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Item Type	Meetings and Proceedings
Authors	Leahy, Martin J.;Liebert, Adam;Maniewski, Roman
Citation	Opto-Ireland 2002: Optics and Photonics Technologies and Applications;4876/ 120-127
Publisher	Society of Photo-Optical Instrumentation Engineers (SPIE)
Download date	2026-05-14 06:38:27
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Link to Item	https://hdl.handle.net/10344/137

EVALUATION OF DIFFERENT SIGNAL PROCESSING ALGORITHMS IN LASER-DOPPLER PERFUSION MEASUREMENTS

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Abstract

The Laser-Doppler (LD) method is extensively used in clinical experiments for microcirculation measurement. The results of LD perfusion measurements depends on technical factors such as laser light wavelength, LD probe arrangement, and signal processing algorithm. The aim of this study is to compare the output of various signal processing algorithms (LDP) with the use of digital spectral analysis of the photodetector current. Comparison is made with the output of a commercial LD instrument (LDF). The results obtained using a simple one-tube physical model confirm that the best linearity of response of the LD instrument for changes of flow velocity is given by the algorithm based on first moment of power spectral density of AC component of photodetector's signal.

1.INTRODUCTION

Laser-Doppler blood perfusion technique has found many clinical applications in the last 20 years [1,12]. It is based on emission of laser light into the tissue and photodetection of remitted light. During interactions of photons with moving red blood cells Doppler scattering occurs. Proper analysis of the photodetected signal gives the information about perfusion, defined as a product of speed and concentration of red blood cells averaged in the sampling volume. The main problems connected with the use of laser-Doppler flowmeters are: relative calibration of the instrument [2] and unknown depth of light penetration in the investigated tissue [3]. It was shown that the LD perfusion signal is proportional to real perfusion of the tissue. The linearity of perfusion characteristics depends on many technical factors. One of most important is the algorithm used for perfusion index calculation [4]. The main goal of this study is to compare various algorithms and compare the linearity of instrument response obtained with different algorithms. The investigations were performed with use of LD flowmeter Oxford Array (Oxford Optronix, U.K.). It allows for 12-channel continuous, non-invasive and real time monitoring of perfusion. The instrument is equipped with two laser diodes operating with wavelength of $\lambda=780\pm 10\text{nm}$ and 12 photodetectors. The instrument is based on PC 486 computer (25Mhz) with 80Mb hard disk. Standard software allows for data acquisition, storage and further analysis. The instrument was modified for our experiments with various data processing algorithms. It allows acquisition of unprocessed AC and DC signals from the photodetectors with simultaneous recording of perfusion signals processed in the instrument using specialised analogue signal processor. The experimental set-up is presented in Fig.1.

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2.LD SIGNAL PROCESSING

The current from photodetector is converted to voltage and amplified. After proper filtering the AC and DC components of the photodetected signal are analysed. The AC component is filtered in the bandwidth of $f_1=10\text{Hz}$ to $f_2=19\text{kHz}$. The DC component is obtained by low-pass filtering with cut-off frequency $f_3=0,5\text{Hz}$.

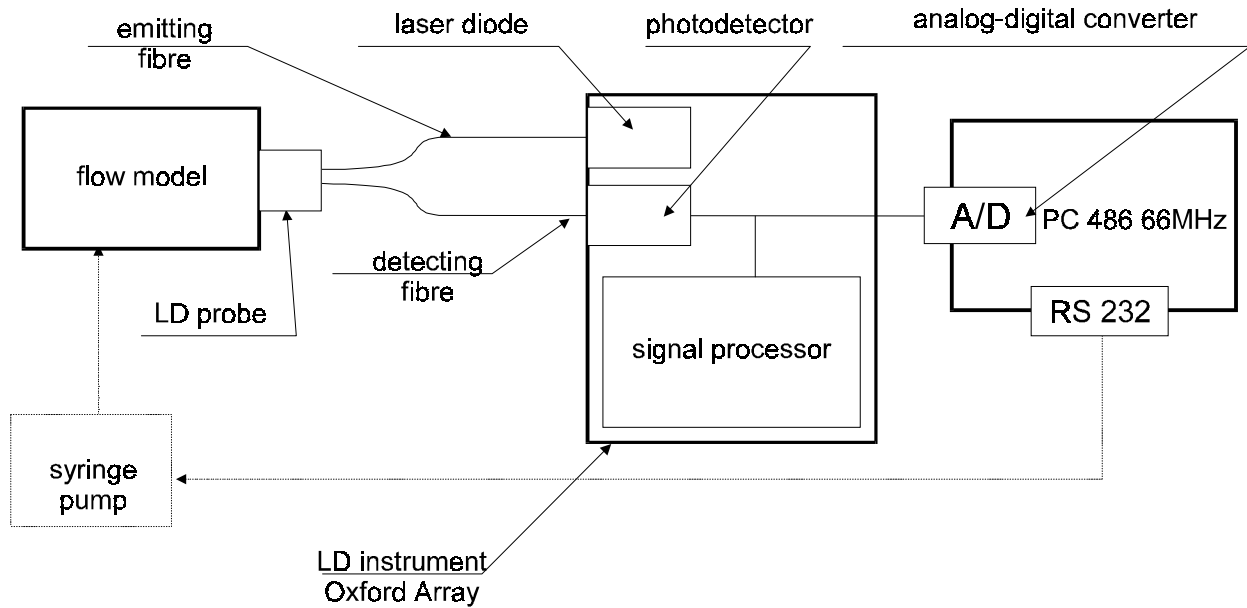


Fig.1. Experimental set-up.

The AC and DC signals are converted to digital form with use of PC486/66MHz computer and two AD boards: PC26AT (Amplicon, UK) for AC conversion with sampling frequency $f_s=40\text{kHz}$; PCL-812 (Advantech Co. Ltd.) for DC conversion with sampling frequency $f_s=2\text{Hz}$. Initial AC signal filter (19kHz) assures that there are no aliasing artefacts during conversion. [4].

For high frequency data acquisition a DMA (direct memory access) channel was used. Analog-digital conversion of the DC signal was performed simultaneously with conversion of AC signal. The data were stored on the hard disk and archived on magneto-optical disks (OB-301, TEAC, Japan). Frequency analysis of the AC signal was applied to estimate the perfusion index. We used Fast Fourier Transformation with data segment length of 4096 giving, 2048 samples of the spectrum with resolution of about 10Hz. To prevent spectral leakage the Blackman Harris window was applied. Finally power spectral density (PSD) function was obtained by averaging of 500 spectra processed every 20ms. A perfusion index was calculated using following formula:

$$LDP = \frac{(ACP - N)}{(DC)^2} CF \quad (1)$$

The ACP represents first moment of the spectrum of AC signal:

$$ACP = \sum_{i=p}^{i=k} f_i \cdot P_{AC}(f_i) \quad (2)$$

where $P_{AC}(f)$ - power spectral density of AC signal.

Values of p and k define the limits of summation ($f_p = 10\text{Hz}$, $f_k = 15\text{kHz}$). Noise N is correlated with DC level and can be estimated experimentally. The perfusion index defined by equation 1 is proportional to the product of averaged concentration and averaged velocity of blood cells:

$$LDP \propto C \cdot V \quad (3)$$

where: C is the averaged concentration of moving blood cells, V is the averaged speed of blood cells in sampling volume.

The C and V signals are obtained using following formulas:

$$C = \frac{CF_C \cdot \sum_{f_1}^{f_2} P_{AC}(f)}{(DC)^2} \quad (4)$$

$$V = \frac{CF_V \cdot \sum_{f_1}^{f_2} f P_{AC}(f)}{\sum_{f_2}^{f_1} P_{AC}(f)} \quad (5)$$

where CF_V and CF_C are calibration factors for speed and concentration, respectively.

Simultaneous recording of the perfusion index LDF , was performed by the commercial LD instrument employing a specialised analogue signal processing unit. The LDF signal was compared with digital signal processing results.

3. VERIFICATION OF ALGORITHMS USING OF FLOW MODEL

Studies by other authors confirmed the linear relationship between perfusion index and flow velocity in various flow models [5] [6] [7] [8] [9]. We constructed very simple one-tube flow model, which consists of one glass tube (OD 1,5mm and ID=1mm) placed between two plates of Plexiglas (Fig.2.).

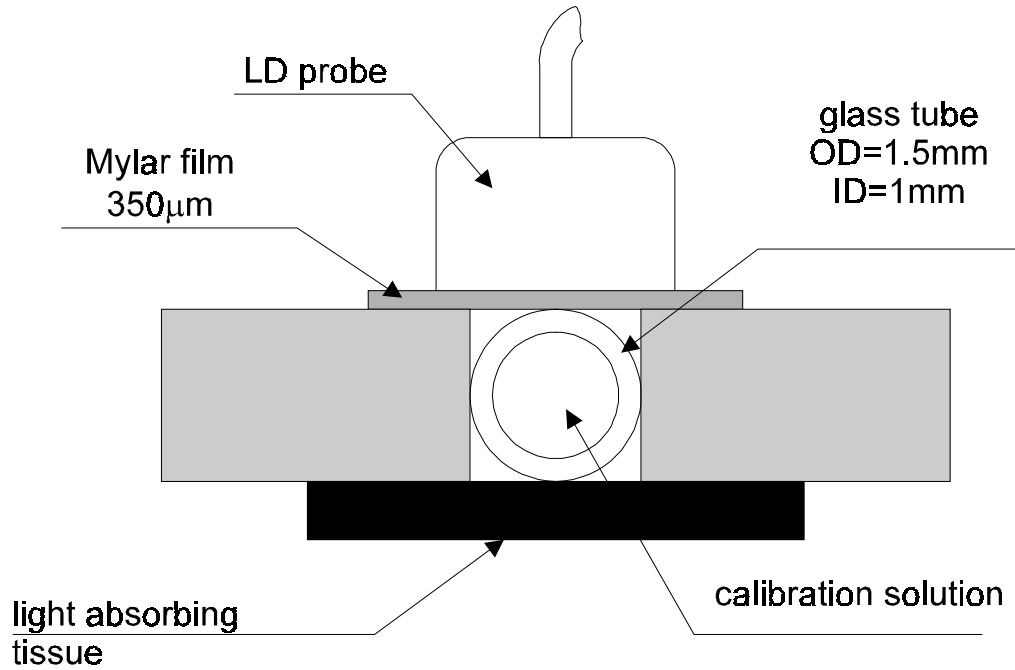


Fig.2. Single-tube flow model

Between the LD probe and the tube a layer of 350µm thick Mylar film was placed. A calibration solution consisting of latex spheres (diameter of $6,4\mu\text{m} \pm 1,9\mu\text{m}$, concentration 0.1% - Seradyn Inc.) in water [10] provides a reproducible scattered, Doppler-shifted signal. The same solution was pumped through tubes with use of syringe pump DUET standard 50 (Kwapisz, Poland). The pump was controlled directly from the computer via an RS-232 interface. The flow was held constant for about 3 minutes before recording of the LD signal to ensure stable conditions. During all experiments the solution was agitated to prevent the influence of sedimentation and aggregation of the spheres. The digital signal processing allowed evaluation of different algorithms for perfusion index calculation. Fig.3 presents changes of flow parameters defined above: LDP , LDF , C and additionally the second moment of PSD_{AC} calculated according to the formula:

$$Mi = \int_{f_1}^{f_2} f^i \cdot P_{AC}(f)df \quad (6)$$

where i - moment order[11].

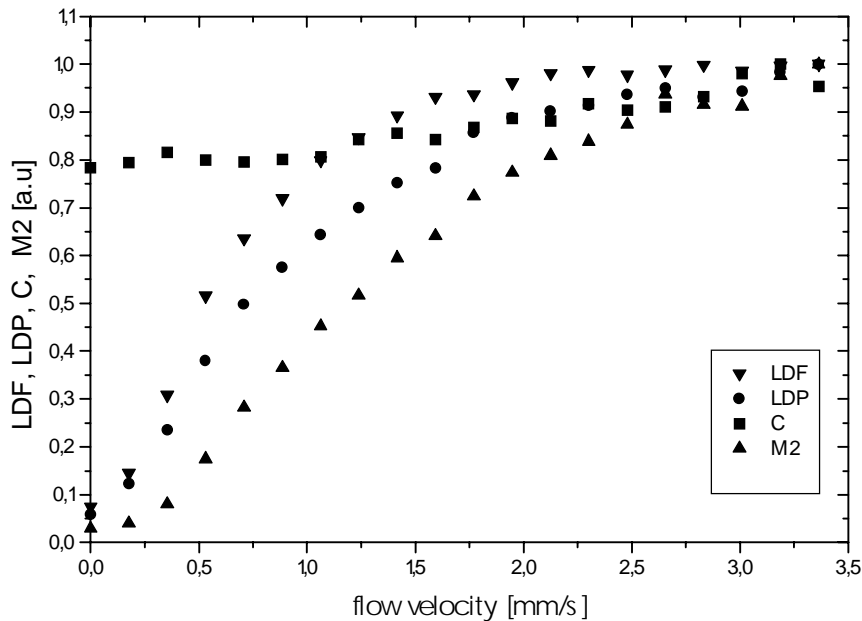


Fig.3. *LDF*, *LDP*, *C*, *M2* vs. flow velocity of the calibration solution in the flow model. The functions shown were normalised by their maximum values.

We observe that *C* does not change with flow velocity due to constant concentration of the solution during this experiment. Parameters *LDF*, *LDP* and *M2* grow with flow velocity but for high flow velocities they underestimate perfusion signal. This is connected with multiple scattering phenomena [12], where multiple Doppler-shifting may exaggerate the actual red blood cell velocity. Second order moment of PSD_{AC} underestimates slow flows, which are caused by the *M2* on f^2 (equation 6). Statistical analysis of the data shows that the relationship between *LDF* and *LDP* vs. flow velocity is linear with statistical significance ($p < 0.05$) in whole range of velocities (0-3.5mm/s). For *M2* we observed that the linearity is not statistically significant in slow flow range (0-0.5mm/s).

4. CLINICAL EVALUATION OF THE ALGORITHMS

A post-occlusive reactive hyperaemia test was used to evaluate the signal processing algorithms. Occlusion was performed with a 2cm wide cuff located on the base of left hand third finger of healthy subjects. The cuff was inflated to the pressure of 200mmHg. Fig.4. presents simultaneously recorded signals *LDF*, *LDP*, *C*, *V* and *M2* normalised by maximum value.

The signals *LDF* and *LDP* are very similar confirming the similarity of the algorithms used for their calculation. Signals *C* and *V* are more noisy causing difficulties in their clinical analysis. The difference between *LDP* and *M2* signals are mostly visible for low perfusion conditions (during occlusion).

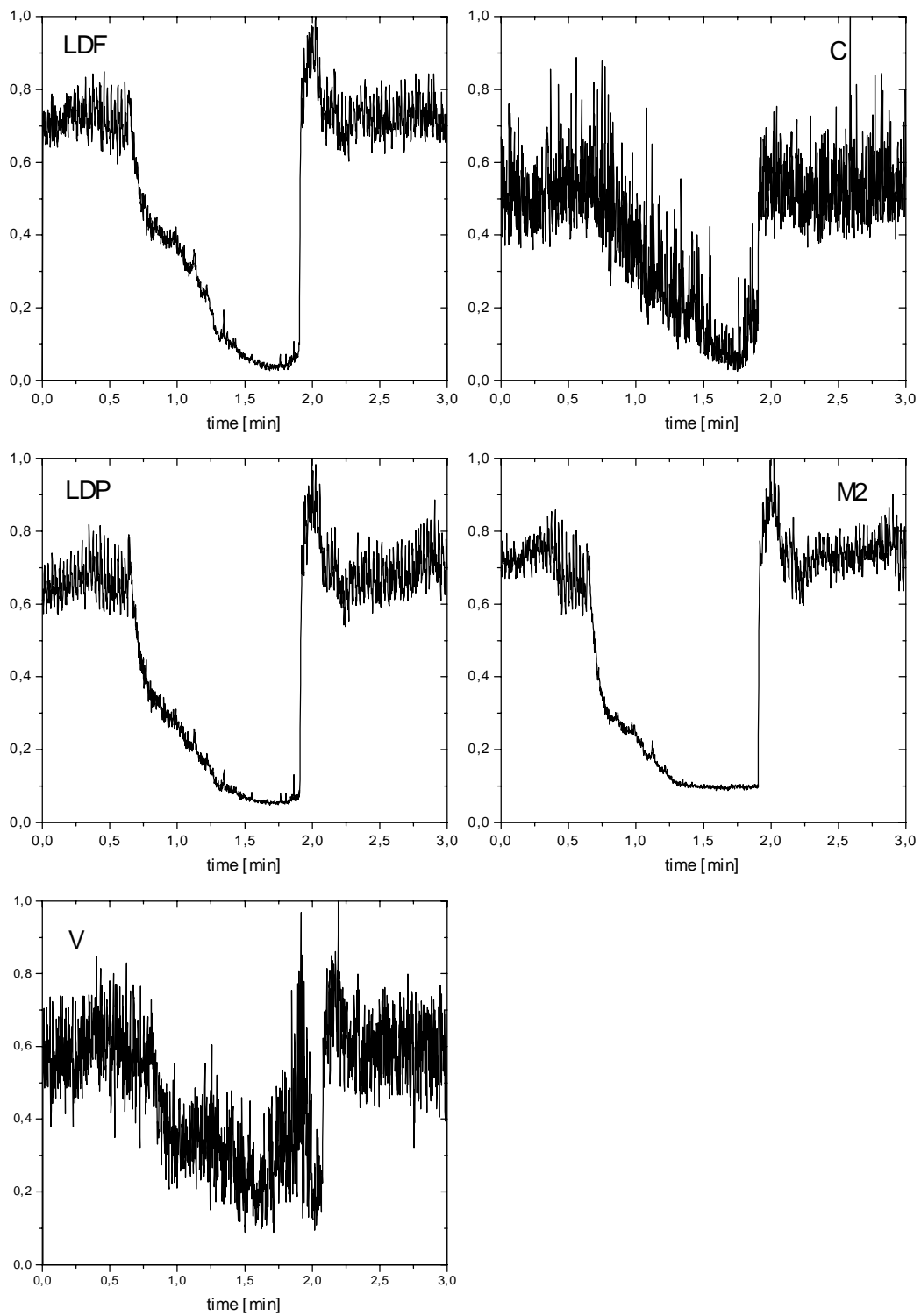


Fig.4. Comparison of signals *LDF*, *LDP*, *M2*, *V*, *C* recorded during post-occlusive reactive hyperaemia test.

5. CONCLUSIONS

- The algorithm based on the first moment of the AC spectrum is sensitive to the high-frequency components of the AC signal and in consequence to higher flow velocities.
- The algorithm based on the second moment of the AC spectrum underestimates the perfusion for low flows, but is linear to much higher velocities than either LDF or LDP.
- A simple one-tube flow model allows for evaluation of the instrument linearity by change of flow velocity and concentration of particles in the calibration solution.
- Comparison of analogue (LDF) and digitally (LDP) processed perfusion indexes suggest that both can be used in LD studies.
- We note that the noise level in LDP signal could be much lower in case of increased number of averaging in spectral estimation. It suggests that the method can be useful in some clinical applications, such as forearm flow or Raynaud's phenomenon, where low flow can cause the noise level to play important role in signal analysis.

6. ACKNOWLEDGMENTS

Authors would like to thank Ms. Anna Zbieæ and Mr. Jerzy Pikowski for their technical assistance. Work is partly supported by the European Commission for Science, Research and Development in BIOMED I Concerted Action: "Laser-Doppler Flowmetry in Microcirculation Monitoring" and by national grant KBN No. 3P40102306.

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