Sensors and Materials, Vol. 17, No. 5 (2005) 249–257
MYU Tokyo

S & M 0603

Noninvasive Optical Transcutaneous pCO_2 Gas Sensor

Do Eok Kim*, Hwan Joo Lee, Eun Jong Cha[1], Dae Hyuk Kwon[2], Seung Ha Lee[3] and Shin Won Kang[3]

Department of Electronic Engineering, Kyungpook National University, Korea
[1]Department of Biomedical Engineering, Chungbuk National University, Korea
[2]Department of Electronic and Information Engineering, Kyungil University, Korea
[3]School of Electrical Engineering and Computer Science, Kyungpook National University, Daegu 702-701, Korea

(Received August 21, 2004; accepted December 5, 2004)

Key words: pCO_2, noninvasive, NDIR, Beer-Lambert's law, Henry's law, pyroelectric

This paper presents an optical transcutaneous pCO_2 gas sensor and analyzer using a noninvasive method. The basic principle of the pCO_2 measurement method adapts Beer-Lambert's law and the embodied system uses the nondispersive Infrared (NDIR) method. Since CO_2 gas reacts to a 4.3 μm wavelength, this wavelength is selected using an optical filter, and used energy decrease by molecule oscillations. The CO_2 concentration is then measured by a mass flow controller (MFC) using basic steps, instead of collecting pCO_2 gas by inflicting heat on the outer skin. The measuring system consists of an IR lamp, optical filter, optical reaction chamber, pyroelectric sensor, and a signal processing part. To make the sensor system portable, the length of the optical reaction chamber is minimized to 1 mm using a Si wafer based on MEMS technology. When CO_2 gas is injected into the optical reaction chamber, a result of 4.3 mV was confirmed when using a photoreaction path of 1 mm with a CO_2 gas reaction. The response time of the system was within 2 s, which we consider to be relatively rapid.

1. Introduction

To estimate the respiratory ventilation ability in the acid-base balance of metabolism, it is essential that the CO_2 partial pressure be measured from arterial blood. However, this is invariably restricted in serious cases or in 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 must be done several times.(1) One noninvasive measurement option is where the pCO_2 is presumed on the basis of the $EtCO_2$, the concentration of CO_2 in the patient's exhaled breath. However, this is also troublesome,

*Corresponding author, e-mail address: dekim@ee.ac.kr

as patients are required to breathe through a duct. The CO_2 that is produced as 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 P_aCO_2-$P_{cap}CO_2$-P_vCO_2. Countercurrent (diffusion), the CO_2 gas transfer mechanism of the venous-arterial system, is one of the major factors. The overall diagram is presented below (Fig. 1).

If it were possible to measure the CO_2 partial pressure in blood that is diffused from the capillary vessels to the skin, this would facilitate noninvasive and nontroublesome measurements of the CO_2 concentration in the blood. 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, because the hypodermal CO_2 concentration increase is accompanied by a rapid arterial blood CO_2 concentration increase.[2] Yet, an invasive method that measures the CO_2 concentration from the arterial blood directly requires a considerably longer time, which is inappropriate in such cases. Technology of measuring the CO_2 gas diffused in the skin capillaries by a noninvasive method would thus be very useful and clinically important .[3] However, this has not yet been studied extensively, even in developed countries, due to technological difficulties.

Currently there is only one product that uses an optical method, the Hewlett-Packard Model 47210A capnometer.[4] However, this product still needs improvement because it produces excessive and highly inaccurate sensor parameters. The use of an electrochemical method has also been studied and developed, although the complexity of the structure, technological difficulties, and troubles with the perception film control has blocked its success. Therefore, the objective of this study is the development of a convenient, noninvasive transcutaneous pCO_2 analysis device that uses an optical method. As such, effective infrared optical sensor cartridges and light sources were investigated, along with

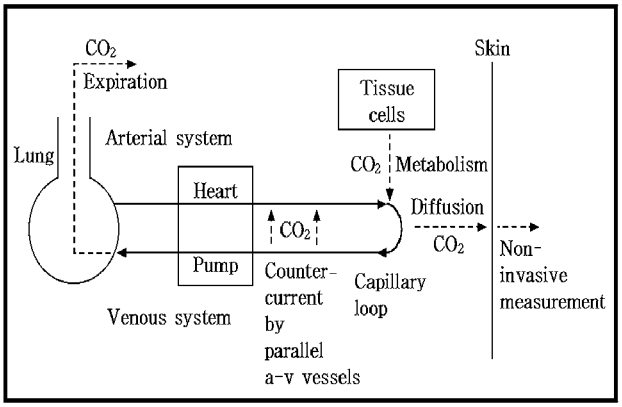

Fig. 1. Schematic diagram of venous-arterial system.

the signal processing. From 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 noninvasive method for the diagnosis and care of emergency patients, including serious cases and patients in surgery who need their CO_2 concentration monitored.

2. Materials and Methods

2.1 *Theory*

2.1.1 *Beer-Lambert's law*

The fundamental theory governing absorption spectroscopy is based on Beer-Lambert's law.[8]

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 ϕ (cm), total pressure P (atm), absorption coefficient for concentration $\in$ ($M^{-1}cm^{-1}$), material concentration C (Mol), and path length L (cm):

$$I_1 = I_0 exp(-S_i \phi P \in CL). \quad (1)$$

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

$$A(\alpha) = In(I_t/I_0) = S_i \phi \in CL. \quad (2)$$

The absorption coefficient is defined as

$$\alpha(v) = S_i \phi P \in CL, \quad (3)$$

From eq. (2), the absorption $A(\alpha)$ can be described as

$$A(\alpha) = \in CL. \quad (4)$$

In this equation, the absorption $A(\alpha)$ is linearly proportional to the concentration C of the measured gas and path length L.[5,6]

2.1.2 *Henry's law*

First, it is essential to confirm Henry's law[7] for converting the CO_2 concentration.

When a liquid and a gas exist in a parallel state, the partial pressure depends on the temperature, and the condition of dissociation is released as the melting state changes. 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,

$$C_g = A\alpha_g P_g. \tag{5}$$

The C_g value is defined as %Vol, which means the quantity of gas per 100 ml 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.

2.1.3 *Pyroelectric effect*

The quantity of electric charge per unit area of a vertical face in a spontaneous polarization direction is called the spontaneous polarization on dipole moment per unit volume. The electric charge is where the insulation that always floats on the surface of a pyroelectric material is absorbed by the surface charge based on spontaneous polarization. The amount of spontaneous polarization is affected by temperature changes, due to the light irradiation on the surface of the material. The neutralization state is broken down and the relaxation time changed between the perception element surface charge and the adsorption floating ionic charge; thus, the electrical equilibrium is broken and a charge occurs that combines the companions. Therefore, the phenomenon whereby the spontaneous polarization of the surface charge changes according to a temperature change is called the pyroelectric effect.[6]

The changes in the charge and current provide a neutralization state. Thus, blocking the light will halt the temperature change in the material. Figure 2 shows the pyroelectric effect when the light is blocked using an electrical chopper. With the chopper, there is no output power from the sensor. Yet, when the chopper is removed, the output power reappears due to the temperature change.

However, an equilibrium occurs because the surface charge is neutralized again by a charge in time. If the chopper is used, the amount of infrared rays decreases and the equilibrium of the charge is broken. As time passes the neutralization state reappears on

Table 1
Bunsen solubility coefficients. [ml (STPD)/ml 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

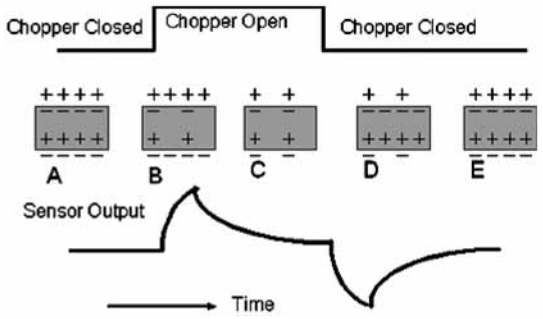

Fig. 2. Pyroelectric effect representation in using chopper.

the superofferings surface. As such, a charge-discharge signal can be received from the sensor by using and removing the electrical chopper.

2.2 Experimental system

Figure 3 shows the composition of the photoreaction system with 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. A mount was then created to collect the CO_2 gas collected in the skin.

In the first step, a 1 light source and 1 sensor system was created. The CO_2 concentration was controlled by the MFC (Mass Flow Controller, PJ KODIVAC Co., Ltd, Japan). Figure 4 shows a schematic diagram of the 1 light source and 1 sensor system.

The experimental system consisted of an optical source, a photoreaction chamber, and photodetection. Figure 5 shows a schematic diagram of the whole system. The optical system used a 4.3 μm IR lamp,[8,9] which CO_2 gas reacts to, while the electrical chopper provided a frequency of about 2 Hz and a light source from a collimator to collect light. A micromachining process was used to fabricate the photoreaction chamber as a microvolume chamber that we used as a simulation for exhausting in a human body. Figure 6 shows a photomask of the photoreaction chamber and manufactured photoreaction chamber with a volume of 20 μl, although diverse photoreaction chambers were fabricated.

The optical detection part detects the output power from the pyroelectric sensor forward to 4.3 μm the optical filter. In the signal processing stage, a circuit was fabricated that can perform the lock-in amp's function, which can perform the fabrication of Chebyshev's Fourth filter. We amplified the output signal using a cascade amplifier circuit.

Instead of detecting the CO_2 gas in the skin, the CO_2 concentration (minimum 1,000 ppm) was controlled using the MFC. The concentration was similar to that in the skin.

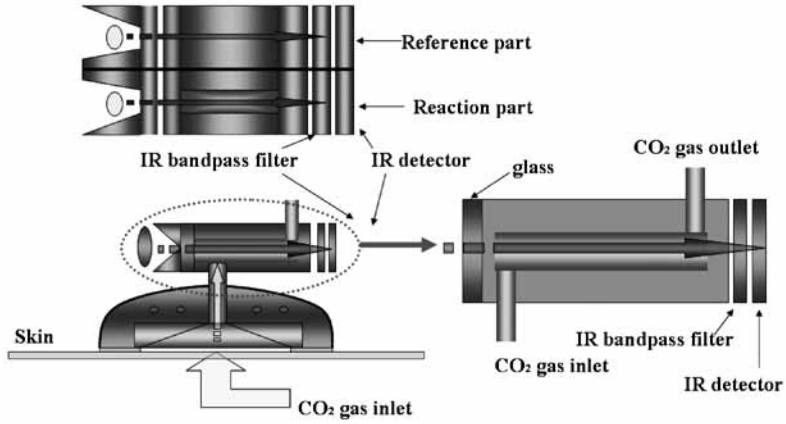

Fig. 3. Schematic diagram of optical reaction chamber.

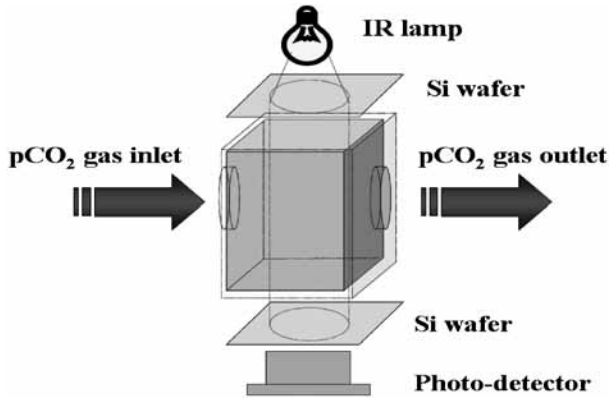

Fig. 4. Condensation model of optical reaction chamber.

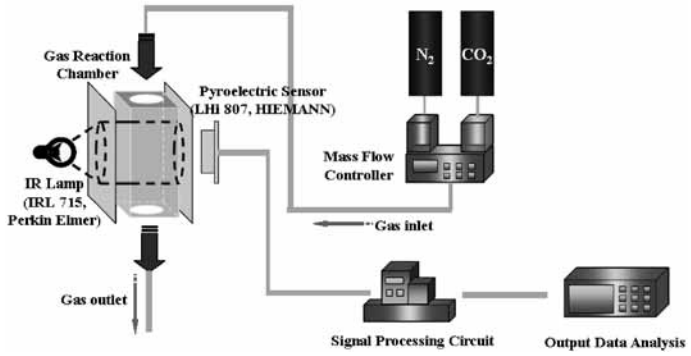

Fig. 5. Schematic diagram of measurement system.

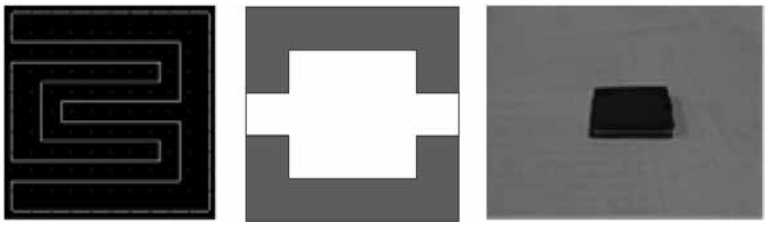

Fig. 6. Mask layout of optical reaction chamber.

3. Results

This study attempted to improve the CO_2 sensor in the existing NDIR method by reducing the photoreaction path and manufacturing a microphotoreaction chamber using a semiconductor process and micromachining technology. As such, the resulting gas sensor system can accurately analyze concentrations based on a very small amount of gas. Figure 7 presents the results with photoreaction paths of various lengths and confirms a sensitivity of 0.43 mV per length. The optimized photoreaction path was 1 mm and the results were confirmed by the gas concentration in a 20 μl gas capacity. Figures 8 and 9 then confirm the measurement conclusion according to the gas concentration. A linear result was confirmed for 2,000 ppm as well as for 20,000 ppm, and is in 10% extent of the amount by exhausted in skin. The response time for CO_2 gas, shown in Fig. 10, was within 2 s, which is fairly fast compared to an invasive method or the $EtCO_2$ method.

4. Discussion

This study showed that the quantity of CO_2 gas in the blood goes through exhalation gas or is improved by the invasive method. Using real-time measurement in a possible transcutaneous noninvasive pCO_2 gas sensor, the first step of the system development, has been achieved. In the first step, a result of 4.3 mV was confirmed using a photoreaction path of1 mm with a CO_2 gas reaction.

A conclusion by verified gas concentration produced a relative linear result, within 10% of $EtCO_2$ gas concentration. This experiment showed that the very small amounts of gas exhausted by real skin can be measured.

In the future, designs are needed for a skin mount that contains a heating wire to collect CO_2 gas from the outer skin, a small-size vacuum pump to inject CO_2 gas into the photoreaction chamber, an optical system and skin mount, packaging technology for a vacuum pump, and a dehumidification method. We believe that the development of a portable transcutaneous noninvasive pCO_2 gas sensor system based such processes is possible.

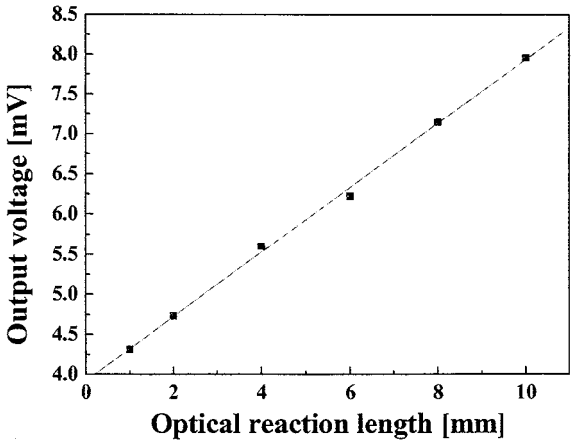

Fig. 7. Output voltage vs optical reaction length.

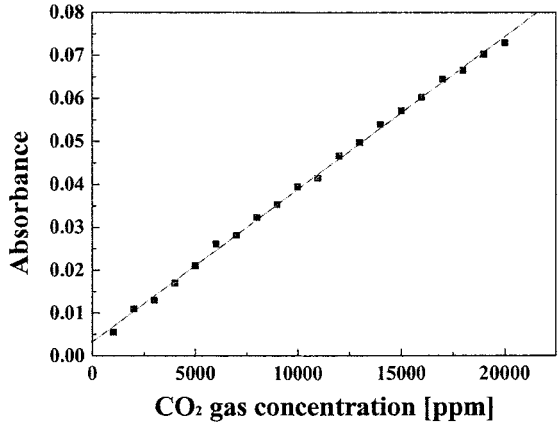

Fig. 8. Absorbance vs EtCO2 gas concentration (1,000ppm–20,000ppm).

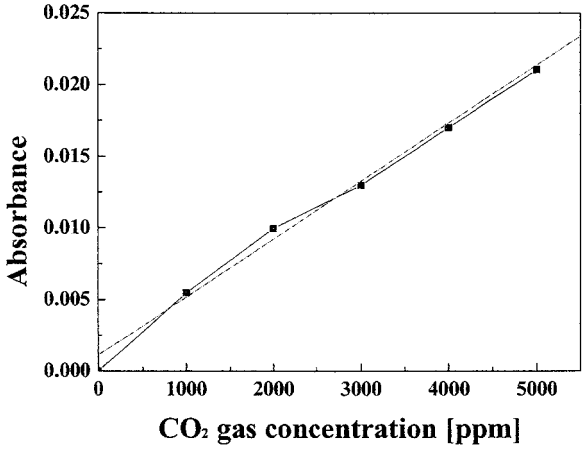

Fig. 9. Absorbance vs pCO_2 gas concentration (1,000 ppm ~ 5,000 ppm).

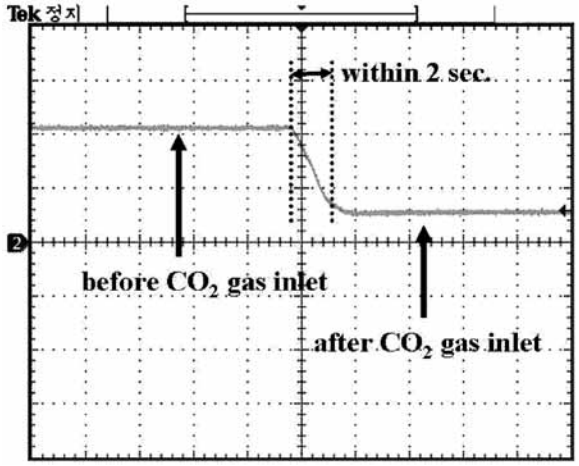

Fig. 10. Response time of fabricated system.

Acknowledgement

This study was supported by a Grant from the Korea Health 21 R&D Project, Ministry of Health & Welfare, Republic of Korea.

References

1 H. Y. Kwon : Koryo Medicine, 2001.
2 J. G. Webster: WILEY, 1999.
3 K. K. Tremper, B. A. Mentelos and W. C. Shoemaker: CRC Crit. Care Med., **8** (1980) 608.
4 G. H. Greenspan, A. J. Block, L. W. Haldeman, S. Lindsey, and C. S. Martin: Chest **80** (1981) 442.
5 H. Ahlberg, S. Lundqvist, R. Tell and T. Andersson: **2112** (1994) 118.
6 N. J. Choi : Master thesis Kyungpook National University, 1998.
7 L. H. Peterson, M.D.: Modern concepts in medical physiology, 1982.
8 J.Quant: Spectrosc. Radiat. Transfer, **60** (1988) 665, HITRAN atmospheric workstation.
9 G. Herzberg: Krieger Publishing Company, Reprint Edition, 1991.