

Hematocrit Measurement – A high precision on-line measurement system based on impedance spectroscopy for use in hemodialysis machines

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Abstract— This paper presents a unique technical approach to measure the proportion of blood volume occupied by red blood cells, the hematocrit value (HCT) on-line and in-line. A practical method has been developed to measure without the need for extracting blood samples out of an existing extracorporeal blood circulation system. The new sensor is based on Impedance Spectroscopy and measures electrical properties of the blood at various frequencies. In order to achieve the required precision resolution the sensor geometry has been optimized by Finite Element Analysis. For sensor readout a digital measurement circuitry based on cheap standard components is developed and allows practical implementation of HCT-sensor devices for the first time. Special care has been taken in order to compensate for the temperature effects.

Keywords— impedance spectroscopy, on-line measurement, hematocrit measurement, hemodialysis, extracorporeal blood flow

I. INTRODUCTION

On-line measurement of hematocrit (HCT) is a major concern in many health care situations such as hemodialysis or surgery [1], [2]. In many of these applications there is an existing external blood circulation system required, based on standard plastic tubing which contains the blood. Traditional methods for HCT-measurement like centrifugation or photometry are well known and precise. The drawback of the first method is the need for extracting a blood sample out of the closed loop system which takes time and gives rise to the cost. The latter requires an optical window to access the blood and therefore it does require an additional precision fabricated disposable component. Therefore these methods are time and cost consumptive. In order to eliminate the disadvantages of the mentioned HCT-measurement methods a new approach for on-line and in-line measurement is presented in this paper. The developed sensor is based on electrical impedance spectroscopy and allows HCT-measurement inside standard plastic tubing without the requirement to open the existing external blood circulation system. No additional device has to be inserted into the circuit and therefore there is no direct contact between the sensor and the blood.

The presented HCT-sensor measures electrical properties of the blood inside the plastic tubing at various frequencies. Since the sensor electrodes are attached to the outer wall of the plastic tubing the sensor measures the properties of the tubing material in addition to the properties of the blood inside the tubing. In consequence of this sensor structure special care has been taken to investigate the effect of temperature drift of the tubing material as well as tolerances in the wall thickness of the tubing. As a result of these investigations a calibration method and a solution for active on-line temperature compensation of the sensor signal is proposed.

Since medical equipment is highly cost sensitive, a cheap practical solution for the sensor readout circuit is mandatory. The electronic sensor circuitry is based on digital standard components and employs a field programmable gate array (FPGA). A unique method for signal digitalization has been developed in order to achieve the required precision resolution of the system.

II. SYSTEM CONCEPT

A. Sensor design

The sensor principle is based on electrical impedance spectroscopy. Since the plastic tubing is an electrical isolator and non conductive a pure capacitive design is proposed. Two sensor electrodes clasp around the tubing [Fig. 1] and form a capacitor. The tubing material itself and the blood inside the tubing serve as laminated dielectrics. The dielectric properties of the blood vary depending on the hematocrit [3] while the properties of the tubing walls remain constant and therefore generate an offset signal in the sensor.

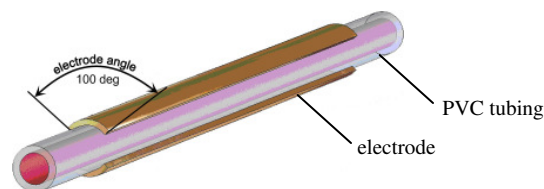


Fig. 1 capacitive Sensor design

B. Sensor simulation and optimization

The sensor principle is based on a capacitor which changes the overall capacitance depending on the hematocrit value of the sample and the applied frequency. In order to maximize the relative dielectric effect of the hematocrit on the sensor output signal, the angular shape covered by the electrodes has to be optimized. The optimization has been done by Finite Element Analysis of the electrical field pattern [Fig. 2] generated by the sensor electrodes.

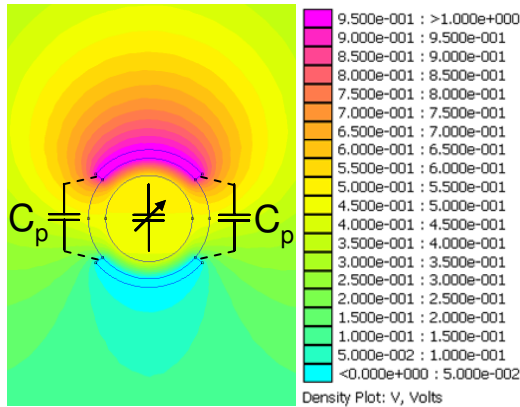


Fig. 2 Electrical field pattern generated by sensor electrodes

The goal of the optimization is to find the angular optimum of the electrodes. If the angular coverage is small the overall capacitance is very small. If the angular coverage is large, parasitic capacitive effect between the electrodes causes a very large sensor offset. [Fig. 2] illustrates this effect by the two parasitic capacitors C_p . Both are connected “in parallel” to the capacitor in the center of the tubing which changes its capacitance depending on the hematocrit. The simulation result is presented in [Fig. 3].

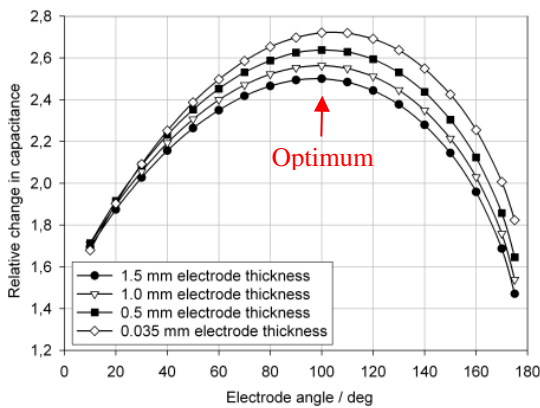


Fig. 3 Relative change in capacitance versus electrode angel

C. Electronic measurement circuitry

Since medical equipment is highly cost sensitive a unique electrical measurement circuitry is developed. The circuitry is based on cheap digital standard components and allows a practical implementation of an on-line HCT-sensor for the first time. The sensor circuitry employs a field programmable gate array (FPGA) which encapsulates most of the required digital modules [Fig. 4].

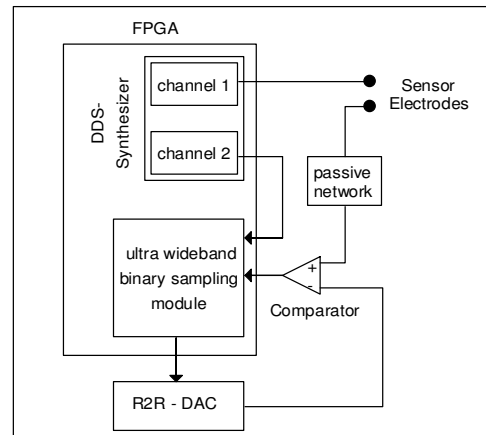


Fig. 4 Digital measurement circuitry block schematic

The FPGA contains a Direct-Digital-Synthesizer (DDS) core which is capable of generating sinusoidal AC-signals over a wide range of frequencies. The signal generated by output channel 1 is filtered and directed to the sensor electrodes. The sensor electrodes output signal is connected to a passive network which has a predictable transfer function. The passive network will cause additional phase shift and amplitude change in the sensor signal. The phase shift is then measured by the high resolution binary sampler module which provides an excellent time resolution of up to 1 picosecond. The signal generated by output channel 2 is used as a trigger signal for the sampling module. The binary sampling module is connected to a Digital to Analog Converter (DAC) built out of a cheap resistor ladder (R2R). The analog output signal from the DAC is fed back to the analog comparator. Depending on the excited measurement frequency internal parameters like trigger frequency or DAC resolution are adjusted. The measured data is then transferred to a computer for further signal processing. A new algorithm calculates the hematocrit value depending on the measured capacitance. Besides the algorithm provides a calibration routine to eliminate a sensor offset signal. In order to compensate for further effects like temperature drift an active compensation algorithm is developed.

III. MEASUREMENT RESULTS

A. Experiments with blood samples

First experiments have been done using blood samples taken from pigs. The blood is warmed up to a defined temperature and circulates in a closed loop system similar to a system used in real hemodialysis machines. A controllable peristaltic pump is adjusted to a flow rate of approximately 200ml/min inside an industry standard PVC tubing with 4.8mm inner diameter and 6.4mm outer diameter. PVC tubings are widely used in hemodialysis machines since the material is very cheap and the tubing is a disposable. The sensor design shown in [Fig. 1] is used for the experiments in the laboratory. The length of the electrodes is 100mm. The following graph [Fig. 5] presents the measured capacitance as a function of the frequency for several hematocrit values. The temperature is stabilized to 33°C.

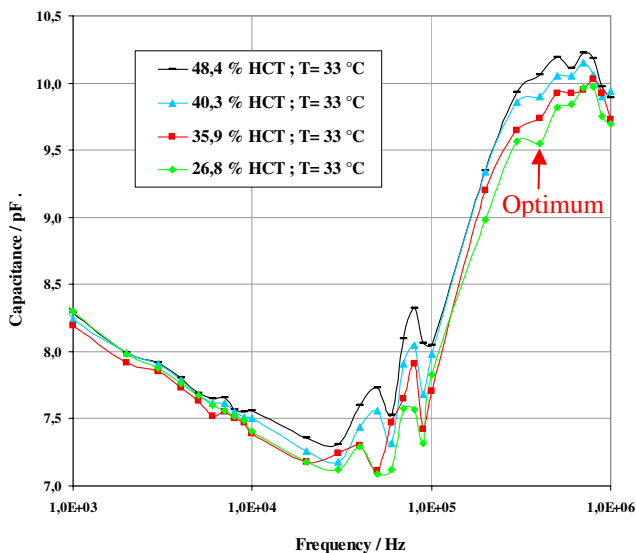


Fig. 5 Capacitance versus Frequency for 4 hematocrit values

The graph shows different values for the measured capacitance at each frequency step depending on the hematocrit. The total sensor capacitance is between 7 pF and 10,5 pF. Because of the small overall capacitance it is desirable to measure at a frequency where the relative effect of the hematocrit on the sensor capacitance has its maximum. A very good measurement frequency in figure 5 is around 400 kHz. Here the relative sensor effect is good and in addition the total sensor capacitance is high (up to 10 pF). The following graph [Fig. 6] shows the relation between capacitance and hematocrit value for a fixed frequency of 400 kHz.

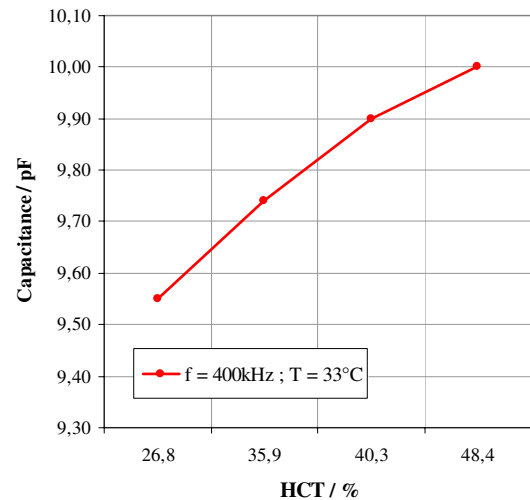


Fig. 6 Capacitance versus hematocrit at 400kHz

According to the graph [Fig. 6] the relation between the measured capacitance and the hematocrit is almost linear. Some nonlinearity may also be caused by measurement errors since the data is unfiltered. In a practical sensor implementation the sensor signal will be measured multiple times and then filtered.

The graph shows a gradient of approximately 20 fF per percent hematocrit. In order to achieve a sensor resolution of 1,0% HCT the electronic circuitry must be able to measure a change in capacitance of 20 fF which represents a relative target effect of around 0,2 %. This relative target effect is very low and special care has to be taken on side effects like temperature drift.

B. Investigating temperature effects

Especially the material of the tubing walls is suspect to drift over temperature. Literature gives already good hints that the permittivity of plastic materials like PVC is not stable [4], [5]. The FEM simulation [Fig. 2] points to other unwanted effects. Due to the relatively low permittivity of approximately 2 for the PVC material and a relatively high permittivity of approximately 80 for the blood, the electrical field distribution inside the sensor is not linear. The highest field strength is found in the tubing wall. In the above mentioned experiment around 85 % of the sensor signal is “generated” in the tubing walls and only the remaining 15 % of the signal amplitude are influenced by the blood. For this reason a temperature test has been done using PVC tubing inside the measurement capacitor. The temperature has been varied and the capacitance was measured as a function of frequency [Fig. 7].

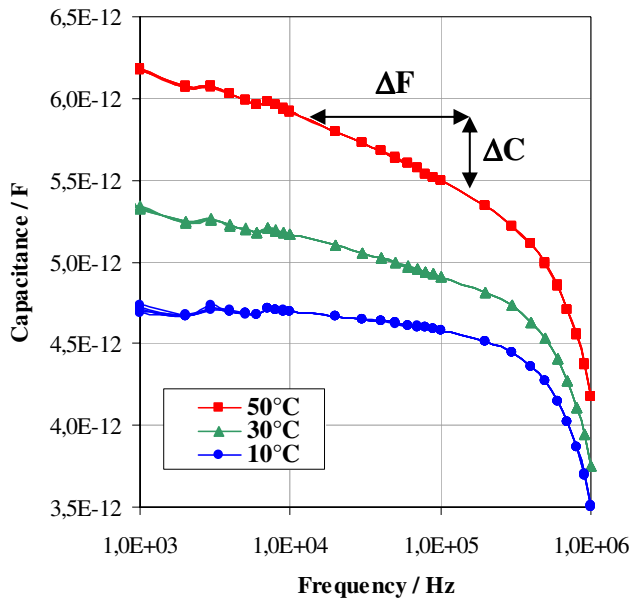


Fig. 7 Capacitance versus Frequency at 3 temperatures

The graph [Fig. 7] shows a large drift in capacitance for the 3 different temperatures. The gradient of the temperature drift at 400 kHz frequency is approximately 18,8 pF per degree Celsius inside the tested range between 10 °C and 50 °C. As a result a ΔT of 1K has approximately the same effect on the sensor output signal like a ΔHCT of 1 % in the blood sample. Since the temperature of the blood may vary between 33 °C and 38 °C during hemodialysis a compensation method is proposed. The 3 different graphs [Fig. 7] have a unique impedance spectrum depending on the temperature. The gradient $\Delta F/\Delta C$ is unique for each temperature. This information can be used to compensate for the temperature drift of the tubing material.

IV. CONCLUSIONS

This paper presents a new approach to measure hematocrit on-line and in-line without the need for extracting a blood sample out of an existing extracorporeal circuit. The sensor principle is based on measuring the capacitive effect of the HCT value. Measurements in the laboratory show the small overall capacitance of less than 10 pF and a target effect of less than 0,2 % of the sensor signal. Because of the very small values a high precision electronic circuitry is required for digitizing the sensor output signal. Besides the requirements for the sensor resolution and accuracy, the sensor must be as cheap as possible due to the cost sensitive field in medical applications. As a solution for both re-

quirements a unique FPGA based measurement circuitry is proposed. The developed circuitry is capable of measuring the sensor output signal by converting the amplitude and phase shift into a time shift. The time shift can then be measured by the digital part of the circuitry. Due to this “transformation” it is now possible to eliminate the need for conventional expensive precision Analog-to-Digital converters.

Special care has been taken on the temperature drift of the sensor. Measurement results show a large effect primarily based on the drift of the PVC tubing. The drift is in the same order of magnitude as the overall sensor output signal and therefore has to be compensated for. This can be achieved using a new compensation algorithm. The input data for this algorithm is based on a frequency sweep.

The HCT estimation is done at a fixed frequency of 400 kHz measuring the overall sensor capacitance. The required temperature compensation is based on a frequency sweep which delivers an impedance spectrum of the sensor. For the first time impedance spectroscopy is introduced to be used as an indirect way of building a high precision sensor for medical applications.

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