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Non-invasive optical transcutaneous pCO_2 monitoring system based on NDIR method

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Abstract

In this study, we developed an optical gas sensor of the non-invasive type for transcutaneous partial pressure of carbon dioxide, pCO_2, monitoring in the arterial blood. In basic step, the measuring system consists of an IR source, an IR detector (pyroelectric sensor), a gas cell made by MEMS technology, and a signal process circuit. The optical reaction length was reduced by 1*mm* using the MEMS technology. We confirmed the result of 2.21×10^{-6} *absorbance/ppm* sensitivity for concentration at 1,040 *mbar*, 2.14×10^{-5} *absorbance/mbar* sensitivity for pressure at 5,000 *ppm(parts per million)* CO_2 and 8.11×10^{-2} *absorbance/mm* sensitivity for optical reaction length.

The fabricated pCO_2 monitoring system showed fast response time about 5 second. The proposed system can be used in optical-bio sensor fields for medical diagnosis such as pCO_2 monitoring system, capnograph system for $EtCO_2$ analysis, and environment monitoring systems.

Keywords: NDIR, pCO_2 gas sensor, Non-invasive, Beer-Lambert law, Henry law, pyroelectric

1 Introduction

To estimate the respiratory ventilation ability in the acid-base balance of metabolism, it is essential that the CO_2 partial pressure, pCO_2, is measured from arterial blood. However, this is invariably restricted in serious cases or surgical patients who depend on artificial respiration, where CO_2 partial pressure monitoring is essential. Measuring the CO_2 by gathering arterial blood is extremely invasive and needs to be done several times [1]. A non-invasive measurement is where the pCO_2 is presumed based on the end tidal CO_2, $EtCO_2$, which is the concentration of CO_2 in the patient's exhaled breathe. But it's troublesome because patients are required to breathe through a duct. The CO_2 that is produced by the metabolic by-product in tissue cells is exhausted into the venous system through a gas exchange process in a capillary vessel, which is supplied blood by the arterial system. Therefore, there is a correlation with $PaCO_2$-$PcapCO_2$-$PvCO_2$. Counter current (diffusion), the CO_2 gas transfer mechanism of the venous-arterial system, is one of the major factors. The overall diagram is presented in Fig.1 [2].

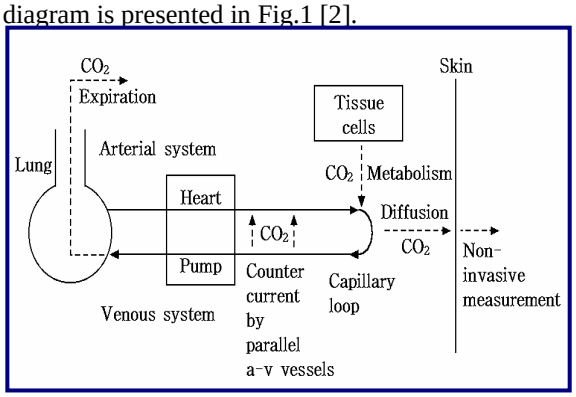

Figure 1: Schematics of venous-arterial system

If it is possible to measure the CO_2 partial pressure in blood that is diffused from the capillary vessels to the skin, this would facilitate non-invasive and non-troublesome measurements of the CO_2 concentration in the blood [3]. This would also be very useful and important clinically, as it would allow a quick estimation in a serious case that depends on an

artificial respiration machine, based on presuming the CO_2 concentration from the arterial blood.

For example, in the early period of respiratory failure, a quick diagnosis and medical care are possible, as the hypodermal CO_2 concentration increase is accompanied by a rapid arterial blood CO_2 concentration increase [4]. Yet, an invasive method that measures the CO_2 concentration from the arterial blood directly needs a considerably longer time, which is inappropriate in such cases. The technology of measuring the CO_2 gas diffused in the skin capillaries by a non-invasive method is thus very useful and important clinically [5, 6]. However, this has not been studied extensively, even in developed countries, because of technological difficulties.

Therefore, the objective of this study is the development of a convenient, non-invasive transcutaneous pCO_2 analysis device that uses an optical method. As such, effective infrared optical sensor cartridges and light sources were investigated, along with the signal processing. Based on previous research on the development of an expiration CO_2 gas sensor and monitoring system, the intent is to develop a hypodermal CO_2 concentration-measuring instrument that uses identical physical principles. This method is expected to be more useful in providing a non-invasive method for the diagnosis and care of emergency patients, including serious cases and patients in surgery who need their CO_2 concentration monitored.

2 Theory

2.1 Henry law

First, it is essential to confirm Henry law [7] for converting the CO_2 concentration. If a liquid and gas exist in a parallel state, then the partial pressure depends on the temperature, and the condition of dissociation is released as the melting state changes.

Table 1: Bunsen Solubility Coefficients

[㎖(STPD)/㎖ solvent, P_g = 760 *mmHg*, T = 37 ℃]

	α_g	
Gas	Plasma	Blood
He	0.0154	0.0149
N_2	0.0117	0.0130
O_2	0.0209	0.0240
CO_2	0.5100	0.4700

As such, when defining the gas concentration in the liquid phase as C_g, the gas partial pressure as P_g, and the constant fixed by each unit as A,

$$C_g = A\alpha_g P_g \tag{1}$$

The C_g value is defined as %Vol, which means the quantity of gas per 100 ㎖ liquid. Thus, the quantity of gas can be obtained using Table 1 when blood is defined as a standard solvent, A as 0.132, and the gas partial pressure as *mmHg*.

2.2 Beer-Lambert law

The fundamental theory governing absorption spectroscopy is based on Beer-Lambert law.

The ratio of the transmitted intensity I_t and initial intensity I_o of the IR radiation through an absorbing medium at a particular frequency relates exponentially to the transition line strength $S_i(cm^{-2}atm^{-1})$, line-shape function $\varphi(cm)$, total pressure $P(atm)$, absorption coefficient for concentration $\in (M^{-1}cm^{-1})$, material concentration $C(Mol)$, and path length $L(cm)$:

$$I_t = I_o \exp(-S_i \varphi \in PCL) \tag{2}$$

The IR intensity can be converted to an absorbance $A(\alpha)$ and is related to the transition parameters by

$$A(\alpha) = -\ln(I_t / I_o) = S_i \varphi \in PCL \tag{3}$$

The absorption coefficient is defined as:

$$\alpha(v) = S_i \varphi \in \tag{4}$$

From equation (2), the absorption $A(\alpha)$ can be described as follows:

$$A(\alpha) = \alpha(v)PCL \tag{5}$$

In this equation, the absorption $A(\alpha)$ is linearly proportional to the concentration C of the measured gas and path length L and pressure P [8, 9].

2.3 Pyroelectric effect

The quantity of electric charge per unit area in a vertical face of the spontaneous polarization direction is called spontaneous polarization on dipole moment per unit volume. The electric charge is where insulation always floats on the surface of a pyroelectric material was absorbed by the surface charge by spontaneous polarization with neutralize. The amount of spontaneous polarization is affected by temperature changes because of the light irradiation on the surface of the material. The neutralization state is broken down and the relaxation time is changed between the perception element surface charge and the adsorption floating ionic charge, so the electrical equilibrium is broken and charge occurs that companion to combine does not exist happens. Thus, the phenomenon whereby the spontaneous

polarization of surface charge changes according to temperature change is called a pyroelectric effect [10].

The changes of charge and current provide for the neutralization state. So, the light must be chopped so that the temperature change of the material may occur. When the light was chopped, the pyroelectric effect clarification [11] can be seen in Figure 2. When the chopper was closed, the output power of the sensor did not appear. But the output power is appear by temperature changed when chopper was opened.

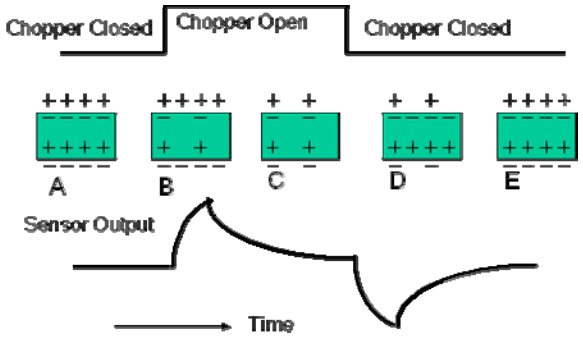

Figure 2: Pyroelectric effect representation in using chopper

There, however, has been equilibrium because the surface charge is neutralized again by a change in time. If the chopper is closed, the amount of infrared rays' decreases and the equilibrium of the charge have been broken. It will return to the neutralization state if time passes again in the super offerings surface. That is, it can receive a charge-discharge signal of the sensor consecutively by the opening and closing of the chopper.

3 An Experiment system composition

In the first step, a 1 light source and 1 sensor system was created. Instead of detecting the CO_2 gas in the arterial blood, the CO_2 concentration was controlled by the MFC (Mass Flow Controller, PJ KODIVAC Co., Ltd, Japan). Its concentration was similar with that in the skin.

The experimental system consisted of an IR source, an IR detector (pyroelectric sensor), a gas cell made by MEMS technology, and a signal process block. Figure 3 shows a schematic diagram of the whole system. The optical system used a 4.26 ㎛ IR lamp[8,9] that CO_2 gas reacts to, while the electrical chopper provided a frequency of about 1.45 *Hz* and a light source by a collimator to collect light. A micro-machining process was used to fabricate the optical reaction chamber as a several micro-volume chamber that we considered that exhausted in human body. Figure 5 shows a photo-mask of the optical reaction chamber and manufactured optical reaction chamber with a volume of 64 ㎕, although diverse optical reaction chambers were made.

The optical detection part detects the output power from the pyroelectric sensor forward to 4.26 ㎛ of the optical filter. The signal processing block manufactured a circuit that can do the Lock-in amp's part which can manufacture Chebyshev's Fourth filter. We amplified the output signal using a cascade amplifier circuit.

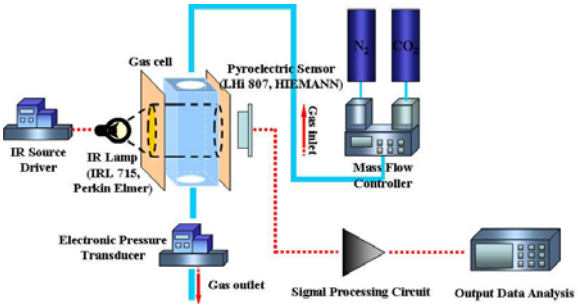

Figure 3: Schematic diagram of the measurement system

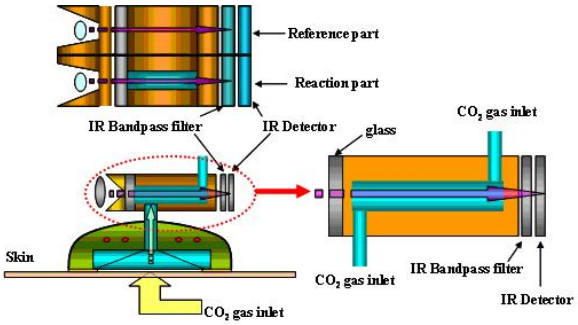

Figure 4: Schematic diagram of optical reaction chamber

Figure 6 is fabricated gas cell and head part of sensor. The gas cell is made by MEMS technology to analysis the small quantity of gases.

Figure 4 was final schematic of non-invasive transcutaneous pCO_2 gas sensing system. That was the composition of the optical reaction system of 2 light sources and 1 sensor used to decrease the effect of noise. That decreasing effect of the noise department, which was thought to be from the outside, is composed of 2 light sources and 1 sensor system with the maximum of the sensing rescue that we wish to embody in this study. The skin mount was then created to collect the CO_2 gas collected in the skin.

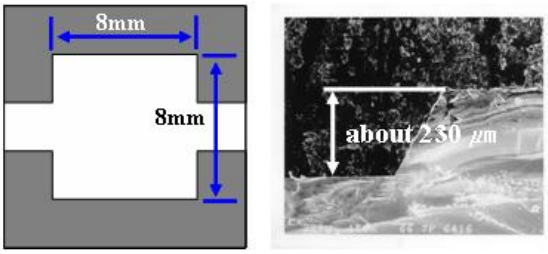

Figure 5: Mask layout of gas cell and SEM image of gas cell's cross section

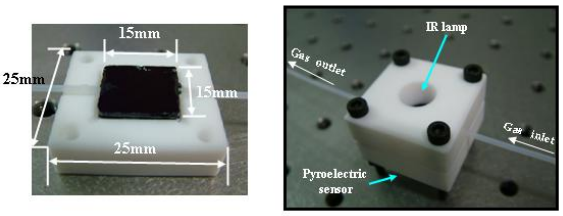

Figure 6: Fabricated gas cell and head part of sensor

4 Results

This study attempted to improve the CO_2 sensor in the existing NDIR method by reducing the optical reaction path and manufacturing a micro-volume optical reaction chamber using a semiconductor process and micro-machining technology. This gas monitoring system can accurately analyze the CO_2 gas concentrations based on a very small amount gases. The optimized optical reaction path was 1 ㎜ and the results were confirmed by the gas concentration in a 64 ㎕ gas capacity. Figure 7 is the sensitivity for the optical reaction length as a result of 8.11×10^{-2} *absorbance/mm*. We can know that the absorbance was directly proportional to the optical reaction length. Figure 8 confirm the measurement conclusion for the CO_2 gas concentration as a result of 2.21×10^{-6} *absorbance/ppm*. A linear result was confirmed in 1,000 *ppm* extent and a linear in 30,000 *ppm* extent, and is in 10 % extent of the amount by exhausted in skin. The response time for CO_2 gas in figure 11 was within 5 seconds, which is fairly fast compared to an invasive method or the $EtCO_2$ method.

Finally, we gather the CO_2 gas in exhausted in skin. So we check pressure vs. absorbance relationship. Figure 9 was showed that influenced absorption about 10%. We confirmed that the pressure is also affect to the absorbance.

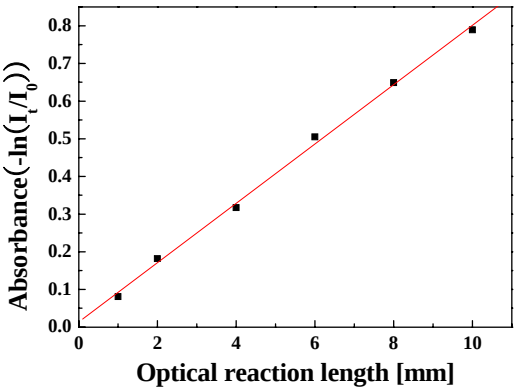

Figure 7: Absorbance vs. optical reaction length

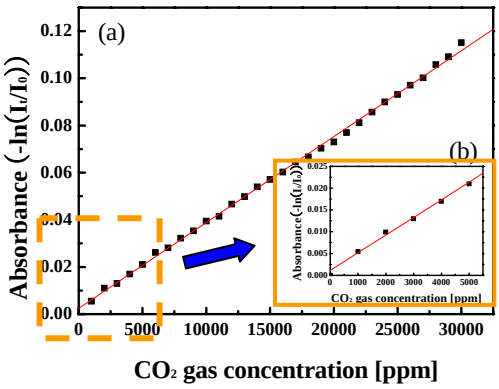

Figure 8: Absorbance vs. (a) wide range and (b) narrow range of CO_2 gas concentration

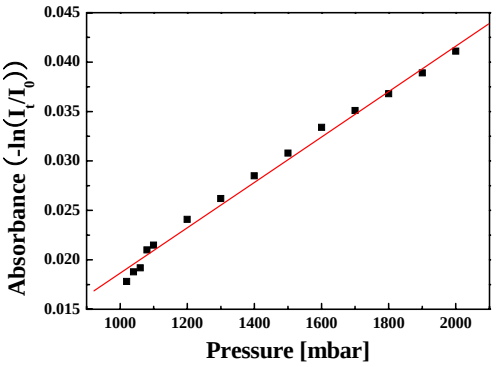

Figure 9: Absorbance vs. pressure of CO_2 gas

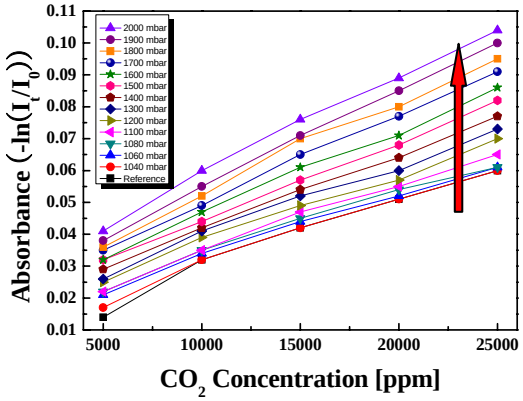

Figure 10: Absorbance vs. concentration and pressure of CO_2 gas

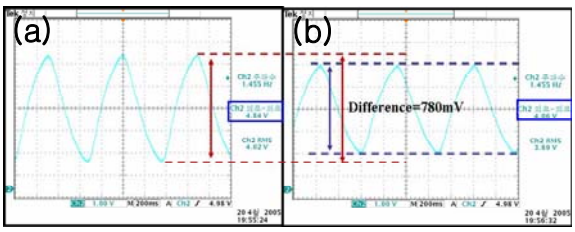

Figure 11: Response characteristic at 10,000*ppm* CO_2 (a) before gas inlet, (b) after gas inlet

5 Discussions

This study showed that the quantity of CO_2 gas in the blood goes through exhalation gas is improved by the invasive method, so in using the real-time measurement for possible transcutaneous non-invasive pCO_2 gas sensor, the basic step of the system development is achieved. In the first step, a result of 8.11×10^{-2} *absorbance/mm* sensitivity for optical reaction length, 2.21×10^{-6} *absorbance/ppm* sensitivity for concentration at 1,040 *mbar*, 2.14×10^{-5} *absorbance/mbar* sensitivity for pressure at 5,000 *ppm* CO_2 was confirmed using a optical reaction path of 1 ㎜ with a CO_2 gas reaction.

A conclusion by verified gas concentration produced a relative linear result in 10 % within of pCO_2 gas concentration. This experiment showed that the very small amounts of gas exhausted by real skin can be measured.

In the future, design of system with the pump and CO_2 collection can make response time extension. But as the absorbance is proportional for the optical reaction length, we can prevent the delay of response time by lengthening optical reaction length. Also considering response time in skin, we need enough heating time because of the small quantity gas from the skin. The CO_2 reaction time from the skin was about 10 sec after skin heating. And the heating pad does not give rise to patient discomfort. Only weak damage of skin by vacuum is expected. According to basis clinical demonstration, about 38~41℃ heating which is imposed to massed capillary do not give rise to patient discomfort.

In future designs are needed for a skin mount that contains a heating pad to collect CO_2 gas from the outer skin, a small-size vacuum pump to inject CO_2 gas into the optical reaction chamber, an optical system and skin mount, packaging technology for a vacuum pump, a dehumidification method and a clinical test. Based on such processes, the development of a portable transcutaneous non-invasive pCO_2 gas sensor system is surely possible.

6 Acknowledgement

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