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Custom Visible to Infrared, Multi-wavelength Light Emitters for Pulse Oximeter Applications

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Abstract— **Compact multi-wavelength light sources have been developed to provide flexibility in pulse oximeter device design. Advantages of these light sources include emitting light of multiple wavelengths with a constant ratio of intensity, and a co-linear light propagation path through tissue. These devices can function as normal pulse oximeter to determine oxygen saturation and pulse rate, but this approach potentially reduces circuit complexity, reduces motion artifacts and enables multi-wavelength sensing of different forms of hemoglobin or other blood products.**

I. Introduction

Dual wavelength pulse oximetry has had a tremendous medical impact and has developed to the point where in addition to being used in clinical environments low cost devices are available for home use. These systems are relatively sophisticated since they need to be robust to motion artifacts and self-calibrating under a wide variety of conditions.

Increasing the number of wavelengths potentially allows more accurate readings to be obtained. Also a multi-wavelength approach can potentially detect the presence of carbon monoxide bound to hemoglobin, abnormal forms of hemoglobin and waste products that are indicative of renal and hepatic function. This additional functionality since it is non-invasive to the body and in a form factor that is acceptable to clinicians is likely to be very useful.

Multispectral pulse oximetry systems incorporating several optical sources using multiple discrete LED components have been successfully demonstrated by industry leaders, such as Masimo Corporation. This discrete component approach results in a complex circuit and signal processing requirements. This complexity was emphasized when tolerances in one LED component changing under while operation lead to unacceptable errors in saturate partial oxygen level leading to a recall of devices. [7]. Thus approaches that reduce the complexity of calibration are expected to have a large system level impact in terms of system reliability and accuracy. In this paper co-linear multi-wavelength sources suitable for pulse oximetry are described.

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II. Background

With revenues of over $200 million/year and growing, pulse oximetry is perhaps the prevailing success story of low-cost, portable, real-time, noninvasive medical diagnostics. In its simplest form, pulse oximetry involves the introduction of 2 wavelengths, red and infrared, through the skin followed by the real-time monitoring of the resulting optical attenuation of the backscattered signal [1]. Oxidized hemoglobin (HbO_2) absorbs light primarily in the infrared wavelengths, and reduced hemoglobin (Hb) absorbs light primarily in the red, Figure 1. In general, by monitoring the changes in optical attenuation during the pulsatile component of arterial blood flow compared to the constant absorption and by factoring the ratio of red to-infrared absorption, one can estimate the oxygen saturation in the blood (SpO_2).

The ability to track oxygen levels in a patient's blood in real-time has revolutionized the area of anesthesia, is widely used for patient monitoring, and is increasingly finding new uses in real-time health monitoring and activity-level monitoring.

The principles of pulse oximetry go back to the 1860s, when investigators realized that optical absorption in blood varied with oxygen content [2]. In the 1980's, researchers developed a small, inexpensive, finger-clip model employing only 2 wavelengths (red and infrared), incorporating specially designed LEDs to replace large lamps and the fiber optic bundles. Thus multiple wavelength capability was given up in exchange for size, form factor and cost.

There is growing clinical interest in adding new functionality to real-time, noninvasive pulse oximetry, with a particular focus on differentiating between the various types of hemoglobin, monitoring more metabolites, monitoring drug dosage, and monitoring the metabolic effects of drugs for a given dosage [3,8]. However, there is a perceived tradeoff between the performance and cost. This has motivated corporations to seek new, low-cost, high performance optical sources, particularly in the near-IR, to make multi-wavelength pulse oximetry a practical and affordable reality.

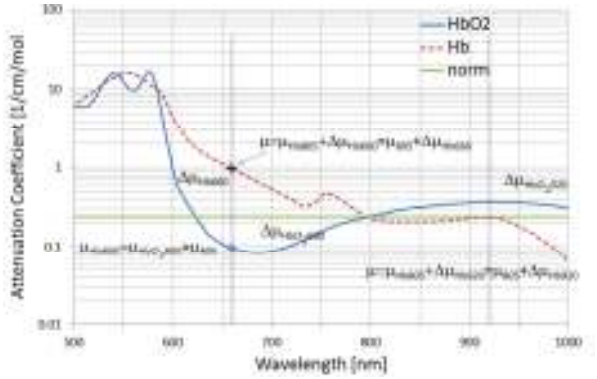

Figure 1. Hemoglobin and Oxyhemoglobin attenuation spectra as a function of wavelength. The conventional spectral position of light sources is shown at 660 and 920 nm.

Recently there has been interest in obtaining more spectroscopic information from the blood noninvasively and in convenient form factors. Masimo Corporation's new "Rainbow SET" technology has up to 8 specially designed red-to-near-InfraRed (nearIR) LEDs (on discrete chips) in a finger-mount pulse oximeter. The goal of this device is to provide real-time differentiation of 2 additional types of hemoglobin: methemoglobin [Hi], and carboxy-hemoglobin [HbCO], through their unique optical absorption spectra [3]. Methemoglobin is a form of the oxygen-carrying metalloprotein hemoglobin, in which the iron in the "heme-group" is not Fe^{2+} of normal hemoglobin but Fe^{3+}. Methemoglobin cannot release bound oxygen, unlike Oxyhemoglobin [HbO_2]. This anomaly is present in approximaterly 2% of the population. Carboxyhemoglobin (carbon monoxide carrying hemoglobin), is particularly important for accurate blood oxygen monitoring in smokers [4]. Additional benefits of multispectral attenuation screening are in detecting: "blue-baby syndrome" in infants [5]; monitoring medication-related methemoglobinemia [6]; and potentially quantifying bilirubin and blood urea nitrogen as well as various other biochemical constituents.

III. Materials and Methods

The wavelengths traditionally selected are in the red (660 nm) and infrared (920) where there is a large change absorption that can be compared with a small change in absorption as shown in Figure 1.

The light fluence (ψ) passing through any absorbing/scattering medium to first order can be expressed as Beer's law given in equation (1):

$$\psi_\lambda = \psi_0 e^{-\mu_\lambda d} \tag{1}$$

where the attenuation coefficient: μ_λ is a function of both wavelength (λ) and oxygen saturation (SpO_2). The radiance fluence intensity ψ_0 entering the tissue is a function of LED current and device geometry. While "d", the path-length traversed by the light, is also a function of wavelength geometry and tissue composition.

The measured attenuation at any time is the combined attenuation from Hemoglobin (Hb), Oxy-hemoglobin (HbO2), Hemiglobin (Hi), Carboxy-hemoglobin (HbCO) and the tissue layers surrounding the blood vessels (skin, muscle, etc.). This is expressed by the attenuation definition in equation (2):

$$\mu_{a,total}(\lambda) = \mu_{a,Hb}[Hb] + \mu_{a,HbO}[HbO_2] + \mu_{a,Hi}[Hi] + \mu_{a,HbCO}[HbCO] + \mu_{a,skin} + \mu_{a,muscle} \tag{2}$$

Where the concentrations of the respective constituents are expressed in square brackets. The presence of Carboxy-hemoglobin is not typically considered to form a significant contribution to the measurement, however anecdotal evidence in the form of discussions with emergency room physicians indicates that that while rare there are relatively serious situations where this can be very misleading. In the case of CO poisoning the absorption peak of Carboxy-hemoglobin and Oxy-hemoglobin coincide, but Carboxy-hemoglobin does not contribute to the oxygenation of the tissue.

The oxygen saturation is defined as the percentage of blood that can release oxygen. This is with respect to the total blood content, both oxygenated and de-oxygenated. In mathematical expression the oxygen saturation is represented by equation (3):

$$SaO_2 = \frac{[HbO_2]}{[HbO_2]+[Hb]} 100\% \tag{3}$$

Since the saturation is directly related to both the concentrations and the attenuation coefficients of the hemoglobin and oxy-hemoglobin, an attenuation measurement can provide a good first order approximation of the Oxygen content of the blood.

The above equations are time dependent since the heart is pumping blood through the body. During systolic blood supply the oxygenation level will rise while fresh oxygenated blood is supplied to the tissue. During diastole the oxygen deposited by the hemoglobin in the blood is reduced consumed by the surrounding cells in the tissue.

Additional time dependence from breathing results in a low frequency breathing rate component. During inhalation the blood oxygenation will be high and during exhaling the oxygenation level of the arterial blood will decrease over time leading up to the ensuing inhalation.

The light can be detected either in transmission or in reflection. In both cases the optical path that is determined by geometry and scattering is important in determining the total absorption for each wavelength. The factors influencing the light collected by a detector in reflection mode when placed besides the multi-wavelength light source which is directed at the tissue surface are expressed by equation (4):

$$\psi_\lambda = \psi_0' \exp\{-\mu_{s_\lambda}(t) C_{onc} d' f_\lambda(C_{onc}, SpO_2, d')\} \quad (4)$$

where C_{onc} represents the general concentration of hemoglobin, also known as the hematocrit. The hematocrit can vary between male and female subjects, while certain blood conditions can have a significant impact on concentration and functionality of the red blood cells. Thus we find that the optical path length is both wavelength and blood composition dependent.

In case the measurements are performed on or near an artery the expansion and contraction during pulsatile flow will changes the geometry of the optical path making measurement very sensitive to location. This is a nonlinear problem and difficult to specify and to solve.

So to make the problem tractable, the factors that influence the radiant fluence rate entering the tissue as well as the ability to detect the light on exit are usually captured as coefficients which can be represented as: $\psi_0' = \Sigma c_i \psi_0$, with ψ_0 the emission radiance of the source. This leads to the expression in (5) where tissue irradiated by custom light source is represented by the following equation:

$$\psi_\lambda = \psi_0' \exp\langle -\{\mu_{s_\lambda}(HbO_2) C_{onc}(t)[d'_{HbO_2}] + \mu_{s_\lambda}(Hb) C_{onc}(t)[d'_{Hb}]\} f_\lambda(C_{onc}, S, d, t)\rangle \quad (5)$$

where all wavelengths are collected simultaneously and expressed as an intensity varying with time.

A pulse-oximeter discriminates steady state blood flow (venous) from pulsating blood flow (arterial). Thus the maximum (I_{max}) and minimum (I_{min}) radiance values detected correspond to the diastolic and systolic arterial pressure respectively. The pulsed oxygen saturation (SpO_2) is expressed in equation (6):

$$SpO_2 = f \frac{\ln\left(\frac{I_{\min}}{I_{\max}}\right)_{660}}{\ln\left(\frac{I_{\min}}{I_{\max}}\right)_{920}} 100\% \quad (6)$$

The factor f in equation (5) is a calibration function, taking into geometric and other effects such as pigmentation into account.

It is because of the above complexity of knowing the true values for the emitted and detected irradiances that having light sources where multiple wavelengths are collinear and have a constant ratio of intensity can have value. While the wavelength dependent scattering still occurs, since the geometry is simplified and the ratio of emitted intensities is precise the complexity of analysis is reduced.

A calibration process is still needed since the determination of oxygen saturation still depends on the determination of absolute values in attenuation between diastole and systole is still needed. However, the magnitude of the errors is reduced leading to improve measurement. One can also use multiple multiwavelength sources to probe tissue with different wavelength bands to calibrate for tissue specific conditions.

To investigate these possibilities several multiple wavelength light emitters were constructed using a single LED as a pump source and using fluorescent materials to generate multiple wavelengths. Both transition metals and rare earth elements in wide band gap matrices were investigated.

IV. Results

Initially narrow band multi-wavelength emission sources were obtained by using epitaxial growth of single-crystal, thin films of β-Ga_2O_3 (gallium oxide) on c-plane (0001) sapphire (α-Al_2O_3) substrates by Pulsed Electron-beam Deposition (PED) [9] and by Pulsed Laser Deposition [10]. Optimally grown films had a bandgap of ~4.96 eV at room temperature with transmission of ~80% throughout the ultraviolet, visible and near infrared (UV-VIS-NIR) spectrum. This and the transparent substrate allow the light to be extracted through the substrates of the materials.

As shown in Figure 2, the rare earth elements chosen when incorporated in gallium oxide [11,12] and aluminum oxide formed sharp emission lines in the visible and near infrared when excited by an ultraviolet LED at 280 nm.

More recently, we have been investigating the use of Blue LEDs with more conventional high efficiency phosphors. The advantage in this approach is that selective electrodeposition of the phosphors can be performed and high brightness blue LEDs are now commercially available. We have also found that the internal scattering within the phosphor helps provide a spatially homogeneous light source.

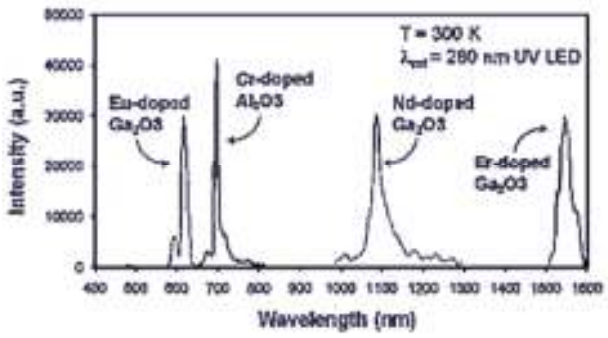

Figure 2. Example of the emission spectra of the room temperature photoluminescence from selected metal-oxides combined with rare-earth doped oxides using Europium, Chromium, Neodymium and Erbium.

As shown in Figure 3, where three phosphors are being optically pumped in unison, the feature in the optical spectra are distinct enough to be exploited.

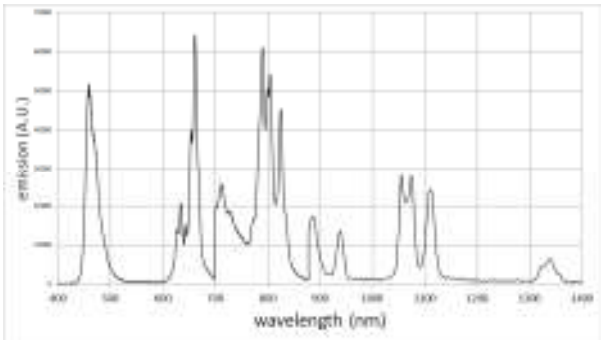

Figure 3. Example of the emission spectra of the constructed multi-wavelength light source based on Phosphor composites excited by 460 nm. Specific emission lines or bands can be modulated in amplitude and emission rate individually or in groups.

As a proof of concept, a small profile pulse oximetry system is being constructed using blue LEDs pumping these phosphors as shown in Figure 4. While not yet in a packaged form suitible for clinical use this simple demonstration indicates the feasibility of this approach.

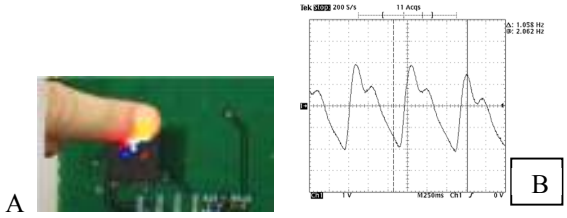

Figure 4. a) Example of the multi-wavelength light source applied to the finger for pulse-oximetry (false-color for near infrared light). b) the experiemental pulsatile pattern of the optical characteristics of blood-flow and associated cross-over between oxyhemoglobine and de-oxyhemoglobin attenuation for each systolic-diastolic component of the heart beat.

V. Conclusions

Collinear multi-wavelength LED devices were constructed and investigated for use in pulse oximetry. The collinear nature of the emission of multiple wavelengths with known ratios of intensity is expected to be beneficial in improving pulse oximetry devices. Adding functionality by increasing the number of wavelengths can be used to differentiate between the various types of hemoglobin, as well as be a noninvasive probe to investigate renal and hepatic function. A simple demonstration of pulse oximetry using these devices was performed.

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