

## Investigation of Metrological Performance of the ISFET-Based pH Sensors

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**Abstract:** The metrological performance characteristics of the dual channel *p*-type ISFET pH sensors with silicon nitride sensitive layer are investigated. The simplified three-lead sensor design and corresponding signal readout circuit are employed and the sensor transform function is evaluated as dependency of the ISFET channel current on the analyzed solution pH. It is shown that achievable pH measurement accuracy is sufficient for use of the developed sensors in typical laboratory applications in either single channel or differential mode of operation, as alternative to the traditional glass electrodes. Copyright © 2014 IFSA Publishing, S. L.

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### 1. Introduction

Ion-sensitive field-effect transistors (ISFET) are a well-known alternative for conventional glass electrodes in pH sensing applications [1]. The potential advantages of these solid state devices as compared to glass electrodes are faster response, small size, better mechanical ruggedness, easier maintenance, and ease of manufacturing in bulk quantities with the use of existing semiconductor technology facilities. However, despite over 40 years of research and development history, ISFET sensors have failed to supersede glass electrodes in all but a few selected application areas, namely industrial monitoring [2] and biochemical research [3, Chapter 26; 4, Chapters 7, 10].

There are several reasons that prevent the ISFETs from coming into widespread use. The major ones are:

- While the standard CMOS process can indeed be used to produce the ISFET, the performance of the

resulting product is likely to be mediocre in comparison with a high-quality glass electrode; in order to improve it, additional non-standard operations must be introduced into production route, such as the use of relatively exotic materials (metal oxides, special composition silicates) for gate dielectric [5; 6], and non-standard thermal and chemical conditions for annealing stages [7–11]; this can make the production economically unfeasible.

- Although the planar semiconductor technology allows for very small active sensing area sizes, the problem of isolation of the non-operational silicon crystal areas from the analyzed liquid leads to necessity of special measures for crystal sealing, including the use of various polymer compounds and special encasement designs. This at least partially levels out the potential possibility of sensor miniaturization.

- Since from electrical standpoint the ISFET acts as a usual field-effect transistor, a variety of different conversion circuits can be used for electrical

signal readout. This can be considered an advantage since the signal conversion is potentially easier compared to the glass electrode that requires a potentiometric circuitry with very high input impedance; ISFET, on the contrary, is a transconductance device and can be used in virtually any signal conversion circuit as conventional transistor. But on the other hand, this freedom of choice for the secondary transducer schematic results in lack of standardized instrumentation; as a result, every commercial ISFET sensor manufacturer has to offer accompanying instruments for their sensors, that have limited or no compatibility with the sensor modules from other suppliers.

The rest of the practical disadvantages of ISFET sensors are similar to that of the glass electrodes: requirement of periodical calibration, presence of the long-term drift, temperature dependence and inability of accurate measurement in solutions with low ionic strength (low conductivity). This is not surprising since the primary transduction principle (equilibrium proton exchange) is the same in both types of sensors.

If we take a look at the application peculiarities in the areas where ISFET did succeed, we shall see that the use of these sensors is advantageous not because of their sensory properties, but rather because of mechanical ruggedness and chemical resistance (in industrial applications) and convenience of biochemical material deposition due to availability of the small active area with a flat surface (in scientific applications). In other words, the actual advantages of ISFET sensors over traditional electrodes lie in the area of sensor design and geometry rather than in primary transducer properties; the latter in practice appear to be very similar.

The dual channel ISFET sensors considered in this work were developed primarily for biosensor applications, namely for estimation of relative changes in enzyme activity upon the influence of various chemical agents. The demands for underlying pH sensor performance characteristics in this kind of application may vary greatly depending on the actual enzymatic reaction used and desirable detection threshold for corresponding substrate or inhibitor, but generally a moderate accuracy pH sensor (with equivalent maximum error of about 0.1 pH) is sufficient in many cases, since the uncertainty caused by variations in enzyme layer properties (enzyme quantity, membrane thickness, layer adhesion etc.) can be much greater than the error of the pH sensor itself [12]. At the same time, the measurement error caused by susceptibility of the sensor to various influencing factors such as temperature, ambient light, changes in chemical composition and pH of the background medium (buffer solution) should be suppressed by subtraction of the reference channel signal in the two-channel configuration, when a dummy membrane that does not contain a reactive enzyme is deposited onto the reference channel active area. In order for the latter approach to work, though, both sensors fabricated on a single crystal must

demonstrate identical response to the mentioned influencing factors; this identity must be ensured by the sensor pair manufacturing process.

The developed sensors were, to various extents, tested for suitability in toxicity analysis of aqueous solutions [13; 14], alkaloid analysis in *solanaceae* agricultural plants [15], and urea and creatinine analysis in blood serum [16–19], with generally satisfactory results. Although the associated measurements were of semi-qualitative or indicative kind, even these researches, however, have shown the necessity for quantitative assessment of the transducer performance, in order to separate the sources of errors originating in the transducer itself, in the electric signal conversion instrumentation, and in the enzymatic membrane, and consequently to improve the sensor preparation and application protocols.

The purpose of the work presented in this paper was to evaluate the ISFET transducers generic performance characteristics, and thus to estimate their suitability for quantitative measurement with defined accuracy limits in both biosensory and conventional pH-metry tasks.

## 2. Sensor Fabrication and Design

The ISFET sensor crystals were fabricated at the “Mikron” production and research facility (Russia) [20]. Each crystal of 8×12 mm in size contains a pair of *p*-channel transistors on the *n*-type silicon substrate. The source and drain areas along with the leads to the contact pads were formed by thermal diffusion of acceptor impurity. The double-layer gate dielectric consists of 30 nm SiO<sub>2</sub> obtained by thermal oxidation, followed by 70 nm of Si<sub>3</sub>N<sub>4</sub> obtained by low-pressure chemical vapor deposition. The annealing in water vapor at 400 °C was used at the final stage of production.

The crystals were mounted on the fiberglass boards of 8×50 mm in size, and the crystal contact pads were connected to the board printed contact leads (gold-plated) with aluminum wires by means of ultrasonic welding. The contact area then was sealed with epoxy compound.

More detailed description of the sensors fabrication process and design can be found in [21].

The crystal bulk lead and the source leads of the both transistors are electrically connected on the printed circuit board. Thus the assembled dual channel sensor has only three contact leads: the first drain, the second drain, and the common source/bulk. This imposes certain limitations upon the transistors mode of operation and the electrical signal conversion circuit; however, as will be shown below, this does not lead to performance degradation once the proper calibration procedure is employed. Additionally, reducing the number of leads decreases the primary cost of sensor manufacturing in a large scale production.

### 3. Experimental Setup and Measurements

Similarly to the glass pH electrode, ISFET transducer converts proton activity in the analyzed liquid solution to the electric potential difference between the reference electrode and the sensor surface [5]. In case of ISFET, however, it is impossible to measure this potential difference directly. Instead, the fact that the ISFET channel current (once the appropriate drain-to-source voltage is applied) is dependent on the gate charge buildup, which in its turn depends on proton activity, is employed. In a sense, being a transconductance device, ISFET incorporates a part of the secondary signal transducer in itself.

In many (probably in most) of the published signal conversion circuits, ISFET operates in the constant channel current mode, the latter being set by an active current stabilizing voltage tracking circuit with negative feedback [22, 23]. The circuit output signal thus tracks the changes in the ISFET gate potential, mimicking the traditional glass electrode operation.

While this approach allows, albeit indirectly, to read out the primary transducer response signal, it also has several drawbacks. Although the negative feedback essentially eliminates the influence of the ISFET transfer characteristic on the static output of the secondary transducer, the dynamic behavior of the circuit (the output signal settling time and possibility of oscillations during the signal transitions) is still affected by the intrinsic transconductance curve in the vicinity of the ISFET operating point. Since the latter can be chosen quite arbitrarily, the dynamic performance becomes rather unpredictable, especially during the abrupt changes of the analyzed solution pH or chemical composition. While such changes are unlikely to occur during the normal course of operation, they are typical for the calibration process. The circuit behavior in this case can directly affect the subsequent measurements accuracy.

Moreover, if the sensor loses contact with the analyzed liquid, the feedback chain breaks and the circuit output becomes undetermined. Special measures must be taken to prevent the output signal latch-up, and to ensure that the circuit returns to the tracking mode once the contact with solution is restored.

If differential mode of operation with one reference and one active sensor channel is required, the separate control loops must be set up for each of the two transistors. Needless to say that this further complicates the secondary transducer schematics.

The gate potential tracking, however, is by no means the only possible way to register the sensor response. Nothing prevents us from measuring the channel current and associating its changes with the solution pH changes directly. Of course this requires the prior characterization of the sensor transfer function, but in practice this is not a significant issue,

since the calibration is required for accurate measurement regardless of the chosen signal readout method, and the parametric form of the transfer function is known.

In this work, the dependence of ISFET channel current on the analyzed solution pH at the fixed drain and gate control voltages relative to transistor source, as an intrinsic transducer characteristic, was investigated. It is worth to note that the direct sensor current readout allows to employ the ISFET pair in differential configuration without the need for separate control loop for each transistor; this, in its turn, allows to use a single programmable voltage source for the operating point setup, as long as the transfer characteristics of the transistors are well matched.

The experimental setup for evaluation of the developed sensors characteristics consisted of the controlled laboratory power supply (Matrix MPS 3003L-3) and digital multimeter (Fluke 287) for the sensor current measurement. A fixed voltage of -5 V was applied to the transistors drain relative to the common source/bulk lead, and controlled variable voltage was applied to the gate via the argentum chloride reference electrode.

The Hamilton DuraCal reference buffer solutions with pH values of 4.01, 7.00, 10.01 ±0.01 and 5.55, 8.37 ±0.02 were used to obtain the pH dependence of the sensor current at the fixed gate voltage. The latter was preset in the pH 7.00 buffer solution so that the initial ISFET channel current was equal to 500 ± 5 microamperes. The control voltage setup and pH dependency measurement was performed separately for each of the two sensor channels.

The gate control voltage value corresponding to the 500 μA channel current was in the range of -(2.6..2.8) volts for all tested sensor samples; the ISFETs thus were guaranteed to operate in saturation (pentode) region.

All measurements were performed at controlled ambient temperature of 24±0.5 °C. All sensor samples for testing were randomly selected from the same production run.

A preliminary series of measurements was conducted to determine the basic characteristics of the ISFETs (threshold voltage and transconductance values) and short-term stability of the transfer characteristic. For this purpose, the transfer characteristics (channel current vs. gate control voltage at the fixed drain voltage) were measured in pH 7 solution, and threshold voltage values were calculated from quadratic approximation of the obtained data.

To estimate the short-term response drift, the changes in channel current (from the initial value of 500 μA) during 1.5 hours with the fixed gate voltage were registered, and the drift rate calculated as the difference between maximum and minimum current magnitudes registered during the last hour of measurement.

The pH sensitivity was evaluated by measuring ISFET channel currents in three reference solutions

(pH 4.01, 7.00 and 10.01) and performing linear approximation of the obtained dependence.

All these measurements were performed for two sets of sensors, 20 sensors without special treatment, and 80 sensors with the dielectric surface cleaned with chromic acid (0.1 M potassium dichromate in concentrated sulfuric acid diluted by 50 % with distilled water) prior to measurement. The data obtained for these two groups of sensors was processed separately.

Since the cleaning with chromic acid have shown to improve the pH sensitivity (see Table 1 in the next section), all sensors used in the subsequent measurements for pH sensitivity and accuracy evaluation were subjected to this treatment.

Additional measurements for pH sensitivity and accuracy evaluation were performed for five newly selected sensors. Four series of measurements at intervals of two weeks were conducted in course of this investigation, in order to determine the long-term stability of the sensor characteristics in addition to the primary characteristic values.

For each of the five tested sensors (ISFET pairs), the channel currents were measured in buffer solutions with five different nominal pH values, uniformly distributed within the pH 4..10 range. To determine the sensor response hysteresis, the measurements were performed in three passes, with the buffer pH changing first from acidic to basic (pH 4 to pH 10), then in the reverse order, and then once more in the initial order.

Before the change of the buffer solution the sensor surface was rinsed with distilled water and then with the subsequent reference solution. The channel current value was registered 5 minutes after the buffer solution change to minimize the error caused by sensor response kinetics.

Between the measurement sessions the sensors were stored in dry condition at room temperature without any special precautions.

#### 4. Results and Discussion

The sensor current values registered in experiment for five different pH buffer solutions were used to determine the quadratic transform function coefficients by means of least squares approximation. A typical experimental sensor transform characteristic along with the approximating curve is shown in Fig. 1.

The obtained quadratic model dependence was then used to calculate the following parameters:

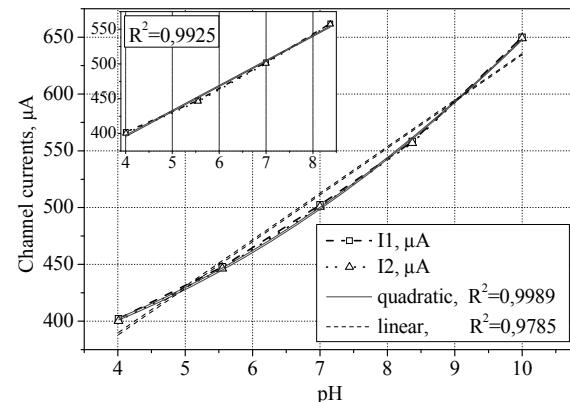
1) pH-sensitivity of each ISFET in the given pH range (pH 4..10);

2) Difference in pH values calculated from the approximation curves between the two sensor channels;

3) Variation of the difference in calculated pH values between the two sensor channels across the entire range of pH 4..10;

4) Hysteresis of the sensor response at the middle point of the pH range (pH 7);

5) pH determination error from the obtained model dependence (assuming zero error in the independent variable, i.e. the reference solution pH).



**Fig. 1.** Characteristics of ISFET as pH-to-current transducer with quadratic and linear approximations in pH 4..10 range; the inset shows linear approximation and its determination coefficient in pH 4..8.5 range.

The pH-sensitivity  $K(pH)$  [ $\mu\text{A}/\text{pH}$ ] of an ISFET was defined as derivative of the obtained quadratic dependence:

$$I(pH) = A \cdot pH^2 + B \cdot pH + C, \quad (1)$$

$$K(pH) = 2A \cdot pH + B,$$

where  $A$ ,  $B$  and  $C$  are the approximation coefficients. The sensor channels reading mismatch is then determined from the expression

$$\Delta pH_{1-2} = \Delta I_{1-2}/K, \quad (2)$$

where the lowest of the  $K_1$  and  $K_2$  values at the corresponding pH is substituted for  $K$ , the numeric indices indicating first and second sensor channel respectively, and  $\Delta I_{1-2}$  is the difference between ISFET channel currents determined from the approximation curves of the first and second channels.

The variation of the sensor readings mismatch between channels across the pH range is determined as

$$\delta pH_{1-2} = d(\Delta pH_{1-2})/d(pH) \quad (3)$$

where  $\Delta pH_{1-2}$  is the difference in pH value readings from the first and second sensor channels caused by mismatch in their transform characteristics (1). This parameter can also be defined via the mismatch of sensor channels sensitivity across the pH range

$$\Delta K_{1-2} = 2(A_1 - A_2)pH + (B_1 - B_2). \quad (4)$$

Hysteresis of the sensor response is determined as the maximum difference between ISFET sensor

currents corresponding to the same pH value in quadratic dependences (1) obtained from the measurements in forward (pH 4 to pH 10) and reverse order of the solution pH change. The equivalent pH reading error then can be determined from (2) with the hysteresis current difference substituted for  $\Delta I_{1-2}$ .

Finally, the pH reading as such is determined from the calibration curve (1) by the measured ISFET channel current  $I$  according to the expression

$$\text{pH} = \frac{-B + \sqrt{B^2 - 4A \cdot (C - I)}}{2A}. \quad (5)$$

The results of the analysis of preliminary measurement series are shown in table 1, and the performance characteristic values obtained from the additional measurement series are shown in table 2 below. Approximately Gaussian distribution was observed for calculated pH-sensitivity values in the first measurement series. Statistical distribution of the parameters obtained from the additional measurement series was not estimated because of the small number of samples. The mean values and standard deviations in Table 2 are calculated over the results obtained for five sensors in one measurement session; two channels of each sensor are accounted as independent samples except for channels mismatch parameters.

**Table 1.** Basic ISFET parameters obtained from preliminary measurements.

Parameter	Units	Measured or calculated parameter value range	
		Without sensor surface cleaning	With chromic acid cleaning
ISFET channel current pH-sensitivity	$\mu\text{A}/\text{pH}$	32..34	36..40
ISFET gate voltage pH-sensitivity*	$\text{mV}/\text{pH}$	28..31	33..35
ISFET transconductance in the current range of 350..650 $\mu\text{A}$	$\mu\text{S}$	867..896	877..920
ISFET threshold voltage	V	1.52..1.64	1.58..1.66
Operating point gate voltage	V	2.61..2.75	2.66..2.75
Channel current drift	$\mu\text{A}/\text{hour}$	2..6 (2..2.6..7 mV/hour, 0.06..0.17 pH)*	
Drift of the difference between sensor channels currents	$\mu\text{A}/\text{hour}$	0.34..1.22 (0.4..1.4 mV/hour, 0.01..0.04 pH)*	

\*Calculated from the measured transconductance curve and sensor pH-current transform function.

**Table 2.** Sensor performance characteristics obtained from additional measurement series.

Characteristic	Units	Mean value			Standard deviation		
		pH 4	pH 7	pH 10	pH 4	pH 7	pH 10
pH-sensitivity $K$	$\mu\text{A}/\text{pH}$	28	44.3	60.63	$\pm 2.9$	$\pm 2.74$	$\pm 5$
pH reading mismatch between sensor channels $\Delta p\text{H}_{1-2}$	pH units	0.102	0.097	0.187	$\pm 0.05$	$\pm 0.048$	$\pm 0.168$
Variation of the pH reading mismatch between sensor channels $\delta p\text{H}_{1-2}$	pH units	0.05	0.032	0.075	$\pm 0.04$	$\pm 0.03$	$\pm 0.065$
Mismatch of pH-sensitivity between the sensor channels $\Delta K_{1-2}$	$\mu\text{A}/\text{pH}$	2.19	1.36	3.3	$\pm 1.5$	$\pm 1.25$	$\pm 3.0$
Hysteresis error, directly measured	$\mu\text{A}$	2.6	9.7	4.9	$\pm 2.1$	$\pm 4.2$	$\pm 4.2$
Hysteresis error	pH units	0.1	0.259	0.09	$\pm 0.08$	$\pm 0.068$	$\pm 0.08$
pH reading error (assuming zero error in calibration solution pH)	pH units	0.125	0.097	0.071	$\pm 0.046$	$\pm 0.054$	$\pm 0.017$

As can be seen from the presented data, the performance of properly calibrated ISFET sensors is on par with the existing traditional pH measurement equipment and generally satisfies the requirements of practical applications that do not demand extraordinary accuracy [24, Section 3.10; 25–27].

However, the dynamic performance of the sensors can substantially affect measurement results as the equivalent pH reading errors caused by response drift and hysteresis can potentially become several times larger than the static calibration error. Unfortunately, there is no universal remedy for this issue. Possible countermeasures include improvements in the sensor

fabrication technology aimed towards stabilization of the mobile charge in dielectric and enhancement of the dielectric surface structure [7–11], and introduction of the additional restrictions to the sensor maintenance and usage protocol. Both technological and methodical approaches have their drawbacks though: the former increases sensor production cost, and the latter complicates the sensor application procedures.

Since the drift rate is likely to diminish with time [28], the negative effect of the drift upon measurement accuracy can be mitigated by prolonging the time between the exposition of the

sensor in analyzed solution and actual pH readout. This, however, severely limits possibility of dynamic pH changes monitoring.

In biosensory applications, the magnitude and direction of expected change of the solution pH are often known *a priori*. In this case it is advisable to perform sensor calibration in the narrower pH range (determined by the expected sensor response magnitude), changing the pH of the reference solution in the direction of the expected response, and keeping the reference solution chemical composition close to that of the working solution (e.g., calibrate sensor with the set of individually prepared phosphate buffer solutions with different pH values, sufficiently close to the nominal pH of the background buffer solution used in the actual experiment).

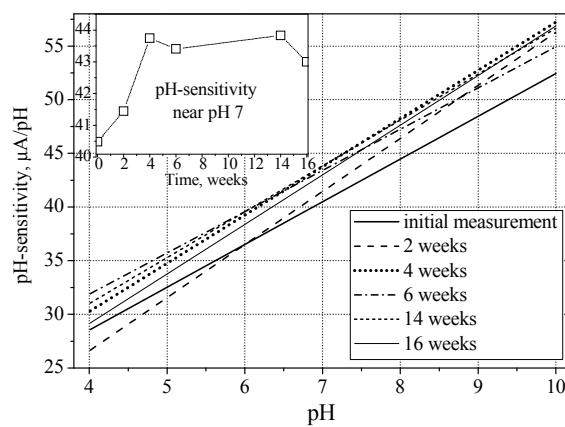
These practices however have limited use in conventional pH-metry, when neither composition of the analyzed solution nor the expected sensor response can be controlled.

It was reported before that preliminary soaking of ISFET sensors in buffer solutions and subsequent wet storage can improve the drift behavior [29]. This method is universally applicable for biosensory as well as other pH measurement applications, however it makes the sensors maintenance much less convenient. As was observed in our preliminary experiments, the treatment of sensors with acidic solution immediately before measurement can substantially decrease the hysteresis effect, and supposedly also positively affects the drift of the sensor response. Particularly, rinsing of sensors with pH 4 reference solution before exposition in the reference solutions with higher pH virtually eliminates the hysteresis effect and apparently accelerates the response kinetics. This can be explained by protonation of the silicon nitride surface, which subsequently accelerates the initial phase of the proton exchange reaction. This hypothesis is indirectly confirmed by the observable increase in pH-sensitivity after the chromic acid surface cleaning (although part of the effect is likely due to removal of organic contaminants). Thus the rinsing in highly acidic solution prior to use can appear to be useful sensor maintenance technique for the measurement accuracy improvement.

Another highly recommended technique of drift counteraction is the differential mode of sensor operation. As can be seen from Table 1, it allows to diminish the drift effect by almost an order of magnitude. This approach is casually employed in biosensory applications, but it also can be used in other pH-metry tasks by suppressing the proton sensitivity of one of the sensor channels with polymer composite layer [30–32]. The ISFET sensor design makes implementation of this mode of operation significantly easier than with the use of traditional glass electrodes.

Results of long-term pH-sensitivity stability test for one of the sensors are shown in Fig. 2. The plot shows the ISFET channel current pH-sensitivity

dependencies recalculated from the calibration data measured at intervals of two weeks (plus additional two points obtained from measurements after a 8 weeks break). These measurements were performed to test for spontaneous sensor surface degradation during the dry storage. It can be seen from the plot that the variation of sensitivity is rather significant (about 4  $\mu\text{A}/\text{pH}$  at the middle point of pH range, which translates to the pH reading error of about 0.1 pH). Some of this variance can be explained by variation in the independent variable, i.e. in the reference solutions pH (which was assumed to be zero when computing the quadratic approximation coefficients). However, the error of 0.02 pH in reference solution can only cause about 0.2  $\mu\text{A}/\text{pH}$  of sensitivity variation. The rest has to be attributed to the reference solutions ageing, ambient temperature variations, measurement errors caused by unaccounted response kinetics, and finally to the changes in sensor dielectric itself (i.e. oxidation).



**Fig. 2.** Sensor pH-sensitivity characteristics in pH 4..10 range calculated from quadratic approximations of experimental data and their variation during the period of 16 weeks.

At least we can state that there is no prominent trend of spontaneous decrease of sensitivity with time. It is thus seems possible to use the sensors without additional recalibration for prolonged time if accuracy of about 0.1 pH or worse is sufficient. Even then, precise temperature compensation is required. If better accuracy is needed, periodic recalibration is mandatory.

Although five calibration points were used to obtain sensor transform functions in this research, in practice three points are sufficient to define the quadratic calibration curve. In this case least squares fitting should be replaced by simple quadratic interpolation. However, it is recommended to repeat the measurement at each point several times and perform interpolation over average current values to improve the curve accuracy. It is also recommended to calibrate for as narrow pH range as possible, if it is known *a priori* that the large pH changes are not expected among the analyzed samples. For the range

of about 1 or 2 pH linear approximation of the calibration curve can be used without significant loss of accuracy.

The electric circuit for sensor signal readout also can be simplified, as two separate voltage sources are actually not necessary for proper sensor operation. A single control voltage (positive relative to the circuit common point) can be applied to the ISFETs bulk/source, with the potential of drain leads and reference electrode set to the common point (ground). In this configuration ISFETs still operate in saturation region (provided the drain and source diffusion resistance is low enough), and only two major components are needed for secondary transducer, namely the stable programmable voltage source and the microamperemeter (dual channel microamperemeter for differential measurement mode). The ability to control the voltage source with high precision is not required, but generated voltage must be stable in time and clear from noise. Likewise, the high absolute accuracy is not required from microamperemeter, but it must provide sufficient resolution depending on the sensor transconductance and pH sensitivity.

It is worth to emphasize that the use of direct current readout mode not only allows to simplify sensor design and secondary transducer schematic, but also allows to use the standard calibrated equipment with well-defined specifications for sensor characterization, as well as for calibration and measurement.

This opens the way to unification of the ISFET sensors application procedures regardless of the particular sensor design.

## 5. Conclusion

The results obtained in this research allow us to conclude that the investigated ISFET sensors, once properly calibrated, are suitable for application in the usual laboratory pH-metry tasks as well as for more specific biosensory applications, with the practically achievable measurement accuracy below 0.1 pH.

While the evaluated performance characteristics do not excel that of the high quality glass electrodes used with traditional pH measurement instrumentation, certain peculiarities of ISFET sensors make them attractive for application in specific areas, such as biosensory analysis of liquid and gaseous samples in food industry, environmental monitoring and medicine.

One of the obstacles that prevent widespread use of ISFET sensors is lack of standardization in primary sensors characteristics as well as in the sensor application protocols. As was shown in this work, it is possible to use the standard well-specified electrical equipment for sensors characterization, by extracting the pH transduction properties from the primary electrical characteristics of the transistors. Similar technique can be used for actual sensor exploitation, resulting in simpler design for both

primary and secondary transducer without compromising the system performance.

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## Digital Sensors and Sensor Systems: Practical Design

Sergey Y. Yurish



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