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Smartphone Pulse Oximetry Using Two-tone Camera-based Photoplethysmography

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***Abstract*—SpO_2 monitoring is a non-invasive measure of blood oxygen saturation levels, particularly important to monitor respiratory and heart circulatory diseases. Incorporating SpO_2 measurement into smartphones has the potential to improve patient access and cost-effective monitoring of health status. This paper presents a new smartphone camera-based photoplethysmography technique for peripheral oxygen saturation measurement without external devices. Most existing methods have limitations for medical use due to a lack of accuracy and insufficient validation. Unlike other approaches, the proposed method uses the smartphone frontal camera and two-tone image patterns to control the emitted light spectrum, allowing for calibration and future clinical application. The method was tested on healthy subjects and demonstrated a strong correlation with reference values ($r = -0.742$). This approach improves camera-based oximetry acquisition and a validation protocol is also proposed for calibration, highlighting its clinical potential.**

***Index Terms*—Pulse Oximetry, Camera-based Photoplethysmography, Smartphone, mHealth**

I. Introduction

The need for frequent SpO_2 measurements as a biomarker of disease, during hospitalization and ambulatory care, especially for silent hypoxemia, has led to a renewed interest in research and development of new methods and devices to pervasively measure this physiological variable [1], [2]. Modern smartphones include hardware that can be explored for the development of sophisticated biomedical applications. SpO_2 levels can be estimated using photoplethysmography (PPG), which is an optical technique that measures small fluctuations of reflected (or transmitted) radiation in vascularised regions of the skin. The light source and detector needed in this technique can be provided by smartphone hardware components, which led to the development of Camera-based PPG (cbPPG) techniques.

Current approaches of cbPPG, use the flash as a light source and the main (rear) camera as the photodetector [3], allowing the measurement of SpO_2 and heart rate [4]. This method presents certain difficulties, for instance, the uncertainty related to the spectral characteristics of the light source (flash) prevents a rigorous calibration process and therefore an accurate measurement of SpO_2, limiting its clinical use.

To address this limitation, we propose a two-tone cbPPG solution based on the frontal camera and the smartphone display. This method is particularly advantageous and innovative, given that it enables fine control of the emitted light wavelength, and constitutes a more cost-effective solution for non-invasive SpO_2 measurements. In this paper, Sections II and III present the background and the state-of-the-art of cbPPG. Section IV details the proposed approach and Section V summarizes the experimental results. Section VI concludes the paper, providing an outline of the main findings.

II. Background

A. Photoplethysmography

PPG is an optical technique that can be used to detect variations in the volume of blood vessels as a result of the cardiac cycle. The measured signal is composed of a static component (DC), resulting from the surrounding tissues, and a pulsatile component (AC) caused by the oxygenated blood flow through the tissue capillaries. The passing blood absorbs part of the emitted radiation and a variation in the light intensity is detectable. PPG devices require a light-emitting diode (LED) and a photodetector, that can be placed either in transmittance PPG mode, where the appendage is between the light source and the detector or in reflectance PPG mode where they are located on the same side.

On a standard smartphone, there are two possibilities to support the implementation of cbPPG: using the back camera and flash [4], [5], or the front camera and screen [6]. In both cases, the camera is used as a photodetector, and the cbPPG signal can be derived from the video signal. When a finger

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is placed on the camera with a white light source on the side, the video feed displays a red image, due to the DC component of the signal, with small fluctuations of the pixel intensities caused by the blood flow.

B. SpO_2 Measurement Principle

Building upon the Beer-Lambert law, the blood concentrations of deoxyhemoglobin (ρ_{Hb}) and oxyhemoglobin ($\rho_{Hb_{O2}}$) can be calculated using two different light wavelengths. Most devices use red (∼660 nm) and infrared (∼940 nm) to maximize the absorptivity difference between Hb and Hb_{O2} [7]. To estimate SpO_2 levels using PPG signals from light sources of frequencies $f1$ and $f2$, a modulation ratio, R, is calculated from the respective AC and DC components [1], [8]:

$$R = \frac{\mathrm{AC_{f1}}}{\mathrm{DC_{f1}}} / \frac{\mathrm{AC_{f2}}}{\mathrm{DC_{f2}}} \tag{1}$$

By calibrating the R values with accurate oxygen saturation levels, the SpO_2 levels can be estimated using the information extracted from the PPG signals.

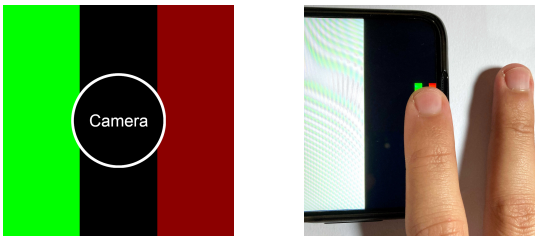
III. Related Work

To tackle the fact that the attenuation of two different frequencies is required to estimate SpO_2, most approaches use two different camera RGB color channels to capture the required frequencies. Many camera-based applications that make use of the camera flash [5], [7] could address the issue by applying wavelength filters to the light source, but this approach would involve the use of an add-on.

In [9], the authors suggest that the green (G) channel absorbs more than the blue (B) channel, so they used both G and red (R) channels. However, even with these approximations, the accuracy is still not comparable to pulse oximeters. Another approach is to use the phone screen illumination and the frontal camera, allowing for a tunable frequency of the emitted light, which improves the signal quality [6].

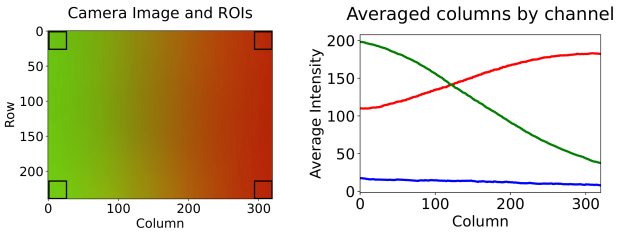
Because the characteristics of the light used are extremely important in pulse oximetry, using the screen as a light source allows greater control over the emitted light than when using the flash LED. Holz used a spectrum analyzer to characterize the spectral intensity of the display, as well as the camera sensitivity [6].

Another challenge is the variability of smartphones, with differences in specifications (e.g., resolution, frame rate), settings (e.g., auto-focus, light exposure control), and sensors. This variability can affect the quality of the acquired data, making it essential to calibrate the method for each device. User behavior also plays a critical role, with proper finger placement and minimal finger movement required to avoid variations in the DC and AC components of the signal. Any changes in pressure or contact area can also affect the model's accuracy.



(a) (b)

Fig. 1: (a) Layout of the dual-band emission of red and green wavelengths. (b) Implementation with the smartphone screen, showing the smartphone display and front-facing camera surrounded by the red and green bands.



(a) (b)

Fig. 2: (a) Image captured by the camera with the identification of the regions of interest (ROIs) where each color-specific PPG signal is calculated. (b) Average intensity by column for each color channel using the proposed approach.

IV. Proposed Approach

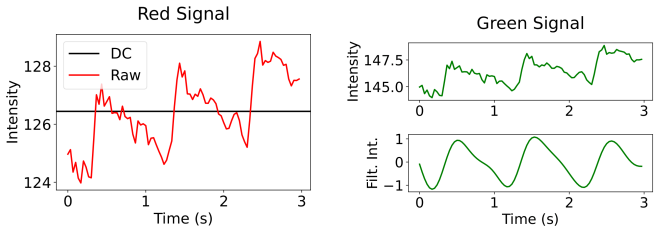
A. Two-tone Camera-based Photoplethysmography (cbPPG)

Unlike previous methods, in our approach, the screen is divided into two sections, each illuminating the finger with a different color, with the camera in the middle (Figure 1). This ensures a greater light wavelength separation on the extremities, getting closer to the values measured when individual colors are displayed (only red or only green). Moreover, the two wavelengths are also isolated, eliminating the typical crosstalk in approaches that use white [4], [5], [7] or yellow (red+green) [6] lights. A black boundary is used between the red and green bands to avoid interference. Lastly, it was found empirically that a darker shade of red (`#8C0000`) was better than pure red (`#ff0000`) to avoid saturation and ensure similarity in amplitude with green.

To properly capture the PPG signals, the user must place a finger above the camera sensor and keep it still, as shown in Figure 1. This ensures that the region of contact between the finger and the colored regions on the screen is maximized and also avoids possible motion artifacts.

Taking this into account, the average intensity value for each color channel is calculated on 25×25 square regions on the corners of each corresponding color (Figure 2). This guarantees that each of the two obtained PPG signals contain the least amount of interference from the other wavelengths, and allows the recording of these two signals simultaneously.

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(a) Red DC component. (b) Green AC component.

Fig. 3: Examples of AC and DC components extracted from the PPG signals using the proposed methodology.

B. Modulation Ratio Calculation

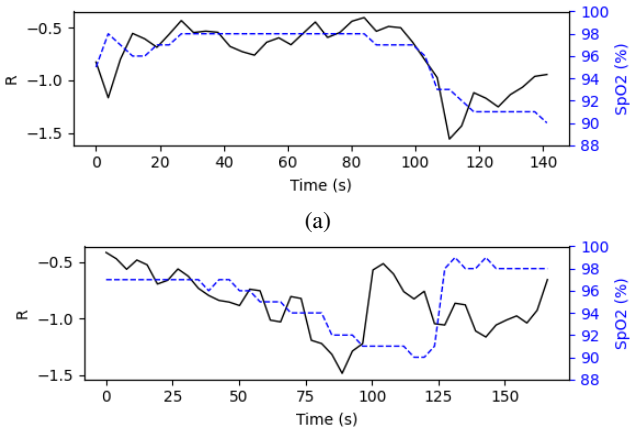
After obtaining the red and green raw signals, the DC and AC components can be computed. This computation occurs every 3 seconds with no overlap over the data recorded in this period. The signal is first inverted by subtracting its value to 255; afterwards, the DC component is computed by calculating average intensity values for each PPG signal [4], [6] (Figure 3). The AC component is obtained by calculating the standard deviation of the filtered signal with the same 3-second window (Figure 3). This filtration assures that the DC component is removed from the PPG signal, as well as most of the undesired noise and signal artifacts. The signal is filtered with a 2nd order Butterworth IIR bandpass filter with cutoff frequencies of 0.67 and 2 Hz (40 and 120 BPM) and the standard deviation is calculated on the filtered signal. The modulation ratio is then calculated using $f1$ as the red PPG and $f2$ as the green one (Equation 1).

C. Calibration Method

The proposed approach was evaluated by comparing two synchronized time series of SpO_2 values: one from a certified oximeter and another from the proposed approach. To this end, a BITalino Core BT device [10] connected to a certified CMS50D+ Pulse Oximeter was used to stream SpO_2 levels every second via Bluetooth. A mobile application was developed to connect to the BITalino device and record its values while recording a video file of the front-facing camera of a Redmi Note 9S smartphone. The data were later analyzed using Python 3. This approach allows for continuous comparison of SpO_2 measurements with a certified device instead of using individual ones for calibration.

D. Acquisition Protocol

Data was collected from 10 participants, resulting in a total of 23 pairs of BITalino signals and video recordings. The participants identified themselves as either female (60%) or male (40%), and their ages ranged from 22 to 57 years (M = 25.7, SD = 5.5). The number of acquisitions per participant ranged from 1 to 3 (M = 2.3, SD = 0.8). Participants were instructed to breathe normally for 30 seconds, followed by holding their breath for as long as possible. Each recording session lasted 3 minutes, during which continuous data was collected using the setup previously described.



(b)

Fig. 4: Comparison of calculated modulation ratio (R) and measured blood oxygenation levels (SpO_2) in two examples. (a) Demonstrates a strong correlation between the signals, while (b) reveals a noticeable disparity in recovery.

V. Results and Discussion

A. Modulation Ratio

A sample of the modulation ratio results is presented in Figure 4, which shows two examples of the calculated ratios and the corresponding measured SpO_2 levels. The plots visually demonstrate a correlation between the R values and the reference SpO_2 levels, indicating that the R values estimated from the proposed approach are a reliable indicator of the subjects' oxygen levels. Also, the plots illustrate a decrease in SpO_2 as the subjects hold their breath, reflected in the decreasing trend of the calculated ratio.

The measurements obtained with the proposed modulation ratio technique appear to be more sensitive than those obtained with the certified oximeter. In Figure 4b it can be observed that, when the calculated ratio starts to rise due to the subject returning to normal breathing, the oximeter readings take longer to recover, indicating a lag in the CMS50D+ device comparatively to the proposed approach. This slower recovery is likely due to the different signal processing methods used by the two devices. Moreover, it should be noted that there is no guarantee that the computed ratio values by both devices are fully synchronized since they might perform different computations. Nevertheless, for the calibration of the proposed method, a linear regression is computed for the recorded values.

B. Calibration with SpO_2 Levels

To perform the calibration procedure for estimating SpO_2 levels using the proposed approach, multiple linear regressions were calculated using the available data. To evaluate the performance of the calibration curve, the Pearson Correlation Coefficient (PCC) was used between the calculated modulation ratios and the oximeter readings.

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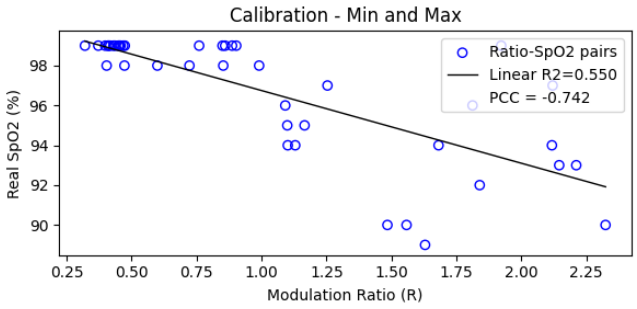


Fig. 5: Calibration line for the modulation ratio and the measured SpO_2 levels using only the minimum and maximum values. The Pearson Correlation Coefficient (PCC) between the calculated ratios and the SpO_2 signals is also presented.

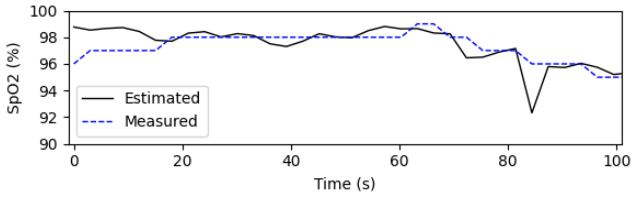


Fig. 6: Estimated SpO_2 levels for a given recording using the chosen calibration line.

Using all data as input for the linear regression resulted in poor performance, with a negative correlation of 0.298. This can be attributed to the influence of the small range of SpO_2 readings and the sensitivity of the method. Subsequently, using the Q1, Q2, and Q3 values resulted in an overall PCC of -0.73 ($p < 0.001$). Even better results were obtained by using only the minimum and maximum R and SpO_2 levels, with a strong PCC of -0.742 ($p < 0.001$). The calibration curve depicting this correlation is shown in Figure 5.

The results obtained from the study demonstrate the need to improve the calibration procedure and expose the acquisitions to a wider range of settings. For instance, the limited range of oxygenation values acquired from the subjects in this study restricted the accuracy of the evaluation. Future studies should aim to include subjects with a wider range of oxygenation levels, including individuals with basal oxygenation levels lower than the healthy range. This would also ensure that the calibration curve is not affected by any lack of synchronization between the compared devices, resulting in more accurate and reliable measurements. Additionally, validating the methodology by including more participants and testing it in other hardware settings could provide more insights into its generalizability.

Despite these limitations, using the calibration curve previously obtained it is possible to accurately estimate the oxygenation value from the modulation ratio (R) signals derived from a recorded video (Figure 6).

VI. Conclusion

In this paper, we propose a novel approach that uses the smartphone display as a light source and the front-facing camera as a detector. The placement of the camera, surrounded by the display, divides the emitted light into two colored regions. Compared to previous approaches, this improvement enhances the quality of the measured PPG signal for the red and green color channels by successfully separating these two wavelengths on the display. Although some limitations were found, preliminary results show that the method accurately estimates SpO_2 levels and even portrays small fluctuations of these measurements. This approach makes cbPPG SpO_2 measurements closer in accuracy to currently commercialized pulse oximeters, which is a significant advancement in the field of non-invasive SpO_2 measurement.

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References

[1] T. Tamura, "Current progress of photoplethysmography and SPO2 for health monitoring," *Biomedical Engineering Letters*, vol. 9, pp. 21–36, 2019.

[2] M. Tobin, F. Laghi, and A. Jubran, "Why COVID-19 silent hypoxemia is baffling to physicians," *American Journal of Respiratory and Critical Care Medicine*, vol. 202, no. 3, pp. 356–360, 2020.

[3] J. Hoffman, V. Viswanath, C. Tian *et al.*, "Smartphone camera oximetry in an induced hypoxemia study," *npj Digital Medicine*, vol. 5, p. 146, 2022.

[4] A. Nemcova, I. Jordanova, M. Varecka, R. Smisek, L. Marsanova, L. Smital, and M. Vitek, "Monitoring of heart rate, blood oxygen saturation, and blood pressure using a smartphone," *Biomedical Signal Processing and Control*, vol. 59, p. 101928, 2020.

[5] X. Ding, D. Nassehi, and E. Larson, "Measuring oxygen saturation with smartphone cameras using convolutional neural networks," *IEEE J. Biomed. Health Inform.*, vol. 23, no. 6, pp. 2603–2610, 2019.

[6] C. Holz and E. Ofek, "Doubling the signal quality of smartphone camera pulse oximetry using the display screen as a controllable selective light source," in *Proc. of the IEEE Engineering in Medicine and Biology Society Annual Int'l Conf. (EMBC)*, 2018, pp. 1–4.

[7] N. Bui, A. Nguyen, P. Nguyen, H. Truong, A. Ashok, T. Dinh, R. Deterding, and T. Vu, "PhO2: Smartphone based blood oxygen level measurement systems using Near-IR and RED wave-guided light," in *Proc. of the ACM Int'l Conf. on Embedded Network Sensor Syst.*, 2017.

[8] E. D. Chan, M. M. Chan, and M. M. Chan, "Pulse oximetry: Understanding its basic principles facilitates appreciation of its limitations," *Respiratory Medicine*, vol. 107, no. 6, pp. 789–799, 2013.

[9] F. Lamonaca, D. Carnì, D. Grimaldi, A. Nastro, M. Riccio, and V. Spagnolo, "Blood oxygen saturation measurement by smartphone camera," in *Proc. of the IEEE Int'l Symp. on Medical Measurements and Applications (MeMeA)*, 2015, pp. 359–364.

[10] D. Batista, H. Plácido da Silva, A. Fred, C. Moreira, M. Reis, and H. Ferreira, "Benchmarking of the bitalino biomedical toolkit against an established gold standard," *Healthc. Technol. Lett.*, vol. 6, pp. 32–36, 2019.

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