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Continuous measurement of hematocrit using an intravascular catheter equipped with a fiberoptic transmission cell

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Abstract

Purpose. The purpose of this study was to measure intravascular hematocrit values continuously by using a fiberoptic probe based on near-infrared photometry.

Methods. We produced a catheter 1.5 mm in diameter that used a pair of plastic fibers. One of the fibers, the measuring fiber, was used to measure the optical density of blood, and the other, the reference fiber, was used to decrease the signal-to-noise ratio. We employed an 805-nm laser diode as the light source. Two photodiodes were used to measure the intensity of the light transmitted through the two fibers, and the output signals were amplified and sent to a personal computer through an analog-to-digital converter.

Results. The hematocrit values obtained by this fiberoptic continuous measurement agreed well with those obtained by microcentrifugation within physiological ranges.

Conclusions. This method is effective for monitoring the rapid changes in hematocrit.

Key words: Intravascular, Hematocrit, Optical density, Monitoring, Fiber-optics

Introduction

The hematocrit is one of the most frequently used values in the evaluation of circulating blood volume in the clinical field. The need for frequent measurements of hematocrit in perioperative patients has generated the desire to develop a new method to measure hematocrit. Continuous measurements of hematocrit by electrical impedance [1] and optical density [2–7] have been undertaken, but these methods have not reached routine use because they require an extracorporeal circuit.

Recent advances in hemodynamic monitoring have made accurate and continuous measurement of mixed venous oxygen saturation ($S\bar{v}O_2$) possible by means of the fiberoptic thermodilution pulmonary artery catheter. The fiberoptic catheter provides continuous on-line measurements of $S\bar{v}O_2$ by means of two optical fibers. Light emitted from diodes is transmitted via one fiber to the catheter tip, and the light reflected from oxyhemoglobin and deoxyhemoglobin in erythrocytes passing by the catheter tip returns via another fiber to the photodetector. Fiberoptic continuous measurements of $S\bar{v}O_2$, however, are subject to error if the hematocrit changes under severe hemorrhage or overhydration during clinical procedures [8], and frequent calibrations are necessary to make this method viable with a wide variance of hematocrit. If hematocrit values can be measured continuously and simultaneously, this problem will be resolved.

We expected that the fiberoptic transmission catheter reported by Iwasaki et al. [5] would be able to provide continuous measurements of hematocrit, thus avoiding the disadvantages of the previous methods. This fiberoptic catheter for the measurement of oxyhemoglobin saturation has a transmission cell composed of face-to-face positioned optic-fiber tips fixed on the catheter, and it can detect the light transmitted through blood samples in vessels. When we applied the fiberoptic probes modeled on transmission catheters to measure hematocrit, it was apparent that these catheters could not measure hematocrit using a single wavelength of light. Then another fiber was fixed parallel to the measuring fiber to exclude modifications of light due to the nature of the plastic optic-fiber.

The purposes of this study were to compare the hematocrit with the optical density measured with our method using an improved fiberoptic transmission catheter in vitro and to report the results of continuous measurement of hematocrit in vivo. Our device showed high performance as a continuous hematocrit monitor. These results suggested that our method improved fiberoptic oximetry.

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Methods

The experimental procedures and the protocols were approved by the Animal Experiment Ethics Committee of the Kyoto Prefectural University of Medicine.

Instruments

A fiberoptic transmission catheter was fabricated as described by Iwasaki et al. [5]. Although plastic fibers are highly flexible, the transmitted light is affected by the bends and movements of the fiber. Fine fluctuations of the source light also cause noisy signals. To eliminate this noise, our catheter was equipped with a reference fiber parallel to the measuring fiber.

A pair of plastic fibers (0.25 mm in diameter, SK-10, Mitsubishi Rayon, Tokyo, Japan) was conducted through one conduit of a triple-lumen polyethylene tubing (1.5 mm in outer diameter, TP-4, Natsume, Tokyo, Japan) (Fig. 1). A stainless-steel pipe (0.4 mm in outer diameter) was inserted in the lumen near the tip of the catheter. The outer wall of the lumen was removed 2 mm along the steel pipe, and one of the plastic fibers (measuring fiber) was exposed. The other (reference fiber) was passed through the pipe. After the measuring fiber had been fixed on the surface of the pipe with epoxy resin, a small piece of the fiber corresponding to the light pathlength was cut off so that the cross sections of the remaining parts faced each other to compose a transmission cell. The optic fibers were bent like a hairpin at the tip of the tubing and conducted through another lumen. The tip was plugged with epoxy resin. The third lumen was used to withdraw blood samples from a pore opened at the transmission cell.

A diode laser device (KLS-3W, Kette, Tokyo, Japan) was used as the source of light, the spectrum of which was centered at 805 nm. Absorption of the light is not affected by the oxidation state of the hemoglobin, because the wavelength is one of the isosbestic points of oxyhemoglobin and deoxyhemoglobin. Moreover, tissue and water are highly permeable to the light.

The intensities of the light transmitted through both the measuring fiber (I_m) and the reference fiber (I_r) were measured by photodiodes combined with amplifiers (S2281 and C2719, Hamamatsu Photonics, Shizuoka, Japan). The output signals were sent to a computer (PC9821Xe10, NEC, Tokyo, Japan) via an A/D converter board (ADXM-98S, Canopus, Kobe, Japan). The sampling rate was 100 kHz (10 ms), and the averaged data were displayed and recorded every second.

Theoretical considerations

The optical density (OD) of a solution is defined as

$$\mathrm{OD} = \log\left(I_0/I\right) \quad (1)$$

where I_0 and I are the intensities of the incident and transmitted light. We defined I_0 as the intensity of the measuring fiber when the catheter was preserved in the control solution without any scattering particles.

As mentioned previously in this section, I_m varied irregularly due to catheter movement or to fluctuations in the source light. I_0 fluttered accordingly, but we could not detect any changes in I_0 during the measurement of samples. In the preparatory experiments, we found that the ratio of I_m to I_r remained fairly constant during measurements in transparent solutions, such as saline, lactated Ringer's solution, and plasma, with the cath-

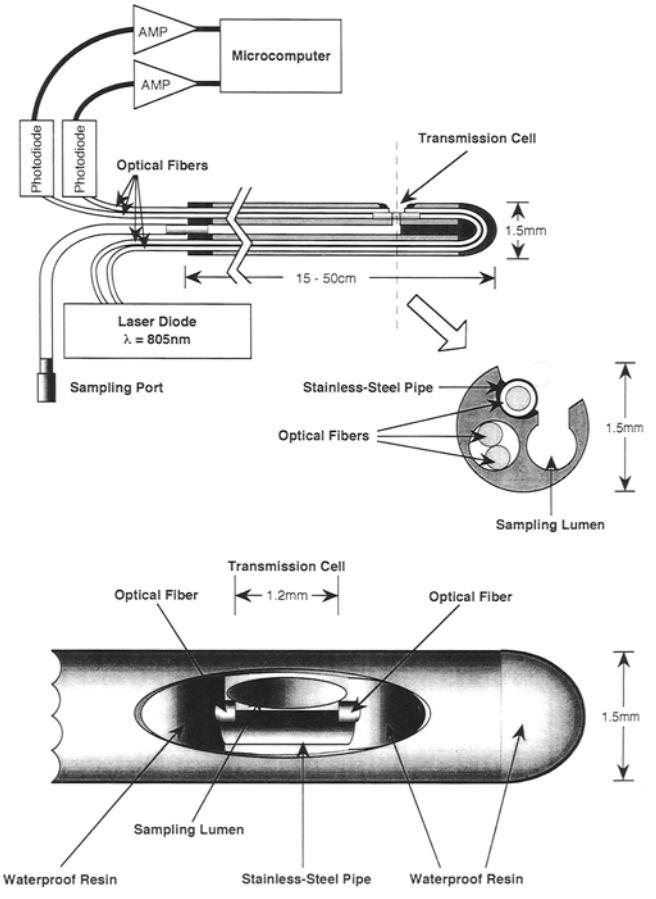

Fig. 1. Schematic arrangement of fiberoptic catheter, a block diagram of the electronics symbolized in the optical monitor, and its connection to the computer. A pair of plastic fibers was bent, conducted through two lumens of a triple-lumen PE tube, and fixed near the catheter tip. A small piece of measuring fiber corresponding to the length of the light path was cut off so that the space between the two fiber tips formed a transmission cell (light pathlength = 1.2 mm). The other fiber, the reference fiber, was not cut off. The third lumen was used to withdraw blood samples from a pore opened at the transmission cell. The intensities of the light transmitted through both the measuring (I_m) and the reference (I_r) fibers were measured by photodiodes combined with amplifiers. The output signals were sent to a computer via an A/D converter board

eter being moved repeatedly. Therefore, before each experiment the initial values of I_m and I_r were measured in lactated Ringer's solution ($I_{m,0}$ and $I_{r,0}$). Then I_0 is given by

$$I_0 = I_r \times \left(I_{m,0}/I_{r,0}\right). \quad (2)$$

Substituting this relation into Eq. 1,

$$\mathrm{OD} = \log\left(I_0/I_m\right) = \log\left(I_r/I_m\right) + \log\left(I_{m,0}/I_{r,0}\right) \quad (3)$$

is obtained. The second term on the right side of Eq. 3 varied among experiments because of the intensity of the source lights, gain of amplifiers, connecting conditions of optic fibers, and so on. Equation 3 demonstrates that OD may be obtained by logarithms of the ratio of I_r to I_m.

It is well known that the light transmitted through suspensions of erythrocytes is influenced by absorption by the hemoglobin molecules, scattering by the red cells, and nonspecific absorption of light by the suspending medium. For the total transmitted flux of light through a suspension of particles, Twersky [9] has developed an expression in which absorption and scattering can be treated independently. According to his theory, the OD of an erythrocyte suspension is expressed as

$$\mathrm{OD} = edH - \log\left[10^{-sH(1-H)d} + q\left(1 - 10^{-sH(1-H)d}\right)\right] \quad (4)$$

where e is a factor depending on the molar extinction coefficient of hemoglobin; d is the optical pathlength; H is the hematocrit; s is a factor depending on wavelength, particle size, and shape; and q is a factor depending on the efficiency of light detection.

The first term on the right side in Eq. 4 is an expression of the Lambert-Beer law, and the second term is the influence of light scattering on the optical density. As indicated in Eq. 1, the scattering term is a complicated function of the hematocrit, size and shape of the erythrocytes, and the geometric parameters of the catheter. It would be virtually impossible to measure all these parameters simultaneously. However, Loewinger et al. [10] measured the OD of whole blood at 805 nm as a function of the hematocrit and showed that the measured values fitted well with the theoretical prediction. By the use of numerical simulation, we found that the scattering term in Eq. 4 is assumed to be constant when the hematocrit varies in the physiological range (Fig. 2).

Anderson and Sekelj [11] reported that s values were almost identical (~150) when the pathlength was greater than 0.25 mm. Since q denotes the probability of the scattered light being projected into the aperture of the receiving fiber, this factor was estimated as approximately the solid angle around the cross section of the face of the receiving fiber obtained from the tip on the opposite side of the transmission cell [$\sim 0.125^2/(4d^2)$].

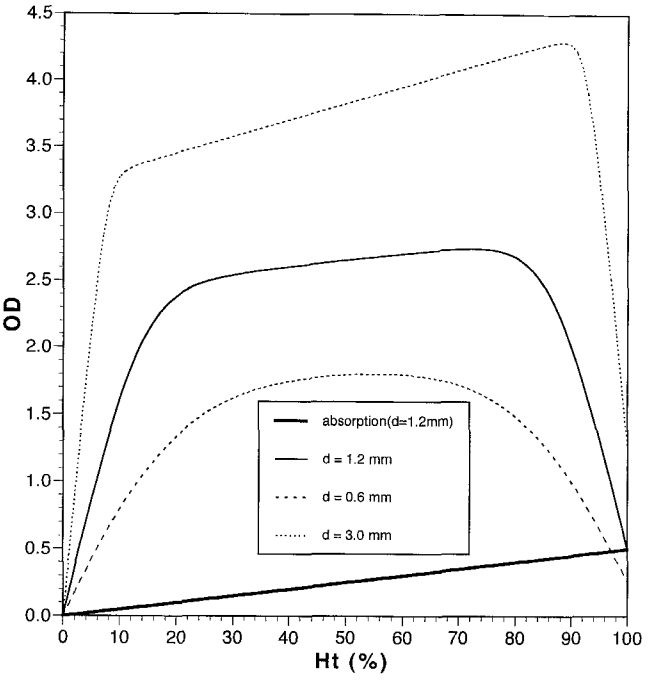

Fig. 2. Optical density (OD) of whole blood as a function of the hematocrit (Ht) according to Twersky's theory (Eq. 4). The curves represent this relationship at fixed different optical pathlengths (d). The straight line at the bottom of this figure represents the absorption term of Eq. 4. s = 150, e = 4.58, q = 0.003

The calculated curves are shown in Fig. 2. The simulation revealed that the linearity between OD and hematocrit increases as the pathlength increases. However, an increase in pathlength causes a decrease in I_m. After several trials of manufacturing, we determined that 1.2 mm was the optimal pathlength of the transmission cell.

Because the second term in Eq. 4 is replaced by a constant in the range of hematocrit values of interest, Eq. 4 may be rewritten as

$$\mathrm{OD} = edH + K \quad (5)$$

where K is a constant depending on the size and shape of the erythrocytes. According to Eq. 5, OD changes linearly with hematocrit. The slope of the line (ed) is characteristic of the catheter, and the intercept (K) depends on the size and shape of the erythrocytes, e.g., the mean corpuscular volume (MCV).

Measurements in vitro

The experimental setup for measuring hematocrit changes in vitro is shown in Fig. 3. The catheter was inserted from the injection port sealed with rubber. The

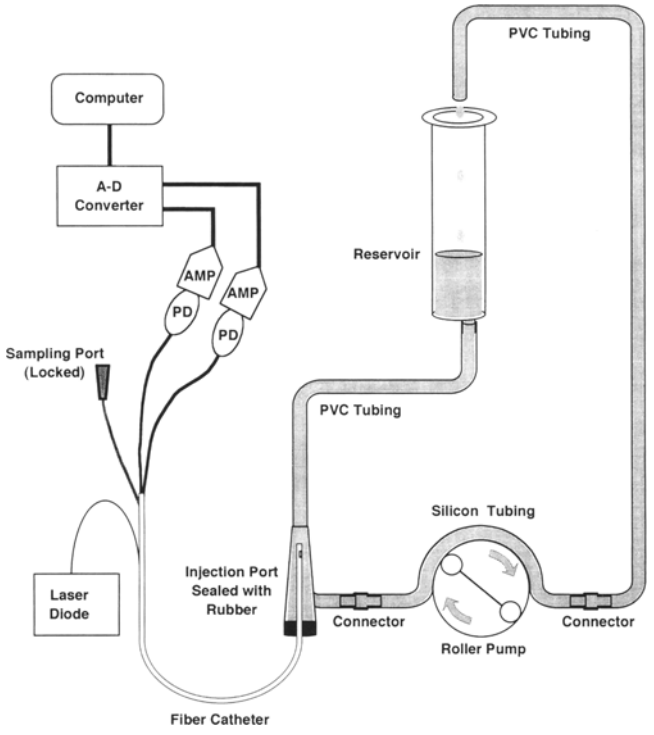

Fig. 3. Experimental setup for measuring hematocrit changes in vitro. The fiber catheter was inserted in the PVC tubing from the injection port sealed with rubber

inner diameter of the tubing where the catheter tip was positioned was 2.1 mm. Suspension was pumped by a roller pump (202U, Watson-Marlow, Cornwall, England). To ensure homogeneous distribution of erythrocytes in the circuit, the total volume of the tubing (~5 ml) and the reservoir volume (~2 ml) were designed to be small.

Human blood with heparin added was washed three times in saline. The erythrocytes were resuspended in lactated Ringer's solution at a hematocrit of 50%. Seven milliliters of the suspension was poured into the reservoir and pumped at 1.6 ml·min^{-1}. The OD of the suspension was calculated according to Eq. 3 by the computer on line. When the OD reached a steady state, the suspension was diluted with 0.7 ml of fluid added in the reservoir to change the hematocrit to 45.5%. This diluting maneuver was repeated until the hematocrit decreased to ~20%. Samples for microcentrifugation were not collected during the course of dilution, because the total volume of the suspension was small. At the end of the experiment, the hematocrit was measured in triplicate by microcentrifugation to compare with the predicted values.

To assess the parameters affecting OD in Eq. 5, the above procedure was carried out using a different catheter, the geometric factors of which were almost identical to the previous one. Samples were donated by different individuals and washed erythrocytes were suspended in autoplasma. The concentrations of the erythrocytes and hemoglobin were measured to calculate MCV and mean corpuscular hemoglobin concentration (MCHC).

The effect of flow on the optical density was investigated by changing the rate (1.6, 2.1, 2.7 and 3.2 ml·min^{-1}) and the direction, when the hematocrit values of the samples were adjusted at 45%, 31%, and 20%.

Measurements in vivo

Male New Zealand white rabbit (2800 g) was anesthetized with urethane (1.5 g·kg body weight, subcutaneously). The left carotid artery was cannulated to monitor the arterial blood pressure and heart rate, and the right femoral vein was cannulated for infusion of solutions and drugs. The catheter was put into the superior vena cava via the right jugular vein, and the rabbit was heparinized with an initial dose of 30 units·kg^{-1} and a maintenance dose of 50 units·kg·h^{-1} throughout the experiment.

After the animal was allowed to stabilize for an hour, 24 ml·kg^{-1} body weight of lactated Ringer's solution, saline with hydroxyethyl starch (SalinHES inj.) and homeostatic solution (KN3B), and 6.5 ml·kg^{-1} of 4% human albumin solution were infused. The duration of the infusion was 15 min at intervals of more than an hour. Blood samples were collected from the withdrawal port of the catheter at intervals of 5 to 15 min. Profiles of the OD recorded continuously through the experiment were transformed into hematocrit by Eq. 5. In regard to the parameters in Eq. 5, the slope e was determined in separate sets of experiments prior to the in vivo experiment, and the intercept k was given by the calculation of the hematocrit value obtained at the initial collection of the blood.

Statistics

Statistical analyses of flow rate and OD were performed using Fisher's Protected Least Significant Difference. Examination of the covariance between hematocrit and OD was accomplished using linear regression analysis. Significance was established when $P < 0.05$.

Results

To measure hematocrit changes in vitro, blood samples, at a volume of <30 ml each time, were collected from a healthy human donor on seven different days within a month. Figure 4 shows the OD values of the erythrocyte suspension as a function of the hematocrit. The average

OD for 1 min at the steady state was plotted on the ordinate. The relationship between OD and hematocrit was linear in the range of hematocrit between 20% and 50%, as described in Eq. 5 (Table 1). The line drawn in the figure represented the least-square regression of all the data obtained on seven days. The slope was 1.83×10^{-2}, the OD intercept was 2.06, and the correlation coefficient was 0.91. At each step of the dilution maneuver, OD decreased immediately after the fluid was added and reached a new steady state within 3 min. Thus, all experiments were done within 1.5 h and no hemolyzation was observed. Although the abscissa showed calculated hematocrit values, the differences between the calculated and the measured hematocrit values at the end of the dilution were less than 2%.

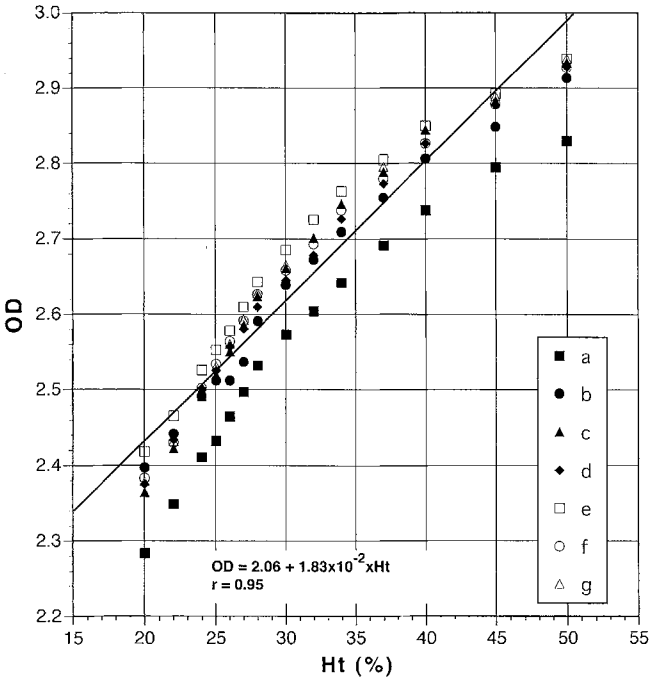

Fig. 4. Comparison of OD measured by the fiberoptic method and the calculated values of the hematocrit (Ht) diluted from the initial concentration. The same donor's erythrocyte suspensions with heparin added were measured seven times on different days. The blood was circulated through the open circuit (Fig. 3) then diluted by adding 0.7 ml of lactated Ringer's solution. Accordingly, the hematocrit levels changed by 14 steps from 50% to 20% (50.0%, 45.5%, 41.7%, 38.5%, . . . , 22.7%, 21.7%, 20.8%, and 20.0%). These experiments were conducted using the same fiberoptic probe. The average OD for 1 min at the steady state was plotted on the ordinate. OD = 2.06 + 1.83×10^{-2} Ht; $n = 98$; $r = 0.98$

Table 1. The hematocrit-OD regression parameters in the in vitro experiments (Fig. 4). There was no significant difference ($P < 0.05$) between slopes or OD intercepts of any two regression lines. OD, optical density

Case	Slope	OD intercept	r
a	0.0183	1.9947	0.9745
b	0.0177	2.0733	0.9850
c	0.0191	2.0502	0.9716
d	0.0184	2.0686	0.9808
e	0.0177	2.1093	0.9744
f	0.0182	2.0813	0.9753
g	0.0189	2.0633	0.9744

The relationship between OD and hematocrit under the influence of different MCV values is shown in Fig. 6. The plotted data for both individuals (MCV, 88.7 and 93.6; MCHC, 32.6 and 38.2) corresponded well to each regression line ($R > 0.97$, Table 2). The slopes were 1.26×10^{-2} and 1.41×10^{-2}, which were smaller than that of the line drawn in Fig. 4, but the differences were not significant. Between the lines in Fig. 6, the difference of the intercepts (1.24 and 1.01) was significant, but the difference of the slopes was not.

The OD measured by the catheter appeared to be insensitive to flow rate (Fig. 5). The average OD for 5 min at the steady state was plotted on the ordinate.

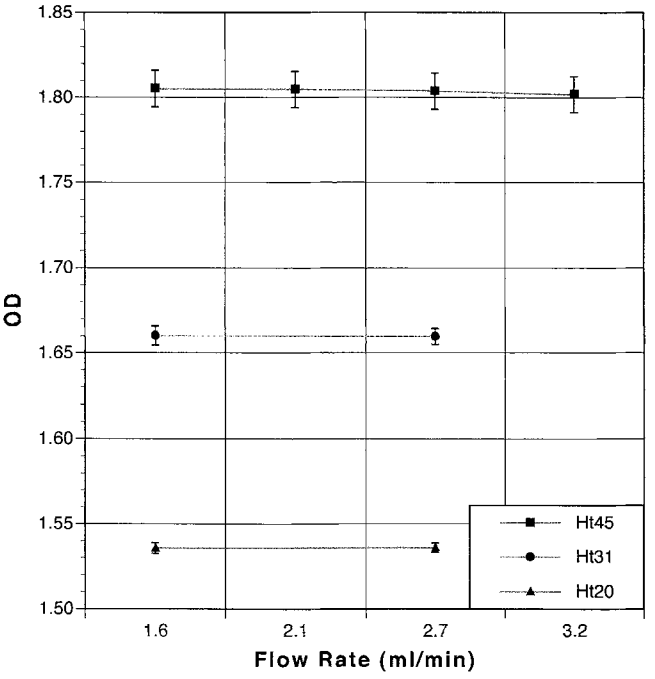

Fig. 5. Effect of the flow rate of blood on OD. Values represent means ± SE. *Solid squares*, hematocrit (Ht) 45%; *solid circles*, Ht 31%; *solid triangles*, Ht 20%. The hematocrit values of the samples were adjusted to 45%, 31%, and 20% by autoplasma and the blood flowed in the open circuit (Fig. 3); then the blood flow rates were changed to 1.6, 2.1, 2.7, and 3.2 ml·min^{-1} every 5 min. These measurements were conducted using the same fiberoptic probe

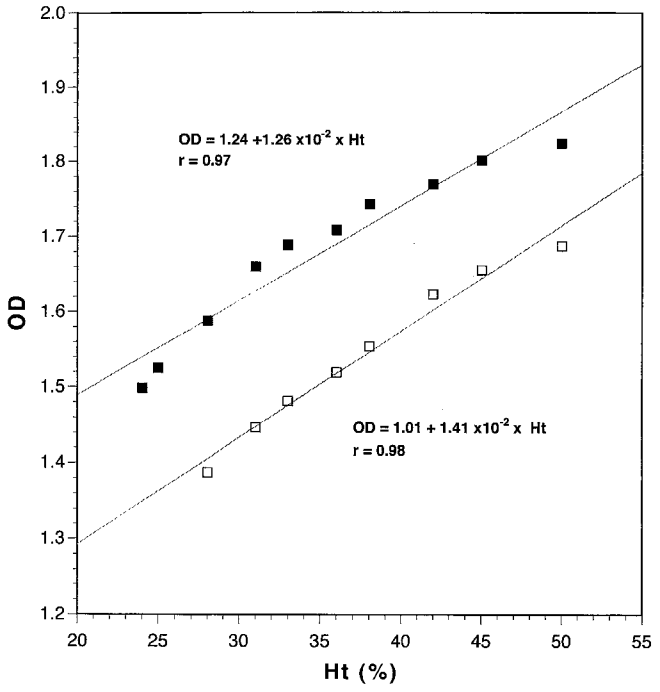

Fig. 6. Effect of mean corpuscular volume (MCV). *Open squares*, MCV 93.6 μm³; *solid squares*, MCV 88.7 μm³. Blood samples obtained from two healthy volunteers were adjusted to hematocrit (Ht) of 50% (7 ml), flowed through the open circuit (Fig. 3), and were then diluted by adding 0.7 ml autoplasma. Accordingly Ht changed by several steps from 50% to about 25% (50.0%, 45.5%, 41.7%, 38.5%, . . .)

Table 2. The hematocrit-OD regression parameters shown in Fig. 6. There was a significant difference ($P < 0.05$) between the intercepts, but there was no significant difference between the slopes. MCV, mean corpuscular volume

MCV (μm³)	Slope	OD intercept	*r*
88.7	0.0126	1.2364*	0.9660
93.6	0.0141	1.0108*	0.9889

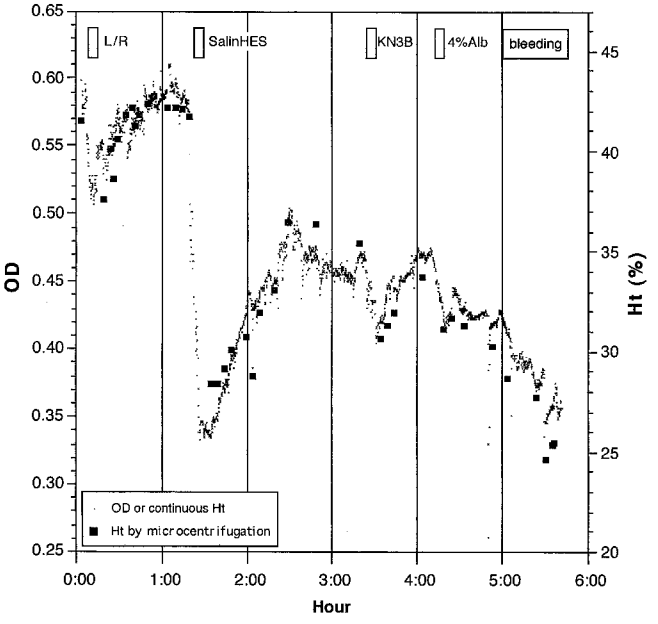

Fig. 7. Changes in OD measured by the fiber-optic method and continuous hematocrit (Ht) calculated from OD against Ht measured by microcentrifugation during infusions and hemorrhage in a male rabbit (2800 g). The left ordinate shows the values of OD measured by the fiberoptic method, and the right ordinate shows the values of Ht measured by microcentrifugation or Ht calculated from OD. The gray points are OD values measured by the fiberoptic method recorded every 10 s, and the closed squares are Ht values measured by microcentrifugation by blood sampling. After the animal was allowed to stabilize for an hour, 24 ml·kg^{-1} body weight of lactated Ringer's solution, saline with hydroxyethyl starch (*SalinHES* inj.) and homeostatic solution (*KN3B*), and 6.5 ml·kg^{-1} 4% human albumin solution were infused. The duration of infusion was 15 min at intervals of more than an hour

Variations of OD were also not affected by the flow rate but increased with hematocrit. When the direction of the flow was altered, the OD fluttered for a few seconds but soon settled at the same value as that before the direction was changed (data not shown).

The validity of our method for continuous measurement of hematocrit in vessels in vivo is demonstrated in Fig. 7. The OD and the calculated hematocrit decreased immediately when the infusion of solutions was started and recovered gradually after the infusion was stopped. The pattern of changes in hematocrit accounted for the hemodilution caused by the infusion. Hematocrit values measured by microcentrifugation fitted well with the calculated curve. Figure 8 shows the relationship between the measured hematocrit and the matching OD, with a correlation coefficient of 0.96. The error limit associated with a hematocrit measurement with this catheter is ±2% (95% confidence interval). Blood was withdrawn (40 ml·50 min^{-1}) at the end of the experiment. Differences in solutions or in the methods used to alter the blood volume did not affect the consistency.

Discussion

We have presented a catheter-based transmission cell composed of optic fibers for determining hematocrit in the vessels in vivo. The validity of the method has been confirmed in a tubing circuit and also in animals. This method applied Twersky's theoretical formulae in regard to multiple scattering of light in suspensions of

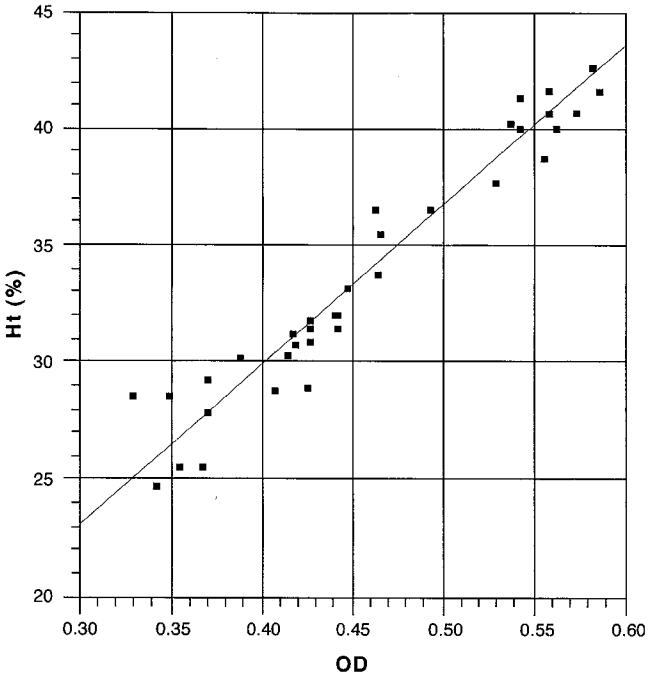

Fig. 8. Relationship between OD measured by the fiberoptic method and hematocrit (Ht) measured by microcentrifugation at the same time during infusions and hemorrhage in vivo. Blood samples for the measurement of Ht by microcentrifugation were obtained from the sampling port of the catheter. Ht = 1.63 + 70.67 × OD; $n = 39$; $r = 0.96$

light-absorbing particles. The error limit associated with a hematocrit measurement with this catheter method was ±2% (95% confidence interval).

The linear relationship between the OD calculated using Eq. 3 and the hematocrit was verified in erythrocyte suspensions (Figs. 4 and 6) and in vessels in vivo (Figs. 7 and 8). The important concept embodied in Eq. 4, which was developed by Twersky [9], is that absorption and scattering can be treated independently. However, the scattering term (the second term on the right side of Eq. 4) is very complex for practical use. The success of our method is based on the idea that the scattering term can be regarded as being constant when the transmission cell is constructed in a certain geometric dimension. Figure 2 shows curves representing the relationship between OD and hematocrit at fixed different optical pathlengths (d). The straight line at the bottom of this figure represents the absorption term of Eq. 4. When $d = 1.2$mm, OD increased linearly with hematocrit in the range of hematocrit from approximately 20% to 60%, and the regression line had the same slope that the absorption term showed.

According to Eq. 5, the slope of the line should be proportional to the optical pathlength of the transmission cell, which was characteristic of the catheter. As shown in Fig. 4, the slopes of the regression lines were almost identical when the same catheter was used. When another catheter was used, the slope changed (Tables 1 and 2). However, the slopes were not unique among different sets of experiments, even if the catheter of almost the same optical quality was used. For example, the slopes in Fig. 6 differed by approximately 10%. The slope is also proportional to the factor e in Eq. 5. This factor depends on the extinction coefficient of hemoglobin, which is responsible for most light absorption at the wavelength used. Because we chose hematocrit as the dependent variable, the absorption coefficient of hemoglobin multiplied by MCHC is an explicit and true expression of the factor e. Dividing the slopes in Fig. 6 by MCHC yielded 3.87×10^{-4} and 3.69×10^{-4}; their difference was <5%.

The intercept of the regression line represents the scattering term, which mainly depends on the factor q in Eq. 4 when the hematocrit is in the physiological range. Since q denotes the probability that the scattered light will be projected into the aperture of the detector, this factor also depends on the size and shape of the scattering particles. Figure 6 predicted that the change in the MCV would alter the intercept. When samples were collected from the same healthy donor (Fig. 4), changes in the intercepts were not significant, perhaps because neither the size nor the shape of the erythrocytes changed during the period.

Changes in the flow rate of the suspensions did not affect the OD (Fig. 5). This was not surprising, because Twersky's treatment of the scattering phenomenon was based on the theory of probability, and the dimensions of the scattering particles, erythrocytes in this case, were much smaller than that of the transmission cell. The relation of OD to hematocrit was also not affected by the concentration of the electrolytes or albumin (Fig. 7). Our method may be independent of other physiological parameters, such as pH, temperature, and oxygen content of the blood, if these parameters do not alter the size or shape of the erythrocytes.

As shown in Fig. 5, OD fluctuations were not affected by the flow rate. The increase in the output of the roller pump was accompanied by an oscillation of the flow and thus movement of the catheter, which was the cause of noise in the intensity of the measured light. An excellent stability in the ratio of signal to noise (S/N) was achieved by the introduction of the reference fiber. The validity of the reference fiber was proved by the unchanged correlation between the calculated and measured hematocrit values during the in vivo experiment lasting about 6h. The increase in the fluctuation of OD with hematocrit (Fig. 5) was explained by the decrease in I_m and thus in S/N due to the absorption and scattering of light by the multiplied erythrocytes. A decrease in I_m also occurred when the pathlength of the transmis-

sion cell (d) was enlarged. The optimal d was determined to maintain the balance between the linearity of the regression line and S/N.

Continuous measurement of hematocrit in vivo was applied to the evaluation of the changes in the circulating blood volume (Fig. 7). Although the amount of lactated Ringer's solution infused was equivalent to 20% of the blood volume , the increase in blood volume calculated using the changes in hematocrit was only 13%, and the retention of the infused volume was about 65% at the end of the infusion. This result agreed well with that of a previous experiment using ^{51}Cr-labeled erythrocytes reported by Tanaka [12]. Although KN3B showed the same course as lactated Ringer's solution, the infusion of SalinHES and albumin caused a temporary decrease in hematocrit, which was more than expected, and the blood volume remained at the increased level.

In recent years, fiberoptic reflectance oximetry has become widely used clinically as an effective method to measure mixed venous oxyhemoglobin saturation ($S\bar{v}O_2$) in flowing whole blood. However, fiberoptic reflectance oximeters are susceptible to changes in hematocrit and measure the relative value of oxyhemoglobin saturation. To calibrate the value of $S\bar{v}O_2$ using fiberoptic oximeters, we have to measure the hematocrit, the hemoglobin concentration, and the absolute value of oxyhemoglobin saturation frequently during a clinical procedure. If pulmonary arterial catheters were equipped with the transmission cell in addition to the reflectance optic fibers, the oximeters would be freed from this frequent calibration. Furthermore, if it is possible to improve the precision of the measurement of hematocrit, we will be able to estimate the cardiac output with our method by injecting a regular volume bolus infusion.

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