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Method and Device for Metrological Control of Laser Doppler Flowmetry Devices

E. A. Zherebtsov*, A. I. Zherebtsova, A. V. Dunaev, and K. V. Podmaster'ev

The main problems of metrological assurance for laser Doppler flowmetry (LDF) devices that inhibit the further development of this diagnostic technology are discussed. The proposed method and device for monitoring the metrological status of LDF devices, characterized by the use of the principle of signal playback of Doppler shift based on probing the oscillating light-diffusing Lambertian surface, allow checking the metrological characteristics of LDF devices.

Introduction

The recent emergence of many new instruments and methods for optical noninvasive diagnostics is accompanied, with rare exceptions, by the fact that their adoption and use in practice are implemented without full metrological assurance [1-3]. Among them is the method of laser Doppler flowmetry (LDF) allowing estimation of blood flow intensity in the microcirculation link of the bloodstream as well as detection and study of collective rhythmic processes of blood microcirculation. One way to increase the level of metrological assurance is to develop optical phantoms (test objects) reproducing the measured quantities and/or registered biomedical parameters [4, 5].

According to its physical meaning, the measurement results in LDF –microcirculation index (MI) – is a value measured in relative (perfusion) units, which is proportional to the average concentration of red blood cell ensemble and their average velocity [6]. In LDF devices, this quantity is determined by probing biological tissue by laser radiation in the wavelength range of 630-1100 nm and measuring the Doppler shift frequency in the range of 20-24,000 Hz arising after the reflection of radiation from an ensemble of red blood cells moving at different speeds in the range 0.1-10 mm/s in small vessels – arterioles, capillaries, and venules [7]. The current method of LDF is in fact the only method that allows local study of capillary tissue perfusion intensity quickly and noninvasively. One of the main diagnostic applications of LDF is analysis of capillary microcirculation rhythms in the range of 0.01-1.8 Hz [8].

The main constraint for development of the LDF method is the unsatisfactory reproduction of the size of the signal recorded by LDF, which is used for the purpose of setup and calibration at the production stage, as well as to check the current metrological state during operation [9]. The most widespread method of LDF signal reproduction using a stabilized suspension of light-scattering particles undergoing Brownian motion [10] has a number of disadvantages that make it of little use for practical applications: low stability, short shelf life, extreme sensitivity to external influencing factors – temperature and vibration especially, and the sample is able to reproduce only one level of signal. The combination of these factors leads to the fact that in the Russian Federation LDF devices are not subject to state metrological control during their operation (procedures carried out to identify the limit state or latent failure), which often leads to decreased confidence of medical doctors in the method.

Thus, of critical importance are theoretical foundations of LDF signal reproduction, as well as its practical implementation in the form of a device (test object) suitable for metrological state control (MSC) not only at the production stage (adjustment, final checking, testing of accuracy how the specified static and dynamic characteristics of the instrument match its individual characteristics), but also during the operational phase in the health

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Scientific-Educational Center "Biomedical Engineering", State University – Education-Science-Production Complex, Orel, Russia; E-mail: zherebzow@gmail.com

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care setting. This article attempts to improve the metrological assurance of laser Doppler flowmetry by developing a method and device for metrological control of the state of LDF devices.

Principle of Operation and Experimental Studies of the Device

Currently, the most common design scheme of LDF devices is the difference scheme having one probing and two receiving optical fibers, the received signal difference from being used for further processing. The only domestic LDF devices of the LAKK series (LAZMA Ltd., Russia) are designed using this scheme [6]. The generalized formula for calculating the microcirculation index is written as an expression [11]:

$$I_m = k_{\text{in}} \frac{\int_{f_{\min}}^{f_{\max}} S_{i_{ac}}(f)|f|\,df}{i_{dc}^2}, \tag{1}$$

where $S_{i_{ac}}(f)$ is spectral density function of the signal power of variable component of photocurrent $i_{ac}(t)$; $i_{dc}(t)$ is a constant component of the photocurrent; k_{in} is an instrumental factor.

Possible ways to reproduce the Doppler shift of the optical frequency were analyzed based on the classification of methods to solve the problem showing a promising method of LDF signal reproduction using a vibrating diffusely scattering surface in direction collinear to the radiation propagation vector [12].

The proposed approach is implemented in the form of the experimental setup (scheme and appearance shown in Fig. 1), the radiation flux $\Phi_{ed.const}$, which is not subject to the Doppler shift, is formed by reflection from a fixed slab translucent plate located above the moving diffuse reflector. The power capacity of this component can be defined by observing the radiation reflection geometry at the “glass-to-air” and “air-to-glass” interfaces.

Power of radiation flow that is subject to the Doppler shift is defined as the flux generated by reflection from a vibrating diffusely scattering Lambertian surface. Probing radiation flux Φ_{es} coming from the transmission fiber forms a cone with an apex half-angle equal tp θ_{as}, height h_r, and base A_s.

The power of flux Φ_{ed} (which is subject to a Doppler shift and is received by one receiving fiber) can be found by integrating the radiation flux incident on the core of the receiving fiber at an angle less than the aperture and created by the light scattering on the area of overlapping ellipses formed by cutting the cones of the probing and receiving radiation of the Lambertian light-scattering surface. Thus, the radiation fluxes $\Phi_{ed.dop1}$ and $\Phi_{ed.dop2}$ can be found for each of the two channels of the difference scheme. When adding components with and without the Doppler shift on the surface of the receiver, the function of power field changes in time having the form

$$\begin{aligned}\Phi_{ed.sum}(t) &= \Phi_{ed.const} + \Phi_{ed.dop} + \\ &+ 2\sqrt{\Phi_{ed.const}\Phi_{ed.dop}}\cos\left[\varphi_d(t)\right].\end{aligned} \tag{2}$$

The expression for the phase $\varphi_d(t)$ of the third component of Eq. (2) for harmonic oscillation of the scattering surface with frequency f_{osc} is defined as follows:

$$\varphi_d(t) = \frac{2V_{\max}\left[\cos\left(\theta_{fl.avg}\right) + \cos\left(\theta_{sc.avg}\right)\right]}{\lambda_0 f_{osc}} \sin\left(2\pi f_{osc} t\right), \tag{3}$$

where $V_{\max} = 2\pi A_0 f_{osc}$ is velocity amplitude; f_{osc} is oscillation frequency; A_0 is oscillations amplitude; $\theta_{fl.avg}$ is average angle of radiation incidence on the scattering surface; $\theta_{sc.avg}$ is average angle of radiation scattering by the scattering surface.

According to the signal processing scheme of the LDF method, the variable component of the signal from the photodetectors for each channel is divided by the corresponding constant component, and the results of the division are subtracted from each other:

$$\begin{aligned}U_{ac}(t) = 2\cdot\frac{K_{ac2}}{K_{dc2}}\cdot\frac{\sqrt{\Phi_{ed_{const2}}\cdot\Phi_{ed_{dop2}}}}{\left(\Phi_{ed_{const2}}+\Phi_{ed_{dop2}}\right)}\cdot\cos\left[\varphi_{d2}(t)\right] - \\ - 2\cdot\frac{K_{ac1}}{K_{dc1}}\cdot\frac{\sqrt{\Phi_{ed_{const1}}\cdot\Phi_{ed_{dop1}}}}{\left(\Phi_{ed_{const1}}+\Phi_{ed_{dop1}}\right)}\cdot\cos\left[\varphi_{d1}(t)\right],\end{aligned} \tag{4}$$

where K_{dc} and K_{ac} are amplification coefficients for the constant and variable components, respectively.

Let $S_{U_{ac}}(f)$ be the spectral density function of signal power of the variable component $U_{ac}(t)$. According to the model of microcirculation index calculation, the value of the reproduced level of MI is calculated as:

$$I_m^{\text{repr}} = \int_{f_{\min}}^{f_{\max}} S_{U_{ac}}(f)|f|\,df. \tag{5}$$

In the assembled experimental setup, a serially manufactured P-602.8SL precision piezo actuator (Physik Instrumente, Germany) is used as an actuator of the light-scattering surface, and a white polytetrafluoroethylene (PTFE) disk (Ocean Optics, USA) was is used as the light-scattering surface. An NDC-100C-4M continuously variable neutral light filter in a round metal rim

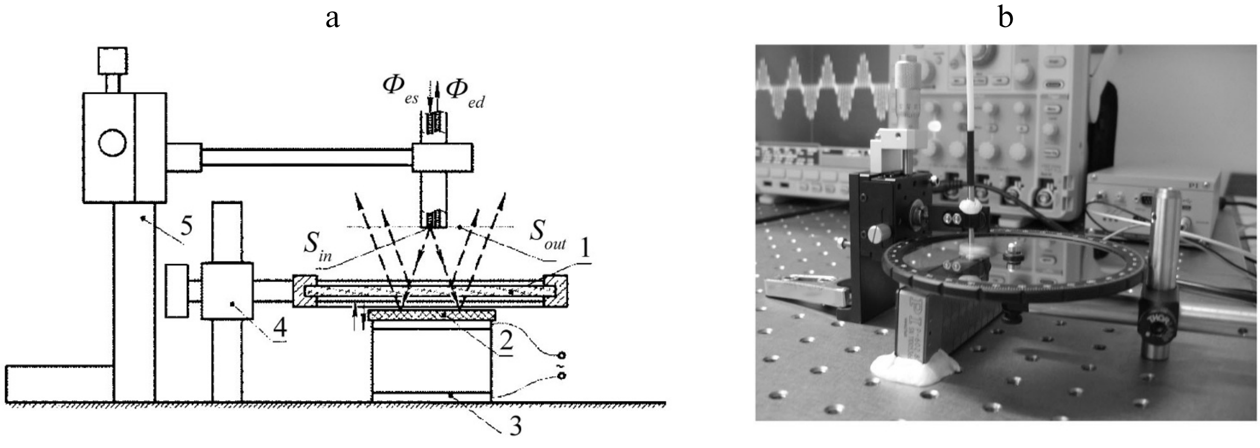

Fig. 1. Scheme (a) and appearance (b) of the experimental setup: 1) continuously variable filter; 2) fluoroplastic disk; 3) serially manufactured piezo actuator; 4) tripod for fixing and adjusting the height of the filter; 5) tripod with a micrometer screw for fixing and adjusting the height of the fiber; S_{in}, aperture of fiber probe; S_{out}, aperture of receiving fiber.

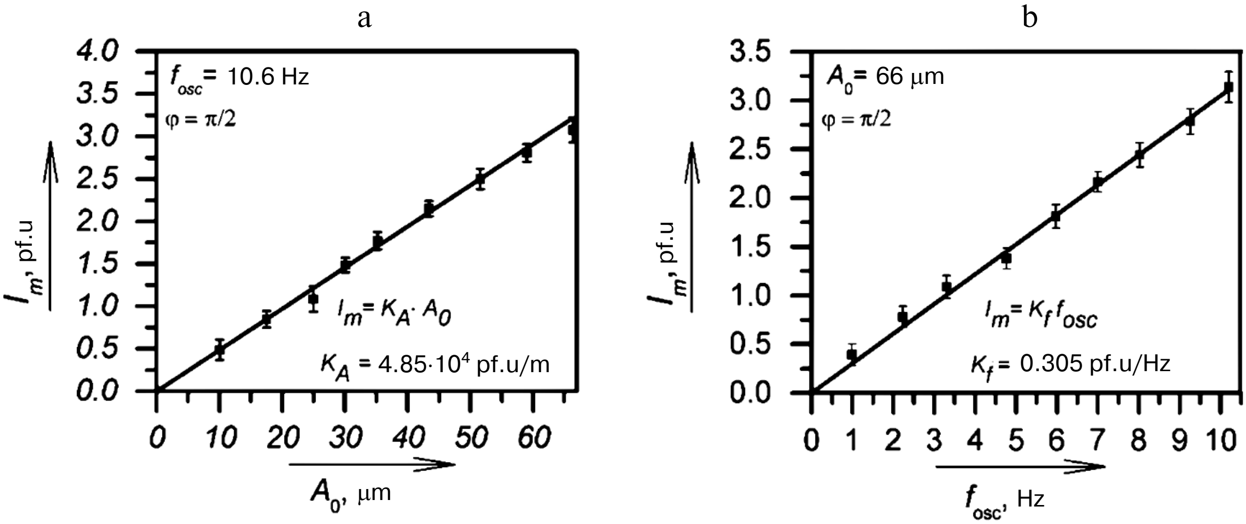

Fig. 2. Experimental and theoretical relations of the reproduced perfusion level from amplitude (a) and frequency (b) of oscillations of the light-scattering surface.

(Thorlabs, USA) was chosen as the slab reflective plate. As an LDF device subjected to MSC, an LAKK-M multifunctional laser noninvasive diagnostic complex (LAZMA Ltd., Russia) with LDF channel of 1064 nm probing wavelength was used in the experimental studies.

Figure 2 shows plots of the reproduced MI level by amplitude (a) and frequency (b) of oscillations of the light-scattering surface (theoretical curve is placed over the experimental points including the standard deviations).

To test the feasibility of the proposed principle to control the reliability of microcirculation rhythm registration and to evaluate dynamic error of the device, an experiment was carried out in which the light-scattering surface was oscillating according to the law of single-tone amplitude modulation. Figure 3 is a graph of the reproduced perfusion signal with modulating frequency of 0.3 Hz (a) and its amplitude–frequency spectrum (b).

Results of the comparison of the reproduced MI signal from the experimental setup and colloidal gel “Motility standard” (Perimed AB, Sweden) for calibration of LDF devices showed that the proposed experimental setup is almost insensitive to vibration of the sup-

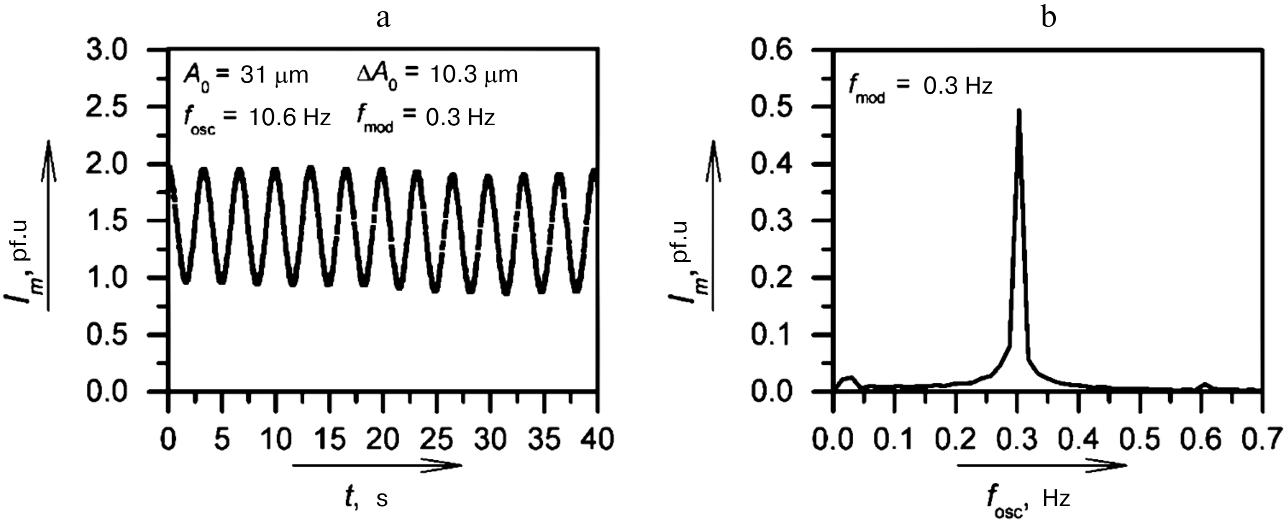

Fig. 3. Experimental plot (a) and an amplitude–frequency spectrum (b) of reproduced alternating perfusion signal (probe angle 3 ± 0.5°).

port (standard deviation of the signal reproduced by the proposed method is four times lower). It should be emphasized that in contrast to the proposed method, the calibration using the gel according to the manufacturer's recommendations requires the use of a table with vibration damping (usually expensive) because of the high sensitivity to external vibrational disturbance.

The electromechanical scheme of the proposed device for MSC of LDF devices is shown in Fig. 4. Its basis is a serially manufactured piezo actuator 6 driven by a control voltage. Light scattering element 5 having Lambertian directional pattern is attached to the piezo actuator. Over the surface of the light-scattering plane, a slab translucent plate 4 is mounted. The fiber-optic probe of the LDF device is installed on the surface of the translucent plate using fixing devices 2 or 3. Fixing device 2 provides normal location of the fiber-optic probe with respect to the surface plane, while device 3 provides the location of the fiber-optic probe at a fixed angle β.

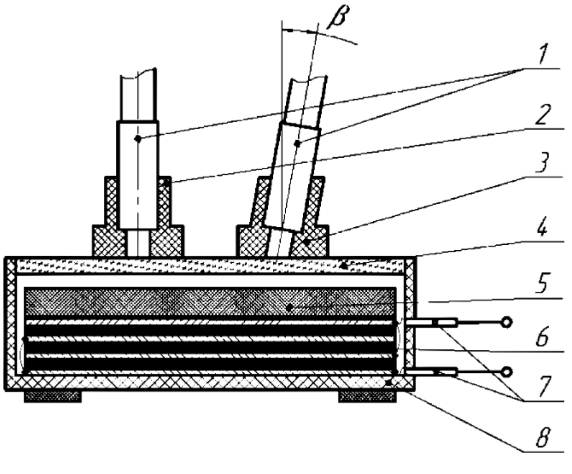

Fig. 4. Scheme of the electromechanical device for MSC of LDF devices: 1) optic probe of LDF device; 2) fixing device for vertical probe mounting; 3) fixing device for angled probe mounting; 4) translucent plate; 5) light-scattering plate; 6) serially manufactured piezo actuator; 7) electrodes; 8) case.

The proposed method for MSC of LDF devices is that the fiber-optic probe of the LDF device is fixed at a certain angle over the fixed translucent and oscillating harmonically light-scattering surfaces arranged one above the other. The reproduced level of MI signal is recorded and compared with the nominal reproduced level of MI signal, which is calculated for given values of the design parameters and operation mode variables. By comparing the results, a conclusion on metrological characteristics (control of correspondence between individual and nominal static characteristics) of the LDF device can be made.

Based on this LDF device, control method techniques for determining static and dynamic errors of an LDF device (complex check of optic–electronic circuit scheme of LDF channel during registration of a varying signal) are suggested. Also, the technique for monitoring transmission coefficients equality of input channels of the LDF device difference scheme in the operating range of the recorded Doppler shift (check of the fiber-optic probe and input stages of the difference scheme) is proposed.

The developed method of determining transmission coefficient equality of the input channels of the difference scheme for LDF method implementation can be used for configuring new devices during the production stage.

Conclusion

The presented method and the device for its technical implementation reflect a promising direction for creating techniques of metrological state control for LDF devices allowing checking of the metrological characteristics in static and dynamic modes and drawing conclusions about their suitability for use at production stage as well as for operation in health care facilities. The reproduced level of microcirculation index has linear dependence on the frequency and amplitude of oscillations of the light-scattering surface. Due to the proposed design solutions underlying this method, the reproduced level of microcirculation index remains almost unaffected by vibration and temperature changes.

The developed method and device for metrological state control of LDF devices increase the level of metrological assurance and bring this noninvasive method to standardized diagnostic technologies used in modern health care.

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