Highly Sensitive Optical Sensor System for Blood Leakage Detection

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ABSTRACT

A highly sensitive method for the detection of blood leakage has been developed, and a practical sensor system for blood concentration measurement has been constructed. The present method is based on the attenuation of laser light by blood cells. The effects of the fluctuations of the incident laser light power are eliminated by normalizing the attenuated light intensity by the incident light intensity. A part of the incident laser light is reflected by a beam splitter mounted at the entrance of the test cell, of which the power is measured to provide base data for normalization. The optical path is extended to enhance sensitivity by using a pair of side mirrors. This multi-reflection method is very effective to increase sensitivity; the maximum sensitivity obtained for blood concentration is about 4×10^{-6} *by volume, which is significantly higher than that of the conventional sensors.*

1 INTRODUCTION

Recently, high sensitivities are needed more and more in the field of medical electronics. One typical application is a blood leak sensing system used in an artificial dialyzer. A number of sensors such as a temperature sensor, flow rate sensor, pressure sensor, negative pressure sensor and bubble sensor are used as well as a blood leak sensor in an artificial dialyzer. A number of dialyzing treatments are normally needed in any one week for a patient suffering from a kidney disorder, and therefore a blood leak could potentially cause fatal damage, even

if leakage in each dialyzing treatment is a very small amount. For this reason numerous research projects have focused on a blood leak sensor.[1]

Most instruments used to detect blood are based on an optical non-intrusive method using an infrared ray or a visible ray from a semi-conductor laser or a diode. In the systems presently used for the dialyzing treatment, an alarm is triggered only when blood concentration exceeds a pre-set threshold level, and even then no quantitative data of the total amount of blood leakage is provided. Furthermore, a patient's life is in danger during the treatment since the sensitivity of the sensor in current use is too low to detect a minor leakage. A patient can be released from the danger if a highly sensitive blood leak sensor is developed which gives a linear sensitivity with concentration in real time.

The main principle of the present method is based on the light attenuation theory. Laser light is attenuated by the presence of blood cells in the optical path in a container, and thus the measurement of light attenuation gives the concentration of blood. The sensitivity and the accuracy may then depend on the sensitivity and the stability of the intensity measurement system, and on the fluctuations of the incident light, respectively. The sensitivity for blood detection of present ranges between 0·01 and 0·001 in weight %[2] (weight ratio of blood to solution), which is too small for recent medical appliances.

The purpose of this study is to propose a simple method for the improvement of both the sensitivity and stability of blood concentration measurement, without using complicated equipment to stabilize the laser output, and to describe the construction of a practical sensor system.

2 PRINCIPLE AND METHOD

Light intensity decreases by absorption or scattering when it propagates in a non-transparent medium. We can then determine the concentration or density of the medium by measuring the light attenuation. Since the intensity decreases in proportion to an optical path length through the medium (if the medium is homogeneous), the path length is extended by the multi-reflection technique, using a pair of side mirrors mounted on the test cell in order to enhance the sensitivity. The present technique is effective for a higher sensitivity, however the use of a narrow-beamed laser is necessary to avoid an over-enlarging beam size at the exit of the cell.

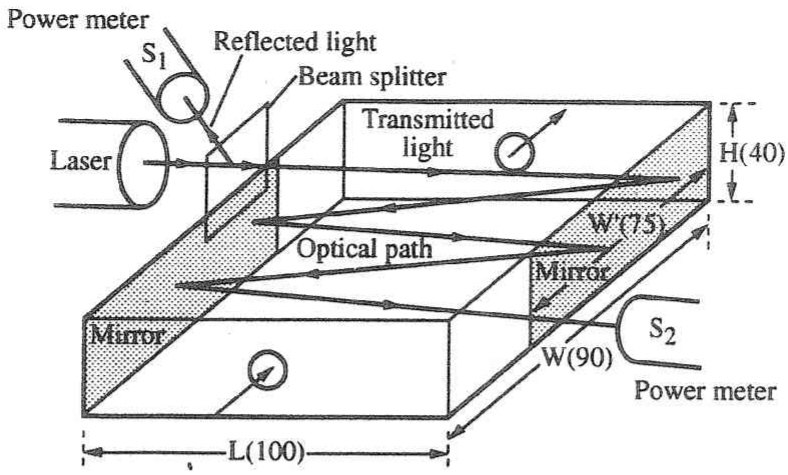

Fig. 1. Experimental setup of the sensor. (Numerical values in this figure are in mm).

Figure 1 shows the experimental setup. Laser light from a semi-conductor laser is divided into two beams by a beam splitter, one is a reflected beam and the other a transmitted beam. The reflected beam directly enters a light power meter (S_1) and its power is measured to provide the incident laser light level to be used for the normalization of the attenuated light. The transmitted beam enters a cell in which a test subject is filled and is led to a light power meter (S_2) after the multi-reflection on a pair of side mirrors inside the cell. The attenuated power is measured by the meter (S_2) and is divided by the power of the incident light given by S_1. This normalization automatically eliminates errors due to the unknown fluctuations in the intensity of the incident beam. The multi-reflection extends the optical path length and thereby increases the sensitivity. A semi-conductor laser with a wavelength of 680 nm has been chosen because it is reasonably priced and is easily available for practical use. A physiological salt solution mixed with blood was used as a test subject in order to evaluate the system sensitivity and stability. The light attenuation of the laser was caused by scattering rather than by absorption, since the laser light irradiates a red corpuscle.

The principle of the method is also shown by a schematic diagram in Fig. 2. Although the light attenuation due to light scattering in a weak solution has been discussed in detail in the literature,[3,4] Lambert's law of simple estimation of attenuation has been used, whereby the light decreases exponentially with the path length in the medium. That is, the light power (I_0) at a distance (x) from the entrance is related to the incident light power (I_i), as follows

$$I_0(x) = I_i \exp(-\alpha x) \tag{1}$$

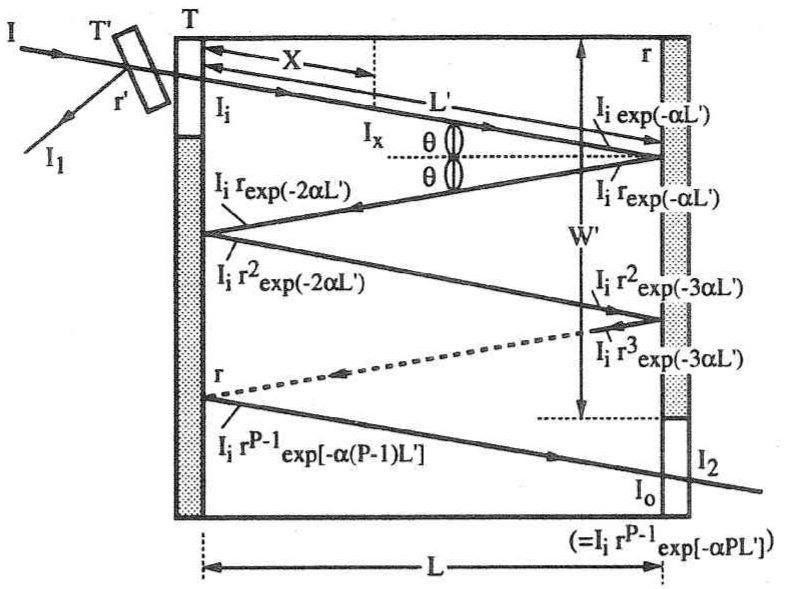

Fig. 2. Schematic diagram of the optics.

where α is an absorption coefficient which depends only on the concentration or the density of the subject (n), to be measured. Applying eqn (1) to the present method (see Fig. 2), the light power, (I_2) received in the second sensor (S_2) can be related to the light power (I_1), received in the first sensor (S_1) as follows

$$I_2/I_1 = (T^2T'/r')\{I_0(x)/I_i\} = (T^2T'/r')\exp(-\alpha pL')r^{P-1} \tag{2}$$

where

$$L' = L/\cos\theta, \quad \alpha \equiv \alpha(n)$$

where T and T' are the transmissivities of the beam splitter and the test cell glass, respectively, r' and r are the reflexibilities of the beam splitter and the side mirrors in the cell, respectively, P is a normalized optical path length defined by $p = x/L'$, where x shows a total optical path length. As is shown in eqn (2), the normalized output light power, I_2/I_1, is directly proportional to $I_0(x)/I_i$. A logarithmic expression of eqn (2) is given as

$$\log(I_2/I_1) = (\log r - \alpha L' \log_{10} e)p + \log(T^2T'/r'r) \tag{3}$$

An absorbance defined by $\log(I_2/I_1)$ is thus directly proportional to p. As shown in Fig. 3, the linearity between $\log(I_2/I_1)$ and p was determined experimentally. The gradient of the straight line, $\log r - \alpha L' \log_{10} e$, shows an apparent absorption coefficient $\alpha'(= \log r - \alpha L' \log_{10} e)$,

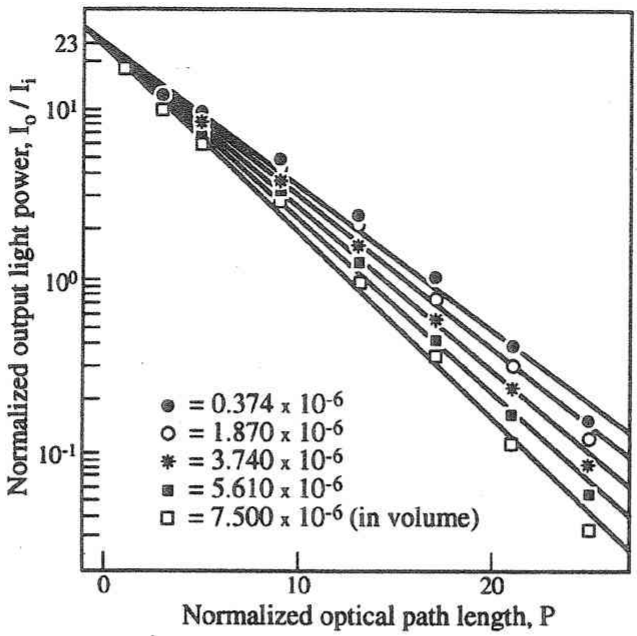

Fig. 3. Effect of the optical path length, p, on the normalized output light power I_2/I_1 (in a logarithmic scale) for blood concentration, n.

from which an absorption coefficient α can be calculated as follows

$$\alpha = [(\log r - \alpha')/(L \log_{10} e)][1 + W'^2/(Lp)^2]^{-1/2} \qquad (4)$$

The multi-reflections technique does not allow us to set the incident angle of the laser beam to be normal to the mirrors in order to maintain its consecutive reflections. This requires the term $[1 + w'^2/(Lp)^2]^{-1/2}$ in eqn (4), which compensates for its effects on α. In this setup, the value of p in the equation is usually larger than three and the magnitude of the effects of the shift is considered to be less than 3%.

The measured data of the light power, I_1 and I_2, are processed by a data processing system, shown in Fig. 4. The continuous analog signals

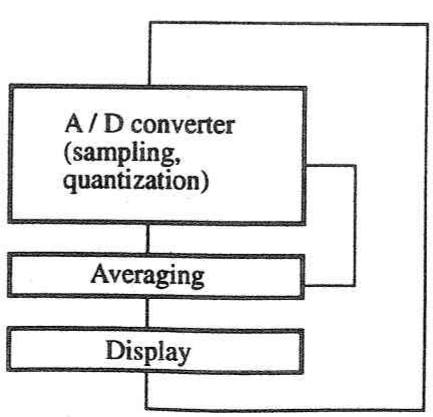

Fig. 4. Block diagram for the data acquisition system.

of I_1 and I_2 are sampled, digitized and stored in a computer where the operation of I_2/I_1 is completed. Results are given in the form of curves after smoothing. The resolution of the data acquisition system is 12 bits and the sampling rate is about 800/s. One data point in the figure is made by averaging 50 measured data, and then requires 63 μs sampling time.

3 EXPERIMENTAL RESULTS AND DISCUSSION

Before taking experimental data, measurements were made in order to determine the system parameters of the optical components, such as the transmissivity (T and T'), and the reflection coefficient, (r and r'). The measurements gave $T = 0{\cdot}93$, $T' = 0{\cdot}90$, $r = 0{\cdot}90$ and $r' = 0{\cdot}04$. This gives us $T^2T'/(r'r) = 21{\cdot}6$, which is in good agreement with the experimental data obtained at $p = 0$ in Fig. 3.

First, the effects of the optical path length were investigated. Experiments were carried out by changing only the optical path length, p, keeping the concentration and laser output constant. Second, a series of experiments were conducted for various concentrations and laser outputs. Results obtained are summarized in Figs. 3 and 5. The normalized output light power, I_2/I_1, in the logarithmic scale decreases linearly with the normalized optical path length, p, as shown in Fig. 3, as well as with the concentration, n, as shown in Fig. 5. The straight

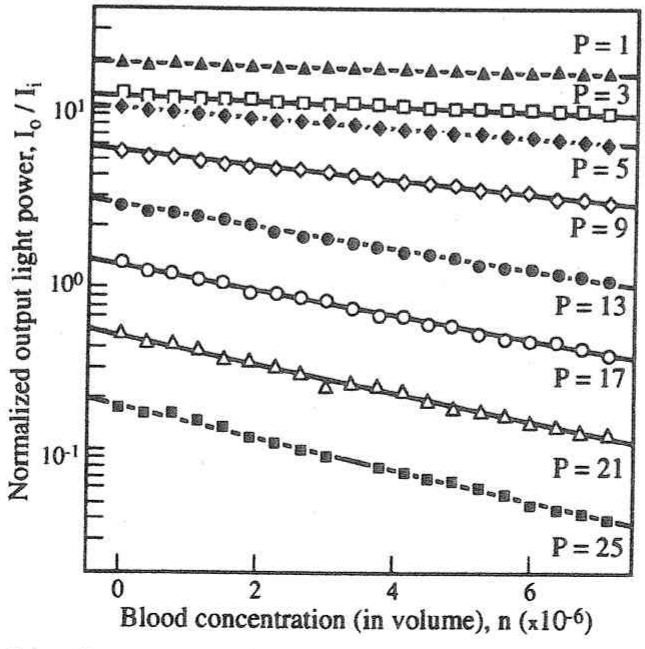

Fig. 5. Effect of the blood concentration, n, on the normalized output light power I_2/I_1 (in a logarithmic scale) for optical path length, p.

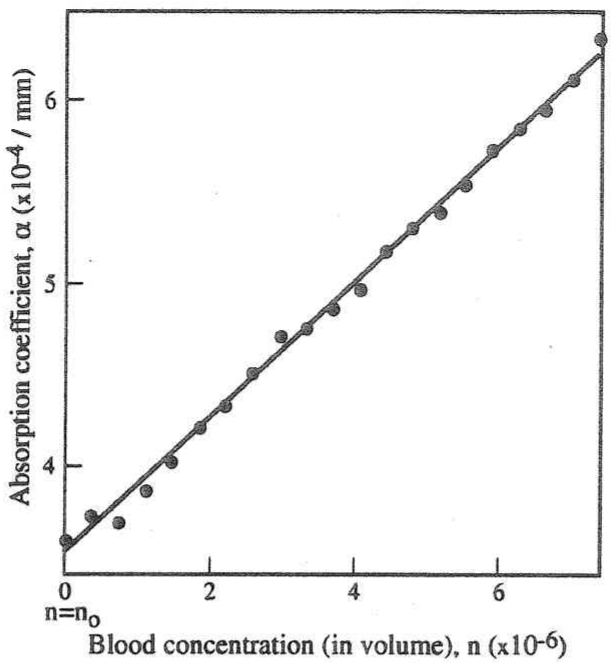

Fig. 6. Effect of the blood concentration, n, on the absorption coefficient, α.

lines in Figs. 3 and 5 correspond to each concentration and normalized optical path length, respectively, and were obtained by the least squares method. It can be seen from these two figures that the normalized output power decreases almost exponentially with the product of the normalized optical path length, p, and the concentration, n. Combining this with eqn (2) implies that the absorption coefficient, α, is to be proportional to n.

The absorption coefficient, α, for each concentration, n, is shown in Fig. 6, which is determined from the gradient of the straight lines in Fig. 3. As seen in the figure, the absorption coefficient increases linearly with concentration as discussed above, and can be expressed as follows

$$\alpha = cn + n_0 \tag{5}$$

where c is a constant independent of concentration (n) and optical path length (p). n_0 is the absorption coefficient of the solution without blood and is estimated experimentally to be $n_0 = 3{\cdot}55 \times 10^{-4}$/mm, as seen in Fig. 6. The constant c can also be determined experimentally by the gradient of the straight line in Fig. 6 and is calculated as 36·4/(mm × concentration in volume). It depends only on the interaction between blood and the laser, and therefore the wavelength of the laser plays an important role in determining c.

When red laser light is used, as in this case, red corpuscles scatter the red light; the constant c is mainly determined not by absorption but by scattering. If a blue laser is used, on the other hand, the attenuation is

caused by absorption. Combining eqn (5) with Lambert's law (eqn (2)) results in the following Lambert-Beer's law[5] which is applicable for a diluted solution, as in this case

$$I_2/I_1 = (T^2T'/r') \exp\{-(cn + n_0)pL'\}r^{P-1} \tag{6}$$

The sensitivity is determined in practice by dividing an increment of the normalized output laser power by the corresponding increment of the blood concentration, n, in a solution, i.e.

$$\begin{aligned} S(p) &= |\Delta(I_2)/I_2|/|\Delta n| \\ &= |\{\Delta(I_2/I_1)/(I_2/I_1)\}/\Delta n| = cpL' \end{aligned} \tag{7}$$

Then the relative sensitivity K divided by $S(1)$ gives,

$$K(p) = S(p)/S(1) = p \tag{8}$$

The sensitivity for $p = 1$ corresponds to one of the conventional sensors. It is apparent that the sensitivity of the present method is p times higher than that of conventional sensors, although there is an advantage with the conventional sensors in that they do not require an expensive laser (an economical laser diode is quite enough for conventional sensors since the path length is shorter).

The normalized optical path length (p) can be multiplied by repeating the reflection on the pair of side mirrors, if necessary, to increase the sensitivity. However, the maximum sensitivity is limited by a cross sectional area of the laser beam after the multi-reflections because the laser beams should not overlap each other on the side mirrors. The maximum sensitivity can then be given by $p_{max} = W'/d$, where W' is a width of the side mirrors as shown in Fig. 1, and d the laser beam diameter.

The normalized output power I_2/I_1 is completely independent of the fluctuations of the laser output power, as can be seen in eqns (2) and (3). This means that a laser can be used without any complicated stabilizing equipment or compensator for the light source, and is another distinguishing feature of the present system, in addition to its high sensitivity. The independence of I_2/I_1 from the laser fluctuations has been confirmed by varying the laser output over an intentionally wide range. The result is shown in Fig. 7. The normalized output laser power is almost constant over the range of the laser output, I/I_{max}, where I_{max} is the maximum output power of the laser.

The cell configuration used in this study is rectangular as shown in Fig. 1. However, a cylindrical cell, as shown in Fig. 8, may be more practical from the view point of commercial production. With this

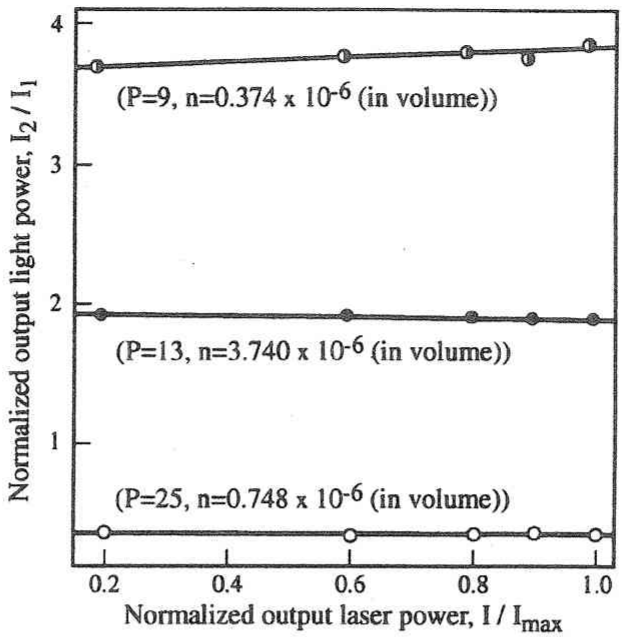

Fig. 7. Fluctuations of the normalized output power, I_2/I_1. (The laser output power, I, is normalized by the maximum output power, I_{max}).

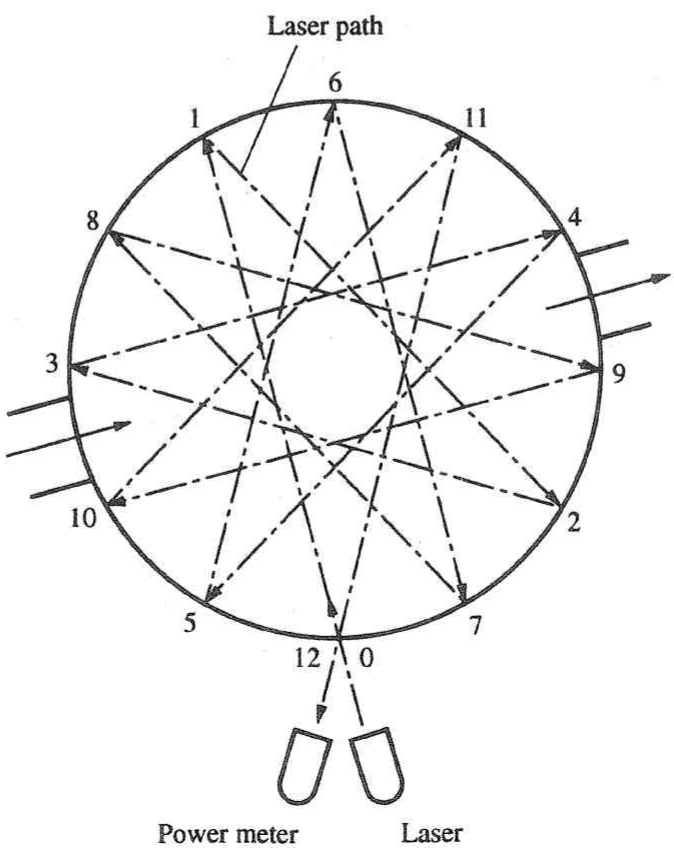

Fig. 8. A proposed cylindrical configulation for a practical sensor.

configuration, star-shape multi-reflections can be applied and then both the inlet and outlet of the laser lights can access the same window.

4 CONCLUSIONS

The sensitivity of the sensor for blood leak detection has been improved by the use of a beam splitter and a pair of side mirrors. This improvement gives a few tens of times higher sensitivity than conventional sensors currently on the market. Furthermore, the fluctuations of the laser power are completely compensated for by using a beam splitter.

ACKNOWLEDGEMENT

Technical support throughout the present study was provided by Mr T. Inaki of Fukui University, Fukui, Japan.

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