Photoplethysmographic Imaging of Hemodynamics and Two-Dimensional Oximetry

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Received December 29, 2021; revised January 30, 2022; accepted February 4, 2022

Abstract—The review of recent papers devoted to actively developing methods of photoplethysmographic imaging (the PPGI) of blood volume pulsations in vessels and non-contact two-dimensional oximetry on the surface of a human body has been carried out. The physical fundamentals and technical aspects of the PPGI and oximetry have been considered. The manifold of the physiological parameters available for the analysis by the PPGI method has been shown. The prospects of the PPGI technology have been discussed. The possibilities of non-contact determination of blood oxygen saturation SpO2 (pulse saturation O2) have been described. The relevance of remote determination of the level of oxygenation in connection with the spread of a new coronavirus infection SARS-CoV-2 (COVID-19) has been emphasized. Most of the works under consideration cover the period 2010–2021.

Keywords: photoplethysmography, photoplethysmographic imaging, remote photoplethysmography, imaging photoplethysmography, oxygenation, oximetry, saturation, blood oxygen saturation, heart rate, SARS-CoV-2, COVID-19

DOI: 10.1134/S0030400X22080057

INTRODUCTION

This work is devoted to the analysis of modern works in a relatively new rapidly developing field of two-dimensional photoplethysmographic imaging of hemodynamic phenomena. In the Russian-language literature, the method of photoplethysmographic imaging (the PPGI) has several synonymous names, which corresponds to the English-language names: “remote photoplethysmography” (rPPG) and “imaging photoplethysmography” (iPPG) or “photoplethysmographic imaging” (the PPGI). In the text of this review, we will use a single name: the PPGI-imaging, consonant with the name of the techniques similar in the problem being solved to such as, e.g., laser Doppler imaging, speckle contrast imaging, and thermographic imaging, which are actively used to study the processes of regulation of peripheral hemodynamics.

The wide spread of the photoplethysmography method is facilitated by its high relevance in the development and design of the wearable devices that control the frequency and variability of the heart rhythm. Currently, most sports watches and bracelets determine heart rate parameters precisely on the basis of the analysis of the photoplethysmographic signal using software processing of the signal reflected from the LED-photodiode optocoupler.

As far as we know, this work is the first attempt to generalize the results of research on the PPGI and two-dimensional non-contact oximetry in the Russian-language scientific literature. The existing review [1] discusses the results of work on the PPGI of the face without describing two-dimensional oximetry. Therefore, it seems relevant to review modern achievements in the field of the PPGI as a method characterized by high spatial resolution, informativity in terms of the diagnostics based on the registration of a complex of optical, biomechanical, and physiological properties of a biological object, and being comparatively simple and inexpensive in hardware and software implementation.

The review structure is constructed in such a way that the method of classical contact photoplethysmography is briefly considered first and then the method of remote two-dimensional the PPGI, as a newer direction of its development. The paper describes the physical basis of the formation of the PPGI signal; the technical aspects of the method (the choice of the emitter spectrum, the camera type, and the area of interest); the physiological parameters determined by the PPGI method; and the possibilities
of non-contact oximetry based on the PPGI technology are considered. In conclusion, the deductions on the main directions of the considered problems of the PPGI and oximetry are formulated.

Thus, the main purpose of this review is to summarize the results of the modern foreign and Russian scientific works related to the development of the non-contact oximetry and the PPGI of hemodynamic phenomena on the surface of living objects with a description of the basic physical, technical, and physiological aspects of the methods. Most of the works under consideration were completed during the period 2010–2021. The authors did not set the task of an exhaustive description of all aspects of the PPGI, so the review can rather be considered as an introduction to this subject area.

**From the Contact Photoplethysmography to the PPGI**

Before the invention of the photoplethysmography as a method of measuring volumetric blood filling in different parts of the body, various designs of the mechanical plethysmographs were used (based on the Greek plethysmos—magnification + grapho—to write, to depict). The first work by A. Hertzman, which describes a photoelectric method for measuring blood pressure, dates back to 1938 [2]. By illuminating a section of tissue with red and infrared light, which after reflection or passage, was recorded by a phototransformer, the authors found its pulsations, which were associated with the pulsating nature of blood flow in microcirculatory vessels. Since blood has optical properties different from those of the surrounding tissue, this ultimately causes a change (modulation) in the intensity of transmitted or reflected light. Initially, the described method was called “photoelectric plethysmography,” but the name “photoplethysmography” became established over time. The further studies have confirmed that the photoplethysmographic (PPGI) signal is linearly related to changes in blood volume in microcirculatory vessels [3]. Depending on the locations of the radiation source and receiver, there are two main types of PPGI sensors: (1) a “transmission” one (transmission sensor) when the source and receiver are located on opposite sides of the object of study, e.g., of a finger; (2) a “reflection” one (reflective sensor) when the source and receiver are located on the same side of the object of study. Progress in the improvement of the PPGI sensors of these types is described in the review [4].

In early 1975, Aoyagi et al. proposed a new noninvasive method for determining arterial blood oxygen saturation based on photoplethysmography [5], which was called pulse oximetry. The pulse oximetry method made it possible to simultaneously assess the pulse rate and the level of oxygen saturation in the blood (SpO₂—saturation pulse O₂, pulse saturation O₂), as a result of which it gained popularity because of its low cost, simplicity, and practicality. In the early 1990s, the pulse oximetry became a mandatory procedure during anesthesia monitoring according to international standards [6].

The first studies indicating the possibility of non-contact photoplethysmographic imaging of the skin blood flow using a video camera were started in 1996 [7]. Later, in 2000, the authors presented one of the first the PPGI systems based on a charge-coupled device (CCD) and demonstrated its applicability for assessing local changes in tissue blood filling [8]. Since then, the number of publications related to the PPGI methods has been rapidly increasing every year. So, according to the Google Scholar search engine (https://scholar.google.ru/) at the end of 2021, the number of publications with the keyword “plethysmography” over the past 6 years exceeds the number of similar publications over the previous 100 years (Fig. 1a), and the number of publications with the keywords “imaging photoplethysmography” and “remote plethysmography” over the past 3 years exceeds the number of similar publications over the previous 10 years. At the same time, the relative weight of the number of the publications on the topic of photoplethysmographic imaging in the total number of the publications devoted to the photoplethysmography has been steadily increasing every year over the past 10 years (Fig. 1b), which indicates an increase in the relevance of the photoplethysmographic imaging.

The first measurements of the PPGI signal were made in 2007 and 2008 by Takano [9] and Verkruysse [10] using a standard camera in the face area. The authors proposed a method that detects fluctuations in the complexion from the set of predefined areas of interest. This method was used for the monochromatic [9] and color images [10]. At the same time, the PPGI signal was formed by simply averaging the intensity of the pixels that make up the area of interest. In 2011, A. Kamshilin and colleagues presented the results of the PPGI in each pixel of the video frame, which provided a high spatial resolution of the method [11, 12]. At the same time, the methods of the non-contact determination of SpO₂-oxygen saturation of blood using a video camera have been developing since 2005 [13, 14]. Currently, the diagnostic capabilities of the methods of the contact and non-contact PPGIs are expanding and include not only the assessment of the heart rate, determination of the vascular wall properties, the level of blood oxygen saturation [15–19], but also the determination of the pulse wave propagation time, as well as its spatial distribution over the skin surface [20, 21].
1. PHYSICAL FOUNDATIONS OF THE PHOTOPLETHYSMOGRAPHY METHOD AND ALTERNATIVE IDEAS ABOUT THE REASONS FOR THE FORMATION OF A PHOTOPLETHYSMOGRAM

In accordance with the modern most common concepts, the PPGI signal is formed as follows: part of the incident radiation of a LED or laser penetrates into biological tissue, undergoing many acts of absorption and scattering, reaches the tissue vascular layer, in which it is partially absorbed by blood (mainly erythrocytes) and partially returns from the depth of the tissue to the surface and is detected by a photodiode or a video camera as a PPGI variable signal [22–24]. As a result, the change in the PPGI signal intensity is associated both with a change in volumetric blood filling and with the absorption and scattering of light by the tissue structures. It is considered in most works on PPGI that the pulsating variable part of the PPGI signal (or the “AC” component) is determined by the volume variation associated with the pulse, whereas the non-pulsating slowly changing part of the PPGI signal (or the “DC” component) is determined by a slow change in the blood filling of the venous and arterial vessels, as well as changes in the intercellular fluid volume and other physiological processes affecting the biological tissue optical properties.

The growing interest in PPG applications and in particular in the PPGI has raised certain questions among researchers related to the physical mechanisms of the formation of the PPGI signal [25, 26]. The factors potentially influencing the PPGI signal shape such as: changes in blood volume, movement of the blood vessel wall, and orientation of erythrocytes, were considered [27–30]. Traditionally, it is believed that the PPGI signal mainly depends on changes in the blood volume in the vessels. However, there have been disagreements on this issue relatively recently [31–33]. For example, in [32, 34], the authors assumed that pulse fluctuations of the transmural arterial pressure in the larger arteries mechanically deform the dermal tissue structure, which leads to periodic changes in the density of the capillaries in the dermis papillary layer and the light scattering coefficient of the skin layers. Thus, the formation of the PPGI signal can be caused precisely by the mechanical compression of the tissue by the blood vessels. This assumption explains why it is possible to obtain an PPGI signal even in the case when optical radiation does not penetrate to the depth of the location of the blood vessels. The depth of the light penetration into the skin for the PPGI in the reflection mode is really of little importance [35], while there are ambiguities in its definition [36].

When performing the PPGI, it is necessary to consider that the depth of the penetration of LED radiation into the skin will be less than when using a contact reflection sensor with the same type of emitters since the power density of the radiation incident on the skin surface decreases. The PPGI with simultaneous regulation of pulse oscillations at wavelengths of 530 and 810 nm showed the possibility of determining the blood perfusion from two functionally and morphologically different layers of the cutaneous microcirculatory bloodstream: the superficial subpapillary plexus and the deeper plexus at the junction of the dermis and hypodermis [37]. The study [38] shows that the difference in the geometry of the contact and non-contact methods does not affect the possibility of the SpO2 calibration based on the camera. A deep understanding of the origin of the PPGI wave form measured distantly is necessary to establish an unambiguous con-
nection of this signal with the variations in the arterial blood volume in the blood vessels and cardiac contractions and to exclude the influence of other physiological factors.

Analyzing the skin spectral properties, it can be concluded that the use of the red and near-infrared radiation in PPGI studies makes it possible to obtain information about the hemodynamics in the blood vessels, while the effect of pulsation in certain areas on the skin surface manifests itself in the form of small mechanical vibrations, i.e., it is a ballistographic effect [33]. These effects can have global and local origin. The global ballistographic effects can be associated, e.g., with the movement of the head caused by the release of blood into the aorta and the local effects, e.g., with the skin surface slope because of the passage of a large artery under the measurement area. The systematic studies of the PPGI signal origin have shown that ballistographic effects occur mainly when using inhomogeneous and non-orthogonal illumination (incident light is not perpendicular to the skin surface). The degree of influence of the global and local effects on the resulting PPGI signal is difficult to quantify.

The ballistographic effects differ significantly from the effects of the fluctuations in blood volume. They also produce a pulsating signal, but the phase of the resulting signal and its morphology can differ from the signals that arise because of the effects of fluctuations in blood volume. Averaging of the areas influenced by the various types of effects can lead to distortion of the resulting PPGI signal. The latter is of particular importance when determining the level of oxygen saturation in the blood when only the change in blood volume should be considered, but not the ballistic effect. Therefore, it is recommended to carefully select the skin surface areas for the oximetry in such a way as to exclude from consideration the areas that perform mechanical vibrations under the influence of the cardiac contractions and the pulse wave propagating through the vessels.

2. TECHNICAL ASPECTS OF THE PPGI

2.1. The Principle of Visual Data Formation in the PPGI

The initial data in the PPGI method are two-dimensional matrices of the values proportional to the intensity of the light reflected from various objects of the background and the object being measured itself, and, in the end, incident on the camera matrix (Fig. 2). The penetration depth of the radiation used to illuminate the object will depend on its wavelength, so the type of the PPGI images can be significantly different when using the sources of different spectral composition, e.g., the green and red ones (Figs. 3a and 3b). A change in the living object reflective properties over time, e.g., with a variation in the skin volumetric blood filling, leads to a corresponding change in the matrix data from the camera. As a result, a set of two-dimensional matrices evolving over time, i.e., an array of three-dimensional data, is formed over a selected period of time. The signal from each element of the two-dimensional matrix recorded for a long time can be considered as a one-dimensional time series, which is mathematically processed by the spectral or statisti-

![Diagram](image-url)
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Fig. 3. PPG imaging of the palmar part of the hand: (a) image of the object using LEDs with a central wavelength of 530 nm (green), (b) image of the object when using LEDs with a central wavelength of 660 nm (red), (c) imaging of the pulsation amplitude in each image pixel, and (d) imaging of the amplitude of pulsations averaged within square areas of 10 × 10 pixels.

Fig. 3. PPG imaging of the palmar part of the hand: (a) image of the object using LEDs with a central wavelength of 530 nm (green), (b) image of the object when using LEDs with a central wavelength of 660 nm (red), (c) imaging of the pulsation amplitude in each image pixel, and (d) imaging of the amplitude of pulsations averaged within square areas of 10 × 10 pixels.

Another common method of the visualization of PPGI data is the construction of a pulsation amplitude map, implemented by calculating the power of the spectral components of the image intensity fluctuations corresponding to the blood pulsation in the vessels in each image pixel (or a group of them). As a result, the imaging parameter in such images is the spatial distribution of the pulsation amplitude either in each pixel (Fig. 3c) or averaged over a group of pixels (Fig. 3d).

In general, the pulsation amplitude map can indicate the areas of the object surface that are the most informative from the point of view of the possibility of determining the heart rate. An important aspect of the high-quality PPGI is the reduction of the mechanical movements of the object during shooting both using the special experimental techniques and the image processing algorithms that compensate for the movements and track the displacements of the object contours from frame to frame.

2.2. Selection of the Emitter Spectrum Area

It was noted in early works [10, 39] that the pulsation amplitudes of the green channel signal of the optical analysis methods in most cases. As a result, each image pixel corresponds to a statistical, spectral, or other numerical parameter, each value of which corresponds to the color (shade) of the palette (pseudo-palette). The use of a pseudo-palette makes it possible to transform an image from shades of gray into a color image with a certain color gradient of the transition of the values from the minimum to the maximum ones, which increases the visibility of the PPGI results.

The spectral processing of photoplethysmographic data using a fast Fourier transform either from each individual pixel or from a group of pixels (zones of interest) is most common. To determine the heart rate, the spectral components corresponding to the cardiac range of 0.5–2 Hz can be distinguished in the amplitude spectrum. The frequency at which the maximum amplitude of the spectral components is detected will correspond to the frequency of heart rate (HR) averaged over the measurement time. Such a method can be implemented either with spatial averaging of the determined heart rate values over the object entire surface or with averaging of the object surface in selected areas characterized by a high signal-to-noise ratio.
color camera (RGB camera) are the largest in comparison with those of the red and blue ones. This is due to the fact that oxyhemoglobin and deoxyhemoglobin of the blood absorb the radiation from the green region more intensively than those from the red and blue ones (Figs. 3a and 3b demonstrate the difference in the images obtained when the skin is illuminated by the radiation from the green and red regions of the spectrum). For this reason, an increase in blood volume in the vessels leads to a more significant decrease in the reflected component in the green channel compared with those the red and blue ones. Thus, the greatest variation of the pulsation signal is observed in the color image green channel, therefore, the digital frame green channel is used in most works on the PPGI using natural light and an RGB camera to determine the pulse and the green lighting is used in the case of a monochrome camera. While the sources in the red and near infra-red spectrum are most often used for tissue probing in the contact PPGI, the use of the radiation from the green spectral region gives the highest signal-to-noise ratio for the PPGI despite the fact that red and near infrared radiation penetrate deeper into the tissue. Using the blue region of the spectrum for the illumination gives a high noise level [39].

2.3. Using Different Types of Cameras

As a rule, the systems based on either a monochrome camera with external illumination (more often green) or systems based on an RGB camera have found the greatest use for the PPGI. To avoid shadows on the object, they provide high intensity and uniformity of the illumination. For this purpose, an annular LED illuminator mounted on the camera lens so that their optical axes are aligned is often used (Fig. 2b) [40, 41]. The use of a polarizing filter on the camera makes it possible to reduce the effect of optical interference on the PPGI signal [42].

In the last 5 years, the possibilities of using budget RGB cameras and webcams (as varieties of RGB cameras with, as a rule, lower resolution and shooting speed) for non-contact heart rate monitoring have been intensively investigated. Compared to the PPGI system based on a monochrome camera, these systems have a low cost and ease of implementation since external lighting is often used for the ambient illumination [10, 43, 44]. The use of the PPGI signal normalization algorithm makes it possible to eliminate the influence of the ambient light intensity instability on the measurement results [45]. When using the data from the RGB cameras, the signal is divided into R- (red), G- (green), and B- (blue) channels, which are analyzed independently making it possible to examine a body part in three different wavelength ranges. The use of the three channels expands the variability of the algorithms for extracting pulse oscillations and ways to reduce motion artifacts on the basis of a combination or independent analysis of R, G, and B components. Some rarer types of artifacts, e.g., the irregularity of the speed of shooting video frames and the influence of the width of the time window for averaging heart rate are considered in [46].

2.4. Selection of the Measurement Area of Pulse Oscillations on the Human Body Surface

In early studies, the area of the wrist or fingers, which were fixed to reduce motion artifacts since they were the ones that introduced significant interference, was used [13, 39, 47].. It was shown later that using a monochrome [9] and RGB camera [10] makes it possible to determine the heart rate from the face area. To measure the pulse on the face, it is recommended to use the cheeks, forehead, and chin or the entire face area [48, 49]. At the same time, recording data from the forehead area leads to the oscillations with the highest amplitude [10, 50, 51]. The area around the lips can also be used to record the pulse [52].

The advantage of using the face area to determine the pulse is that this area is always open. In addition, the algorithms for identifying informative areas on the face that reduce movement artifacts, e.g., the Viola–Jones object detection method [53] or algorithms that use Haar signs [54] have now been developed while it is necessary to manually isolate informative areas on other parts of the body [55].

2.5. Defining the Area of Interest

When manually selecting the region of interest and its size [11, 49, 56], the resulting PPGI signal is usually determined by averaging the values from the group of pixels that make up this area. The disadvantages of this method include: firstly, the influence of a subjective factor, because it is difficult to ensure the reproducibility of the location of the area because of the object micro-movements and secondly, the problem of choosing the optimal size of the area of interest when averaging over small spatial areas is that the signal will be more stepwise and when averaging over a large area, the resulting signal can include both informative and uninformative pixels or the pixels that give a signal of the motion artifacts or spatial reflection. Testing of the existing automatic algorithms for identifying the area of interest, e.g., such as the Viola–Jones or Lucas–Canade face detectors, shows that these algorithms can not accurately track the selected area when the face is rotated. The only best solution to these problems does not yet exist, so various compromise options are offered.

A method for determining the informative pixels using the simultaneous analysis of the entire face area and automatic detection of the areas of interest with the highest pulse amplitude and using adaptive matrices [57] or variations in the brightness of the areas [58] is proposed. Earlier, the same authors proposed using a continuous wavelet transform to identify the pixels
with a pronounced pulse signal [59]. The concept of “super pixels” is proposed for the automatic selection and tracking of the most informative sections [60–63]. The nonparametric Bayesian algorithm can be used for image segmentation and the autoregressive models can be used to remove artifacts associated with light flickering [64]. A stochastic method for selecting points from the cheek region is proposed to estimate the PPGI waveform using the Bayesian approach [65]. One of the modern approaches for automatic identification of areas of interest is the use of convolutional neural networks [66].

3. ASSESSMENT OF THE HUMAN BODY MAIN PHYSIOLOGICAL PARAMETERS BY THE PPGI METHOD

3.1. Extraction of the Pulse Signal from the PPGI Data

Heart rate is one of the most important markers of the health of the cardiovascular system, the control of which is vital for some patients, so the issue of continuous and non-contact heart rate monitoring remains relevant [67]. A relatively high signal-to-noise ratio in the cardiac oscillation spectral region (0.5–2 Hz), which makes it possible to isolate a useful signal against the background of optical interference and video camera matrix noise, contributes to the increased interest in non-contact determination of heart rate parameters. Contactless heart rate monitoring using the PPGI has already had success in clinical application, e.g., for bedside monitoring of infants and premature infants [68], as well as for monitoring sleep disorders.

After the signal have been recorded, as a rule, pre-processing of frames is performed in order to isolate the pulse wave, for which band-pass signal filtering [69], Fourier filtering [49], continuous wavelet filtering [59], or adaptive filtering techniques [70] are used as standard. Another approach is to purposefully isolate a weak pulse signal using the Eulerian amplification algorithm [71, 72].

The authors in [56] proposed to minimize the influence of motion artifacts and highlights in the original signal by removing the trend and separating the RGB data using independent analysis of the R, G, and B components. This method is based on the separation of a multidimensional signal into its independent initial components, while it is assumed that the initial signals are statistically independent of each other and are not Gaussian. The CHROM method, in which the RGB data vectors are combined into two orthogonal signals, normalizing the color channel and reducing the level of interference associated with the highlights on the skin surface, has been developed on the basis of the selection of the main components [18, 56, 73–75].

A normalized RGB space with an orthogonal plane of the signal tone is used in other works [69, 76]. It is possible to exclude the components that are not related to the pulsations of the RGB data caused by the heart rate and the passage of a pulse wave through the vessels on the basis of the analysis of RGB data in such a space.

In addition to the use of various signal filtering methods and algorithms, the use of neural networks is also becoming increasingly popular for determining the heart rate. For this purpose, the deep or shallow learning is used. The input data for this method are either the original video recording of the signal (spatial-temporal data) or preprocessed one-dimensional data from the selected area. The deep learning approach was used for the first time in [48], in which the data was pre-processed by the method of selecting the main components and transmitted to the neural network. Later, more advanced deep learning neural networks, in which the input data was the pre-selected signals in the time domain [77–79], as well as neural networks where the original video recording is used as the input signal [80], appeared. In a number of studies, the influence of various external conditions and interference was modeled in order to assess the degree of distortion of the PPGI data and ways to minimize them [81, 82].

As a response to the change in the current epidemiological situation associated with the spread of the new coronavirus infection SARS-CoV-2, there are works in which the pulse is determined for people in a mask [83], while the works in which the issue of the minimum required area on the face necessary to determine the pulse with a high degree of certainty was studied [84].

3.2. Two-Dimensional Mapping of the Pulsation Amplitudes

In addition to the task of extracting the heart rate signal using the video recording, the spatial mapping of the pulse wave signal amplitudes on the body selected part is also an urgent task. The laser speckle contrast imaging [85] and laser Doppler imaging devices are commercially available for measuring blood perfusion maps [86]. The limitations of such devices include the need for complex measurement protocols and the high cost of hardware and software. The PPGI method does not have these disadvantages, therefore, it can take in the future a firm place as a method of two-dimensional mapping of peripheral hemodynamics parameters.

The first algorithms for constructing a spatial distribution map of the amplitude of blood flow fluctuations included splitting the original image into separate regions. Next, a three-dimensional matrix of size $x \times y \times z$ was formed with the subsequent execution of the Fourier transform (see Section 2.1). One of the
first who used the two-dimensional imaging of blood flow fluctuations were the authors [13]. When performing visualization, the spatial resolution of this method decreases in inverse proportion to the size of the square within which the data is averaged. A new method for obtaining the PPGI images with a high spatial resolution equal to the resolution of the camera used was presented [11]. This method is based on the synchronous amplification of recorded video frames (coherent de-modulation method). It is shown that the method can be used to study the regulation of peripheral blood circulation [87]. To form a reference signal, it is recommended to use an area of 1 cm² or more [88]. The possibility of the PPGI and determination of the heart rate in newborns under room lighting has been demonstrated [89].

Another approach for constructing perfusion maps is the multisensory PulseCam method, in which the signal from each pixel is compared with the reference PPGI signal from the finger sensor. Thus, the method also allows obtaining perfusion maps with a spatial resolution equal to the full resolution of the camera matrix [90, 91]. The coherent demodulation method and the multisensory method use a reference pulse signal, which in the first method is obtained by averaging the signals from the pixels in the selected area and in the second method, — using an independent PPGI sensor. A high noise level is recorded in each individual pixel, so the reference signal is used to increase the signal-to-noise ratio.

In [92], the PPGI unit with a green backlight was used to photograph the face and detect systemic scleroderma. The results of the study of the healthy and sick subjects revealed that the asymmetry of the pulse amplitude on the face is manifested in systemic scleroderma, which is clearly demonstrated by the presented two-dimensional images of amplitude maps. Also, the two-dimensional PPGI demonstrates an increase in blood flow at local heating [93].

The possibilities of the PPGI in the study of cerebral microcirculation of the rat brain were demonstrated [94]. The authors used in experiments an anesthetized rat in which an autopsy of the cranium was previously performed without damage to the meninges. In response to painful stimulation, an increase in the pulse amplitude was recorded in various areas of the rat’s open brain. Further studies have demonstrated the possibilities of the PPGI of pulsation amplitude and pulse wave delay time for the analysis of cerebral blood flow during open brain surgery. It was found that this procedure well visualizes changes in the cerebral blood supply caused by surgical intervention [95]. A decrease in the uniformity of the pulsation amplitude maps in migraine was demonstrated [96].

The PPGI data can be used in conjunction with the electrocardiography to measure the pulse wave propagation time and its spatial distribution, e.g., in the facial region [20, 97]. For three-dimensional mapping of pulse amplitudes, an alternating illumination at several wavelengths, e.g., 660 and 880 nm is used, which makes it possible to visualize the distribution of the pulse wave signal along the depth [41].

### 3.3. Low-Frequency Rhythms of the PPGI Signal Oscillations

The relevance of the study of the photoplethysmogram low-frequency oscillation rhythms of less than 0.5 Hz is due to the fact that the corresponding spectral components can characterize changes in the vascular tone determined by the influence of various physiological mechanisms of regulation such as: endothelial, neurogenic, myogenic, and respiratory. Thus, a decrease in the amplitude of the PPGI signal spectral components in the neurogenic, myogenic, and respiratory ranges was found in a group of patients with diabetes mellitus [98]. In [99], a decrease in the amplitude of the vasomotion with a frequency of about 0.1 Hz in the presence of allergy was recorded with the PPGI. The analysis of the PPGI low-frequency component demonstrated the informativeness in the study of the respiratory waves [100] and vasomotion [101–103]. The high informativeness of the PPGI low-frequency region for determining the functions of the sympathetic nerves and the sympathetic part of the autonomic nervous system is emphasized [104]. It is shown that the hemodynamics slow fluctuations in the frequency range from 0.003 to 0.04 Hz have a high level of correlation of more than 0.9 with a very low-frequency component of the spectrum of the heart rate variability over most of the observation interval. The possibility of using the PPGI system to detect changes in a person’s psychoemotional and energy-deficient states is discussed [105, 106]. Devices are being designed and methods are being developed for the PPGI diagnostics of the disorders of the low-frequency rhythms of the peripheral vascular tone regulation [107, 108].

Recent studies found a high phase coherence of the PPGI signal low-frequency component and the pressure with the heart rate variability low-frequency component [109, 110]. There is a significant phase synchronization of the PPGI signal and the laser Doppler flowmetry (LDF) signal in the low-frequency range (0.0095–0.1 Hz) [111]. A high coherence of the PPGI signal phases between the arm and leg is recorded especially in the endothelial range [112] despite the fact that this range corresponds to a local rather than a central mechanism of vascular tone regulation. Setting a constant breathing rhythm of 0.04, 0.1, and 0.25 Hz led to an increase in the phase synchronization of the PPGI and LDF signals [113, 114]. The results in [112] indicate that the PPGI signal low-frequency part is influenced or interrelated with the heart rate variability and respiratory rate. At the same time, as shown in [115, 116], the PPGI signal low-frequency part largely causes fluctuations in the skin temperature of the
was also assessed using the PPGI in the process of
tolic peaks [121, 122]. The cardiac rhythm variability
accuracy of measuring the time intervals between sys-
tole of the webcam video image, which limits the
disadvantages of this method include the low sampling
produced in 2011 [49] and confirmed in 2013 [121]. The
non-contact assessment of the heart rate variability
based on the PPGI using a webcam was first intro-
duced in 2011 [49] and confirmed in 2013 [121]. The
zones located near the nose and the bridge of the
note are highly informative from the point of
view of the possibilities of determining the heart rate
when using a webcam [124].

3.4.3. Vascular disorders and allergies. The ability
of the PPGI to characterize vascular skin lesions, e.g.,
wine stains, was demonstrated in [41, 125]. Other
authors demonstrated the use of the PPGI to assess
the skin allergic reactions to the use of antihistamines
[126] and to assess mechanical injuries [127]. the
PPGI of the near-surface arteries and veins is imple-
mented for the skin of varying degrees of pigmentation
using color and monochrome cameras and an illumin-
ator operating alternately at two wavelengths [128].

3.4.4. Merging of photoplethysmographic imaging
and infrared thermography methods. Since the consid-
ered PPGI method provides information about the
hemodynamic processes on the skin surface, it is of
interest to compare it with alternative methods of
hemodynamic imaging primarily such as laser Dop-
pler imaging, speckle contrast imaging, and infrared
thermography. Currently, there are only a few works in
this area, e.g., [118]. At the same time, in our opinion,
it looks promising not only to compare the results of
the PPGI and other hemodynamic imaging tech-
niques as independent methods, but also to detect the
possibilities of their fusion when the whole becomes
larger than the sum of its parts and the so-called emerg-
cence manifests itself, e.g., the fusion of the PPGI and
infrared thermography methods. Considering the
found relationships between the PPGI signal fluctua-
tions and the fluctuations in skin temperature [115,
116], it is possible to calibrate the PPGI data express-
ing them not in relative units, as is usually done, but in
absolute units related to the temperature measured in
degrees by the calibrated thermal imaging cameras.
The fusion of the PPGI and thermographic imaging
methods can also make it possible to determine the
skin thermophysical properties by the delay time of the
temperature signal spectral components relative to
those of the PPGI signal [116] and thus detect and

![Fig. 4. Photoplethysmographic imaging of low-frequency rhythms of blood flow oscillations (wavelength of illumination 530 nm) [119, 120].](image-url)
monitor therapy of, e.g., trophic lesions on the skin surface.

Currently, the PPGI and thermography methods in most cases are considered to a greater extent as independent ways of obtaining information about the living object properties [129, 130]. Various schemes of the mutual arrangement of the thermal imaging camera and the camera for the PPGI are proposed (Fig. 5) [131].

Noteworthy is [132], in which a significant number of calculations related to statistical and spectral processing of the dynamic thermograms and two-dimensional photoplethysmograms was performed in order to build a map of the informative features of the object of study. The main idea of this work was to study the spatial-temporal and spectral characteristics of images. The dynamics of the signal in each pixel of the images formed a time series, which at the processing stage was subjected either to the calculation of the signal statistical characteristics in a given pixel (standard deviation, mathematical expectation, coefficients of asymmetry, kurtosis, etc.) or spectral processing using the Fourier transform.

As a result, the video image was compressed for the selected time interval into a single frame, which is a map of the spatial distribution of statistical or spectral features characteristic of this time window. Next, the spatial characteristics of the constructed maps such as the contrast of the selected area of interest or the intersection of histograms (the sum of the minimum column of the matching ones of the model histogram and the histogram of the analyzed image area) were evaluated. A “similarity map” was built on the basis of the results of the analysis of the intersection of the histograms. Thus, first, the processing of the data from each pixel in the time domain was performed and the maps of statistical or spectral features were constructed, and then the features of the spatial distribution of such maps were evaluated, i.e., a spatial-temporal approach to the processing of the PPGI frames and dynamic thermograms was implemented.

As a result, it was found that the determination of the spatial contrast of the sign map makes it possible to identify the informative areas a pulsating component of the photoplethysmogram and an area with a noise component characteristic of the background or uninformative areas of interest. The intersection of the histograms and the similarity map can be used to segment parts of an object and separate it from the background. At the same time, the maps based on the PPGI data are more suitable for highlighting the silhouette of a living object and the dynamic infrared image data is more suitable for highlighting its contour. Nevertheless, despite the considerable number of the calculations carried out, so far the advantages of merging the two technologies of the PPGI and thermographic imaging have not been demonstrated, since the results of processing the corresponding images were considered independently.

4. NON-CONTACT DETERMINATION
OF BLOOD OXYGEN SATURATION LEVEL

4.1. The First Mention of Non-Contact Oximetry

Over the past 15 years, various approaches have been developing to implement non-contact oximetry or to determine the level of saturation of blood with oxygen using a video camera. In most cases, SpO₂, which is pulse oxygen saturation of blood, is determined. The first results, which describes a method for recording pulse oscillations using a monochrome CMOS camera and sequential shooting at several wavelengths (660, 810, and 940 nm) in order to calculate the SpO₂ level, were published in 2005 [13]. In the same year, using synchronous activation of a monochrome CMOS camera and sequential shooting at several wavelengths (660, 810, and 940 nm) in order to calculate the SpO₂ level, were published in 2005 [13]. In the same year, using synchronous activation of a monochrome CMOS camera and sequential shooting at several wavelengths (660, 810, and 940 nm) in order to calculate the SpO₂ level, were published in 2005 [13]. In the same year, using synchronous activation of a monochrome CMOS camera and sequential shooting at several wavelengths (660, 810, and 940 nm) in order to calculate the SpO₂ level, were published in 2005 [13].
the values for the venous one by 3–10% compared with those in the contact pulse oximeter readings. In [133], the applicability of the wavelengths of 880 and 760 nm for non-contact determination of the oxygenation level was investigated: the influence of the pulse rate and the oxygenation level on the measurement results was determined.

4.2. Physical Principle of Noninvasive Oximetry, Determination of SpO₂

The physical basis of the method is the presence of differences in the dependence of the light absorption coefficient of oxygenated (HbO₂) and deoxygenated (Hb) forms of hemoglobin of erythrocytes, which make up the bulk of the blood corpuscles. In accordance with the Bouguer–Lambert–Beer law, the absorption of light by a substance in a solution is proportional to its concentration, therefore, when the blood oxygen saturation level changes, the amplitude of the PPGI signal at the selected wavelength also changes. The presence of other blood and skin chromophores can lead to a decrease in the accuracy of determining the oxygenation level by optical methods. The absorption spectra of the oxygenated (HbO₂) and deoxygenated (Hb) blood in the visible and near infrared wavelength ranges are shown in Fig. 6. When choosing a pair of wavelengths for implementing non-invasive oximetry, it should be considered that the camera registering reflected and backscattered radiation has its own nonlinear spectral dependence of sensitivity usually significantly decreasing at the more than 800 nm. In the simplest case, two wavelengths λ₁ and λ₂ are chosen to implement an oximeter in the absorption spectrum (Fig. 6) so that the Hb absorption coefficient exceeds the HbO₂ absorption coefficient at the wavelength λ₁ and the inverse ratio is true at the wavelength λ₂. With a decrease in the SpO₂ oxygenation, the HbO₂ concentration decreases and Hb increases, which is accompanied by a decrease in the reflection coefficient of the radiation at the wavelength λ₁ and its increase at the wavelength λ₂. When SpO₂ increases, opposite changes in the reflection coefficient are expected.

To determine the percentage of oxygen in the blood using a video camera, as well as for a pulse oximeter, the classical method of the ratio of the pulse signals of the photoplethysmograms recorded simultaneously at two wavelengths is used:

\[ RR = \frac{AC_{λ_1}/DC_{λ_1}}{AC_{λ_2}/DC_{λ_2}} \]

\[ \text{SpO₂} = A \cdot RR + B, \]

where \( RR \) is the parameter of the spectral pulsations, \( AC_{λ_1} \) and \( DC_{λ_1} \) are the variable and constant components of the PPGI signal at the wavelength \( λ_1 \), \( AC_{λ_2} \) and \( DC_{λ_2} \) are similar components at the wavelength \( λ_2 \), and \( A \) and \( B \) are the coefficients of Eq. (2). During the production and adjustment of the oximeter, the parameter \( RR \) is calibrated (coefficients \( A \) and \( B \)) as a percentage of the oxyhemoglobin saturation, determined by direct readings of gas analysis of blood samples, in the arterial blood (SaO₂) [36]. Thus, the value of the pulse saturation SpO₂ can be obtained by measuring \( RR \) and substituting the measured value into the calibration equation (Eq. (2)). The noninvasive oximetry method is considered in more detail in the lecture article by D.A. Rogatkin [36].

4.3. Non-Contact Imaging of the Oxygenation Level

Considering the change in the epidemiological situation caused by the emergence of a new coronavirus infection SARS-CoV-2 (COVID-19), characterized by a decrease in blood oxygenation in the pulmonary form of the disease, non-contact determination of the SpO₂ level becomes particularly relevant. The appropriate systems can be used, e.g., at the checkpoints with high traffic of people to identify potential carriers of infection.

For the first time, the non-contact determination of the oxygenation level in the facial area using a visible range camera and using room lighting was demonstrated in a 2013 paper [134]. For this purpose, two monochrome CCD cameras, on the lens of each of which narrow bandpass filters were fixed to capture the PPGI signals at a wavelength of 520 and 660 nm, were used. Fourier filtering was used to isolate the pulse oscillations, and these signals were averaged over the entire area of interest to determine the saturation level. After having calibrated the camera on the basis of the results of the simultaneous measurements with a pulse oximeter and a camera during breath retention...
of one of the subjects, it was possible to obtain fairly accurate data on the blood oxygen saturation level using the camera on a group of subjects.

4.3.1. Variants of hardware implementation of imaging of the oxygenation level. To measure oxygenation, it is mandatory to record the pulse at least at 2 wavelengths of the optical range. When using monochrome cameras, the authors apply discrete switching on of the camera with each alternating switching of lighting [14, 47, 125, 133, 135]. To do this, a hardware trigger that switches the camera and lighting sources, as which semiconductor LEDs are used, is used. The disadvantage of this method is the low frame rate for each lighting channel, which at the moment is 20 frames per second. Recording speed limit is due to the delay in switching the trigger to turn on the camera and, first of all, to the launch of an array of LEDs. However, it should be noted that the attempts to increase the frame recording frequency for this recording method have not been purposefully carried out, although it is discussed that the recording frequency affects the SpO2 evaluation result [133]. In addition, it is important to assess the effect of the switching time of the LEDs on their brightness, which is associated with the warming up of the LEDs. To exclude changes in the LED brightness, some authors turn on the LEDs before the experiment and let them warm up at the selected switching frequency [125]. An alternative option can be the use of mechanical shutters of various types, providing luminous flux amplitude modulation.

Another approach is to use several monochrome cameras with narrow-band light filters. In this method, a narrow-band filter is attached to each camera to isolate the necessary wavelength in the spectrum of the reflected light flux. At least 2 monochrome cameras identical in technical characteristics are used to measure oxygenation. White light [38], indoor or natural lighting [134, 136] can be used for illumination. There is also a method for processing data from 3 monochrome cameras in order to increase the accuracy of determining SpO2 [25, 26, 137, 138] or 4 cameras [139]. Multiple camera application allows you to simultaneously record a signal with a frame rate that is limited only by the parameters of the camera itself, which is an advantage over a single camera with switching lighting. At the same time, the processing of the results becomes more complicated because of the difference in the spatial location of the cameras and the need to use the algorithms to combine two images into one. The cost of a measuring system based on the use of several monochrome cameras also increases. The influence of the location of cameras on the accuracy of the SpO2 determination is investigated [138].

In addition to monochrome cameras, RGB cameras with indoor or natural lighting are used [140–145]. The increasing popularity of RGB cameras is associated with their high prevalence and low cost. To implement this method, it is enough to have an inexpensive, compared to monochrome one, RGB camera or even a webcam. External lighting can be used as probing radiation. To determine the oxygenation level, the original video data is divided into separate R, G, and B components, by which the amplitude variation of the pulse oscillations is determined. Further, using the ratio of the amplitudes mainly between the R and G components, the RR (Eq. (1)) is calculated by which the oxygen saturation of the blood is determined (Eq. (2)). For the sake of the method simplicity, one has to face up to the relatively low camera sensitivity and the limitation of the choice of wavelengths of incident radiation, since only broadband components of red, green, or blue illumination can be analyzed.

4.3.2. Selection of probing radiation wavelengths. When determining the oxygenation level, the choice of wavelength has the character of a compromise between the camera technical characteristics and the optimal absorption of oxygenated (HbO2) and deoxygenated (Hb) forms of hemoglobin to achieve the maximum amplitude of the blood flow optical signal. Most monochrome cameras have a high spectral sensitivity in the 500–700 nm band, which decreases sharply outside this area. This type of the spectral characteristics of monochrome cameras imposes a restriction on the use of a traditional pair of wavelengths of 660 and 940 nm because of the low signal-to-noise ratio in the region of 940 nm [13]. Because of this limitation, the specified pair of wavelengths is practically not used for non-contact determination of the oxygenation level.

The use of green light makes it possible to record the pulse oscillations of the highest amplitude and with a sufficiently pronounced shape, which provides an acceptable level of accuracy in determining SpO2, since it is in the wavelength range of 530–550 nm that the blood hemoglobin has maximum absorption [11]. Thus, the authors of [136] reported an experimentally detected difference in the pulse amplitudes for a pair of blue (460 nm) and green (520 nm) illumination, and the authors of [134], using a pair of the wavelengths of green (530 nm) and red (660 nm) radiation, determined the variation of oxygen in the blood in the range of 92–98% with a deviation of several percent compared to that of the contact finger pulse oximeter. When using a color camera, the green and red channels were also used in most works to determine SpO2. The theoretical models considered in [26, 139] allowed us to plot the change in the PPGI signal relative amplitude both on the incident radiation wavelength and on the percentage of oxygen in the blood. Theoretical calculations confirm that in the range of 530–550 nm (green), the pulse signal amplitude has a maximum value compared to those at other wavelengths. At the same time, a change in the percentage of oxygen in the blood leads to a slight change in the pulse amplitude at green light, which is somewhat inconsistent with the interpretation of the results in [11]. In this case, it is recommended to use the radia-
tion in the range from 600 to 1000 nm for non-contact oximetry. At the same time, the pulse amplitude recorded by the camera is several times lower, but the percentage change in the oxygen level in the blood leads to a more significant change in it compared to that in the green area. Thus, to increase the accuracy of determining SpO₂, it is necessary to use such pairs of wavelengths at which changes in the oxygen content cause significant changes in the reflected signal amplitude and at the same time, fall into the region of the greatest spectral sensitivity of the camera.

Also earlier in [135], the authors, using modeling based on the Bouguer–Lambert–Beer law, calculated the dependence of the RR change on saturation for several combinations of wavelengths and demonstrated that the pairs: red (660 nm) and infrared (IR) (880 nm), orange (610 nm) and IR (880 nm) demonstrate a change in RR by several times in the saturation range of 70–100%, while the pairs: green (528 nm) and IR (880 nm), blue (470 nm) and IR (880 nm) demonstrate 10 times smaller changes in RR. The authors themselves [135] chose a pair of orange (λ = 610 nm) and IR (λ = 880 nm) for their research, because in orange light, the amplitude of the pulsations is greater than that in red one. For practical purposes, e.g., in clinical settings, it is necessary to be able to measure SpO₂ in a wider range (at least 80–100%). In the study [38], it was experimentally demonstrated that using a pair of the radiation of wavelengths of 675 and 842 nm at low temperature or with low oxygen content in the blood, it is possible to determine the oxygenation level with high accuracy.

CONCLUSIONS

1. Advantages and limitations of the PPGI method compared to the contact photoplethysmography method.

1.1. Advantages lie in the following features:
—recoding and color imaging of two-dimensional spatial distribution of blood pulsation amplitudes in surface vessels, spatial distribution of blood oxygenation SpO₂, and pulse wave propagation time, which is relevant, e.g., in the study of hemodynamics in cerebral cortex vessels or hemodynamics in the area of the face and extremities;
—non-contact determination of heart rate and heart rate variability averaged over a large surface area, which is relevant, e.g., when monitoring vital parameters of newborns and patients with allergic, trophic, and thermal skin lesions for whom contact of the sensor with the skin is not desirable.

1.2. The limitations of the PPGI method are:
—in a lower signal-to-noise ratio when recording a reflected signal by a camera compared to recording using a photodiode;
—in the possible influence of the reflected background radiation detected by the camera, but not related to the hemodynamics of the object of study on the results;
—there is less versatility in choosing a camera detector (compared to that in a photodiode case), which as a rule has uneven and lower spectral sensitivity in the near infrared wavelength range compared to that in the visible one.

2. Physical and technical aspects of the PPGI:

2.1. The main reasons for the PPGI signal modulation can be: a change in the volume of circulating arterial and/or venous blood; movement of the wall of blood vessels, which creates a different level of compression of the skin and changes its dispersing properties; a change in the orientation of red blood cells in the blood flow depending on the speed of its movement into the systole and diastole; and a change in the ratio between the concentrations of oxygenated and deoxygenated blood hemoglobin. The mechanical movements of the body as a result of heartbeats and respiration can be considered either as an informative component for the ballistography based on the photoplethysmography or as a hindrance.

2.2. To visualize the pulsations of arterial blood flow, it is recommended to use the spectrum green region, because in this case, a more intense absorption of green light by the blood (compared with that when using the illumination in the red and blue regions of the spectrum) leads to a greater decrease in the signal amplitude with the arrival of each successive volume pulsation. A detailed consideration of the features of the propagation of light of various wavelengths in biological tissue is carried out in the monograph [146].

2.3. One of the disadvantages of the PPGI method is the lack of the binding of the signal amplitude to the absolute units of measurement. The solution to this problem can be the development of an optimal PPGI signal calibration method used by all researchers and allowing us to compare the results obtained in various scientific groups. In this case, one of the calibration options can be a comparison of the temperature and PPGI data.

3. Medical application of the PPGI method.

Among the areas of medical application of the PPGI method, it is possible to note studies devoted to the diagnosis of systemic scleroderma, changes in cerebral blood flow, migraines, and allergies on the skin surface. Considering the change in the epidemiological situation associated with the spread of the new coronavirus infection SARS-CoV-2 (COVID-19) and characterized by a decrease in the blood oxygenation level in the pulmonary form of the disease, the non-contact determination of the level of the blood pulse saturation with oxygen SpO₂ by the PPGI methods becomes particularly relevant.

Most of the work on the PPGI is aimed at developing methods for analyzing the frequency, heart rate variability, and amplitude of blood pulsations in the
frequency range of 0.5—2 Hz. The use of the low-frequency range (0.005—0.5 Hz) in the field of biomedical diagnostics remains relatively little studied today, therefore, a further increase in the number of works in this direction is predicted.

4. Ways of further development of the PPGI technology.

4.1. The main ways to improve the PPGI method include increasing the sensitivity and signal-to-noise ratio for the cameras in the visible and near infrared ranges; increasing the incident radiation monochromaticity; reducing the mirror-reflected component and the influence of optical interference unrelated to the object of study; improving the algorithms for tracking the object of study compensating for motion artifacts; and development of the algorithms for the mathematical analysis, methods for calculating statistical and spectral parameters of the PPGI images in the time, spatial, and spatial-time domains with color mapping of these parameters.

4.2. One of the promising directions is the combination of the PPGI method with other methods of the imaging of the hemodynamic phenomena such as: infrared thermography, laser Doppler imaging, or speckle contrast imaging, or the contact methods (pulse oximetry, electrocardiography, rheography, etc.). Moreover, the joint synchronized use of the methods can lead to the appearance of new features that are not available to each of the methods separately. For example, the determination of the biological tissue thermophysical properties by the delay of the temperature signal spectral components relative to the PPGI signal or the method of coherent demodulation, described in Section 3.2.

4.3. The development of the PPGI and oximetry methods using webcams is likely to lead to their widespread distribution and implementation on the basis of the cameras of smartphones and other mobile devices.

FUNDING

A review of the possibilities of determining oxygenation based on the photoplethysmographic imaging technology was carried out with the support of a grant from the President of the Russian Federation for state support of young Russian scientists, Candidates of Sciences of the Russian Federation MK-140.2021.4; a review of the possibilities of thermal imaging data verification by the photoplethysmographic imaging method was supported by the Russian Science Foundation (grant no. 21-75-00035).

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Translated by N. Petrov