

Optical methods to quantify tissue blood supply in clinical settings: experiment and interpretation (review)

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In modern surgery, intraoperative assessment of blood perfusion is an urgent but still unresolved problem. Due to the fact that light has minimal impact on biological tissue, optical methods are the most promising for non-contact measurements of blood flow in vivo. Nevertheless, despite many years of extensive studies of blood flow parameters by means of optics, the fundamental question of the interaction of light with living biological tissue containing blood vessels remains insufficiently studied. For example, the reasons why light, after its interaction with tissue, is modulated at the heart rate are still a matter of debate. This review will present a number of experimental observations that cannot be explained within the framework of the generally accepted models of the light interaction with living tissue. The question of how the observed modulation of light is related to the parameters of blood flow and its changes during pharmacological or surgical interventions will be discussed. Special attention will be paid to the discussion of systems of imaging photoplethysmography synchronized with an electrocardiography and laser speckle contrast imaging, which allow the assessment of tissue perfusion in a wide angle of view and with high resolution in both space and time. Since imaging photoplethysmography and laser speckle contrast imaging rely on fundamentally different principles of the interaction of light with biological tissue in vivo, their complementary (multimodal) use could provide an in-depth comprehensive assessment of hemodynamics in clinical setting.

Keywords: light interaction with tissue, intraoperative monitoring, blood supply, imaging photoplethysmography, laser speckle contrast imaging.

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1. Introduction

Changes in tissue perfusion during surgical operations can cause complications such as hypoxia, ischemia, followed by necrosis. Therefore, intraoperative monitoring of regional perfusion is essential for surgery optimization and reduction of postoperative complications. However, a standard microcirculation monitoring procedure is not available yet. High requirements are imposed on the blood flow measurement/imaging system: it shall be easy to use and non-contact, and shall provide real-time quantitative evaluation of change in tissue perfusion simultaneously throughout the field of study. Among the variety of known methods, optical techniques are very promising for non-contact measurement of blood flow parameters. The following techniques are seen as candidates for non-contact intraoperative measurement of tissue perfusion: indocyanine green fluorescence angiography (ICG-FA) [1–3], laser speckle contrast imaging (LSCI) [3–5], imaging photoplethysmography (iPPG) [6–8], laser Doppler flowmetry (LDF) [9,10], hyperspectral imaging (HSI) [11,12] and optical coherence tomography (OCT) [3,13]. From the above-mentioned techniques, LDF and OCT have significant restrictions for intraoperative applications because they measure blood flow only in small regions from which information about spatial blood flow nonuniformity, being a critical microcirculation property, cannot be obtained.

Other four techniques can evaluate microcirculation in a wide field of view and simultaneously for the entire field of study because a standard digital camera is their main component and the blood flow information is generated from video stream processing. ICG-FA system is a commercially available well-proven imaging technique which is widely used for intraoperative applications. Nevertheless, surgeons predominantly use only qualitative visual perfusion distribution obtained using ICG-FA, while quantitative evaluation of these distributions remains at the research and development stage [14]. In addition, employment of a fluorescent substance with potential side effects in patients and failure to perform continuous or repeated perfusion evaluation are the main disadvantages of the technique.

HSI is a noninvasive optical technique combining the principles of absorption spectroscopy and digital imaging to obtain quantitative data both on perfusion parameters and tissue condition. Moreover, this technique requires high-tech instrumentation — an imaging spectrometer, and processing of a large amount of spatial and spectral components of the obtained data. There are currently only a few studies that describe the use of this blood flow imaging technique in real clinical conditions [15].

LSCI and iPPG can be currently considered as the most promising methods for intraoperative perfusion evaluation because only a digital camera and illuminator are sufficient for implementing both of them. Systems based on these techniques are simple, easy to use, don't require

contrast medium administration and can evaluate blood flow parameters simultaneously in a wide field of view. However, despite positive and repeatable results of pilot studies of LSCI [4,5,16–21] and iPPG [6–8,22–26] in clinical conditions, interpretation of results is not always sufficiently justified due to undergoing discussion about the causes of modulation of light that has interacted with biological tissue *in vivo*. It is shown in the above-mentioned literature that various functional impacts on the body lead to a significant change of LSCI and iPPG signals. Nevertheless, the question on how reliably the observed changes in optical signals can be interpreted as changes of organ perfusion remains open in both cases.

The current critical review describes some experimental facts obtained during hemodynamic parameter response monitoring via the LSCI and iPPG techniques. Explanation of these facts within a conventional model of interaction between light and living tissue is problematic.

2. Light penetration depth and its modulation at the heart rate

Perfusion of each living body organ is carried out by pulses, frequency and amplitude of which is defined by cardiac muscle contractions. Therefore, it is quite natural to expect that parameters of light that has interacted with the organ of interest will obtain modulation at the heart rate. Both imaging systems, LSCI and iPPG, use a reflective scheme where the light source and receiving camera are situated on the same side of the region under study. A key role in such configuration is played by the light penetration depth that defines interaction between light and highly heterogeneous structure of biological tissue where blood vessels with different lumen and purpose are located at different depths. Since physical principles behind the LSCI and iPPG techniques differ dramatically, generation of a signal associated with hemodynamic changes in these systems will be addressed separately.

2.1. Imaging photoplethysmography

In an iPPG system, the biological tissue under study is illuminated by incoherent light. Photoelectric plethysmography (more often called photoplethysmography (PPG) afterwards) was proposed in the late 1930s [27] and is the first optical technique used to evaluate the perfusion of various parts of the body. The very first studies reported that after interaction between light and living tissue containing vessels, perfusion intensity appeared to be modulated at the heart rate [27,28]. Since the optical properties of blood differ from those of surrounding tissue, a change (modulation) of the returned light correlating with the heart rate carries the information about perfusion of the region of interest. Traditional PPG signal generation model implies that the main cause of light intensity modulation is the change of blood volume in the tissue [27,29,30]. It

is hereinafter referred to as the „blood volume model“, abbreviated to BVM. However, this model cannot explain the well-established experimental fact that the highest modulation amplitude of the PPG signal at the heart rate is observed when green lighting is used [31–33]. The point is that green light is actively absorbed by red blood cells (RBCs) of the capillary layer and its penetration depth is less than 0.5 mm [34]. At this depth, there are no pulsating arteries and arterioles, while pulsating pressure in capillaries [35,36] and time variation of the RBC velocity profile [37] cannot lead to blood volume modulation (and, accordingly, to green light absorption modulation) due to rigidity of capillary walls [38,39]. Theoretical estimates have confirmed that the PPG signal amplitude at the heart rate within BVM is much lower when using green light than at other wavelengths [40,41].

To explain the observed abnormally high amplitude of green light modulation at the heart rate, our group proposed an alternative PPG theory in 2015 [42]. The essence of the theory is in that the arterial wall pulse pressure transmurally compresses and releases the capillary network at the tissue surface: this reversibly changes the distance between capillaries which actively absorb green light, modulating the amount of light-interacting RBCs, which leads to the light intensity modulation at the heart rate. Thus, the capillary network serves as a natural distributed converter of the blood pressure pulse wave propagating in arteries into changes of the return light power. This PPG signal generation model will be hereinafter referred to as the tissue compression model (TCM). BVM and TCM concur in that light modulation is caused by changes in light absorption due to a change in blood volume with which it interacts [41]. The difference between them is in where exactly these changes of blood volume occur. While BVM doesn't specify the cause of light modulation at the heart rate, claiming that this is likely the interaction between light and pulsating arteries and arterioles, then TCM clarifies in detail the causes of this modulation.

In terms of using iPPG in intraoperative systems for organ perfusion dynamics imaging and evaluation, TCM shows that the PPG signal carries useful information concerning the pulse wave parameters in deep-seated arterial vessels notwithstanding that light doesn't directly interact with these vessels.

2.2. Laser speckle contrast imaging

In the LSCI technique, the coherent light source (laser) illuminates the biological tissue, and the back-scattered light is focused on the receiving matrix of the digital camera. When light is scattered on a fixed object, a stationary high-contrast speckle pattern is formed. When scatterers are mobile as in the case of RBCs moving in vessels, the time-integrated speckle pattern recorded during the finite camera exposure time becomes blurred. According to the generally accepted theory behind LSCI, speckle contrast defined as the ratio of root-mean-square (rms) deviation σ

to mean intensity (or squared ratio) is inversely proportional to the velocity of particles that scatter the coherent light [43,44]. The higher the scatterer velocity relative to illuminating beams, the lower the speckle pattern contrast in the corresponding image area. This makes it possible to visualize blood-carrying vessels in the form of darker regions against a brighter background [44], assuming that it is traveling RBCs that make the greatest contribution to coherent light scattering and corresponding speckle pattern generation. Red ($\lambda = 632\text{--}660\text{ nm}$) [16,17,43] or infrared ($\lambda = 785\text{--}808\text{ nm}$) [3,4,18,19,21,45,46] laser light is more often used in LSCI systems. These light wavelengths penetrate deeper into tissue with vessels than green light, but due to fundamental peculiarities of speckle pattern generation, the red and infrared LSCI signal carries the information on RBCs velocity only in a near-surface layer less than 1 mm in depth [47–49]. This feature is considered to be one of key constraints of LSCI as a method for imaging functional vessels [5,16,48–50] because it doesn't allow a surgeon to evaluate the changes of blood flow in deeper vessels.

It should be noted that many researchers report a strongly pronounced LSCI signal modulation at the heart rate [4,16,17,20,21,45,46,51]. This observation is generally explained by the fact that the RBC velocity in capillaries is pulse-wave modulated [37,52], this pulse wave causes time modulation of speckle pattern contrast. However, various groups also reported experimental studies where LSCI of vessels showed the most significant reduction of the speckle contrast (i.e. the highest velocity of coherent-light scattering particles) exactly at large veins [46,51,53–56]. Whereas high modulation amplitude of LSCI signal at the heart rate was reported throughout the brain surface, including the venous sinus and large veins [46].

3. Causes of LSCI signal modulation at the heart rate in veins

Interpretation of abnormally high LSCI signal modulation in large vein region becomes a real challenge. RBC motion in the vascular system is mainly caused by the pressure pulse wave generated by the cardiac muscle. The largest difference between the systolic and diastolic pressures in this wave is naturally observed in arteries. Therefore, it should be expected that it is arteries where the highest RBC velocity shall be gained. Vascular system is closed: the amount of blood flowing through the arterial region shall be equal to that flowing through the venous region. As blood enters veins after passing through the capillary network, which is characterized by the maximum hydrodynamic resistance, the pulse wave amplitude in veins shall be much lower than in arteries. Moreover, the lumen of the largest veins is much greater than an arterial lumen. Consequently, the lower pulse pressure in veins cannot accelerate RBCs to velocities even approaching the RBC velocities in arteries. The question arises now: on what

traveling particles or structures can light scattering lead to such significant reduction of the speckle contrast in large veins?

To answer this question, our group has recently experimentally compared cerebral blood flow imaging and response to sodium adenosine triphosphate (ATP), administration of which expands cerebral arteries [57]. Animal (rats) studies were performed. This study has confirmed once again that LSCI visualizes blood flow predominantly in large veins provided that blood flow is associated with RBC velocity calculated inversely proportional to the speckle contrast. Blood flow distribution over the cerebral cortex at different times is shown in Figure 1, *a*. Variation of blood flow response to ATP administration (started at 17th second) is shown in Figure 1, *b*, where the inverse speckle contrast variation in the regions of interest on the left and right hemispheres is shown by red and blue curves, respectively.

It should be emphasized that broadening of curves in Figure 1, *b* is not noise-related, but is the speckle contrast modulation which correlates very well with the heart rhythm, which is shown in Figure 2, where the inverse speckle contrast variation is displayed on a more expanded time scale together with the ECG signal during ATP administration. As shown in Figure 2, the speckle contrast is time modulated with a period equal to the cardiac cycle length. Note that this modulation is well correlated with R peaks on the ECG not only in initial conditions before vasodilator administration, but also when the cardiac cycle length changes in case of severe bradycardia induced by ATP administration.

Spatial distribution of the amplitude of the pulsatile component (APC) of the PPG signal measured on the same rat with the identical ATP administration protocol as that for LSCI (Figure 1) is shown in Figure 3, *a*. Variation of the APC response to ATP infusion evaluated in the same regions of interest is shown in Figure 3, *b*. Details of the experimental comparison of LSCI and iPPG signal responses to ATP infusion are described in [57]. It can be clearly seen that spatial distributions of the cerebral perfusion parameters obtained using LSCI and iPPG differ dramatically: for LSCI, the most intense change in the inverse speckle contrast is observed in the sagittal and transverse sinuses (Figure 1, *a*), while the blood pulsation amplitude detected in large veins using iPPG is much lower than in other cerebral cortex regions (Figure 3, *a*). The difference between blood flow parameters evaluated using LSCI and iPPG becomes even more apparent when their responses to ATP infusion are compared (Figure 1, *b* and Figure 3, *b*). Immediately after vasodilator infusion, LSCI shows the decrease in the speckle contrast, whereas iPPG shows the increase in the pulsation amplitude.

Solution to the observed discrepancies was proposed in [57] where an assumption is made that the LSCI signal modulation in veins results from the arterial pulse wave, but

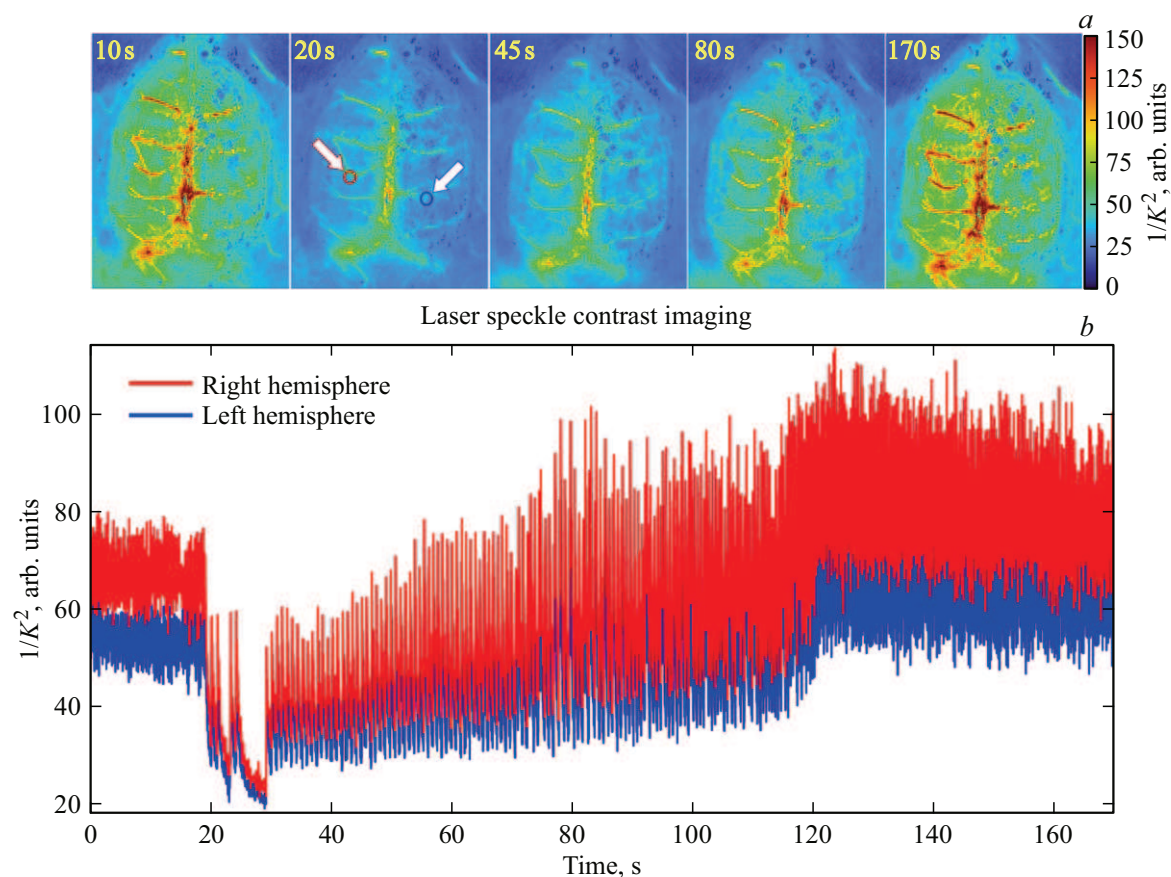


Figure 1. Example of cerebral blood flow response to ATP infusion measured by LSCI. (a) Inverse speckle contrast maps calculated at various session times. (b) Time variations of inverse speckle contrast in small regions of interest on the right and left hemispheres, positions of which are marked with red and blue circles, respectively, in the second image, upper panel.

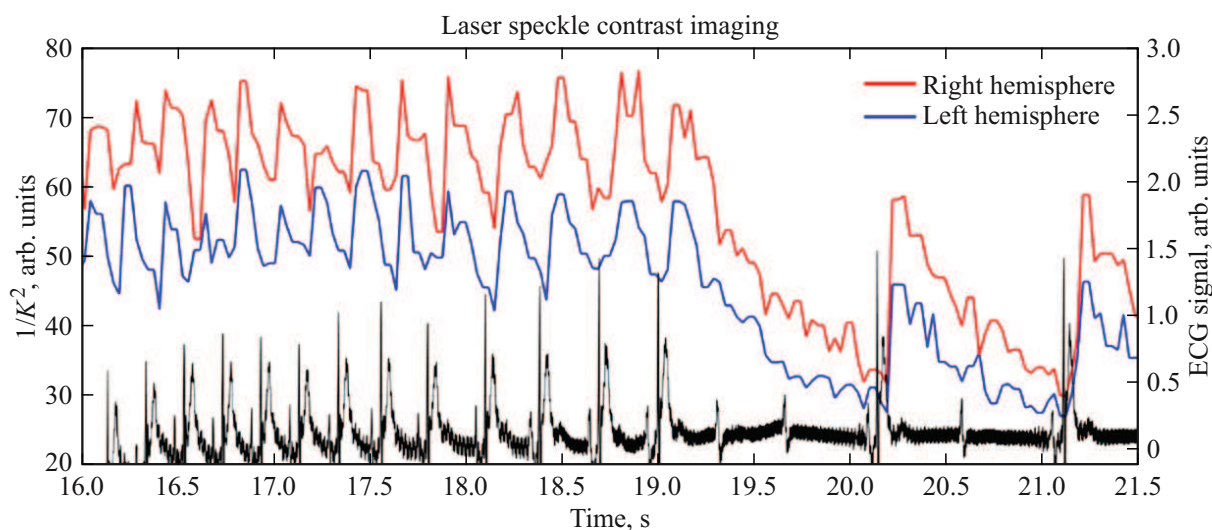


Figure 2. Diagram of speckle contrast variation in the regions of interest on the right (red curve) and left (blue curve) hemispheres recorded simultaneously with the ECG signal (black curve).

transmitted into the venous region by acoustical-mechanical means through the parenchyma. Taking into account the apparent change of the LSCI signal modulation at the heart

rate with time during the ATP infusion session (Figure 1, b), the modulation amplitude of the inverse speckle contrast was calculated in each cardiac cycle. Such calculation

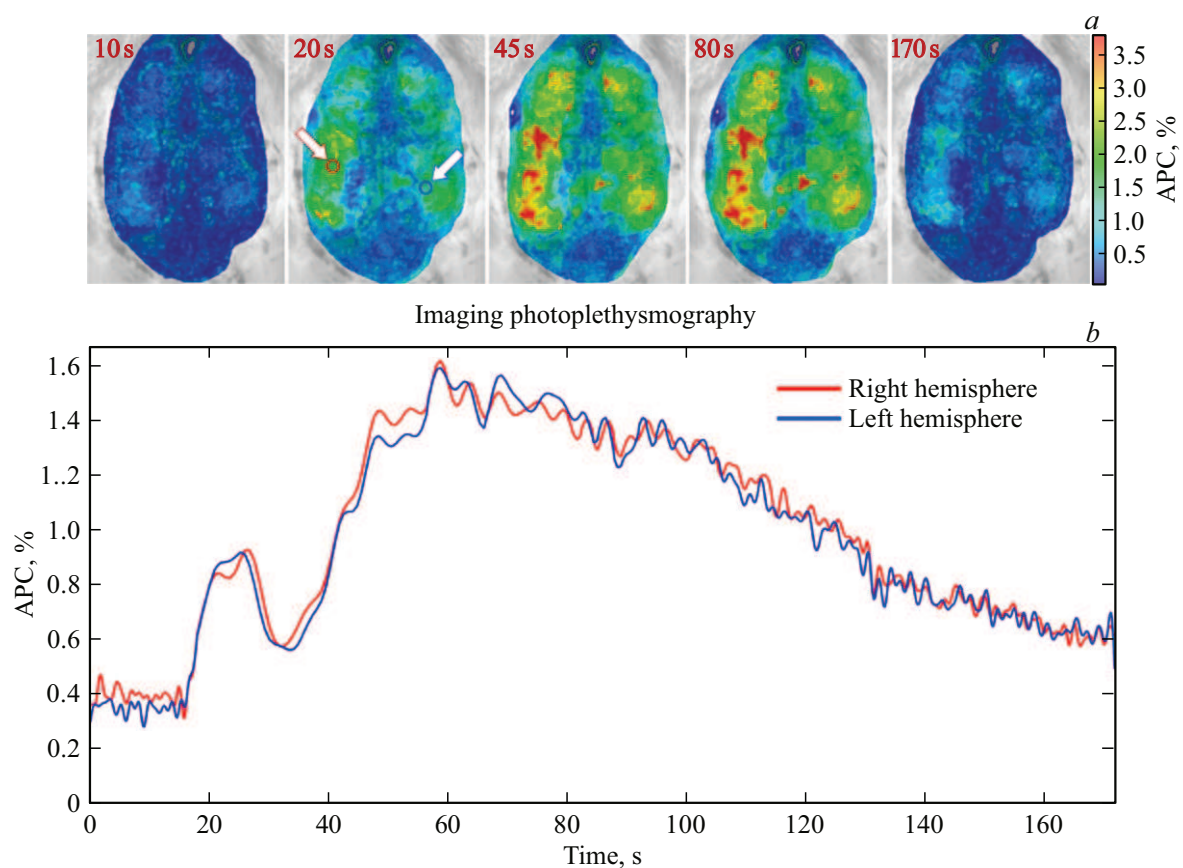


Figure 3. Cerebral blood flow response to ATP infusion measured by iPPG on the same rat as in Figure 1. (a) Blood flow pulsation distribution maps. (b) APC time variations in the regions of interest arranged in the same way as during LSCI measurements and marked by red and blue circles in the second frame, upper panel.

converts the dependences shown in Figure 1, *b* into the curves shown in Figure 4.

As can be seen from comparison of curves in Figures 4 and 3, *b*, after the calculation, the inverse speckle contrast pulsation amplitude variation response to ATP infusion now looks very similar to the iPPG APC signal variation pattern.

Causes behind the speckle contrast modulation at the heart rate in veins were discussed by Postnov et al. [45]. This modulation was assumed to be a consequence of three types of pulse wave in veins:

- (1) forward-propagating pressure pulse wave,
- (2) wave generated by the motion of surrounding tissues,
- (3) „pulling-back“ cardiac wave.

However, cause № 3 hardly can be used to interpret the nature of the LSCI signal shown in Figure 2. This figure shows that a pressure peak associated with the maximum RBC velocity (i.e. with the maximum on the inverse speckle contrast curve) is observed with a quite short delay after the R peak. But for the back pressure wave, this delay shall be longer [58–60]. As for cause № 1, we repeat the above statement from paragraph 1 in this section: low pulse pressure in veins and their large lumen prevent from accelerating RBCs in veins to significant velocities. Thus, there is only the effect of surrounding tissue

motion (cause № 2) which unfortunately wasn't sufficiently specified in [45].

4. Discussion

Coincidence of the response curve to ATP infusion (Figure 3, *b* and 4) after calculation of the LSCI signal in the form of the speckle contrast modulation amplitude in each cardiac cycle as detected in [57] suggests that TCM discussed in 2.1 also can shed light on the causes behind the abnormal modulation of the inverse speckle contrast at the heart rate in veins. The largest amplitude of heart-beat-related pressure pulsations is observed in arteries. These pulsations are transmitted to the parenchyma in a natural way and excite acoustic vibrations in the parenchyma which cause parenchyma boundary displacement. Venous wall displacements appear to be particularly strong because blood pressure in veins is fairly low. This wall displacement causes RBC motion in veins, but the motion vector is perpendicular to vein axes because venous walls are displaced orthogonally to the axis. At the maximum arterial pulse pressure, the RBC velocity in direction orthogonal to the vein axis is the highest, and the maximum speckle pattern decorrelation is observed, i.e. the maximum inverse

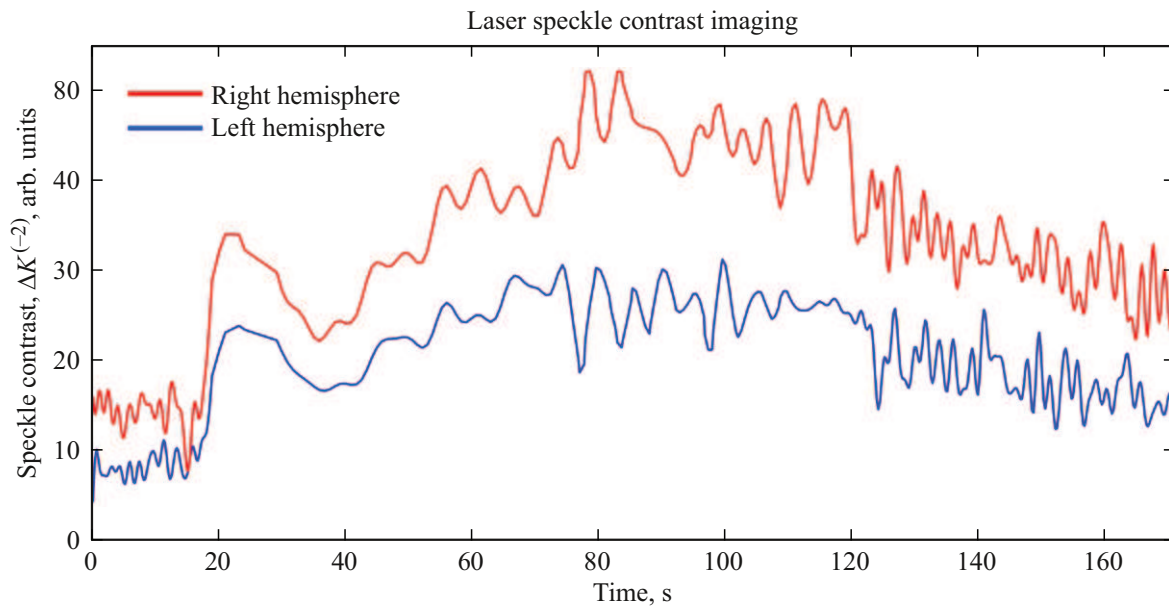


Figure 4. Variation of the inverse speckle contrast pulsation amplitude calculated from the input data shown in Figure 1, *b*.

speckle contrast (Figure 2). Consequently, though speckle contrast variations are most prominent in the sagittal and transverse sinus region (Figure 1, *a*), it is the pulsating arterial pressure that causes the speckle decorrelation in these vessels.

It is important to emphasize that not every movement of RBCs can be considered as contributing to brain perfusion. Perfusion is intended to deliver blood to the capillary network to ensure effective metabolism. Therefore, perfusion evaluation shall take into account only those particles that move in and along arteries. It is incorrect to interpret RBC motion in veins perpendicular to vein axes which is detected by LSCI with the normally used calculation of the inverse speckle contrast as perfusion evaluation. On the other hand, a new algorithm proposed in [57] for calculating the inverse speckle contrast modulation amplitude in each cardiac cycle actually provides indirect parenchyma-mediated evaluation of changes in arterial tone of vessels located much deeper than the light penetration depth commonly assumed in the LSCI technique. Therefore, the speckle contrast modulation amplitude at the heart rate can be referred to as the perfusion index similarly to iPPG. It should be emphasized that parenchyma topology and acoustical-mechanical properties define the ratio of the light modulation amplitude at the heart rate to the pulse pressure range in arteries feeding the test area. This ratio can vary both between types of biological tissue and between participants. Therefore, correct conclusions about the regional perfusion behavior can be made only by comparing optical parameters at the same points of study and provided that mechanical properties of connective tissue are relatively constant.

Cerebral blood flow study in an animal model was used to compare the generation of iPPG and LSCI signals, since in this case the location and type of vessels are adequately

identified. Nevertheless, the conclusions concerning non-trivial interpretation of pulsations in venous region and significant effect of mechanical compression of connective tissues can be also extended to cover other organs. For example, in a study of peripheral blood supply to skin by optical methods, different groups reported observing a pulse wave, the shape of which is inverted in time compared with that observed in the adjacent regions [58,61–65]. Note that the inverted pulse wave is observed in the wrist [61,62,64] or neck [58,63,65] region where arteries and veins are spaced closely to each other and relatively near the skin surface. Within TCM, the inverted phase effect has its logically consistent explanation [42,64]. In the case when studies are carried out in regions where arteries and veins are spaced quite apart, we expect that spatial distributions of the perfusion index calculated via iPPG and LSCI (using a new data processing algorithm [57]) will coincide.

Note that perfusion index distribution maps calculated from the LSCI data using the new algorithm will slightly differ from those shown in Figure 1, *a* because the speckle contrast modulation at the heart rate is inversely proportional to mean decorrelation during the cardiac cycle time. Maximum speckle contrast modulation amplitude will be observed in the same regions where the most significant decorrelation is observed, i.e. on large veins. Consequently, blood supply change (for example, due to reaction to functional or surgical intervention) in these regions will be also measured with a better signal-to-noise ratio than in the adjacent regions. In the case of iPPG-evaluated perfusion index distributions (Figure 3, *a*), regions with the highest PPG signal amplitude are located in other zones, which allows the perfusion variation to be qualitatively evaluated in regions where the LSCI signal is insufficiently high. Due to different physical principles behind LSCI and

iPPG, these methods supplement each other in the case of quantitative evaluation of perfusion variations. Since only digital cameras and corresponding light sources are required for implementing both techniques, the development of a system providing simultaneous recording of iPPG, LSCI and ECG signals seems to be quite realistic. Such system will provide more correct intraoperative evaluation of quantitative changes in perfusion of various zones of the test organ than iPPG or LSCI taken separately.

5. Conclusion

Among the existing optical techniques, LSCI and iPPG are the most promising for quantitative evaluation of tissue perfusion variation in response to functional or surgical interventions because simplicity of both systems requiring only standard cameras and corresponding lighting makes them especially suitable for clinical application. They provide continuous imaging of spatial distribution of the perfusion index of the organ under study and temporal-spatial variation of the perfusion index in a wide field of view and in real time, without requiring contrast medium administration.

As the above analysis shows, despite a shallow depth of light penetration into biological tissue with blood vessels, both techniques yield a signal associated with the vascular tone of arteries supplying the region under study and located deep enough provided that the LSCI data are processed using a new algorithm proposed in [57]. Since the LSCI and iPPG techniques are based on physically different principles of interaction between light and vessels *in vivo* (scattering and absorption, respectively), they offer different and complementary concepts of hemodynamics. Therefore, employment of a multimodal system that simultaneously records the iPPG, LSCI and ECG is even more promising and will allow in-depth comprehensive evaluation of hemodynamics. It is believed that further experimental and theoretical studies of simultaneous use of these systems for perfusion analysis will allow the findings to be extended to all organs in terms of the degree of manifestation of the effects under discussion.

At present, neither of the optical techniques can measure absolute perfusion values in tissue. Nevertheless, evaluation of relative perfusion variations with high spatial and temporal resolution can be used to compare dynamic changes in organ perfusion between pathophysiological groups and provide physicians with important unbiased information about hemodynamics.

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Compliance with ethical standards

All experiments were performed in accordance with the ethical recommendations of the International Association for the Study of Pain, Directive 2010/63/EC of the European Parliament and Council on the protection of animals used for scientific purposes, and were reported in accordance with the ARRIVE guidelines 2.0. The study was approved by the local ethical committee of Sechenov University, minutes № 23–22 as of November 17, 2022 [57].

Conflict of interest

The authors declare no conflict of interest.

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