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Noninvasive Hematocrit Monitoring Based on Parameter-optimization of a LED Finger Probe

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An optical method of measuring hematocrit noninvasively is presented. An LED Light with multiple wavelengths was irradiated on fingernail and transmitted light from the finger was measured to predict hematocrit. A finger probe contained an LED array and detector. Our previous experience showed that prediction accuracy was sensitive to reliability of the finger probe hardware and we optimized the finger probe parameters such as the internal color, detector area and the emission area of a light source based on Design of Experiment. Using the optimized finger probe, we developed a hematocrit monitoring system and tested with 549 persons. For the calibration model with 368 persons, a regression coefficient of 0.74 and a standard deviation of 3.67 and the mean percent error of 8% were obtained. Hematocrits for 181 persons were predicted. We achieved a mean percent error of 8.2% where the regression coefficient was 0.68 and the standard deviation was 3.69.

OCIS codes : 170.1470, 170.3890, 170.4580, 170.6510

I. INTRODUCTION

Noninvasive measurement of substances in the body fluids such as blood provides much convenience to people since taking out blood causes pain and has a potential risk of contamination. Optics, which uses light as an energy source and is often based on various optical spectroscopic techniques, is an excellent tool for noninvasive diagnosis [1-6]. Light scattering in the body and interference by other substances generate technical challenges. Future development of noninvasive techniques will depend on how to overcome these problems.

So far, pulse oximetry is a successful example. Pulse oximetry measures saturated oxygen in arterial blood (S_pO_2), that is the ratio between oxyhemoglobin (H_bO_2) and total hemoglobin in the peripheral artery [2]. Figure 1 shows absorption spectra of H_bO_2, reduced hemoglobin (H_b) and water. There are spectral regions where there are distinct differences between H_bO_2 and H_b. For example, 660 nm shows higher absorption for H_b compared to H_bO_2 and 940nm shows higher absorption for H_bO_2. The ratio of transmitted or backscattered light between 660 and 940nm is used as a parameter to compute S_pO_2. Pulse oximetry has been widely used in critical care and associated technologies are well documented. Another example is bilirubin. A high concentration of blood bilirubin causes jaundice especially for newborns. Noninvasive bilirubin monitoring has been available based on measurements , for example, at 450 and 550 nm [3]. A bilirubin has an absorption peak at

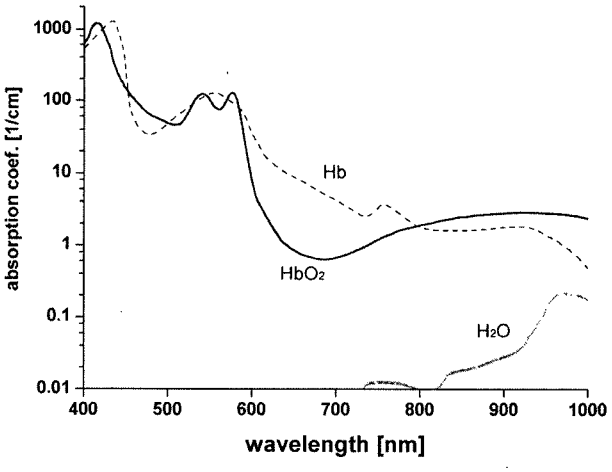

FIG. 1. Absorption spectra of oxyhemoglobin (H_bO_2), reduced hemoglobin (H_b) and water (H_2O) between 400 and 1000 nm.

450 nm and 550 nm is used as the reference wavelength.

Recently, anemia is another area for optical diagnosis. Anemia is caused by deficiency in the oxygen-carrying hemoglobin. Volume concentrations of hemoglobin, red blood cell volume or red blood cell number are used as parameters to present the degrees of anemia [7]. 40% of females suffer anemia to a greater or less degree. Monitoring hemoglobin is demanded prior to blood donation. For example, total hemoglobin level should be higher than 12.5 g/dl. Real time monitoring of total hemoglobin is recommended for bleeding patients or during surgery. Noninvasive monitoring is useful at the doctor's office, especially gynecology, pediatrics and others. All these needs arouse interests in noninvasive monitoring of hemoglobin.

Portable devices that are commercially available for total hemoglobin measurement requires blood extraction and have been used as point-of-care devices [8]. On the other hand, a noninvasive method has advantages of painless real time monitoring and has been investigated [4, 9-10]. However light scattering and interference by other substances are major technical issues. Furthermore, accuracy and reliability also depend on the experimental conditions such as the interface between a sensing unit and the human body. Previously, we reported on the development of a finger probe that contains a light source and detector [11]. In this paper, we developed a noninvasive hematocrit monitor that contained that finger probe and a clinical test was performed.

II. THEORY ON HEMATOCRIT PREDICTION

Transmitted light through a finger was measured and hematocrit was predicted based on multiple linear regression analysis. Use of 569, 805, 940 and 975 nm wavelengths produced the best prediction results. Two isobestic wavelengths of 569 and 805 nm were used as the key variables of predicting total hemoglobin (Fig. 1). The best accuracy was obtained when additional wavelengths of 940 and 975 nm were used [4].

We define a parameter related with optical signals, R, the ratio between arterial pulsatile component and non-pulsatile components that are described as AC and DC respectively in Figure 2.

$$R = \mathrm{AC}/\mathrm{DC} \tag{1}$$

Real time monitoring and compensation of the individual variation in vessel thickness and finger thickness is difficult. To deal with this problem, the ratio between R's was suggested in the pulse oximeter case [12].

$$R_{\lambda 1,\lambda 2} = R_{\lambda 1}/R_{\lambda 2} \tag{2}$$

The subscripts λ_1 and λ_2 indicate different wavelengths. We derived the relation between the parameter defined in Eq. (2) and the hematocrit is expressed by [13]

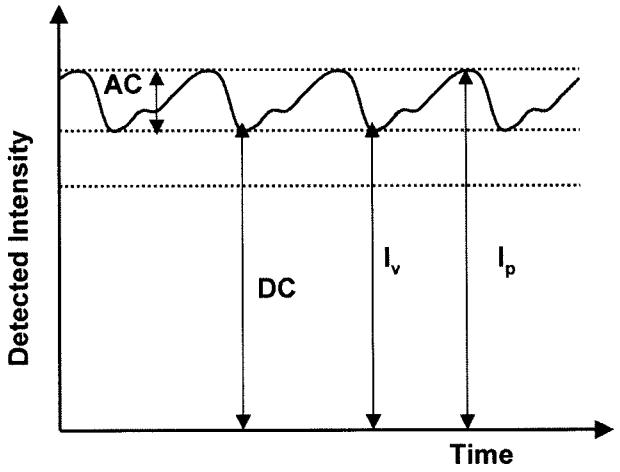

FIG. 2. A typical photoplethysmography (PPG).

$$R_{\lambda 1,\lambda 2} = [\mathrm{AC}_{\lambda 1}/\mathrm{DC}_{\lambda 1}]/[\mathrm{AC}_{\lambda 2}/\mathrm{DC}_{\lambda 2}] \sim [35\varepsilon_{\lambda 1} + k_{\lambda 1}a_{\lambda 1}(1-H)]/[35\varepsilon_{\lambda 2} + k_{\lambda 2}a_{\lambda 2}(1-H)] \tag{3}$$

where AC_λ : pulsatile component of PPG at λ wavelength
AC = I_p - I_v (I is the light intensity as shown in Fig. 1)
DC_λ : non-pulsatile component of PPG at λ wavelength
DC = $(I_p + I_v)/2$
λ : wavelength - 569, 805, 940 and 975 nm were used
ε : extinction coefficient
k : value depends on the wavelength, the configurations and geometry of the light source, detector and sample.
a : shape function of red blood cells
H : hematocrit

Three variables of $R_{569,805}$, $R_{569,940}$ and $R_{569,975}$ were used for calibration and prediction models.

III. A PARAMETER-OPTIMIZED FINGER PROBE AND HEMATOCRIT MONITOR

Figure 3 depicts the block diagram of a noninvasive hematocrit monitor that contains a finger probe. This finger probe contains the LED array and detector along with a preliminary amplifier. The LED array, custom-designed by Epitex Inc, contained LEDs of 569, 805, 940 and 975 nm. The finger probe has the followings parameters and detailed analysis was given in [11].

Probe color, the substrate color inside the finger probe, is black. Our initial color was white [4]. Using a dark substrate, multiple reflections between finger and substrate can be reduced. Measured signal at the photo detector (PD) is reduced, however, the relative noise level is what is important to the measurement. The active PD

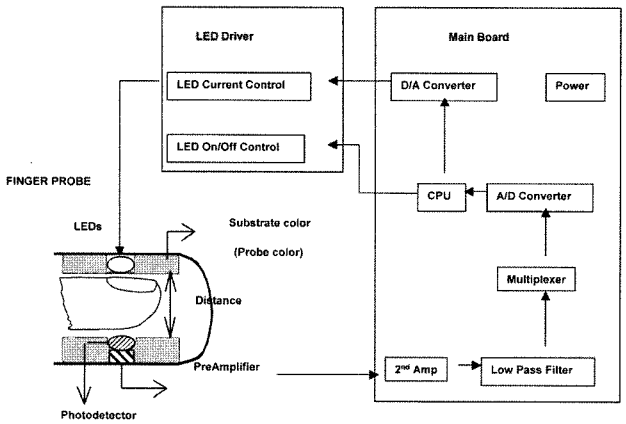

FIG. 3. Block diagram of non-invasive total hemoglobin measurement.

size is reduced from 3.2 × 3.2 mm to 2.3 × 2.3 mm by placing an aperture in front of the PD. A smaller PD size may reduce light shunting even though the background level decreases. This will increase the relative amplitude of pulsatile component with respect to the DC level. The emission area of the LED array is decreased from 3 × 2 mm to 3 × 1 mm. The LED array had three 569 nm LEDs since commercially available 569 nm LED chips have lower power compared with longer-wavelength LEDs.

Each wavelength in the LED array was turned on and turned off consecutively and its intensity level was controlled. Sensor signals were analogue-to-digital (A/D) converted with a resolution of 16 bits. An ARM^{TM} controller performed most data processing including filtering and computation as well as control of the device (Fig. 3).

IV. RESULTS AND DISCUSSION

A total of 549 persons at Samsung Medical Center, Seoul, Korea were tested over the period of five days. Each person inserted his or her finger into the probe and the variables stated in Eq. (1) were measured. Immediately, blood was drawn for checking the actual hematocrit value that was obtained from a clinical laboratory instrument in the hospital. Each measurement took eight seconds and five measurements were repeated for the same individual.

368 data out of 549 persons were used to compute the calibration model. The variables used in multiple linear regression were $R_{569,805}$, $R_{569,940}$ and $R_{569,975}$ as given in Eq. (3). Multiple linear regression analysis was done using MINITAB^{TM} (Minitab Inc). The results are shown in Fig. 4. A regression coefficient of 0.74, a standard deviation of 3.67 and the mean percent error of 8% were obtained. The computed regression coefficients from the calibration model were stored in the hematocrit

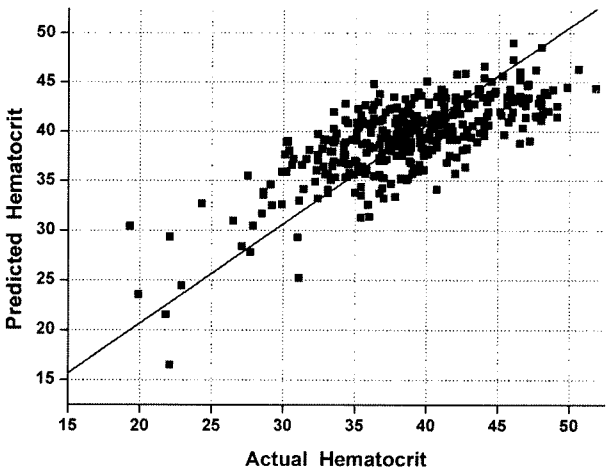

FIG. 4. The calibration model with 368 persons is shown. A regression coefficient of 0.74, a standard deviation of 3.67 and the mean percent error of 8% were obtained.

monitor. Then, hematocrits for 181 persons were predicted by inserting measured R values into the regression equation. The results for the prediction model were compared with actual hematocrit values that were measured from the clinical blood analyzer. The predicted results are illustrated in Fig. 5. The mean percent error was 8.2% with a regression coefficient of 0.68 and a standard deviation of 3.69.

V. CONCLUSIONS

Development of a noninvasive hematocrit monitor that is intended for applications in telemedicine care was investigated. We used an optical sensing method for this purpose. Measurement was made on a finger

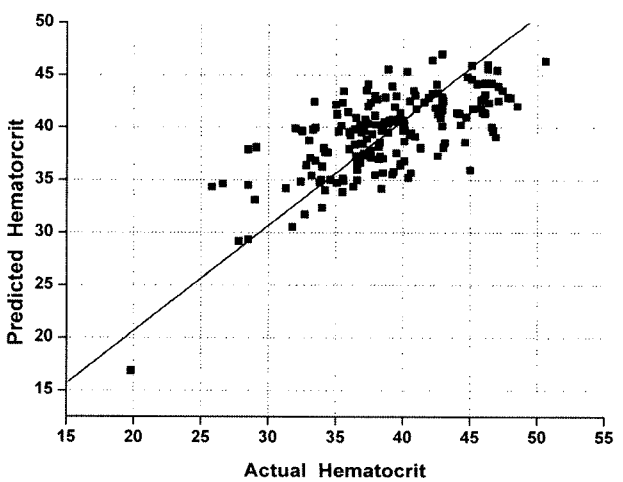

FIG. 5. Hematocrits for 181 persons were predicted. A regression coefficient of 0.68, a standard deviation of 3.69 and a mean percent error of 8.2% were achieved.

that allows much convenience for users. A single LED sensor makes the system compact and inexpensive. The achieved absolute mean percent error of hematocrit was 8.2 %. Our work proves that the method and algorithm for noninvasive hematocrit measurement based a single LED sensor are valid.

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