Continuous Non-Invasive Sodium Monitoring in Extracorporeal Circuits

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Abstract:
Dialysis is a blood purification therapy indicated by kidney failure, where blood is pumped via an extracorporeal circuit into a dialyzer consisting of a semi-permeable membrane separating blood and the dialysate. Due to the concentration gradients across this membrane, toxins can be removed from the blood by diffusion and the electrolyte balance can be regulated [1]. However, excessive changes of osmotic substances such as electrolytes can be critical. For instance, a rapid or excessive loss of sodium in blood, and thus a drop in plasma osmolarity, can cause overhydration of cells, cardiovascular instability and disequilibrium syndrome with headache, muscle cramps and fatigue symptoms. On the other hand, an inefficient loss of sodium can cause increased thirst, hypertension and risk of pulmonary edema [2–5]. Especially for critically ill patients with acute kidney injury, any additional stress during dialysis has to be avoided. A dialysate with individually adjusted electrolytic concentration can prevent such complications. Thus, it is necessary to monitor the plasma electrolyte concentrations and other osmotic substances during dialysis treatment [6]. However, the required hemocompatibility is often difficult with invasive in-line measurement methods [7]. In this work, we present a new approach to non-invasively monitor sodium based on a differential transformer. The output voltage of this transformer depends on the conductivity of the medium [8]. First measurements show a linear correlation between sodium concentration and output voltage.

Key words: continuous sodium monitoring, non-invasive in-line monitoring, differential transformer, individualized hemodialysis, extracorporeal circuits

Introduction and Experimental Setup

A differential transformer usually consists of three coils on a ferrite core. We use a ferrite core with a relative permeability of \( \mu = 300 \). Additionally, we add a fourth compensation coil \( L_c \) to the differential transformer, which eliminates asymmetries in the setup by active compensation. Fig. 1 shows the experimental setup. The primary coil \( L_p \) excited with an alternating voltage \( U_p \) generates a primary flux through the secondary coils \( L_{s1} \) and \( L_{s2} \) as well as the medium. Due to the opposite direction of winding and the same inductivity of the two secondary coils, no output voltage \( U_s \) should be induced by the primary flux. However, small asymmetries in the setup lead to an induced voltage, which can be actively reduced by the additional compensation coil \( L_c \). Inside the medium, the primary flux generates displacement and eddy currents causing a secondary flux. Since the medium is closer to the secondary coil \( L_{s1} \) compared to \( L_{s2} \), a higher voltage is induced into \( L_{s1} \) resulting in an output voltage \( U_s \). The voltage \( U_s \) can be calculated according to Eq. (1):

\[
U_s = K_2(\omega^2) \cdot \varepsilon''_r - j \cdot K_2(\omega) \cdot (\omega \cdot \varepsilon''_r + \kappa)
\]

where \( j \) is the imaginary unit, \( \omega \) is the angular frequency, \( K_1 \) and \( K_2 \) are frequency-dependent constants, \( \varepsilon' \) is the polarizability of the medium, \( \varepsilon'' \) describes the dielectric losses and \( \kappa \) is the electrical conductivity [8].

Fig. 1. Schematic setup of the differential transformer.
Assuming a linear correlation between the conductivity of blood plasma and the sodium concentration, which is a sound assumption due to the significantly higher sodium concentration compared to the other electrolytes, it is possible to determine the sodium concentration from the inductively measured electrical conductivity of blood, when the dielectric losses $\varepsilon''$ are zero [3,9,10].

**Results**

In preliminary experiments, the sensor response is investigated for different sodium concentrations in DI-water. In addition, the test solutions contain 5 mmol/l urea comparable to blood urea concentrations to account for interfering effects. Fig. 2 illustrates the imaginary part of the output voltage $U_s$. As predicted by Eq. (1), a linear correlation between the sodium concentration and output voltage $U_s$, with a coefficient of determination $R^2 = 0.99$ and a sensitivity $S = 2.08 \text{ mV/mmol}$, results after an initial step. The initial step only occurs at very low concentrations and may be the result of frequency and concentration dependent dielectric losses [11].

![Fig. 2. Imaginary part of the output voltage $U_s$ versus sodium concentration in a 5 mmol/l urea solution.](image)

In order to exclude possible cross-sensitivities, the influence of other substances on the imaginary part of $U_s$, which can significantly change during dialysis treatment, was investigated. For example, Fig. 3 shows the influence of urea in a 140 mmol/l sodium chloride solution with a cross-sensitivity of just 0.07 mV/mmol leading to an absolute error of approximately ±1.35 mmol/l for possible changes of urea of 40 mmol/l during dialysis treatment. This gives a relative error of just 1%. Similar results apply for glucose.

![Fig. 3. Imaginary part of the output voltage $U_s$ versus urea concentration in a 140 mmol/l NaCl solution.](image)

We conclude that the actively compensated differential transformer is a promising approach for non-invasive monitoring of the sodium concentration in extracorporeal circuits, which is especially interesting during hemodialysis.

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**References**


