

# Relationship Between the Conductivity of Human Blood and Blood Counts

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**Abstract**—Achieving a better characterization of human blood conductivity is of high relevance for medical applications. In this study we measured the complex impedance of  $N=10$  human whole blood samples (from  $N=10$  oncology patients) at room temperature ( $T=22.6\pm0.8$  °C) and at body temperature ( $T=36.6\pm0.4$  °C). The complex impedance was measured using the measurement setup consisting of a custom made four-electrode probe and a commercially available galvanostat. The measured complex impedance data were used to calculate the conductivity of whole blood over the 631 Hz–100 kHz frequency range. The calculated conductivity data is presented and was compared with the literature data. The data from our study is in good agreement with the data available in the literature. Additionally, full blood counts were provided for  $N=8$  samples and Pearson correlation coefficient was calculated between the conductivity and blood counts at different frequencies. The three blood count parameters with the highest correlation coefficient are haematocrit (Hct), haemoglobin (Hgb) and red cell count (RBC). The correlation coefficient was shown to decrease as the frequency increases and was the highest at  $f=631$  Hz, which is the lowest reported frequency. To our knowledge this is the first study to measure low-frequency (i.e. below 1 MHz) conductivity of whole human blood at body temperature using the four-electrode technique. The results of this study represent an important contribution to the literature, which is currently limited in this area and will help further medical device design.

**Keywords**—Conductivity, Bioimpedance, Electrochemical impedance spectroscopy, Conductivity measurement, Biological materials.

## I. INTRODUCTION

**K**NOWLEDGE of the electrical conductivity of human blood at low frequencies (i.e. below 1 MHz) is fundamental for medical electromagnetic applications. Blood conductivity measurements have the capacity to be used to quickly estimate important biological parameters such as haematocrit (Hct) [1], erythrocyte sedimentation rate (ESR or sed rate) [2], cardiac output [3]. In addition, low-frequency human blood conductivity is one of the key parameters used as in the design of numerical simulations of radiofrequency (RF) [4] and pulsed electric field (PEF) [5] ablation treatments. RF ablation has become widely accepted treatment for most atrial and ventricular arrhythmias, including atrial fibrillation (AF) [6]. It relies on electrical conduction through the tissue, where

RF current is able to pass through the tissue because of the abundance of ionic fluid present in the tissue and since the tissue is not a perfect conductor, RF current causes resistive heating (the Joule effect) [7]. Pulsed field ablation (PFA) exploits the delivery of short high-voltage shocks to induce cells death via irreversible electroporation (IRE) [5]. The use of numerical models provides advantages in the progression of RF ablation, IRE ablation and other applications to human clinical trials, and aid in the design of optimised preclinical trials [8]. Parameters such as blood flow, dielectric and thermal properties of the tissues all need to be incorporated in the development of accurate numerical models [9].

There are numerous studies in the literature reporting conductivity measurements of blood. However, knowledge of the conductivity of human blood at frequencies below 1 MHz and in particular below 100 kHz is limited due to the measurement challenges, including electrode polarization (EP), limited availability of human blood and difficulties in maintaining the temperature of blood. The two most commonly used methods of eliminating EP are the use of so-called platinized platinum electrodes (in a conventional two-electrode setup) and four-electrode technique [10]. platinized platinum method minimises the effect of EP by maximizing the electrode–electrolyte interface area, which is inversely proportional to the impedance [11]. This increase in area can be achieved either by mechanically roughening the electrode surface, or by using electrochemical treatments that produce a porous or fractal electrode surface with a large effective surface area [11]. It was empirically shown [10] that platinized platinum electrodes demonstrated extremely rough surfaces and shifted the EP to lower frequencies. Four-electrode techniques were first described by Schwan [10] to eliminate the problem of EP by providing a second pair of electrodes with which to measure the voltage across the sample [11]. Recent studies usually implement four-electrode technique [12]–[14].

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This study presents the measurements of low-frequency (i.e. below 1 MHz) conductivity of whole human blood at body temperature using the four-electrode technique.

## II. BACKGROUND

### A. Blood Conductivity Studies

Fricke *et al.* [15] in their 1925 study measured electrical resistance and capacity of calf blood at the frequency range from 800 Hz to 4.5 MHz using a custom two-electrode electrolytic cell [15], [16]. The effect of polarization was mitigated by the use of the platinum electrodes coated in platinum black [15]. They found that at lower frequencies, which they specified as being from 3.6 kHz to 8.7 kHz, the resistance and the capacity of blood are frequency independent [15]. At frequencies from 8.7 kHz to 4.5 MHz however, they found that both resistance and capacity begin to decrease [15]. Hirsch *et al.* [17] and Texter *et al.* [18] used a similar two-electrode electrolytic cell with platinized platinum electrodes in their 1950 studies. They found that the conductivity of blood can be used to measure red cell counts [18]. They used a variable frequency oscillator operating at approximately 1 kHz, 5 kHz, 11 kHz and 19 kHz as their alternating current source [17].

Both Fricke *et al.* [15] and Hirsch *et al.* [17] stressed the importance of stirring blood samples before measuring the conductivity to ensure the homogeneity, as letting blood settle can increase conductivity.

In 1973, Geddes and Sandler [19] measured the specific resistance of human, canine, bovine and equine blood at 25 kHz, at body temperature with a Hct range extending from 0 to 0.7 L/L. They eliminated the electrode impedance errors by varying the length of their two-electrode measurement cell. They found the resistivity of blood increases with increasing haematocrit [19]. In a 1975 study by Mohapatra and Hill [20] it was confirmed that resistivity of blood increases with increase in haematocrit. The same study also shows resistivity increase with decrease in temperature [20].

Beving *et al.* [21] in 1994 measured the dielectric properties of human blood at radiofrequencies (200 kHz–10 MHz) and investigated the dependence of the properties on cell volume fraction. Beving *et al.* [21] were using a four-electrode technique with the four electrodes being made of solid 24-carat gold pins in order to mitigate the effect of electrode polarization. In 2002, Chelidze *et al.* [22] covered the frequency range from 30 Hz to 30 MHz. Chelidze *et al.* [22] used a two-electrode technique at lower frequencies. They excluded the effects of electrode impedance by using the platinized platinum electrodes and varying the distance between the electrodes [22]. Jaspard *et al.* [23] performed measurements on bovine and ovine blood at body temperature at frequencies from 1 MHz to 1 GHz in 2003. They used an open-ended coaxial probe method. They found that the dielectric properties of blood are strongly dependent on haematocrit [23]. The conductivity was found to decrease with the increase of haematocrit while the relative permittivity would increase with the increase of haematocrit [23].

In 2008 study, Chang *et al.* [12] find the four-electrode technique superior to the two-electrode technique for an accurate characterization of blood impedance in the low-frequency range (100 Hz up to 20 kHz). The four-electrode technique also has been used to characterize the blood impedance by Gabriel *et al.* [13] in 2009 and Constantinou *et al.* [14] in 2017.

There are also a number of studies examining the dielectric properties of blood at microwave frequencies. Alison and Sheppard [24] characterized human blood dielectric properties at frequencies from 29 GHz to 90 GHz in a 1993 study. Wolf *et al.* [25] measured dielectric spectra of human blood over a broad frequency range from 1 Hz to 40 GHz using several different techniques across the frequency range. In the frequency range from 1 Hz to 10 MHz, they applied the AC voltage to a parallel plate capacitor made of platinum containing the sample material [25]. In a more recent study from 2018 Santorelli *et al.* [26] found that at microwave frequencies (i.e. 500 MHz–8.5 GHz) haemoglobin is the biggest predictor of changes in complex permittivity of blood. They demonstrated that blood permittivity at a single microwave frequency (i.e. 1 GHz) can detect anaemia with high sensitivity and specificity [26]. All studies at microwave frequencies were conducted by using the open-ended coaxial probe method [24]–[27].

This study employs the method that was used by Gabriel *et al.* in their 2009 study [13]. However, Gabriel *et al.* measured porcine blood in vivo while in this study we measured human blood samples in vitro. This study is thus the first to our knowledge to measure the conductivity of:

- (a) human blood,
- (b) at body temperature (37 °C),
- (c) at frequencies below 1 MHz, and
- (d) using the four-electrode technique.

Table 1 presents an overview of studies investigating blood conductivity. The table is comprehensive and includes the aforementioned studies as well as some studies that were not mentioned in the text. The fields in the table are highlighted where the corresponding study has fulfilled one or more of the criteria mentioned above. Of note is that none of the studies in the table fulfils all four criteria.

## III. MATERIALS AND METHODS

### A. Samples

Thirteen ( $N=13$ ) venous blood samples were taken from the antecubital veins of thirteen oncology patients ( $N=13$ ) approximately 30 minutes before performing the measurements. Whole blood was collected in ethylenediaminetetraacetic acid (EDTA) tubes, and transferred directly to the laboratory for measurement. The patients were scheduled for blood tests, and in accordance to ethical guidelines, one EDTA tube was set aside for the experiment. The blood samples were sourced by University Hospital Galway, who also secured the ethical approval for the study. Ten tubes ( $N=10$ ) contained approximately 6 ml of blood, which was enough to fill the cell culture well. The results of measurements on these ten samples were used to assess the mean conductivity of blood.

TABLE I  
OVERVIEW OF STUDIES ON THE CONDUCTIVITY OF BLOOD: HIGHLIGHTED FIELDS FULFIL ONE OF THE FOUR CRITERIA: A) HUMAN BLOOD, B) AT BODY TEMPERATURE, C) AT FREQUENCIES BELOW 1 MHz AND D) USING THE FOUR-ELECTRODE TECHNIQUE. NO STUDY SATISFIES ALL FOUR CRITERIA.

Year	Study	Method	Frequency	Electrode polarization compensation/reduction	Species	Temperature [°C]	Blood count dependency
1925	Fricke and Morse [15]	Two-electrode electrolytic cell	(800 Hz) 87 kHz–4.5 MHz	Platinized platinum	Calf	21.6	
1950	Hirsch <i>et al.</i> [17]	Two-electrode glass conductivity cell	1075 Hz, 5.2 kHz, 11 kHz, 19 kHz	Platinized platinum, 5 kHz results reported	Human	30	Conductance decreasing with increase in red cell count
1973	Geddes and Sadler [19]	Two-electrode conductivity cell	25 kHz	Variable length	Human	37	Resistivity increasing with percentage haematocrit
					Canine Bovine Equine		
1975	Mohapatra and Hill [20]	Two-electrode conductivity cell	100 kHz	Variable length	Human	22–40	Resistivity increasing with percentage haematocrit
1980	Hahn <i>et al.</i> [28]	Nine needle probe, two-electrode configuration	(0.1 MHz) 1 MHz–110 MHz (100 MHz)		Pig	34–36	
1993	Alison and Sheppard [24]	Waveguide system	29 GHz–90 GHz	N/A	Human	25, 37	
1994	Beving <i>et al.</i> [21]	Four-electrode	200 kHz–10 MHz	Four-electrode technique 24 carat gold pins electrodes	Human	25	
2002	Chelidze <i>et al.</i> [22]	Two-electrode	30 Hz–30 MHz	Platinized platinum Variable length	Human	22–28 (15–75)	
2003	Jaspard <i>et al.</i> [23]	Open-ended coaxial probe	1 MHz–1 GHz	N/A	Cow	37	The conductivity increases in the whole frequency range when the haematocrit decreases
					Sheep		
2008	Chang <i>et al.</i> [12]	Two-electrode (12 MHz - 100 MHz) Four-electrode (100 Hz–20 kHz)	100 Hz - 100 MHz	Four-electrode technique	Human	24	
2009	Gabriel <i>et al.</i> [13]	Four-electrode probe	40 Hz - 1 MHz	Four-electrode technique	Pig	37	
2011	Wolf <i>et al.</i> [25]	Parallel plate capacitor (1 Hz–10 MHz) Open-ended coaxial probe	1 Hz–40 GHz	Platinum coating	Human	37	
2017	Constantinou <i>et al.</i> [14]	Four-electrode technique	30 kHz–300 kHz	Four-electrode technique	Human	22	
2017	Salahuddin <i>et al.</i> [27]	Open-ended coaxial probe	400 MHz–20 GHz	N/A	Human	37	
2018	Santorelli <i>et al.</i> [26]	Open-ended coaxial probe	500 MHz–8.5 GHz	N/A	Human	24.1	Haemoglobin is the biggest predictor of changes in complex permittivity of blood

Three tubes ( $N=3$ ) contained less than 6 ml of blood, which was insufficient for use.

A full blood count of eight ( $N=8$ ) out of remaining ten samples was provided by the Irish National Accreditation Board accredited testing laboratory (Registration Number '239MT'). This full blood count included the total white cell count (WCC), red blood cell count (RBC), haemoglobin concentration (Hgb), haematocrit (Hct), mean corpuscular volume (MCV), mean corpuscular haemoglobin (MCH), mean corpuscular haemoglobin concentration (MCHC), platelet count, and differential white cell count (as neutrophil, lymphocyte, monocyte, eosinophils and basophils levels). The mean WCC and lymphocytes values were increased. Mean WCC was  $13.6 \times 10^9/L$ , while the normal range is 4 to  $10 \times 10^9/L$ . The mean lymphocyte count was  $4.2 \times 10^9/L$  and the normal range is 1 to  $3 \times 10^9/L$ . The mean values of other blood counts were within the normal range.

The samples were delivered to the laboratory at room temperature. The first measurement on each sample was performed at room temperature ( $22.6^\circ\text{C}$ ,  $SD=0.8^\circ\text{C}$ ). The second measurement on each sample was performed at close to body temperature ( $36.6^\circ\text{C}$ ,  $SD=0.4^\circ\text{C}$ ). The samples were brought to body temperature using a thermal water bath.

### B. Measurement Method

Measurements of electrical impedance of human blood were performed with a four-electrode probe method. Electrical impedance was measured on  $N=10$  human blood samples. Electrical impedance data was measured using the PGSTAT204 potentiostat/galvanostat (Metrohm Autolab B.V., Utrecht, The Netherlands) in a galvanostatic mode. The measured electrical impedance was used to calculate electrical conductivity of the samples.

The measurement setup consisted of the PGSTAT204, connected to a custom made four-electrode probe. A personal computer running NOVA 2.1.1 software including FRAM32M impedance analyses module (Metrohm Autolab B.V., Utrecht, The Netherlands) was used to capture the measurement data. A cell culture tray was used as a sample holder and the temperature of the samples was controlled by the water thermal bath (Fisher Scientific, Isotemp R, Waltham, MA, USA).

Right before each measurement, the temperature of the samples was measured using the digital stick thermometer. The thermometer was also used to stir the sample before the measurement in order to avoid the effects of aggregation and sedimentation of RBCs, which can affect the conductivity measurement [15], [17], [29].

The four-electrode probe was assembled with the parts from a standard electronics development kit and is shown in Figure 1. Four pins of a gold plated pin header (Harvin, Portsmouth, United Kingdom) were used as four electrodes in a linear array. Four male-to-female breadboard jumper cables (MikroElektronika, Belgrade, Serbia) were used to connect the pins to the PGSTAT204. The pins were connected to the PGSTAT204 in a four-electrode configuration.

In a galvanostatic mode, constant current is injected between the two outer electrodes called the working electrode (WE,

red) and counter electrode (CE, black). The value of current was set to  $10 \mu\text{A}$  RMS in order to operate in a pseudo-linear region, where the complex impedance does not depend on the value of current [30]. Voltage drop is measured between the inner two electrodes, which are called the working sensing electrode (WSE, purple) and reference electrode (RE, blue). The measurements were performed at  $N=51$  frequency points with a logarithmic distribution across the range from 1 Hz to 100 kHz ( $N=10$  frequency points per decade).

At lower end of the frequency range the negative effects of EP were not completely eliminated which resulted in corrupt measurement results. The results that were corrupted by EP were obvious as at frequencies below 631 Hz the measurement results varied drastically between the neighbouring frequency points and some of the data points would have negative real parts. Therefore, since all the data points with negative real parts were at frequencies below 631 Hz, a low cut-off frequency of 631 Hz was selected.

Electrical impedance data for each sample was verified using the Kramers-Kronig (KK) relations. The KK relations consist of a set of transformations that can be used to predict one component of the impedance from the other over the frequency limits from zero to infinity [32]. In practice, it is impossible to measure impedance over the infinite frequency range. The lin-KK method from Schönleber *et al.* [33] is a quick test for checking the validity of the electrical impedance spectroscopy (EIS) data. The validity of an impedance spectrum is analysed by its reproducibility by a KK compliant equivalent circuit. The residual error between the predicted and measured impedance can then be used to determine consistency with the KK relations [32].

The average error of the fit over the frequency range in worse case was less than 0.3% for both resistance and reactance. The maximum value of the fit error at any single frequency never exceeded 0.8%. The low error of the fit values signify a good fit and KK compliance of the impedance data.

### C. Cell Constant and Measurement Uncertainty

Impedance data was converted to electrical conductivity of the samples in two steps. The first step is determining the cell constant, which is the factor that relates measured conductance and the corresponding reference conductivity [13]. The second step is calculating the electrical conductivity from the measured impedance data and the cell constant. The cell constant is determined by measuring the impedance of a standard liquids with known electrical conductivity and relative permittivity. In this study we determined the cell constant by using 0.01, 0.05, and 0.1 mol/L aqueous NaCl solutions at room temperature as standard liquids. The cell constant  $k$  (in m) is calculated as:

$$k = G/\sigma, \quad (1)$$

where  $G$  is the measured conductance (in S) and  $\sigma$  (in S/m) is electrical conductivity of the standard liquid. The conductance for each of the three aqueous NaCl solutions was measured  $N=10$  times (total  $N=30$  measurements). In theory, there should be no dielectric dispersion in ionic aqueous solutions at frequencies below 1 MHz [6]. Therefore, the conductivity is

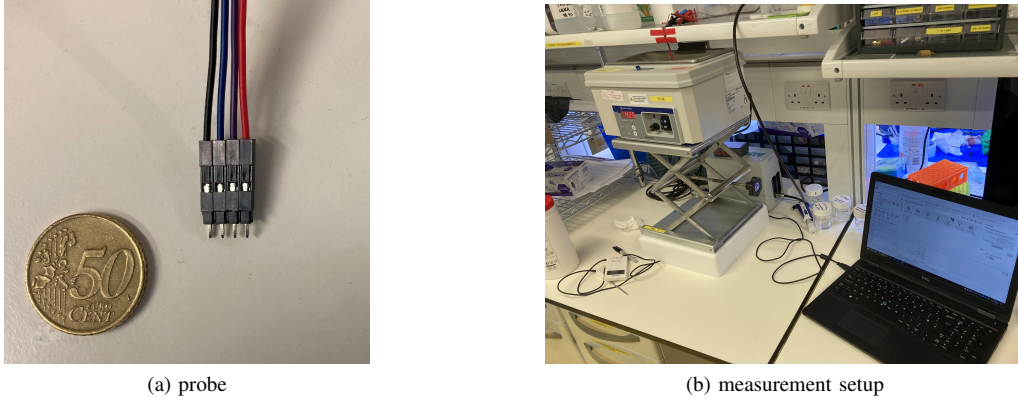


Fig. 1. The probe is sufficiently small to measure small samples ( $< 6$  ml), is cheap and easy to make and has shown good characteristics down to 100 Hz (a). The probe is designed to be connected to a four-terminal impedance measurement device in a four-electrode configuration (b). The connections are: working electrode (red), working sensing electrode (purple), reference electrode (blue), and counter electrode (black). The distance between the electrodes is 2.54 mm and the length is 3 mm.

TABLE II

MEAN CONDUCTIVITY [S/M], STANDARD DEVIATION [S/M], STANDARD DEVIATION AS A PERCENTAGE OF THE MEAN [%], REFERENCE CONDUCTIVITY [S/M] [31] AND PERCENTAGE DIFFERENCE BETWEEN THE MEAN MEASURED CONDUCTIVITY AND REFERENCE VALUES [%].

NaCl [mol/L]	$\sigma$ [S/m] average	SD ( $\sigma$ ) [S/m]	SD ( $\sigma$ ) [%]	$\sigma$ [S/m] literature [31]	Error [%]
0.01	0.097	0.006	6.05	0.094	3.27
0.05	0.457	0.026	5.79	0.466	-1.98
0.1	0.934	0.032	3.40	0.932	0.24

frequency independent and only varies with the concentrations of the aqueous NaCl solution. The cell constant is determined as the coefficient that relates the measured conductance and reference conductivity [13]. The cell constant of the probe in this experiment is 0.028 m which is suitable for measurements of biological tissues and is similar in value to the value of cell constant in the Gabriel *et al.* [13] 2009 study (0.02 m). The conductivity is calculated by dividing the measured conductance by the cell constant [13].

The same  $N=30$  measurements on standard liquids were used to quantify the systematic and random error [13], [34]. In ideal case there should be no variation in measured conductance as a function of frequency [13]. We used the cell constant and the measured conductance to calculate the ionic conductivity of each of the aqueous saline solutions. The deviation of the mean calculated conductivity from the true conductivity of the standards corresponds to systematic error, while the standard deviation of the mean represents random error. Table II gives both the random and the systematic error values. If we consider just the values for 0.05 mol/L saline as it is the standard with conductivity closest to that of human blood, we get combined uncertainty of 5.90% [34]–[36]

#### IV. RESULTS

The measurements were compared to the data from the literature. Figure 2 shows the mean conductivity of the  $N=10$  measurements of  $N=10$  samples (one measurement per each

sample) across the frequency range from 631 Hz to 100 kHz. The dark blue line is the mean of measurements performed at room temperature and dark red line is the mean at body temperature. The shaded areas around the lines represents the variability between the samples conductivity by showing the plus or minus one standard deviation of the mean interval.

The measured conductivity is in good agreement with the data from the literature [15], [17], [18], [20], [21], [37], [38]. The mean conductivity measured at room temperature is lower than conductivity measured at body temperature. The difference can be explained by blood conductivity increasing with temperature as the mobility of the ions that transport the current increases with temperature and the viscosity of the extracellular fluid decreases [20], [39]. The data from this study at body temperature matches the data from Hirsch *et al.* [17] and Texter *et al.* [18] at 30°C, as well as the data from Rosenthal and Tobias [38] at body temperature. The average difference between our data and data from Hirsch *et al.* [17] is 2.55%. The average difference between our data and data from Texter *et al.* [18] is 13.56%. The difference between our data at  $f=1$  kHz and data from Rosenthal and Tobias [38] at  $f=1$  kHz is 12.65%. The data from this study at  $T=23$  °C and at  $f=79$  kHz matches well with the data from Fricke and Morse [15] at  $T=21.6$  °C and at  $f=87$  kHz. The difference between the two values is 1.35%. Other data from literature is at either higher or lower frequencies [20], [21].

Conductivity relation to red cell count and erythrocyte concentration has been demonstrated in previous studies by Hirsch *et al.* [17] and Texter *et al.* [18]. In this experiment, the three blood count parameters that were most highly correlated with the conductivity were Hct, Hgb and RBC. Conductivity is most determined by the Hct, where at  $f=631$  Hz, 80.7% of variation in conductivity between the samples can be explained by the difference in Hct between the samples. Since Hgb and RBC are also correlated with Hct, this explains the high correlation between the conductivity and Hgb and RBC.

Figure 3 shows the coefficient of determination between the conductivity and the most correlated three blood counts (Hct, Hgb and RBC) for different frequencies. We see that at lower

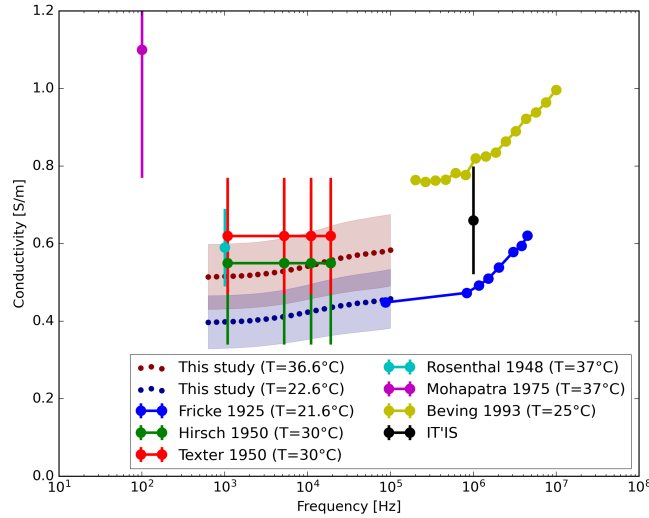


Fig. 2. Mean conductivity in S/m of: human blood samples at room temperature (dark blue is data points, shaded is mean  $\pm$  SD); human blood samples at body temperature (dark red is data points, shaded is mean  $\pm$  SD); and blood conductivity value from literature for human blood conductivity (mean  $\pm$  SD) [15], [17], [18], [20], [21], [37], [38]. The data from this study at body temperature matches very closely the data from Hirsch et al. [17] (green line) and Texter et al. [18] at 30°C (red line) as well as the data from Rosenthal and Tobias [38] at body temperature (light blue line). The average difference between our data and data from Hirsch et al. [17] is 2.55%. The average difference between our data and data from Texter et al. [18] is 13.56%. The difference between the data from this study at 1 kHz and the data from Rosenthal and Tobias [38] at 1 kHz is 12.65%. The data from this study at  $T=23^\circ\text{C}$  and at  $f=79$  kHz matches well with the data from Fricke and Morse [15] (blue line) at  $T=21.6^\circ\text{C}$  and at  $f=87$  kHz, with 1.35% difference between the two values. Other data from literature is at either higher or lower frequencies and are shown to illustrate the extent of human blood conductivity data available in the literature.

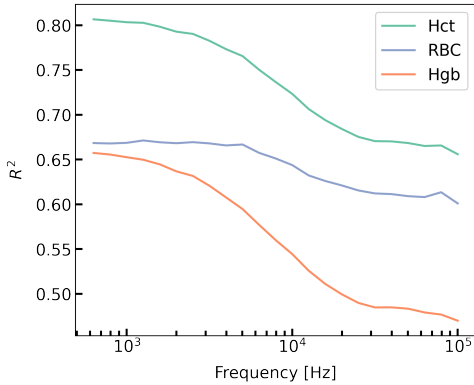


Fig. 3. Coefficient of determination for conductivity and the three most highly correlated blood counts for different frequencies. The blood counts are Hct, Hgb and RBC. The figure shows that the coefficient of determination drops with frequency and its values are highest for Hct ( $T=36.6^\circ\text{C}$ ).

frequencies there is higher correlation between the conductivity and all three blood counts. The highest correlation is with the Hct and at lower frequencies. The high correlation would be key in applications that rely on the relationship between the conductivity and blood count parameters focused on red blood cells, such as the detection of low Hct for example. The fact that the correlation is higher at low frequencies highlights the feasibility of using low frequency conductivity measurements as a diagnostic tool.

## V. DISCUSSION

### A. Limitations

One of the challenges in measurement of conductivity of human blood is sourcing the human blood samples. In

this study we only had  $N=8$  samples which satisfied the conditions too be included in this study (size of the sample and availability of blood counts) that were coming from oncology patients. The small number of samples as well as the fact that the conductivity of blood plasma was not considered separately limits the scope of this study. This study does not claim the blood counts can be predicted by measuring the conductivity of the whole blood, but it indicates that if such application was desired, working at lower frequencies would be preferred.

## VI. CONCLUSION

This study introduces a simple and low-cost four-electrode probe as a practical tool for measurements of electrical conductivity of human blood at lower frequencies. This study is first to measure the conductivity of human blood at low frequencies (i.e. below 1 MHz) and at body temperature using the four-electrode technique.

The results of this study are the values of conductivity of whole human blood (at  $25^\circ\text{C}$  and at  $37^\circ\text{C}$ ) and the Pearson correlation coefficient between the measured conductivity and available blood counts. Both the conductivity values and the correlation coefficient values are given at frequencies from 631 Hz to 100 kHz. The values of the measured conductivity are in good agreement with the data from the literature and have also been validated using Kramer-Kronig relations. These agreements in measurement data also validates our methodology. The correlation coefficient between the conductivity and blood counts decreases with frequency, highlighting the advantage of using low-frequency conductivity measurements in applications that rely on strong relationship between the conductivity and blood counts. These results provide valuable additional data to the literature, which is currently limited and will help further medical device design.



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