

PulseCam: High-resolution blood perfusion imaging using a camera and a pulse oximeter

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Abstract

Measuring blood perfusion is important in medical care as an indicator of injury and disease. However, currently available devices to measure blood perfusion like laser Doppler flowmetry are bulky, expensive, and cumbersome to use. An alternative low-cost and portable camera-based blood perfusion measurement system has recently been proposed, but such camera-only system produces noisy low-resolution blood perfusion maps. In this paper, we propose a new multi-sensor modality, named PulseCam, for measuring blood perfusion by combining a traditional pulse oximeter with a video camera in a unique way to provide low noise and high-resolution blood perfusion maps. Our proposed multi-sensor modality improves per pixel signal to noise ratio of measured perfusion map by up to 3 dB and improves the spatial resolution by 2 – 3 times compared to best known camera-only methods. Blood perfusion measured in the palm using our PulseCam setup during a post-occlusive reactive hyperemia (PORH) test replicates standard PORH response curve measured using laser Doppler flowmetry device but with much lower cost and a portable setup making it suitable for further development as a clinical device.

1. Introduction

Blood perfusion is the flow of blood to the end organs and tissues through the blood vessels in the body. Blood flow (or perfusion) is vital in ensuring oxygen delivery to the cells and in maintaining metabolic homeostasis. Measuring peripheral perfusion, i.e. perfusion of the blood just underneath the skin surface is important in both medical and surgical fields [1] including as-

essment of peripheral perfusion in critical care, tissue viability in plastic, reconstructive, and burn surgery as well as for wound assessment.

Blood perfusion varies from one tissue site to another, and can also change over time due to varying metabolic demands, and so a spatial map of blood perfusion over time, i.e. a three-dimensional quantity, is usually measured. Laser speckle contrast imaging [2] and laser Doppler imaging [3] devices are commercially available for measuring blood perfusion maps. But, these devices are (i) bulky and require specialized measurement protocols, (ii) are not generally used in operating rooms or at the bedside in the intensive care units (ICUs) as they may cause interference to the ongoing care and discomfort to the patients, and (iii) are too expensive to be routinely used for outpatient care. Recently, an alternative camera-based blood perfusion imaging system has been proposed [4] which has the potential to be a portable, cost effective and non-contact blood perfusion imaging system. However, current camera-only blood perfusion imaging algorithms produce noisy and low-resolution perfusion maps and thus are rarely used in clinical settings.

In this paper, we propose a new multi-sensor modality, named PulseCam, for measuring blood perfusion by combining a traditional pulse oximeter and a video camera in a unique way to provide low-noise and high-resolution peripheral blood perfusion maps. Blood flow (or perfusion) is the rate of change of blood volume at any tissue segment over time. Both a pulse oximeter and a camera measures the blood volume change over time at an external site through optical means. A pulse oximeter is a simple spot measurement device which can measure blood volume waveform reliably from one body location, but cannot simultaneously take spatial

measurements from a large region of the skin surface. On the other hand, a camera provides noisy measurements of blood volume waveform, but a camera can simultaneously take spatial measurements from a large region of the imaged skin surface owing to its unique spatial dimension: each pixel on the image sensor can be considered as a pulse oximeter which is virtually (from a distance) attached to the corresponding location on the imaged skin surface and provides an independent but noisy measurement of the blood volume waveform from that location. PulseCam combines these two disparate devices by using the reliable blood volume waveform from a pulse oximeter as a reference, and then correlating this reference waveform with the noisy blood volume waveform obtained from each pixel in the camera to produce low noise and high-resolution perfusion maps of any imaged skin surface.

Several researchers have recently shown the feasibility of camera-only blood perfusion imaging [4, 5, 6]. Most of these works, however, rely on excessive spatial averaging (around 20×20 to 100×100 pixel block) to reduce camera's quantization noise, shot noise and readout noise, and suppress motion artifact before estimating the blood perfusion maps, thereby compromising the achievable spatial resolution. In this paper, our proposed PulseCam requires a minimal spatial averaging over only 4×4 pixel block, and produces blood perfusion maps with $0.5 - 3$ dB higher signal to noise ratio (SNR) per pixel block compared to the state-of-the-art technique that only relies on camera recordings [4] to measure blood perfusion maps. We also validated PulseCam functionally by conducting a standardized post-occlusive reactive hyperemia (PORH) test on 4 healthy individuals and found the derived blood perfusion measurements to be in agreement with published PORH-test response curve measured using a laser Doppler flowmetry device [7].

Our main contributions in this paper are: (i) a novel multi-sensor approach for measuring blood perfusion maps by combining a camera and a pulse oximeter, (ii) a signal model for blood perfusion imaging taking into account differences in camera operating parameters, and (iii) a maximum likelihood (ML) estimator for three-dimensional blood perfusion maps which combines the pulse oximeter and the camera measurements.

In Section 2, we propose a signal model for multi-sensor blood perfusion imaging and in Section 3 we develop an ML-estimator for estimating three dimensional blood perfusion maps. In Section 4 we discuss the experimental setup and data collection protocol used for validating PulseCam, in Section 5 we summarize our results, and in Section 6 we discuss our key insights, major challenges, and future direction of research.

2. Blood perfusion signal model

Blood flow (or perfusion) is the rate of change of blood volume at any tissue segment over time. The rate of change of blood volume is proportional to the amplitude of the blood volume waveform. Therefore, to measure spatial blood perfusion maps, we estimate the amplitude of the blood volume waveform at any tissue location over time. In this section, we will develop a blood perfusion signal model based on light-tissue interaction.

When light falls on the skin surface, it is partly absorbed by the skin and the underlying tissue, and is partly reflected back and recorded by the camera sensor imaging the skin surface. Let us assume that the incident light intensity $I(\vec{x})$ does not change over time. Here, \vec{x} denote the location on the skin surface corresponding to pixel $\vec{x} = \{x, y\}$ on the camera. Then, the camera recorded video signal over time can be modeled as

$$V(\vec{x}, t) = I(\vec{x})(b(\vec{x}) + c(\vec{x}, t)) + w(\vec{x}, t) \quad (1)$$

where the skin reflectance is separated into two component: first component $b(\vec{x})$ is due to light absorption by skin surface and tissue underneath and is time invariant, and the second component $c(\vec{x}, t)$ is due to light absorption by the chromophores in the blood, and is time varying due to pulsatile changes in the blood volume in the microvasculature underneath the skin surface. Finally, $w(\vec{x}, t)$ is the noise added during the camera acquisition process [8].

The subsurface light absorption component due to pulsatile changes in blood volume can be decoupled as $c(\vec{x}, t) = a(\vec{x}, t)p(t - \tau(\vec{x}))$ where $a(\vec{x}, t)$ is the amplitude of the blood volume waveform $p(t)$ and is different at different locations and can also change over time due to temporal variations in blood perfusion. The blood volume waveform signal $p(t)$ is assumed to be delayed by different time $\tau(\vec{x})$ at different locations \vec{x} on the skin surface. The noise term $w(\vec{x}, t)$ is dominated by (i) camera's quantization noise, readout noise and photon shot noise, and (ii) motion artifact. Therefore, the camera-recorded video signal can be modeled as

$$V(\vec{x}, t) = QI_0(b(\vec{x}) + a(\vec{x}, t)p(t - \tau(\vec{x}))) + w(t) \quad (2)$$

where Q is the multiplication factor due to camera's exposure and aperture settings. Also, to simplify the model, we have assumed the spatial variation in light intensity to be minimal, and replaced it with mean illumination value I_0 over the imaged skin region.

Based on the proposed signal model, in the next section, we will present an estimator for the blood

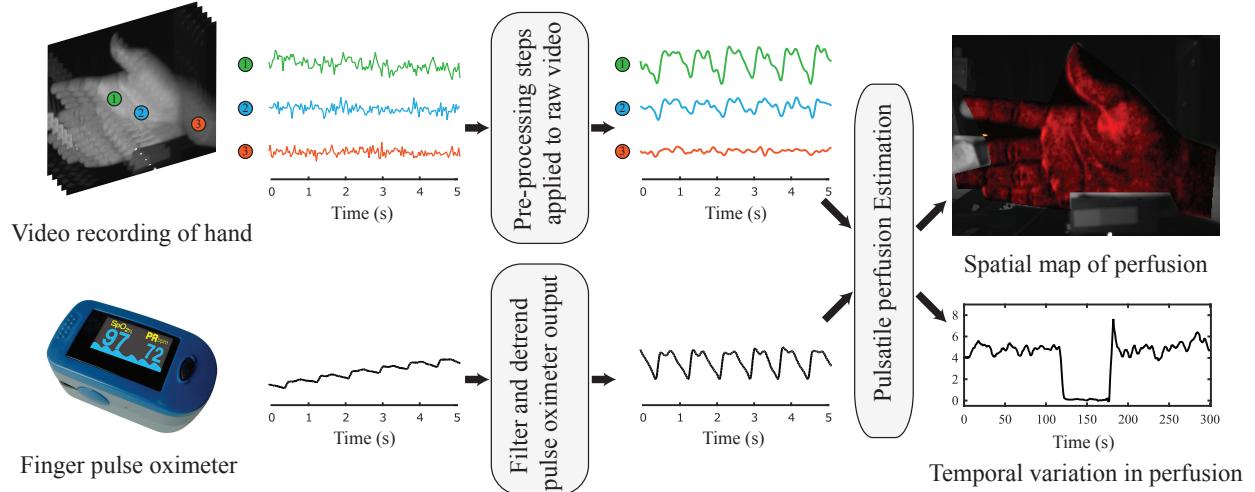


Figure 1: Processing steps involved in PulseCam: Raw video frames from the camera are processed to estimate noisy blood volume waveform from each camera pixel, and the simultaneously recorded reliable blood volume waveform signal $p(t)$ which is reliably measured using a pulse oximeter.

perfusion map $a(\vec{x}, t)$ given that we have noisy camera recording $V(\vec{x}, t)$, and the underlying blood volume waveform signal $p(t)$ which is reliably measured using a pulse oximeter.

3. PulseCam blood perfusion estimation

Blood perfusion estimation using PulseCam involves a series of three preprocessing steps to be performed on the raw video recordings from the camera. After the preprocessing steps, the processed video from the camera and pulse oximeter recordings are combined together to estimate the blood perfusion map. The processing steps involved in PulseCam is illustrated in Figure 1.

As a first pre-processing step, we apply an $M \times M$ spatial mean filter on the raw video recording $V(\vec{x}, t)$ from the camera to reduce the contribution of camera sensor noise on the measurement from each pixel. The choice of M will be a trade-off between the spatial resolution of resulting perfusion map and the desired SNR of the perfusion estimate per $M \times M$ pixel block.

Then, as a second preprocessing step, the spatially averaged video is filtered temporally using a bandpass filter having passband between 0.5 Hz - 5 Hz with unity passband gain to obtain $V_{AC}(\vec{x}, t)$. This temporal filtering removes the slowly varying surface reflection component in the video recordings, and only allows the subsurface reflection component due to blood volume changes to be retained. The simultaneously recorded

blood volume waveform from the pulse oximeter is also filtered temporally using the same bandpass filter with passband between 0.5 Hz - 5 Hz.

In the third preprocessing step, we use a standard Grayscale chart having reflectance b_{std} placed adjacent to the skin surface and then we compute a normalization factor N by averaging the camera recording from the pixels imaging the Grayscale chart. Then, we divide $V_{AC}(\vec{x}, t)$ by N to get a normalized video signal $V_N(\vec{x}, t)$. The normalization factor N is equal to QI_0b_{std} and it removes the effect of changes in camera exposure and aperture settings Q and intensity of incident light I_0 on the eventual perfusion estimate.

If we assume that the noise in the camera measurement is white and Gaussian, then the maximum likelihood estimator for the perfusion $a(\vec{x}, t)$ will be

$$\hat{a}(\vec{x}, t)_{ML} = \max_{D(\vec{x})} \langle V_N(\vec{x}, t - D(\vec{x})), p(t) \rangle_T \quad (3)$$

where $\langle \cdot, \cdot \rangle$ is the inner product between vectors. The delay $D(\vec{x})$ is used to align the reference blood volume waveform signal $p(t)$ with the blood volume waveform signal at location \vec{x} . Based on the proposed blood perfusion imaging model, $\hat{a}(\vec{x}, t)_{ML} = \frac{a(\vec{x}, t)}{b_{std}}$, where b_{std} is the reflectance of the standard grayscale chart used in the normalization step. The inner product in the above equation is defined over the time window T over which the perfusion is assumed to be constant.

Readout or thermal noise generally follows a Gaussian distributed, and under sufficient illumination, camera's shot noise also follows a Gaussian distribution.

Quantization noise is uniformly distributed, but if we assume a minimum of $10 - 20$ pixel in a pixel block over which spatial averaging is done ($M \approx 4$), then due to central limit theorem, quantization noise can also be modeled as Gaussian. Noise due to motion artifact is generally spiky and difficult to model, and is not considered here.

The signal-to-noise ratio of the above ML estimate is same as the SNR of the signal of interest and can be estimated as

$$\text{SNR}(\vec{x}, t) = \frac{\hat{a}_{ML}^2(\vec{x}, t)}{\widehat{\text{Var}}(V_N(\vec{x}, t) - \hat{a}_{ML}(\vec{x})p(t - D(\vec{x}))_T} . \quad (4)$$

4. Experiments

We present two sets of experiments. In the first set, we did a controlled experiment to characterize the average SNR per pixel block for blood perfusion imaging using our proposed PulseCam and using camera-only method [4] as a function of ADC quantization level and spatial mean filter size M . In the second set of experiment, we conducted a standard post occlusive reactive hyperemia (PORH) test on 4 healthy individuals to measure the change in their blood perfusion in the palm before, during and after an occlusion event. The experimental procedures involving human subjects described in this paper were approved by the institute review board (Rice IRB Reference number 841313 – 1).

The experimental setup consists of a monochromatic CMOS camera (Grasshopper GS3-U3-23S6M-C from Point Grey) on which we put a green optical filter having an optical passband between 520 nm to 560 nm. Using green optical filter improves the SNR of camera-based blood volume waveform as the absorption spectra of hemoglobin peaks at a wavelength of around 530 nm. The camera is operated at 30 fps, with automatic gain control and gamma correction turned off, and exposure time is set at 12 ms. For all the experiments, the camera records a video of the palm rested on a hand support. Occlusion of blood flowing to the palm is done using a standard pressure cuff put on the arm of the same hand. We simultaneously record a blood volume waveform from the middle finger of the other hand using Biopac system's MP150 data acquisition unit as a reference pulse oximeter.

5. Results

5.1. Perfusion imaging SNR

Figure 2a shows the variation of average SNR (in dB) per pixel block (computed using Equation (4)) of

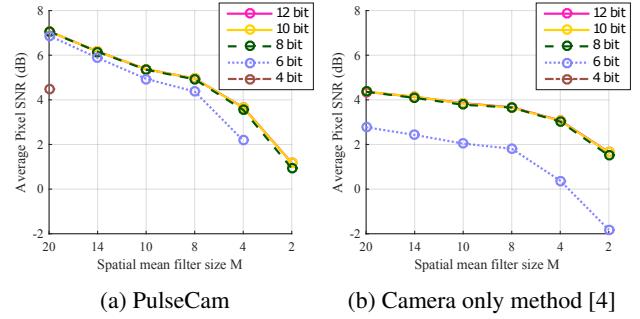


Figure 2: **PulseCam SNR improvement:** Variation of average per pixel SNR (in dB) as a function of the spatial mean filter size M using (a) PulseCam and (b) state-of-the-art camera-only method [4]. PulseCam provides an SNR improvement of around 0.5 – 3 dB per pixel block.

the estimated blood perfusion map using PulseCam as spatial averaging filter size M^2 is varied from 20×20 down to 2×2 , and Figure 2b shows the corresponding per pixel block SNR of the estimated blood perfusion map using the camera-only method [4]. The time window T for perfusion estimate is set to 10 sec for both methods. During this controlled experiment, the motion artifact due to hand movement is kept small so that the only source of noise are due to the camera acquisition process. On an average, we see an SNR improvement of 0.5 – 3 dB per pixel block in the blood perfusion map derived from PulseCam compared to camera-only method.

5.2. PORH Results

For this experiment, we recorded 2 min video of the palm before the occlusion to get a baseline estimate of perfusion, a total of 1 min video under occlusion, and a 2 min video post occlusion for 4 individuals (3 male and 1 female) having different skin tones (1 Caucasian, 1 Asian, and 2 brown).

Figure 3 shows the temporal variations of the average perfusion in the palm (averaged over all pixel block in the palm region excluding fingers) during the PORH test for 4 healthy individuals before, during and after the occlusion. The time window T for perfusion estimate is set to 5 sec with a 4 sec overlap (sliding window) to track sudden changes in perfusion due to occlusion. These PORH response curve agrees well with PORH curve estimated using laser Doppler perfusion monitoring [7]. The average perfusion before the occlusion (marked as RF) is lower than the average perfusion just after the release of the occlusion (marked as MF), and

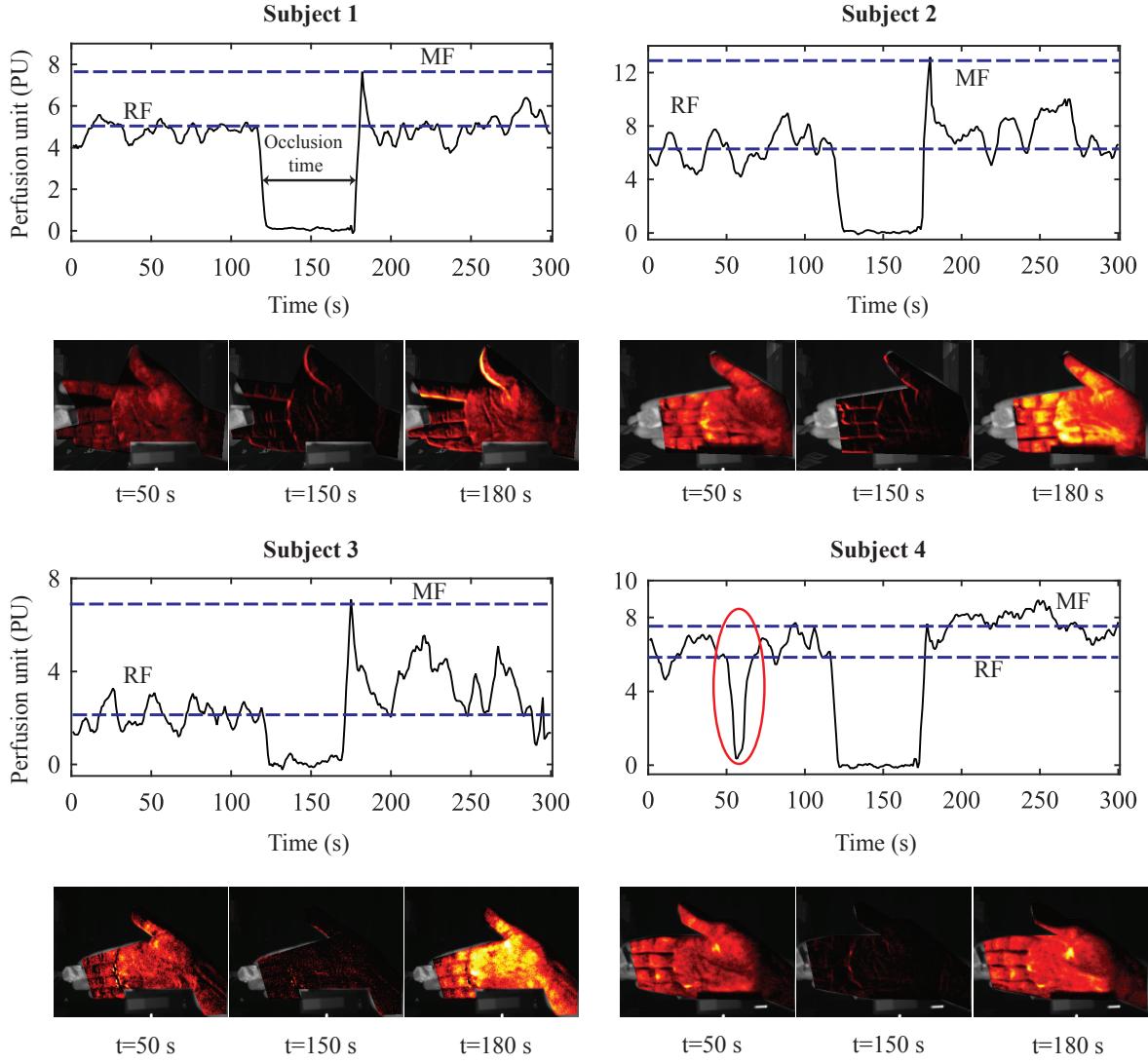


Figure 3: PORH Response using PulseCam: Temporal variation of average blood perfusion in the palm of 4 subjects before, during and after an occlusion (post occlusive reactive hyperemia or PORH), RF is the resting flux (perfusion) before the occlusion, MF is the maximum flux just after the occlusion; Below the PORH curve for each subject are image inlets showing spatial map of blood perfusion using false color over the palm computed at different time instances [$t = 50$ s (before occlusion), $t = 150$ s (during occlusion), $t = 180$ s (just after the occlusion)], brighter regions signify higher blood perfusion. The sudden dip in estimated blood perfusion within the red marking (Subject 4) is due to recording artifact in the reference pulse oximeter derived blood volume waveform.

the ratio MF/RF also has diagnostic value for assessing arterial health. The variations in the average perfusion around the baseline could be due to rhythmic oscillations in the vascular tone caused by changes in smooth muscle constriction and dilation, and are usually 4 – 10 cycles per minute (cpm). For subject 4, the sudden dip in estimated blood perfusion at around $t = 60$ sec is due to recording artifact in the reference pulse oximeter derived blood volume waveform.

Below PORH response curve of each subject are image inlets showing the two dimensional spatial map of the blood perfusion using false color at specific times during the occlusion and release cycle. These spatial maps are generated with $M^2 = 4 \times 4$ spatial averaging filter. Darker pixel blocks in the spatial map show lower blood perfusion, whereas brighter pixel blocks show higher blood perfusion. Evidently, using PulseCam, one can easily visualize the temporal dynamics of blood flow in the hand before, during and after the occlusion.

6. Discussion

Since a pulse oximeter is readily available as a low-cost medical device, and CMOS cameras of various form factors, sizes, and specifications are available commercially, therefore our proposed PulseCam can be used as a portable and low-cost blood perfusion monitor. PulseCam could have applications in monitoring wound healing in diabetic patients, in plastic surgery to monitor skin flap perfusion after a microvascular reconstructive procedure, and to assess the skin's endothelial function. In many scenarios, like in intensive care unit (ICU) and operating rooms (OR), existing systems like laser Doppler imaging cannot be readily used, whereas PulseCam, by virtue of being passive and operable from a distance, is suited for such scenarios. Thus, successful validation of PulseCam for critical care may open up the possibility of real-time blood perfusion and microcirculatory monitoring at the bedside during surgery and in ICU care — the need of which has been identified by several critical care researchers [9].

Though, in this paper we proposed using pulse oximeters as a reference blood volume waveform, other waveforms such as electrocardiograph (ECG), arterial blood pressure (ABP) waveform, or even a camera-based imaging photoplethysmography (iPPG) can be used as a reference. The accuracy of the estimated blood perfusion maps depends significantly on how reliably the reference waveform is acquired. For camera-only methods like [4], reliable estimation of reference blood volume waveform is challenging due to motion artifacts, or due to limited blood perfusion in the imaged

region (e.g. due to occlusion), and therefore the overall accuracy of perfusion maps obtained using camera-only methods suffer.

In this work, we have shown the feasibility of using PulseCam to obtain low noise and high-resolution perfusion maps by combining a camera and a pulse oximeter. There are widespread applications for real-time perfusion monitoring, and as a next step, we will extend PulseCam to more challenging scenarios like monitoring patients in ICU and for monitoring wound recovery.

References

- [1] J. Allen and K. Howell, "Microvascular imaging: techniques and opportunities for clinical physiological measurements." *Physiological measurement*, vol. 35, no. 7, pp. R91–R141, Jul. 2014.
- [2] K. R. Forrester, J. Tulip, C. Leonard, C. Stewart, and R. C. Bray, "A laser speckle imaging technique for measuring tissue perfusion." *IEEE transactions on bio-medical engineering*, vol. 51, no. 11, pp. 2074–84, Nov. 2004.
- [3] A. Serov, B. Steinacher, and T. Lasser, "Full-field laser Doppler perfusion imaging and monitoring with an intelligent CMOS camera." *Optics express*, vol. 13, no. 10, pp. 3681–9, may 2005.
- [4] A. A. Kamshilin, V. Teplov, E. Nippolainen, S. Miridonov, and R. Giniatullin, "Variability of microcirculation detected by blood pulsation imaging." *PloS one*, vol. 8, no. 2, p. e57117, Jan. 2013.
- [5] A. A. Kamshilin, S. Miridonov, V. Teplov, R. Saarenheimo, and E. Nippolainen, "Photoplethysmographic imaging of high spatial resolution," *Biomedical Optics Express*, vol. 2, no. 4, pp. 996–1006, Mar. 2011.
- [6] U. Rubins, V. Upmalis, O. Rubenis, D. Jakovels, and J. Spigulis, "Real-time photoplethysmography imaging system," in *15th Nordic-Baltic Conference on Biomedical Engineering and Medical Physics (NBC 2011)*. Springer, 2011, pp. 183–186.
- [7] F. Morales, R. Graaff, A. J. Smit, S. Bertuglia, A. L. Petoukhova, W. Steenbergen, P. Leger, and G. Rakhorst, "How to assess post-occlusive reactive hyperaemia by means of laser Doppler perfusion monitoring: application of a standardised protocol to patients with peripheral arterial obstructive disease." *Microvascular research*, vol. 69, no. 1-2, pp. 17–23, jan 2005.
- [8] M. Kumar, A. Veeraraghavan, and A. Sabharwal, "DistancePPG: Robust non-contact vital signs monitoring using a camera," *Biomedical Optics Express*, vol. 6, no. 5, p. 1565, May 2015.
- [9] C. A. den Uil, E. Klijn, W. K. Lagrand, J. J. Brugts, C. Ince, P. E. Spronk, and M. L. Simoons, "The microcirculation in health and critical disease." *Progress in cardiovascular diseases*, vol. 51, no. 2, pp. 161–70, Jan. 2008.