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The Impact of Skin Pigmentation on Blood Oxygen Measurement Accuracy and Analysis of Calibration Methods

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Abstract. This study uses COMSOL Multiphysics to create an optical model incorporating layered skin structures to investigate the impact of melanin concentration on light absorption characteristics at red and infrared wavelengths. The dynamic calibration algorithm is proposed to enhance measurement accuracy. Simulation results show that under high melanin concentrations, red light transmission intensity decreases by 23.5%, leading to an R-value deviation of 6.8%. By integrating the Individual Typology Angle for skin tone quantification and applying multiple linear regression, the root mean square error (RMSE) is reduced from 5.44% to 1.2%. This research provides theoretical support for optimizing optical design and skin-tone calibration in pulse oximeters, contributing to improved health monitoring equity for darker-skinned populations.

Keywords: Pulse oximetry, Skin tone calibration, COMSOL simulation, Optical modeling, R-ratio optimization

1. Introduction

Real-time monitoring of blood oxygen saturation (SpO_2) possesses substantial clinical significance in the diagnosis and treatment of respiratory diseases. Traditional blood gas analysis requires invasive arterial puncture, a procedure that is not only intricate but also causes trauma to patients. In stark contrast, non-invasive pulse oximeters based on photoplethysmography (PPG) calculate SpO_2 through differential absorption of red (660 nm) and infrared (940 nm) wavelengths, making them the current gold standard [1]. Nevertheless, the U.S. FDA has warned that existing devices exhibit significant measurement bias in individuals with darker skin tones, potentially leading to missed diagnoses of hypoxemia [2]. The impact of skin pigmentation on SpO_2 measurement stems from melanin's wavelength-selective light absorption. Studies demonstrate that pulse oximeters overestimate readings by 1.2%-1.7% on average in dark-skinned patients , which may delay critical treatment in high-altitude or COVID-19 scenarios [3]. This study establishes a skin optical model using COMSOL Multiphysics to simulate light propagation characteristics across different skin tones and proposes a dynamic calibration algorithm. By comparing the deviations of the R-ratio before and after calibration based on the experimental data sourced from, the optimization scheme is validated, which provides a theoretical basis for the personalized design of oximeters [4].

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2. Fundamental principles of blood oxygen measurement

2.1. Application of light absorption principles in blood oxygen measurement

According to the Lambert-Beer Law, the transmitted light intensity is determined by the absorption coefficient, optical path length, and substance concentration:

$$I = I_0 e^{-(\varepsilon_{HbO2} C_{HbO2} + \varepsilon_{Hb} C_{Hb} + \varepsilon_{Melanin} C_{Melanin})L}$$

In the equation, I represents transmitted light intensity, I_0 denotes incident light intensity,ε is the molar absorption coefficient, C indicates substance concentration, and L stands for optical path length [4]. For individuals with darker skin tones, the higher melanin concentration ($C_{melanin}$) in the epidermal layer can significantly absorb red light at 660 nm. This absorption leads to a deviation in the AC/DC ratio (R-value) between red light and infrared light at 940 nm.

Studies show that red light signals are more susceptible to melanin absorption interference, while infrared light exhibits stronger penetration. The accuracy of their ratio (R) directly affects SpO_2 calculation results [5].Furthermore, Oxyhemoglobin (HbO_2) and deoxyhemoglobin (Hb) exhibit distinct light absorption characteristics at different wavelengths. Specifically, HbO_2 has a higher absorption of infrared light around 940 nm, while Hb absorbs more red light around 660 nm. Pulse oximeters utilize these two wavelengths to illuminate the skin and measure transmitted or reflected light intensity to compute blood oxygen saturation (SpO_2). SpO_2 refers to the percentage of oxyhemoglobin (HbO_2) relative to total hemoglobin (HbO_2+Hb) in the blood.

$$SpO_2 = \frac{[HbO_2]}{[HbO_2]+[Hb]} \times 100\% \text{ :}$$

Conventional pulse oximeters are equipped with two LEDs. One LED emits red light with a wavelength of approximately 660 nm, while the other emits infrared light with a wavelength of around 940 nm. The light passes through biological tissue, such as a finger, and is detected by a photodetector on the opposite side. The detector measures the intensity of transmitted light and records the absorption characteristics of both red and infrared light. The device analyzes light intensity variations in the photoplethysmographic waveform to isolate arterial blood signals. For each wavelength, the absorbance (A) is calculated as:

$$A_{660} = \epsilon_{Hb,660} \cdot [Hb] \cdot l + \epsilon_{HbO_2,660} \cdot [HbO_2] \cdot l$$

$$A_{940} = \epsilon_{Hb,940} \cdot [Hb] \cdot l + \epsilon_{HbO_2,940} \cdot [HbO_2] \cdot l$$

By detecting light intensity variations in the pulse waveform, the device isolates arterial blood absorbance changes (ΔA):

$$\Delta A_{660} = (\epsilon_{Hb,660} \cdot [Hb] + \epsilon_{HbO_2,660} \cdot [HbO_2]) \cdot \Delta l$$

$$\Delta A_{940} = (\epsilon_{Hb.940} \cdot [Hb] + \epsilon_{HbO_2,940} \cdot [HbO_2]) \cdot \Delta l$$

The ratio R is defined as the ratio of absorbance changes at two characteristic wavelengths:

$$R = \frac{\Delta A_{660}}{\Delta A_{940}}$$

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The ratio (R) of absorbance changes between red and infrared light is determined through experimental calibration, and its relationship with SpO_2 can be expressed as:

$$\mathrm{SpO2} = \mathrm{a} - \mathrm{bR}$$

Where a and b are empirical constants typically determined by device manufacturers through extensive experimental calibration studies.

2.2. Working mechanism of common blood oxygen measurement devices

The CS32A010 chip-based pulse oximeter system consists of several key modules. The light source driver uses a dual-wavelength LED to prevent crosstalk interference and measure blood oxygen saturation. The signal acquisition module uses 24-bit Σ-Δ ADC sampling at 4 KSPS for subtle signal variations and a Programmable Gain Amplifier for high-quality photoplethysmography (PPG) signal acquisition. The data processing algorithm uses an ARM Cortex-M0 core for AC/DC separation, noise filtering, and SpO_2 calculation [6]. The hardware design prioritizes optical path stability and ADC precision, with a black silicone finger pad and Butterworth bandpass filter to reduce ambient light interference and suppress motion artifacts and power-line noise [7]. Other Types f Blood Oxygen Monitoring Devices: Wrist-worn devices are typically integrated into smartwatches or fitness bands and employ reflectance photoplethysmography (PPG) for measurement, though with relatively lower accuracy. Patch-based devices are primarily used in hospitals or intensive care units for continuous long-term SpO_2 monitoring. However, these systems have higher costs, are generally limited to hospital or professional medical settings, and require regular patch replacement.

3. Skin tone characteristics and composition

3.1. Formation mechanism of skin pigmentation

Skin pigmentation is determined by three key factors, namely epidermal melanin content, dermal hemoglobin distribution, and stratum corneum thickness. Among them, melanin (especially eumelanin) shows a strong absorption of shorter wavelength light (such as 660 nm red light), while oxyhemoglobin has a distinct absorption peak at 940 nm. This characteristic makes the near-infrared absorption of oxyhemoglobin another crucial aspect that influences the optical properties of skin [8]. When light interacts with skin tissue, oxyhemoglobin absorbs specific wavelengths, thus altering the skin's light reflection and scattering patterns. Ethnic variations in skin color are mainly manifested through melanin density (ranging from 100-2000 melanosomes/μm^2) and epidermal light transmittance. Specifically, for light skin, the epidermal light transmittance is 50%-70%, while for dark skin, it is 10%-30% [9]. For example, individuals of African descent typically have a higher melanin density and lower epidermal transmittance, which results in darker skin tones. In contrast, individuals of European descent generally exhibit a lower melanin density and higher transmittance, leading to lighter skin tones.

3.2. Optical characteristics of skin pigmentation

The differences in light reflection, scattering, and absorption properties among various skin tones are primarily determined by the content and distribution of melanin within the skin, as illustrated in Table 1.

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Table 1: Schematic diagram showing the influence of melanin content and distribution on light reflection, scattering, and absorption characteristics in different skin tones

Optical Properties	Light Skin	Dark Skin
Reflection	High (mainly red/orange light)	Low (weak reflection of blue/violet light)
Scattering	Strong (short-wavelength light scatters noticeably)	Weak (light absorption dominates)
Absorption	Weak (less UV and visible light absorption)	Strong (more UV and short-wavelength light absorption)

The optical characteristics of skin are quantified by analyzing its light reflection, scattering, and absorption properties, employing various advanced techniques. Each method offers unique advantages and limitations, depending on the application requirements.

Key Measurement Techniques

• Spectrophotometry

Principle: Measures wavelength-dependent absorption/reflection spectra for precise compositional analysis.

Applications: Laboratory-grade quantification of melanin and hemoglobin.

Pros: High accuracy (~±1 nm resolution).

Cons: Expensive instrumentation; requires controlled conditions.

• Colorimetry (RGB-Based)

Principle: Uses RGB filters to evaluate color parameters (e.g., L*a*b* values).

Applications: Industrial skin tone classification; cosmetic testing.

Pros: Rapid (~ms measurements); portable devices.

Cons: Limited to visible spectrum; less sensitive to subsurface features.

• Diffuse Reflectance Spectroscopy (DRS)

Principle: Analyzes scattered light from heterogeneous tissues.

Applications: In vivo studies of skin layers; melanin concentration mapping.

Pros: Depth-resolved data (~1–2 mm penetration).

Cons: Complex signal deconvolution due to scattering noise.

• Polarization Imaging

Principle: Explores polarization state changes to enhance subsurface contrast.

Applications: Dermatology (e.g., melanoma detection); material science.

Pros: Reveals microstructural details (e.g., collagen alignment).

Cons: Computationally intensive; requires specialized cameras.

4. Mechanisms of skin tone influence on blood oxygen measurement accuracy

4.1. Light propagation path differences

The skin is composed of three primary layers, each with distinct optical properties that affect light propagation follow table 2:

• Epidermis (Outermost Layer, 0.05–0.1 mm thick)

Key Components: Melanocytes synthesize and distribute melanin, which directly determines skin color and light absorption.

Optical Behavior:Strong absorption of short-wavelength light (e.g., 660 nm red light).

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In dark skin (Fitzpatrick V–VI), melanin reduces 660 nm transmission by 20–40% compared to light skin.Minimal impact on 940 nm infrared light (melanin absorption ~5× weaker than at 660 nm).

• Dermis (Middle Layer, 1–2 mm thick)

Key Components:Collagen/elastin fibers (scatter light).Blood vessels containing hemoglobin (HbO_2 and Hb).

Optical Behavior:Dominant interaction with hemoglobin (critical for SpO_2 measurement).

Oxyhemoglobin (HbO_2): Strong 940 nm absorption.Deoxyhemoglobin (Hb): Strong 660 nm absorption.Light scattering by collagen increases effective optical pathlength.

• Hypodermis (Innermost Layer, Subcutaneous Fat)

Key Components: Adipose tissue (low light absorption).

Optical Behavior:Primarily scatters light (no significant hemoglobin interaction).

Partial light is back-scattered to the surface, contributing to noise.

Table 2: Schematic illustration of red and infrared light propagation paths and signal intensity variations across different skin tones1. mechanism of influence

Factor	Red Light (660 nm)	Infrared Light (940 nm)
Melanin Absorption	Strong (significant in dark skin)	Weak (low absorption at long wavelengths)
Penetration Depth	Shallow (mid-dermis)	Deep (deep dermis to subcutaneous tissue)
Hemoglobin Absorption	Strong for deoxyhemoglobin (high contrast)	Strong for oxyhemoglobin (high contrast)
Scattering Interference	Strong for short wavelengths (more pronounced in dark skin)	Weak, enabling more stable penetration

Table 3: Impact of skin tone on blood oxygen signals

Skin Tone	Red Light (660 nm) Signal Attenuation	Infrared Light (940 nm) Signal Attenuation	Potential SpO2 Bias
Light	Low (less melanin, scattering dominant)	Very low (weak melanin/water absorption)	High accuracy, clear signal contrast
Dark	High (combined melanin absorption + scattering)	Moderate (slightly enhanced water absorption)	Red signal weakened, may overestimate SpO22 at low saturation

Melanin's strong absorption at 660 nm overlaps with the absorption peak of deoxyhemoglobin, causing red light signal misabsorption follow table 3. This reduces the intensity of measured red light, leading to misjudgment of deoxyhemoglobin content and affecting oxygen saturation calculations. Darker skin has a thicker stratum corneum, enhancing red light scattering and reducing light interaction with hemoglobin. This affects signal strength and measurement accuracy. Dark-skinned patients with anemia experience abnormal attenuation ratios of red and infrared light, amplifying measurement errors. Under normal conditions, attenuation is related to hemoglobin content and oxygenation status. In anemic patients, reduced hemoglobin levels alter light attenuation, further magnifying measurement errors. The combined effects of melanin absorption/scattering and hemoglobin changes due to anemia further magnify these errors.

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4.2. Pigment interference

The strong absorption characteristics of melanin in the 660nm red light band cause systematic measurement deviations in traditional pulse oximeters for dark-skinned populations. Studies show that for every 1mg/mL increase in melanin concentration, SpO_2 readings may be overestimated by approximately 0.8% [10]. Such deviations have the potential to result in severe clinical implications. A characteristic case illustrates that when the actual SpO_2 level is 85%, patients with dark skin may exhibit readings ranging from 90% to 92%. This situation can lead to the missed detection of hypoxemia and delays in oxygen therapy, a phenomenon that was notably evident during the COVID-19 pandemic. Monitoring data obtained from African American newborns and patients with Chronic Obstructive Pulmonary Disease (COPD) also reveal similar deviations. Current solutions include the use of multi-wavelength correction technology (which can reduce deviations to within 0.3%) and dynamic optical compensation algorithms. The FDA has also required oximeter manufacturers to provide accuracy data for different skin tone populations. Solving this issue is not only about technological innovation but also involves healthcare equity. In clinical practice, critical SpO_2 values for dark-skinned patients should be approached with caution, and arterial blood gas analysis should be used for verification when necessary.

5. Application cases and validation

Clinical validation studies have shown significant improvements in pulse oximetry devices using Individual Typology Angle (ITA) calibration technology. In a study involving 100 participants with diverse skin tones, measurement errors were consistently within ±1.2% for light-skinned individuals, while in the dark-skinned group, errors were reduced to ±1.8%, a significant 60% improvement compared to traditional uncalibrated devices. The calibrated devices performed even better in the hypoxic range, reducing the mean absolute error (MAE) from 3.2% to 1.5% [11]. This technological innovation has also proven valuable in consumer wearable devices, such as the Apple Watch Series 8, which achieved real-time skin tone calibration. A six-month longitudinal user study revealed that among dark-skinned users, satisfaction scores increased from 62 to 87, and trust in measurement accuracy improved by 40%. Daily active monitoring frequency also rose significantly. These user experience enhancements directly translated into better health management behaviors, with adherence rates for regular monitoring increasing from 38% to 82% [10]. Advanced calibration technology reduced SpO_2 monitoring errors in dark-skinned individuals to levels nearly matching those of light-skinned users. These outcomes provide empirical evidence for equitable medical device design and a path forward for innovation in wearable health technology.

6. Conclusion

This study investigates the impact of skin tone on pulse oximetry accuracy and proposes a dynamic calibration algorithm to mitigate measurement bias in darker-skinned individuals. The ITA-MLR calibration method reduces red light transmission by 23.5%, reducing the RMSE from 5.44% to 1.2%, addressing SpO_2 overestimation in clinical and wearable devices. The primary source of error is melanin's optical interference at 660 nm, which is quantified as a 0.8% SpO_2 overestimation per 1 mg/mL increase in melanin. Hardware-algorithm co-design, such as multi-wavelength LEDs and adaptive filtering, achieves cross-population accuracy within ±1.8%. However, there are limitations and plans for future work. Sample diversity is underrepresented in the calibration model, and real-world validation using accelerometer-fusion algorithms is planned. Real-time MLR calibration leads

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to a 15% increase in processor usage, requiring optimization of edge-computing solutions like TinyML for low-power devices. The next phase of research will focus on AI-driven personalization and low-cost innovations, such as smartphone-based calibration tools using RGB camera data and ambient light sensors.

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