor photodiode. At first approximation, the signal induced by the scattering of a single particle is a Dirac-delta function at frequency $\omega_D$, depending on its position and speed (Eq.2). The whole signal spectral density is the integral of all the contributions and, with the constant particles density approximation inside the fluid, it is expressed as:

$$S(\omega) \approx \int \int \delta(\omega_{D(\ell,y,z)} - \omega) \cdot \eta(\ell,y,z) \cdot I(\ell,y,z) \, dx \, dy \, dz$$

where $I$ is the relative laser intensity, modeled as a Gaussian beam, and $\eta$ is the single particle feedback fraction, calculated as described in [14], equal to the portion of light back-scattered into laser cavity. For the numerical simulation Eq.3 was rewritten in cylindrical coordinates as:

$$S(\omega) \approx \int_0^{R} \int_0^{2\pi} \int_0^L \delta(\omega_{D(\ell,y,z)} - \omega) \cdot \eta(\ell,y,z) \cdot I(\ell,y,z) \, dr \, d\ell \, dz$$

The Doppler frequency $\omega_D$ is:

$$\omega_{D(\ell,y,z)} = \frac{2\pi \cdot v(\ell,y,z) \cdot \cos(\ell,y,z)}{\lambda/2}$$

and the fluid speed can be described as a laminar flow:

$$v(\ell,y,z) = v_0 \left[ 1 - \left( \frac{R}{R_0} \right)^2 \right]$$

The integral in Eq.4 has been numerically computed with some approximations. The most critical is the discretization step: along $z$ the number of point should be large enough to finely sample the laser Gaussian shape, but the least possible to limit the computational cost. The final choice was 1020 point along $z$ and 180 points along $x$ and $y$, a good trade-off between accuracy and computation time.

Fig. 3 shows an example of simulated spectra, for different flow speed, ranging between 0.1 l/min and 0.6 l/min in a duct with diameter 5 mm. By means of the simulation tool it is possible to evaluate the mechanical misalignments and blood parameters sensitivity, especially the hematocrit level. The spectral shape strongly depends on the attenuation coefficient, an hematocrit level function.

![Simulated spectra, for blood flow ranging between 0.1 l/min and 0.6 l/min.](image)

**IV. ACQUIRED SPECTRA**

The optical probe has been realized as shown in Fig. 1. The laser wavelength is 780 nm and the emitted laser power is about 20 mW. The signal acquisition and elaboration has been implemented on a DSP, with a sampling frequency equal to 6 MHz. In 1 s the system acquires and elaborates 1000 FFT over 1024 samples. The output average over 1000 spectra can be read on a personal computer. Fig. 4 shows an example of measurements made on a duct with 9 mm diameter, with flows ranging between 1 l/min and 6 l/min. The liquid is human blood, with level 35% hematocrit. In the DSP elaboration the measurement floor has been clamped to a constant value, higher than noise and disturbances, in order to remove spurious peaks. The experimental measurements confirm the simulation results and the effectiveness of the proposed optical architecture. Fig. 5 shows a comparison of a measured and a simulated spectrum, for hematocrit 35%, duct diameter 9 mm and blood flow 4 l/min. Noise floor is also indicated.
V. SIGNAL ELABORATION

The goal of the signal spectrum elaboration is to calculate the flow value. To achieve it, the first step consists in finding the mean frequency $\bar{f}$, as a weighted average of the power spectrum $S(f)$, using the logarithm of $S(f)$ as weights:

$$\bar{f} = \frac{\int_{f_{\text{noise}}}^{f_{\text{max}}} \log(S(f)) \cdot f \cdot df}{\int_{f_{\text{noise}}}^{f_{\text{max}}} \log(S(f)) df} = \frac{\sum_{n=f_{\text{noise}}}^{f_{\text{max}}} \log(S(f)) \cdot f}{\sum_{n=f_{\text{noise}}}^{f_{\text{max}}} \log(S(f))}$$

The integral upper limit is the frequency $f_{\text{noise}}$, corresponding to the intercept between the signal spectrum and the noise floor. The logarithmic weight in (6) enhances the system sensitivity, because the high-frequency components are more significant for the speed measurement. The drawback of this procedure is the integral upper limit needed accuracy: the higher than the noise floor. The noise floor intercept is obtained through a linear approximation for $\log(S(f))$, after several averages. The simulation tool predicts a linear relation between the blood flow and the calculated mean frequency $\bar{f}$. The linearity has been also verified by different experimental sessions, referenced to calibrated commercial instruments. After the calibration, a series of experiments was performed on human blood, with different hematocrit levels, yielding an accuracy better than 0.1 l/min over the whole range of clinic interest (0-6 l/min).

Finally, the simulation tool confirmed the sensor linearity even in the case of parameter variations, and a very-low system sensitivity to mechanical misalignments, such as errors in laser positioning: distance, inclination or rotation.

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REFERENCES


