Monitoring of the dilution rate during continuous in vivo blood sampling with a double lumen catheter

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Summary. A simple and non-destructive method for monitoring the dilution rate during continuous blood sampling is described. When the dilution rate is not constant, the proposed method based on electrical resistivity measurement provides a correction factor for further analysis.

Continuous chemical analysis on venous blood is generally performed using a double lumen catheter supplying anticlotting substances, through one of the lumen, and drawing the blood so diluted via the other lumen. The dilution rate, ideally constant, may be subjected to variations (due to partial clotting of the catheter or changes of its position in the vein), providing a source of error in the subsequent analysis. Since, during the time of the analysis, the circulating blood hematocrit (i.e. the relative volume of red cells in the whole blood) is virtually constant, the hematocrit value of the diluted blood allows the calculation of the dilution rate provided by the double lumen catheter, thus yelding the needed correction factor.

The proposed method for the measurement of the hematocrit is based on the electrical resistivity of the diluted blood. Indeed, for a low frequency current, the hematocrit of motionless blood samples is related to the blood (r_b) and plasmatic (r_p) resistivities[1]:

$$\text{hematocrit} = \frac{\gamma(r_b - r_p)}{\gamma r_b + r_p} \qquad (1)$$

γ being a factor depending on the shape of the red cells. We describe here an application of this principle to continuous blood analysis and its application to an artificial pancreas.

Methods. The hematocrit cell (figure 1) is composed of 4 ring-shaped electrodes of 1 mm inner diameter and 15 mm length connected by PVC transmission tubing. The whole, including leads, is moulded in rigid araldyte. The blood is continuously pumped through the hematocrit cell at 0.4 ml/min flow rate. The device consists of an oscillator supplying a constant current of 1 μArms at 500 Hz to the current electrodes (outer electrodes) and an amplifier picking-up from the inner electrodes a voltage proportional to the blood resistivity. In connection with the artificial pancreas, the hematocrit can be calculated from eqn (1) by a digital computer. However, analogical calculation can be made from the logarithm of the blood resistivity[2].

Results and discussion. The stability of the measurement was first tested by running sodium chloride 9‰ for 24 h. With this 4-electrode system, no significant drift is observed (1.7% between the higher and the lower value). By contrast, with a 2-electrode system, such as that described by Tanaka et al.[3], a drift of 9.5% between the higher and the lower value has been observed during a 14-h period. When dog blood is pumped through the system, a very straight correlation (figure 2) has been found between the value of hematocrit determined by the impedance method and the standard centrifugation method, when a value of 1.32 is given to the shape factor γ. This is in good agreement with the value obtained for motionless blood samples[1,4] and corresponds to a well accepted axial ratio of 1/3.2 for dog red cells. Furthermore, physiological modifications of the plasmatic resistivity r_p, (considered here as constant) do not significantly affect the blood resistivity[5,6]. Blood resistivity measurement can thus be applied to continuous blood analysis and provides an easy and fast method of hematocrit assessment for which no chemical reagents are required.

Application to the artificial pancreas. The hematocrit cell is incorporated in the artificial pancreas developed by our group[7], based on the principles described by Albisser et al.[8] but including significant technological advances. Blood is sampled continuously from either a peripheral or a jugular vein and carried through the hematocrit cell to a glucose sensitive electrode. Glycemia and hematocrit are monitored by a real time digital computer (PDP 11-45, DEC), which determines the optimal insulin requirement and activates the pump delivering its injection. The dilution rate is initially determined by comparison between diluted and non-diluted blood samples. Later on, any variation of this ratio is detected by a change in the hematocrit value and correction can be applied using:

$$\text{corrected glycemia} = \text{apparent glycemia} \times \frac{\text{calibration hematocrit}}{\text{actual hematocrit}} \qquad (2)$$

Eqn. (2) was verified by variations of the dilution rate of the catheter between 1.4:1 and 3:1. For example, when the ratio of blood sampled to diluting fluid is increased by 38%, the apparent glycemia and the actual hematocrit increase

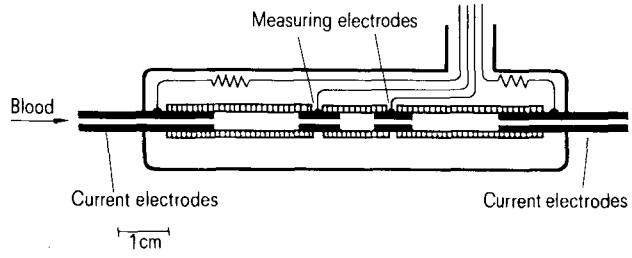

Fig. 1. 4 electrodes measuring cell for continuous determination of blood resistivity.

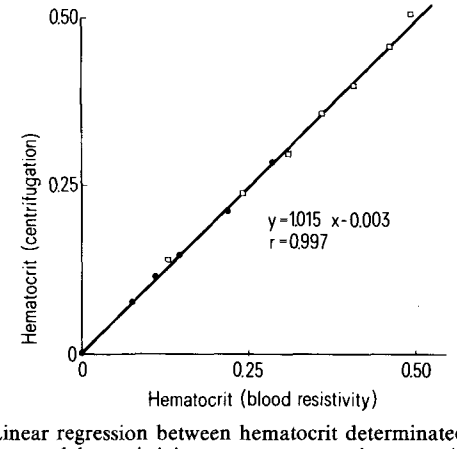

Fig. 2. Linear regression between hematocrit determinated by centrifugation and by resistivity measurement using eqn. (1). Blood samples diluted in their own plasma from 2 different dogs.

by respectively 36% and 38%, so that their ratio, and thus the corrected glycemia remains practically constant.
Conclusion. The proposed method makes it possible to monitor continuously, and immediately on changes in the dilution rate, to apply adequate instantaneous correction. It is simple, does not require chemical reagents, is not damaging and thus does not interfere with other concomitant analysis.

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CORRIGENDUM

T. Iso, H. Yamauchi, H. Suda, N. Nakajima, K. Nishimura and J. Iwao, Experientia *34*, 1202 (1978). The title should correctly read: *Potentiative effects of sulfhydryl compounds on carrageenin-induced oedema in rats and relationship to their potencies as inhibitors of angiotensin-converting enzyme in vivo.* In the 'Results and discussion' section, the 5th line reads: sulfhydryl compound, ...

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