Compact Laser-Diode Instrument for Flow Measurement

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***Abstract*— We demonstrate a flow-meter sensor using a bare laser diode (LD) as a self-mixing interferometer and no additional lens nor optical component. The diode is placed directly on the edge of the duct, exploiting the natural laser divergence. The flow measurement is based on the Doppler shift induced by the scattering particles inside the fluid and read by the self-mixing effect inside the LD. The proposed approach has been modeled by a numerical simulator in order to find the best signal processing algorithm. A set of measurements on blood proves good performances as a flow meter, while simulations confirm the system robustness to mechanical misalignments and changes in liquid properties.**

***Index Terms*— Fluid flow measurement, laser velocimetry, optical interferometry, optical sensors, optoelectronic and photonic sensors.**

I. Introduction

In THE literature [1], a number of different techniques are proposed for flow measurement and monitoring. The simplest is based on pressure measurements. With regard to the measurement without contact, the ultrasonic technique is widely employed [2]. An alternative approach is offered by the optical methods, such as particle image velocimetry [3], laser Doppler anemometry [4], and also with variations for high-speed flow [5]. These techniques are extensively used for biomedical applications, such as blood-flow measurement [6]–[9]. For high-speed applications, all these approaches are rather complex and expensive. The aim of this paper is to develop a novel optical instrument, one working well and with the simplest possible setup. The working principle we propose is an evolution of the system presented in [10], based on a self-mixing interferometer [11], [12]. The total absence of any optical components, such as lenses, represents the main system novelty, alongside the self-alignment. The Doppler signal is induced by the natural divergence of the laser beam and is managed by a dedicated data processing system. To assess sensitivity to mechanical positioning, the system has been studied by numerical simulations and successfully compared with experimental results. After the first application to blood-flow measurement, several other fluids of industrial interest (specifically milk emulsion and lubricating oil) are under evaluation as areas of application.

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II. Optical Setup

The proposed sensor is based on self-mixing interferometry: a laser diode (LD), subjected to back-reflection of light, suffers an amplitude and frequency modulation of the oscillating field [12]. Self-mixing interferometry takes advantage of this feedback effect in a simple measurement configuration. The induced amplitude modulation depends on the phase $\phi = 2\underline{k}\underline{s} = 2\ ks\cos\theta$ of the back-reflected optical field and, hence, on the remote target position, where $\underline{k}$ is the wavevector (and $k = 2\pi/\lambda$ with λ the wavelength), $\underline{s}$ is the displacement vector, and θ is the angle between directions of wavevector $\underline{k}$ and displacement $\underline{s}$. The power emitted by the LD, because of the optical feedback, is modulated by $F(\phi)$, a periodic function of the interferometric phase ϕ. The modulation index m and the waveform of $F(\phi)$ both depend on the feedback parameter C [11], [12], and for weak feedback, ($C \ll 1$) $F(\phi)$ becomes the usual interferometric function $\cos(\phi)$ of optical pathlength phase ϕ.

The amplitude modulation signal is detected by the built-in (rear) monitor photodiode, but it can also be detected by an external photodiode, placed at any position, for example, in front of the laser beam. The back reflection from a diffusing target induces a modulation index m around 10^{-3} for a collimated beam. With such a reflection level, the self-mixing signal is suitable for measuring displacement [13]–[16], absolute distances [17]–[19], and vibrations [20], [21].

In order to measure liquid flow inside a duct, we propose the simple optical setup shown in Fig. 1, a preliminary account of which was presented in [22].

Laser configurations described in the literature [23]–[29] are based on sending a collimated beam into the medium and looking to the flow from a fixed angle θ, so as to develop a Doppler frequency written as

$$f_{\text{Doppler}} = \frac{2}{\lambda} \cdot v \cdot \cos\theta. \qquad (1)$$

Our hint is that a fixed θ is not actually necessary, if we just need to measure the average flux speed.

Thus, we use a bare LD placed very close to the wall of the flow-carrying tube, as shown in Fig. 1, and let the flow be illuminated with a spread of θ angles. Then, if the liquid moves

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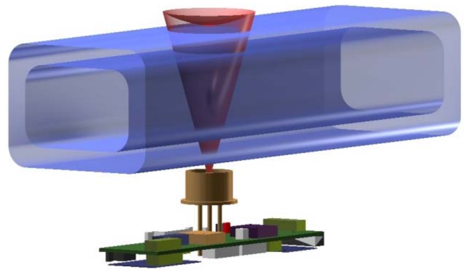

Fig. 1. Setup of the optical instrument.

at speed v, the back-diffused light will contain a distribution of Doppler shift contributions, each weighted on the cosine of angle θ, as indicated by (1). However, we will show how to get rid of this dependence and obtain v.

Thus, with the configuration of Fig. 1, we no longer need the collimating optics necessary to get a defined θ, and also eliminate alignment constraints and associated mechanical mountings.

The self-mixing interferometer sums up the contributions containing the individual Doppler frequencies (1) linearly, introducing an amplitude modulation of the laser power, conveniently measured by the monitor photodiode already in place in the LD package. For this particular configuration, the resulting signal is a broadened spectrum resulting from the summation of contributions at different frequencies due to the local particle speed and to angle θ.

In our lensless configuration, the optical backscattering returning to the laser is very small (typ. 10^{-6} in power), and therefore, the system works in the very-weak back-injection regime ($C \ll 1$), where the $F(\phi) = \cos(\phi)$, so that the harmonic distortion of the Doppler signal is negligible [10].

III. Experimental Measurements

The optical probe realized (Fig. 1) employs an LD emitting at $\lambda = 780$-nm, biased to supply a continuous power of ~20 mW, and with a beam divergence of $\theta_Y = 8°$ and $\theta_Z = 22°$ along the major and minor spot axes. The z-axis is taken parallel to the flow, so divergence (full angle half-width) of the beam shown in Figs. 1 and 4 is 22°.

To process data in real time with reasonable resolution and averaging, it turns out that we need to acquire and process 1000 FFT over 1024 samples in 1 s: for this rather heavy computation task, we need a DSP or an high-speed microcontroller (e.g., the STM32F4).

The prototype we have realized employs a DSP (model TMS320F28044) with a sampling frequency of 6 MHz and a resolution of 12 b. The signal output, averaged over 1000 spectra, can be elaborated directly by the DSP, or transmitted to a personal computer.

The first measurement with our flow meter was on blood flow in extracorporeal circuits, one very useful for dialysis machine control. Important to note, because of the large beam divergence, local heating of blood flow is negligible, so the system is safe.

All the experimental data collected have shown a strong dependence of the signal frequency content on the flow speed,

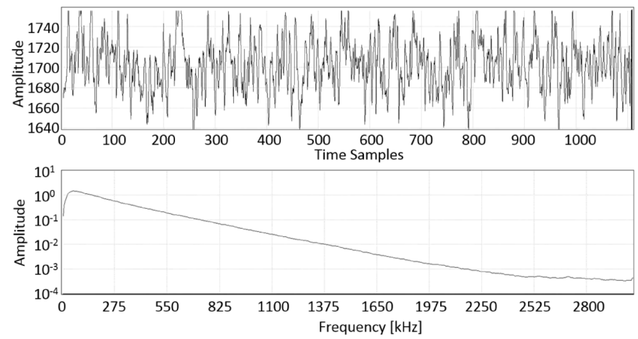

Fig. 2. Experimental signal output from photodiode (top), and resulting spectrum computed by the DSP (bottom), for a blood-flow speed of 4 L/min. Amplitudes are reported in levels of the 12-b converter.

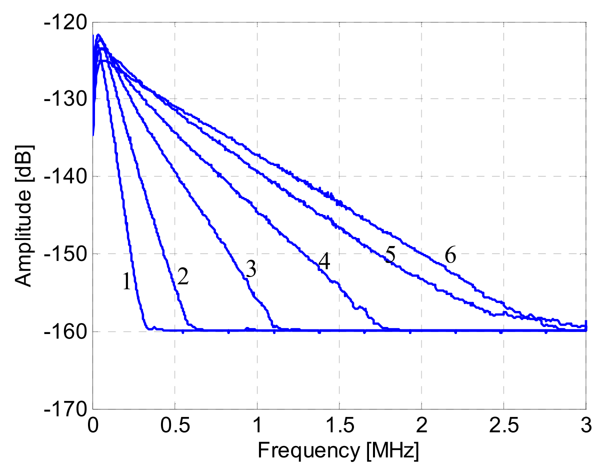

Fig. 3. Measured spectra for blood flow of 1–6 L/min.

and a good signal-to-noise ratio, also for fluid with moderate concentration of scattering particles. Fig. 2 shows an example of measurements made on a duct with 9-mm diameter, with flow speed equal to 4 L/min.

In Fig. 2, the top diagram is the sampled (self-mixing) signal of the photodiode output, and the lower curve is the frequency spectrum computed by the DSP, after averaging on 1000 FFTs of the sampled signal and filtering by a Savitzky–Golay filter [30]. All functions are implemented by the DSP in real time.

In the experiment, the liquid was human blood with 35% hematocrit. The signal in the time domain looks rather noisy, whereas in the frequency domain [Fig. 2 (bottom)], it exhibits a smooth, well-defined trend, function of the flow speed. In the DSP routines, a threshold in amplitude was introduced, set at level a little bit higher than noise and disturbances, in order to remove spurious peaks. Fig. 3 shows the signal spectra evolution with measured flow ranging from 1 to 6 L/min.

As can be seen, the signal spectrum is well correlated with the flow speed. In Section IV, we introduce the proposed signal processing to extract the velocity.

IV. System Simulation

In order to understand the system behavior, estimate sensitivity to mechanical tolerances, and devise an appropriate signal processing, we have developed a numerical simulator.

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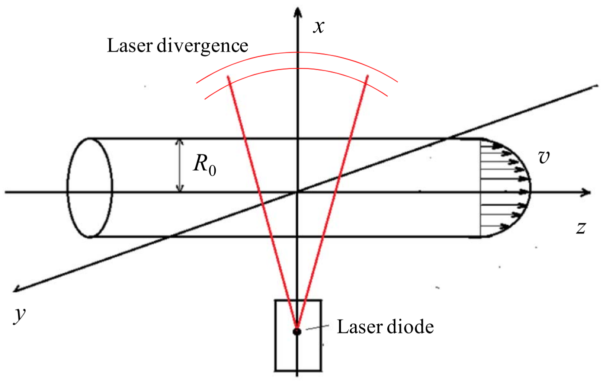

Fig. 4. System schematization and coordinate axes.

The mathematical model follows the one described in [26], but dropping the first-order approximation of paraxial rays (i.e., no angular divergence). Indeed, in our new setup, we shall account for the large spread of Doppler frequency due to the laser beam divergence. The coordinate system is chosen, as shown in Fig. 4, and reasonably, we assumed a laminar regime for the flow, so that the radial velocity distribution is parabolic. The approximation of laminar flow is often valid for small tubes, up to very-high flow. For example, for a 9-mm diameter, the Reynolds number is evaluated equal to 2000 (limit for laminar flow) for a blood flow of ~4 L/min.

The aim of the model is to calculate the signal read by the monitor photodiode, in correspondence to a laminar fluid motion. The signal induced by the scattering of a single particle is a Dirac-delta function at frequency ω_D, depending on its position and speed (1). By assuming a constant density of particles inside the duct, the total signal spectral density is the integral of contributions summed on the elemental volumes dV, and it can be expressed as

$$S(\omega) \propto \int_0^{R_0}\int_0^{2\pi}\int_0^{\infty} \delta(\omega_{D(r,\theta,z)} - \omega) \cdot \eta_{(r,\theta,z)} \cdot I_{(r,\theta,z)} r\,dr\,d\theta\,dz \tag{2}$$

where I is the relative laser intensity, modeled as a Gaussian beam, and η is the single particle feedback fraction, equal to the portion of light back-scattered into laser cavity and calculated as in [25], and $dV = r\,dr\,d\theta dz$ in polar coordinates.

The quantity $\omega_D/2\pi$ is the Doppler frequency expressed as

$$\omega_{D(r,\theta,z)} = \frac{2\pi \cdot v_{(r,\theta,z)} \cdot \cos(\alpha_{(r,\theta,z)})}{\lambda/2}. \tag{3}$$

Assuming a laminar flow, the fluid velocity is a parabolic function of the radial position r on the duct radius R_0, considering the speed v_0 at the duct center

$$v_{(r,\theta,z)} = v_0\left[1 - \left(\frac{r}{R_0}\right)^2\right]. \tag{4}$$

The integral in (5) was computed numerically with a discretization step limited by the computational cost, but large enough to finely sample the laser Gaussian shape. The final choice was 1020 point along z and 180 points along ρ and θ, a good tradeoff between accuracy and computation time.

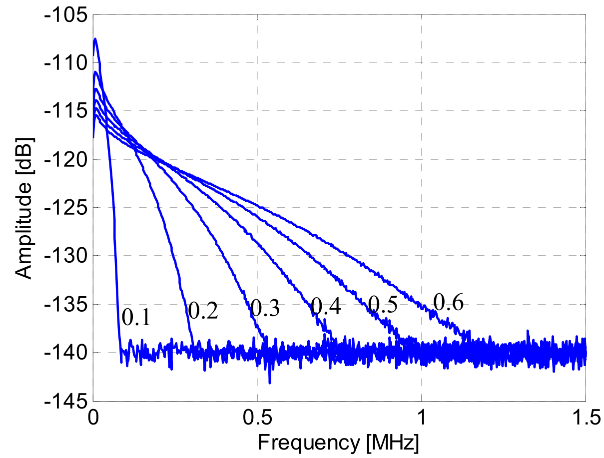

Fig. 5. Simulated spectra for blood flow ranging between 0.1 and 0.6 L/min.

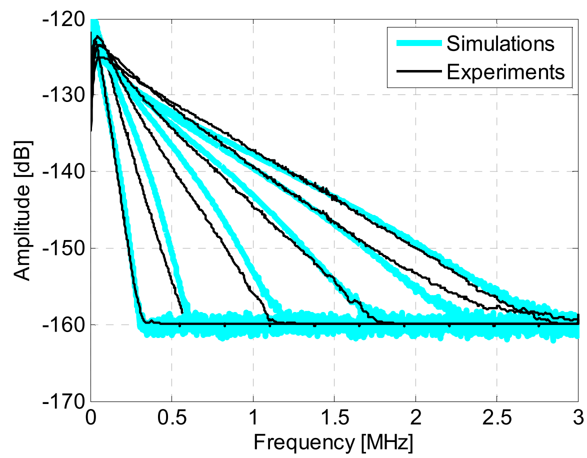

Fig. 6. Comparison of measured and simulated spectrum for blood flow of 1–6 L/min.

Fig. 5 shows an example of simulated spectra, for different flow speeds, ranging between 0.1 and 0.6 L/min in a duct of 5-mm diameter. By means of the simulation tool, it is possible to evaluate the errors due to mechanical misalignments and the sensitivity to parameters of the fluid. The simulation tool is also useful for studying the best signal processing procedure, in order to limit the dependence on the scattering parameter of the considered liquid. For example, the spectrum diagram may be dependent on the attenuation coefficient (for the blood, a function of the hematocrit level), and we wish the estimate of flow speed be insensitive to that parameter.

In order to test the simulator, in the calculations, we introduced the same parameters of the experimental measurements, and obtained a very good agreement: the only parameter we need to fit was the amplitude level. Fig. 6 shows a comparison of measured and simulated spectra for human blood with hematocrit 35%, duct diameter 9 mm, and blood flow between 1 and 6 L/min (the same data of Fig. 3). The simulated noise floor was added to be comparable with the experimental one.

V. Signal Processing

In the literature [6], the simplest data processing proposed for laser Doppler flow measurement, in view if the $\cos\theta$ dependence of the Doppler signal (1), is that of estimating

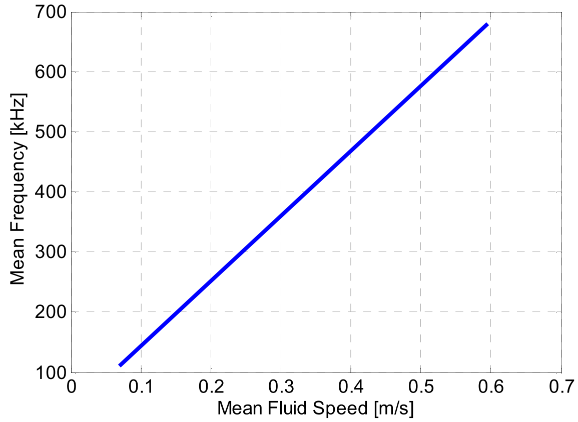

Fig. 7. Simulation results: linear dependence of the mean frequency, calculated by (6), from the fluid speed.

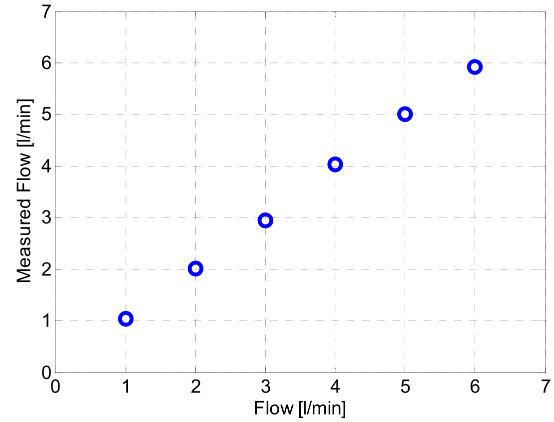

Fig. 8. Experimental measurements of blood flow.

the highest frequency content of the detected signal, i.e., the frequency at which the tail of the spectrum is lost in noise, and using it to compute the flow velocity from (1). This approach is satisfactory but does not exploit the entire information content of the signal and, therefore, does not attain the best accuracy.

The algorithm with best performances we have found consists in calculating the mean frequency $\bar{f}$ as an average of the power spectrum $S(f)$, weighted on the logarithm of $S(f)$

$$\bar{f} = \frac{\int_0^{f_{\text{noise}}} \text{Log}(S(f)) \cdot f \cdot df}{\int_0^{f_{\text{noise}}} \text{Log}(S(f)) df} \cong \frac{\sum_0^{f_{\text{noise}}} \text{Log}(S(f)) \cdot f}{\sum_0^{f_{\text{noise}}} \text{Log}(S(f))}. \tag{5}$$

In (6), $S(f)$ is the spectral density normalized to the minimum value read in the full spectrum, that is $S_{\min}(f)$.

The upper limit of the integral is the frequency f_{noise} corresponding to the intercept between the signal spectrum and the noise floor. The logarithmic weight enhances the system sensitivity, because the high-frequency components are more significant for the speed measurement. The simulation tool supplies a linear relation between the flow speed and the calculated mean frequency $\bar{f}$, as shown in Fig. 7.

The drawback is the accuracy in finding the integral upper limit: the power spectrum should only be summed up until the signal reaches the noise floor. The noise floor intercept is obtained through a linear approximation for $\log(S(f))$, after 1000 averages and Savitzky–Golay filtering [30].

The sensor calibration was implemented on measured data after some experimental sessions referenced to calibrated commercial instruments. The experimental measurements confirmed the linear dependence of the flow speed with the mean frequency $\bar{f}$, as predicted by the simulations. Fig. 8 shows the result of flow measurement obtained by the acquisitions shown in Fig. 3. The maximum error with respect to the ideally linear curve is <0.1 L/min in the range between 1 and 6 L/min.

VI. System Sensitivity to Misalignments

For industrial application, it is mandatory to reach a good robustness against mechanical misalignments and mounting tolerances. Therefore, simulations were carried out to evaluate sensor sensitivity to positioning errors. The first parameter is the angular alignment of the laser, i.e., the deviation respect to correct direction, perpendicular to the tube. Fig. 9 shows the simulated spectra as a function of the laser inclination between 0° and 5°.

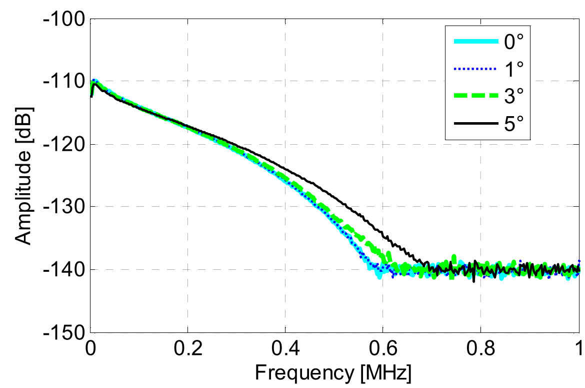

Fig. 9. Simulated spectra as a function of the laser inclination angle.

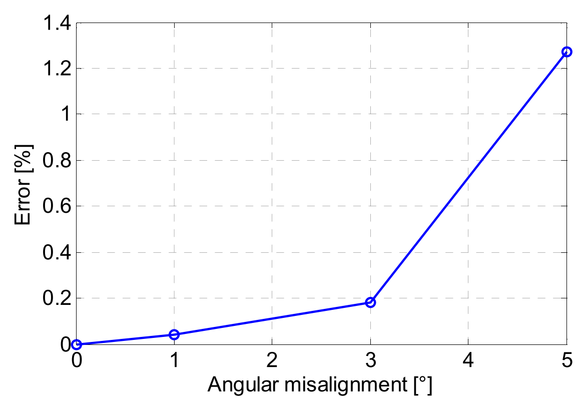

Fig. 10. Error in the flow measurement due to an angular misalignment of the LD.

The spectra are very similar for small angles, and the error in the flow measurement is limited to about 1% for misalignment up to 5°, as shown in Fig. 10.

The second error considered is the angular rotation of the LD with respect to its axis. It can induce an error because of the ellipticity of the LD emission spot and different values of θ_Y and θ_Z. Fig. 11 shows some simulated spectra for rotation

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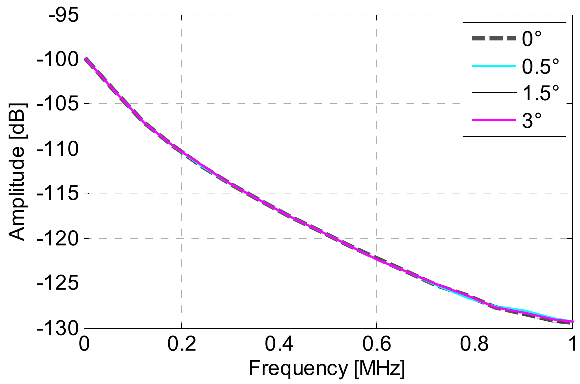

Fig. 11. Simulated spectra as a function of the laser rotation.

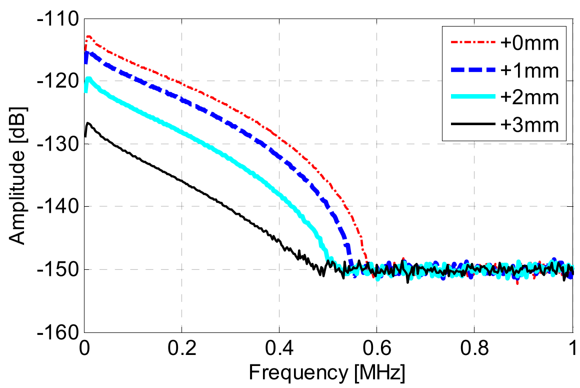

Fig. 12. Simulated spectra as a function of the laser-tube distance.

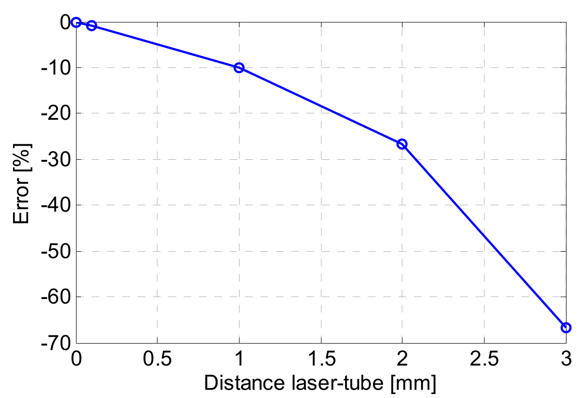

Fig. 13. Error in the flow measurement due to the laser-tube distance.

up to 3°. As can be seen, the relative error induced by 3° of rotation is in the order of 10^{-5}, and therefore, the influence on the flow measurement is negligible.

Last, we considered the positioning error of the LD, that is, the distance of the laser optical window from the tube. The normal operation corresponds to the laser package placed directly in contact with the tube. Fig. 12 shows the variation of the self-mixing spectrum for different laser-tube distances (the zero is when laser window and tube external surface are in contact).

For this parameter, the error is no more negligible for distance variation higher than 1 mm, as shown in Fig. 13, and as the error is negative, the flow is underestimated.

For having error lower than 1% with respect to the contact position, the maximum allowed distance variation between laser package and tube should be 0.1 mm. This value is certainly satisfied with standard mechanical mounting, and therefore, we can conclude that the proposed sensor can be produced in series without the need for a very precise and high-cost mechanical mounts.

VII. Results and Conclusion

A simple flow meter based on a bare LD with no optics has been developed and tested. After calibration, measurements performed at different hematocrit levels have shown good accuracy, better than 0.1 L/min in the range 0–6 L/min of clinical interest. By means of a simulation tool, we have compared different signal processing schemes, and found that the logarithm-weighted mean optimizes sensor linearity even in the case of parameter variations and minimizes sensitivity to mechanical misalignments. The simulations also indicate a good robustness to errors, such as distance, inclination, and rotation.

The flow meter can also be applied to all liquids exhibiting some optical scattering (whereas it will not operate well on pure water). For example, a system dedicated to milk flow measurement and to lubricant oil is under development. The strengths of the proposed approach are: direct application on the duct (if transparent), no need for dedicated cuvettes, no optical alignment, low-cost, if compared with ultrasonic sensors, and good robustness to error in the mechanical positioning. The main limitations of the method are as follows.

1) It is a measurement inherently limited to the outer part of the flow (about 1 mm), the best performance being for ducts between 2 and 9 mm in diameter.
2) The flow should be laminar; therefore, curves or folds just before the measurement point must be avoided.

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