Passive, Wireless ECG-Sensor Based on Impedance Loaded SAW-Transponders

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Abstract— **A remotely requestable, passive short-range sensor network for measuring electrical potentials is presented. The sensor system is able to simultaneously monitor six voltages, and is intended to be used for Holter-electrocardiogram (ECG) and other biopotential monitoring, or in industrial applications. The sensors are based on a surface acoustic wave (SAW) delay line with voltage-dependent impedance loading on a reflector interdigital transducer (IDT). The load circuit impedance is varied by the measurand through the capacitance of a varactor. High resolution is achieved by developing an MOS-capacitor with a thin oxide and capacitance tuning at zero-voltage region, as well as a high-Q-microcoil by UV-LIGA-process. The circuit is constructed planar, and further miniaturization is achieved by using Flip-Chip technology. Simultaneous monitoring of multiple potentials is realized by time-division-multiplexing of different sensor signals.**

Keywords— **SAW-sensor; wireless; biopotential; long-term monitoring**

I. INTRODUCTION

The health care system in Western Countries is currently based on the healing of existing diseases. This system is expensive, and the costs could be decreased by developing preventive methods. Cardiovascular diseases are the main cause of death withing the population in the age of 44-64 years, and the second most frequent cause of death of people between 24 and 44 years. In Germany alone, about 300 000 people suffer annually from a heart attack. An early recognition of possible attack symptoms and warning the patient or the doctor would enable them to take preventive action to avoid the attack. This would reduce the risk of irreparable damage to the organs, or even death. To lower the number of heart attacks, preventive methods must be improved. Monitoring risk groups, such as people recently having a bypass or pacemaker implantation surgery done, has proven to be a good way to achieve results. The continuous monitoring should be performed in real time, since the threat of an attack often requires urgent action.

Long term recording of ECG is a standard procedure in current cardiac medicine, but the devices are not capable of performing real-time analysis due to restrictions in size and energy consumption of the recorder. A further problem of the current technology is that the devices are uncomfortable for the patients, limiting their mobility. The chest-electrodes are connected via cables to the recording unit, which must be worn on the body during all of the daily activities and sleep. If these cables between the electrodes and the recording unit were replaced by a radio link, the patient comfort would be increased.

In recent years there has been research for solving the problems of wireless data transmission [1, 2, 3] and for improving the signal evaluation [4, 5, 6]. The goal of the project TEDIANET is the development and evaluation of a telemetric diagnosis-network for monitoring risk-patients by using surface acoustic wave (SAW) transponder sensors. The network consists of three levels, as shown in Figure 1. The sensor is responsible for data acquisition at the patient. It is interrogated by a RF-impulse sent from a portable reader/recorder unit [7, 8], which is to remain in immediate vicinity such as pocket or nightstand of the patient. The data collected and pre-processed at the reader unit will be sent to a stationary analysis unit where the diagnostics takes place. The link between the reader unit and the analysis unit is flexible, using for example blue tooth in home-care setting, or GSM link in ambulatory situations. In this paper we focus on the conception and development of data-acquisition and transmission to the portable reader unit, marked with a circle in Figure 1. SAW sensors have been realized for various mechanical [9], physical [10, 11], and chemical [12, 13] parameters, but this is the first passive and wireless sensor under development specifically for medical applications and for measuring potentials.

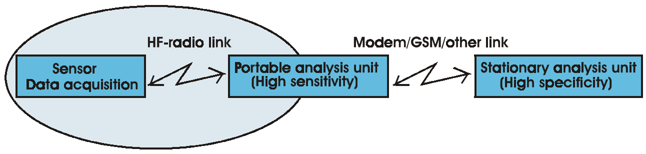

Figure 1 ECG-monitoring system with sensor, portable reader and stationary central analysis unit.

II. Sensor

A total set of three leads is usually required to obtain useful data of the heart function in a standard ambulatory Holter-monitoring, and 12 leads are used to get the full dataset. Our system can simultaneously monitor up to six leads. Each sensor consists of a SAW delay line, a load circuit, and electrodes attached to the patients chest, as illustrated in Figure 2.

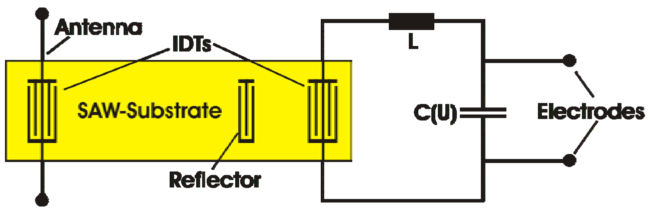

Figure 2 Design principle of the ECG-sensor.

To interrogate the sensor a short, high-frequency read impulse is sent out from the reader unit. The signal is captured at the antenna-IDT of the sensor, where it is converted to an acoustic signal, which then travels as a surface wave through the delay line to the reflector and to the load-IDT. At the reflector, part of the signal is reflected and delivers a reference-value used for calibrating the signal path. The rest of the signal travels to the load-IDT, where it is modulated as a result of the sensor function. The sensor effect relies on the variation of the complex impedance Z_{load} of the load circuit. At the instant of measurement the heart-potential has a certain value U, which determines the capacitance C(U) of the varactor diode. The impedance $Z_{load}(C(U))$ of the load circuit thus depends on the C(U), and causes the amplitude and phase of the signal to be modulated [9]. The scattering parameter S_{11} of the reflected signal can be estimated by formula (1), which is derived from the P-matrix formalism.

$$S_{11}(Z_{load}) = \frac{P_{13}^2}{P_{33} + \frac{1}{Z_{load}(C(U)) + Z_{match}}}, (1)$$

where P_{13} is the electroacoustic coupling factor of the IDT and P_{33} is the electrical admittance, and Z_{match} is the impedance component of the matching element L in Figure 2. The phase and amplitude modulated signal is then reflected back from the load IDT to the evaluation unit, where the evaluation electronics calculates the desired physical values.

A special feature about heart- and other biopotentials is that they are very small: An ECG-signal is in the range of a few millivolts and less. The change of impedance of the load circuit is limited by the capacitance ratio of the used varactor diode as well as the quality factor of the load-circuit. As the resolution of the system is critical because of the extremely low voltages, the system sensitivity can be increased by evaluating the phase modulation of the sensor-response. This way better signal to noise ratio (SNR) can be achieved [11]. The signal evaluation by phase modulation requires accurate matching of the resonant circuit to the driving frequency of the SAW-device.

Figure 3 shows the response of a single sensor. The partial wave reflected at the reflector can be seen in the time domain as a peak at 1,3 µs. The peak at 2,3 µs is the wave reflected at the load-IDT at a certain Z_{load}. Peaks caused by multiple reflections at the reflector and IDTs follow the two main peaks. Figure 4 combines the signals from all six sensors. It illustrates the importance of the reference peak. The response of different sensors varies according to the attenuation in the transmission path, so that for example the sensor number 2 has a higher attenuation than sensor 1 caused by for example longer distance from the sensor to the reader unit. To separate the signals from individual sensors, the reflector and the measurement IDT on each sensor are placed on the SAW-chip surface with a fixed increasing offset distance resulting in a time division multiplexed system response.

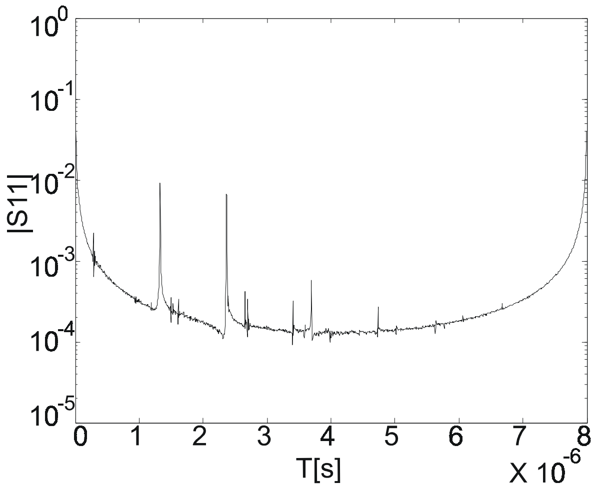

Figure 3 Response of first sensor.

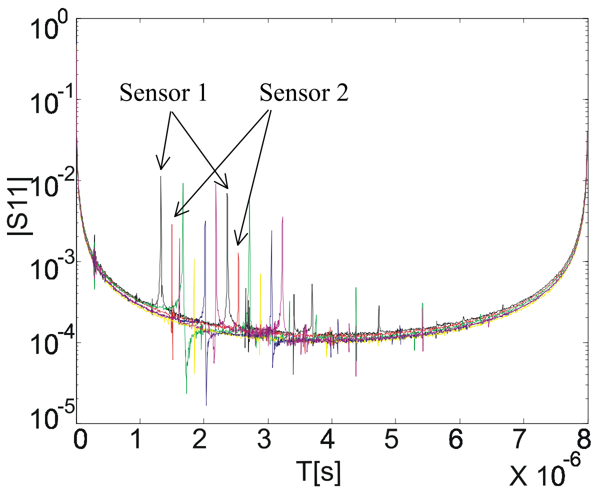

Figure 4 Response of all six sensors.

A. Varactor

The capacitor functions as the detuning element of the load circuit, where the capacitance is changed by the heart-potential. The capacitor must have a steep capacitance-voltage curve about the zero-voltage region, as well as a high Q-factor. There are several ways to select a capacitor, the most obvious being a Schottky-diode or an MOS-capacitor because of their inherent high Q-factors. Both have been tested for their suitability in the device. The capacitance of the MOS-diode is determined by the serial connection of the oxide capacitance and the capacitance of the space-charge region.

The capacitance depends on the thickness of the oxide, since at the maximum negative voltage in the accumulation region the differential capacitance of the semiconductor is very high, and the total capacitance is therefore at its maximum. As the voltage increases, a depletion region is formed at the semiconductor surface, and the total capacitance decreases, Figure 5. The minimum achievable capacitance is determined by the oxide thickness as well as the doping density [14].

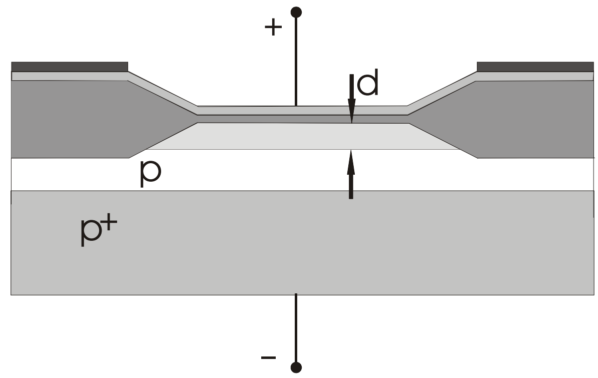

Figure 5 The capacitance of an MOS-diode depends on the width (d) of the space charge region as a function of the applied voltage.

We have developed a new type of MOS-capacitor with capacitance tuning at the zero-voltage region. The capacitor produced with a standard MOS-process has a very thin oxide-layer and a low flat band voltage, thus resulting in a high capacitance ratio [15]. The capacitance of the diode between –0.5 and 0.5 volts is shown in Figure 6.

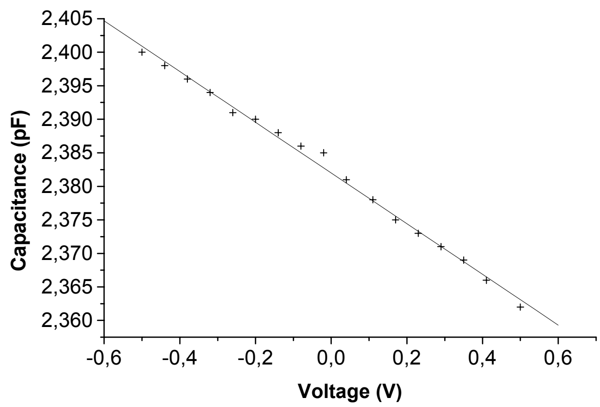

Figure 6 Capacitance-voltage curve of the developed MOS-capacitor.

The capacitance of a Schottky-diode also depends on the width of the depletion region, which is determined by the doping density and the applied voltage. A C-V-characteristic of a Schottky-diode of model MGR704 (Micrometrics, NH, USA) is shown in Figure 7. This diode type is evidently also well suited for the system, because of its high parallel resistance, high capacitance ratio and low capacitance.

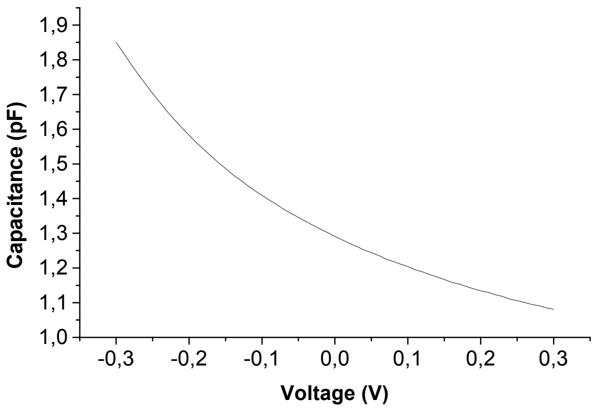

Figure 7 Capacitance-voltage curve of the developed MOS-capacitor.

B. Matching Circuit

To achieve the necessary resolution with the system, a very high Q-factor of the load circuit is required. This has been achieved by producing a microcoil with a high quality-factor by thick metal electroplating [16]. By constructing high structures with a desired geometry, the serial resistance of the coil can be reduced to a minimum. The dielectric losses are minimized by using glass or highly doped silicon as substrate material. The stray capacitances are further minimized by optimizing the geometry of the inductor.

The microcoils were manufactured with a UV-LIGA process, which is a low-cost, ultraviolet alternative to the standard LIGA-technique. The production comprises six steps, which are shown in Figure 8. Thin layers of tungsten and gold were deposited on the substrate for electrical connection. A 50 µm thick layer of photoresist was spread on the substrate, and patterned by photolitography to form a negative of the coil structure. Copper was electroplated into the structure to form the coil, and finally the resist and the thin metallization layer were removed.

The produced coils have smooth and vertical sidewalls. Figure 9 shows the photoresist before the electroplating of copper, and Figure 10 the finished copper coil. The height of the structure of these coils is 50 µm, and the line width is 25 µm.

C. Resistors

The conductors of the ECG measurement-electrodes are brought to the poles of the varactor, as can be seen in Figures 11 and 12. These conductors are long respective to the wavelength of the RF-signal, so that they form an antenna, which radiates power from the resonant circuit. This disturbs the measurement or may avert it completely. Therefore, thin-film resistors were integrated between the electrodes and the varactor. They are fabricated by reactively sputtering titanium oxynitride on the substrate and patterning with photolithography and etching. The resistance of the resistors must be high as compared to the impedance at the high-frequency side of the circuit. On the other hand, it must not be too high, because it reduces the sensitivity of the sensor by causing the measured voltage to drop at the capacitor. The resistivity of titanium oxynitride can be adjusted in a wide range by changing the amount of oxygen in the process, i.e. the stoichiometric composition of the deposited TiON layer, or by varying the deposition time or the geometry of the resistors.

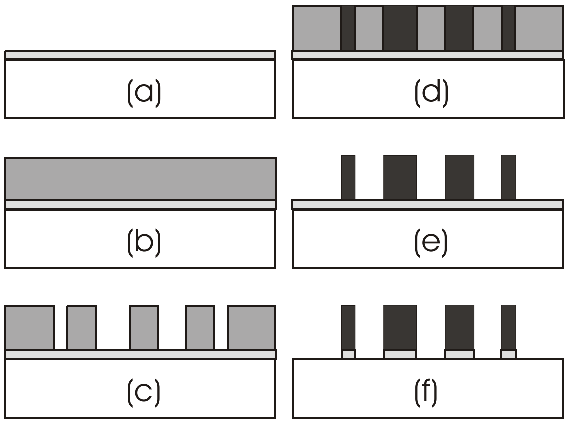

Figure 8 Coil manufacturing process. White: substrate, black: copper, light gray: tungsten and gold, dark gray: photoresist. (a) Deposition of conducting layer of tungsten and gold (b) Spin coating thick photoresist (c) Photolitography (d) Copper electroplating (e) Removal of resist (f) Removal of conducting layer.

III. INTEGRATION

Building an LC-resonant circuit at 2,4 GHz requires special attention to be paid on parasitic effects. The component values are very small in the range of 1pF or a few nH. We have built our circuit planar in order to minimize the parasitics, Figures 11 and 12.

Both of the varactors investigated are available in a chip-size package with dimensions of about W350 x L350 x H150 μm^3. They were attached on the substrate with a silver-filled adhesive because of the ease of the procedure as compared to reflow-soldering. The additional advantage of adhesive bonding is that the whole system can be assembled lead-free to be compatible with the EU-directions regarding lead-free electronics.

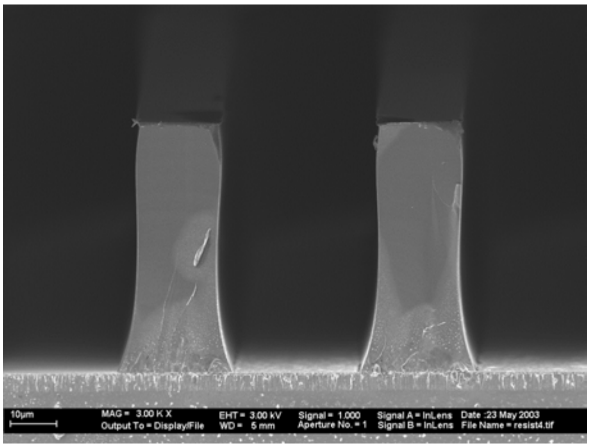

Figure 9 SEM image of the resist before copper electroplating.

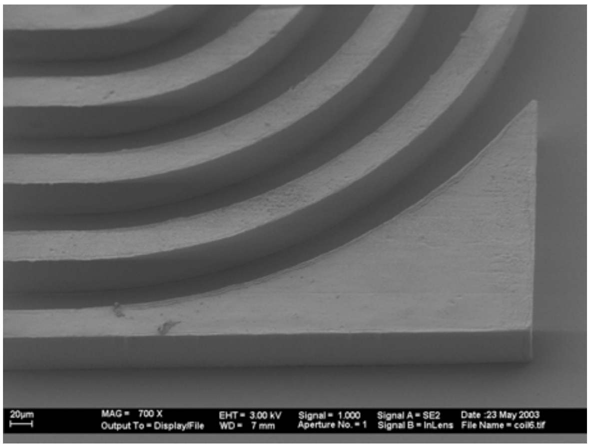

Figure 10 SEM image of a finished microcoil.

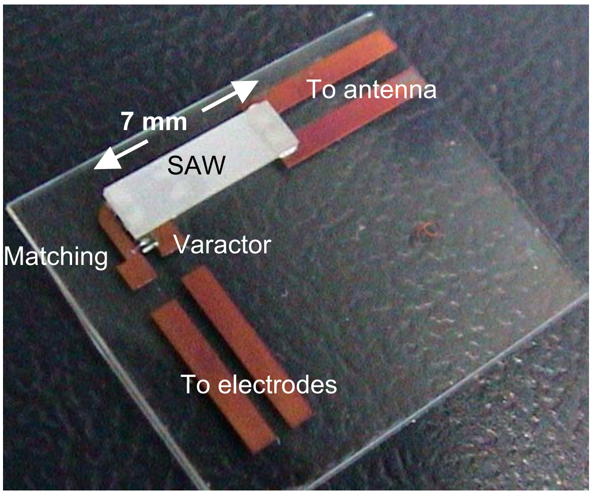

Figure 11 Sensor constructed on a ceramic substrate with electroplating and Flip-Chip technologies.

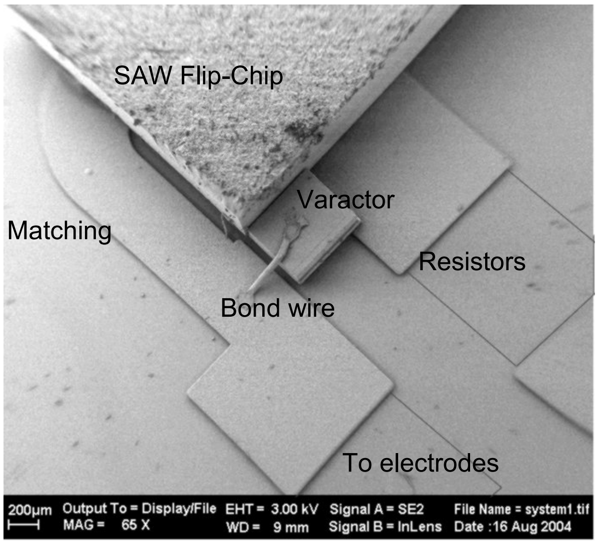

Figure 12 Detailed image of the sensor assembly.

The top-side of the varactor is connected with a bond-wire to the circuit. The length of the bond wire is less than 500 µm causing an inductance of about 0,5 nH. The rest of the required inductance, 2-3 nH, is the same as the matching circuit. This inductor is an electroplated wire loop connecting the two poles of the IDT.

The SAW-chip is connected with state of the art Flip-Chip technology. Today's standard technology is to wire-bond the IDTs to the substrate, but our system cannot tolerate the inductances of these bond-wires.

IV. Measurement Results

The sensor was tested for function. The amplitude of the reflected signal is carried over the load-circuit capacitance in Figure 13. At the resonant frequency the impedance of load circuit, and the attenuation of the reflected signal are at their minimum, as depicted in formula 1.

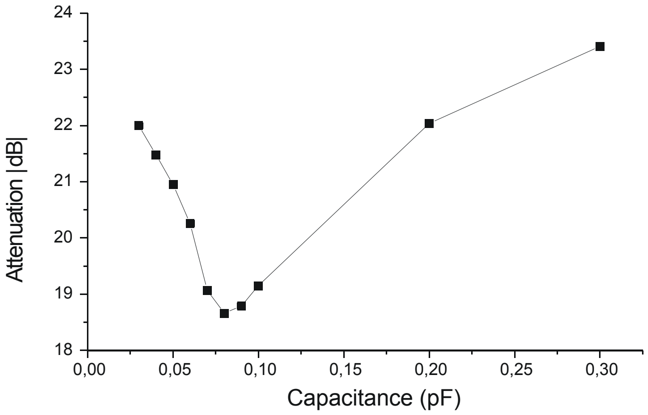

Figure 13 Sensor response to change of load-circuit impedance.

V. Future Work

In the present work the sensor function was tested with a cable connecting the read-device and the sensor. As can be seen in Figure 13, the dynamic range of the measured assembly is only a few decibel, which is not large enough to deliver the signal via an antenna to the reader unit for evaluation. The future goal is to improve the sensor response by reducing the real impedance of the load-circuit further, and thus to improve the resolution of the system.

VI. Conclusions

A concept of SAW transponder sensor for measurement of biopotentials such as ECG and other small voltages has been presented. The functioning of the system has been proven. The sensor effect relies on varying the impedance of the load circuit consisting of an inductance and a capacitance, which causes the reflected impulse to be modulated in phase and amplitude. The impedance is dependent on the heart potential at the instant of measurement through a varactor diode. Because of the scantiness of the measured signal and the limited capacitance ratio of the diode, the sensor requires an LC-load-circuit with a high Q-factor. An MOS-diode with capacitance tuning in the zero-voltage region has been developed. High-Q-microcoils have been manufactured by an UV-LIGA process. The separation of signals from different sensors is achieved by time-division multiplexing.

Acknowledgment

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