Aiming for the cloud - a study of implanted battery-free temperature sensors using NFC

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***Abstract*—In this paper we present results based on measurements of implantable devices which can be powered externally and communicated with using the near-field communication (NFC) infrastructure. NFC allows us to not have a dedicated gateway and intra-body communication to bridge the data from sensors to phone. In our trials, we have used commercially available sub-components and mounted them on a thin plastic with printed interconnections and coated them for bio-compatibility. Devices were implanted in porcine models during one week. We could during this time measure the in-vivo body temperature through skin and subcutaneous tissue ranging in thickness from some mm to a couple of cm. The implanted sensor devices are mounted on thin, printed-electronics plastic sheets where the coils and conductors are designed with different types of materials. The choice of materials is done in order to offer a low-cost solution to read out data from in-vivo sensors. We compile measured data, practical results and guidelines, together with theoretical results referring to the design of the implanted inductive NFC coil as well as the energy transfer from one mobile device to another.**

I. Introduction

The billions and billions of electronic devices that within a few decades will be interconnected will require reliable energy sources. Smaller devices (mobile phones, and similar aside) offer no or little chances for regular battery replacements. Instead we have to rely on energy harvesting principles, either through dedicated actions (such as winding, shaking, solar panels, etc.) to charge the batteries or on ambient energy around us through thermal variations, vibrations, radio "pollution", etc. Inductive power transfer to implanted, medical devices is nothing new. Quite early in the development of pacemakers there were means to charge the batteries from the outside of the body without having to replace the batteries through surgery. Schuder, et al., in 1971 [1] had also refined the system and demonstrated that very high power levels (1 kW!) could be transferred through tissue without significant raise of temperature or other harmful effects. The power is transferred wirelessly through inductive coupling between two coils placed close to each other. The applications mature and the demand increases [2] [3]. The intelligent pill has also been suggested [4], and is actively used in diagnostics today, as a monitoring device that could be ingested and communicated with, through the digestive system. Wirelessly, devices can be powered and communicated with, through an inductive coil. The Qi standard has recently matured as a standard to wirelessly (inductively) charge mobile devices by placing them on top of a charger containing a coil. The standard also, as a mean of safety, includes handshaking to set power levels at the most suitable level for optimum efficiency. Qi is aimed at transferring higher amounts of power but the near-field communication (NFC) standard (ISO/IEC 18000-3:2010) is however also a candidate to perform similar task. For implanted sensors where the power levels are not that high, NFC may provide sufficient energy and at the same time be used as a means of communicating with the sensor, to read out data. Data is therefore captured and read out "on demand" and not stored in the background. NFC is readily available in many mobile phones already today and offers means for clinical staff - or patients - to easily read out data from sensors capturing medical information and send to care provider during for example post-operative care.

We have investigated the use of thin plastics on which we mount active components (integrated circuits) and passive structures, mainly coils. We have developed two generations, one for porcine models (larger) and one for rats which can be slided in under e.g. the skin on the hind leg. For the porcine model we read out data manually by applying a mobile phone close to the implanted device. For rats, the data will be read out automatically once they pass through a special gate. These experiments will be commenced during the end of 2016. For optimum power transfer the coil that picks up the inductively transferred energy must be carefully designed with respect to resistive, inductive and capacitive components. The quality factor, the Q-factor, of the coil should be high, implying low resistance, implying short total length of the coil, etc. In order to implant the device, it also have to have a small area. These parameters are all contradicting each other and finding a near-optimum design is a non-trivial task.

II. Background and theory

The bionic human has been part of, if not other things, popular culture for quite some time and obtaining immortality has been appealing. It does not have to be that dramatic though. With bionic ideas we can help the body to measure, recover, and sustain given that it has been damaged through injury or other critical impacts. It has already been widely applied: for example pacemakers, diabetes injectors, glucose sensors, and other types of shunts. Within the surgical field, there is a great demand for safe and reliable post-operative monitoring of tissue. Small, bio-monitoring sensors could here be used instead of larger, more expensive equipment.

Most existing systems for bio-monitoring are using equipment developed for a highly specific demand. This makes both the development costly and the final system expensive. The use of standardized components will radically lower these

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costs and thereby contribute to a safer and more effective health care that can be afforded by more patients even in less developed countries. In our research we are investigating these opportunities. In the work presented here, we use a commercially available IC which offers a battery-free, built-in temperature sensors. When a battery is not present, the chip can still be powered through the NFC interface and also modulate, back-scatter, the impedance such that information can be transmitted. In our experiments we have not focused on optimizing the NFC circuitry as such, but rather only the inductive coil and plastic offering a smooth sensor that can be implanted easily and offers bio-compatibility. By investigating this method and how well it works in an implanted model we can pave the way forward for more advanced solutions.

A. *Inductive coils and efficient power transfer*

A symmetrical, circular planar coil (with turns in only one "layer") will have an approximate inductance [5]

$$L \approx n^2 \cdot R^2/(30R - 22r) \tag{1}$$

where the outer radius of the coil, R, is determined by the inner radius, r, and the width w of and spacing s between the coil inductors, as well as the number of turns n,

$$R \approx r + n(w + s)/2. \tag{2}$$

The conductor width is assumed constant. Examples of the planar coils used in our work are presented in Fig. 1 for different number of turns, conductor widths and other physical parameters. Communication ports and access for other possible auxiliary components are placed in the center of the coils in various sizes. The conductive material is silver ink. The wiring is for measurement purposes. The number of windings n range from 3 to 7, and wire width from 250 to 1000 μm. In our process the plastic tags were printed with a DuPont 5000 (120-34) and were dried at 130 degrees for four minutes. The thickness of the wire is around 16 to 18 um and resistivity approximately 30 mOhm/square. If the wire width is 0.1 mm and the spacing is 0.1 mm, the total maximum inductance in an square-inch area, $R = 25.4$ mm (and an ASIC in the center, i.e., sufficient inner radius r), is approximately 28 μH. The number of turns will be approximately 37 but the total conductive length becomes approximately 2 metres. Given the ink we use above, the total resistance would be approximately $R_{tot} = 20000 \cdot 0.03 = 600$ Ohms. A high resistance will directly influence the quality (Q factor) of the inductor:

$$Q = \omega_0 \cdot L/R_{tot}. \tag{3}$$

Assuming that the frequency of interest (for NFC it is 13.56 MHz) we get an approximate Q factor to

$$Q = 82.5 \cdot 10^6 \cdot 28 \cdot 10^{-6}/600 \approx 4 \tag{4}$$

in our example, which is a rather low value. Further on, the resonance frequency of the inductor, i.e., dependent on the total capacitance C_p loading the inductor through external components and through its own physical appearance should preferably be tuned to the center frequency of interest, i.e., 13.56 MHz. This implies that, in our example,

$$C_p = 1/L \cdot (2\pi f)^2 \approx 5.4pF. \tag{5}$$

The larger the inductance is the less tolerant we will be against parasitic capacitance. From a design perspective one can aim for lower inductance and insert additional capacitance. Combining the equations also conclude that the Q increases with increased conductor width and fixed spacing. It will also decrease with fixed width and increasing distance. The higher the Q value, the higher amount of energy will be transferred to the tag. It can be shown that the power transfer efficiency is dependent on the geometric average of the Q factor of the transmitting device and the Q factor of the receiver [6]. Since the Q factor of the read-out device (a commercially available smart phone) is fixed, the target is to maximize the Q given the one square inch area we have at hand.

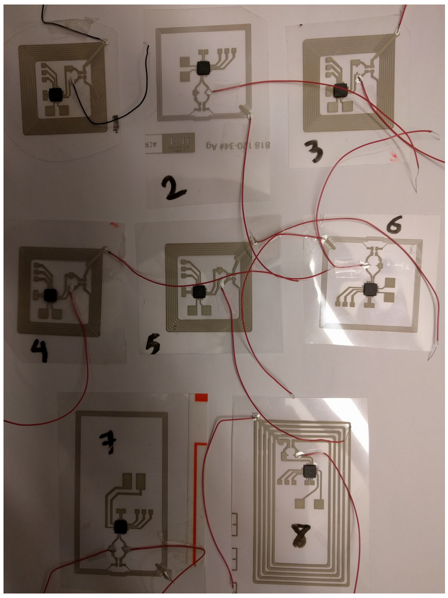

Fig. 1. Example the tags implemented on plastics (with wires for measuring the coil inductance). The number of windings n range from 3 to 9, and wire width from 550 to 1000 μm.

B. *Substrate and biocompatibility*

A flexible substrate is required to be implanted with a minimum impact on the tissue and the components cannot be too large, both due to aesthetic reasons and to regrowth of tissue. We have chosen to keep the components less than one inch in size. In order to allow implantation two biocompatible materials are explored in our experiments. For the porcine model we used silicon that was manually coated on the tags. For the rodent model we will use parylene which will be vaporized on the tags. The silicon coating will be thicker and prevent the tag from bend, and will, in the case of being implanted in a patient, more obstructive. Both these materials have been widely used for implantable sensors and could be described as fully biocompatible [7]–[9].

III. Method

We have conducted a theoretical study together with simulations of different coil parameters, such as area, wire width, and size. Various sizes were then implemented on plastic using silver ink for measurements. Less were implanted due to ethical reasons. The coils were experimentally verified to verify that theoretical and experimental results. To derive the coil inductance we applied two mathematical methods based

TABLE I. CALCULATED, MEASURED AND SIMULATED INDUCTANCE.

Tag	Turns	Width μm	Space μm	Shape	Calc 1 μH	Calc 2 μH	Sim μH	Meas. μH
1	9	550	250	Sq	4.4	4.4	7.8	4.6
2	5	250	250	Sq	2.3	2.6	2.9	2.9
3	9	550	250	Sq	4.4	4.4	7.8	5.1
4,5	7	550	250	Sq	3.2	3.3	4.7	3.5
6	5	250	250	Sq	2.3	2.7	3.2	3.2
7	5	250	1000	Rect	-	-	5.1	7.3
8	5	1000	1000	Rect	-	-	3.4	2.2

on the equations in the previous section. ADS was used to simulate the inductance of the structure based on the layout. The commercially available sensor chips were from AMS, SL13A, [10] for which a default coil design was provided (credit-card size and five turns with copper conductive traces).

Examples of the tags used in the study are shown in Fig. 1. Different number of turns, shapes, and diameters were tested on a transparent 150-μm thick plastic. For bench tests, a Texas Instruments kit was used to evaluate the tags with respect to e.g. distances, configuration, and dielectricum (influences the magnetic coupling coefficient), i.e., not a phone. As a quality measure we used the received signal strenght indicator (RSSI) value as detected by the Texas Instrument kit. We also manually measured the inductance in the coils and approximated the capacitance by searching for the resonance frequency.

The sensors were used in a parallel flap study in a porcine model [11] to monitor temperature. The animal test was conducted on mixed breed pigs (Swedish landrace pigs). The advantage of the porcine model is the similarity between human and porcine tissue. The flaps were raised on the buttock of the pig including skin, subcutis and muscle fascia. The average thickness of the flap was 10 mm. The sensor was sutured to the underlying tissue and left in place after the surgery for four days before the pigs were euthanized. The pigs showed no additional discomfort from the sensors and there were no sign of irritation or inflammation in the adjacent tissue. For data read out from the implanted devices, we used a Nexus 5X phone and an android app. The phone has a 2-by-2 cm large coil placed around its camera.

IV. RESULTS

To verify the simulated results against the mathematical models we swept the parameters of the coils (number of turns, widths, and spacing) and measured the power efficiency by estimating the Q value and coupling given an fixed transmitter coil. A 3D field solver tool was used. We could verify that, as predicted, the Q factor and efficiency decreased with increased conductor width as shown in Fig. 2. This verified that minimum distance is attractive. Simulations also show an increased efficiency (Fig. 3) with the width of the coil conductor for a fixed separation. Notice, however, that a maximum is obtained beyond which the efficiency decreases due to too large coils and too much capacitance. The results indicate an optimum point at around 550 μm which was used for printing the tags. Given the simulated results, we designed tags (Fig. 1) with different configurations and measured them. Table I compiles the results and some values deviate within the same order of magnitude. Error sources are typically the approximate estimates of the dielectricum.

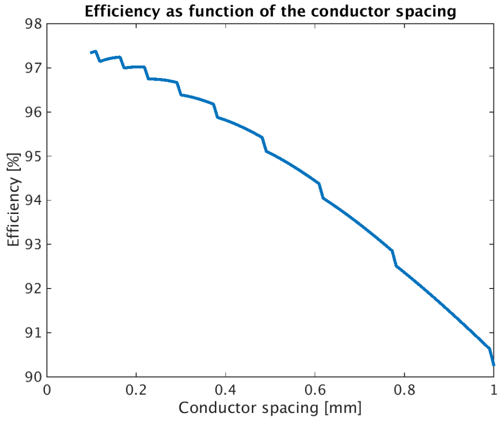

Fig. 2. Simulated efficiency as function of spacing between coil wires.

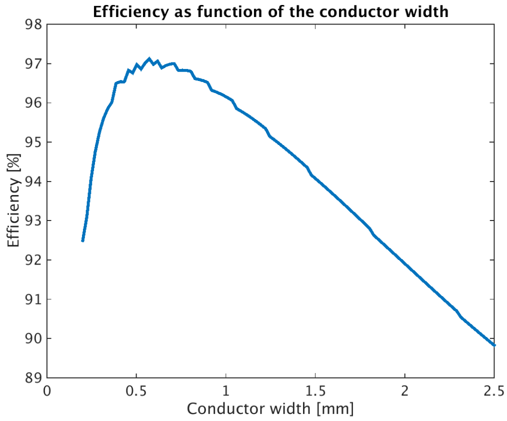

Fig. 3. Simulated efficiency as function of width of coil wires.

To evaluate how efficient the designs were we conducted experiments where the received signal strength indicator (RSSI) of the power transmitter (listening to the echo from the tag) was recorded as function of distance, decoupling, and tissue between reader and tag. The results are compiled in [12], and an excerpt of the results are shown in Figs. 4 and 5. For the AMS pre-designed board (Fig. 4) we see that we can communicate up to 10 cm in open air. Once the RSSI is significantly low (below 60dB) the communication is lost. With a 6-mm thick slice of meat, mimicking living tissue, between the devices we see that the maximum distance decreases by some 4cm to 5.5cm. Using an additional decoupling capacitor on the AMS tag, the distance could be increased, since more energy could be stored locally. Similar experimental results, but with our plastic tags without storage capacitors, are shown in Fig. 5. The tags with most turns, with widths closest to the optimum point of Fig. 3, and minimum spacing provides the longest distance. The combined results indicate that tags of type 1 should be tried. They provide the longest distance and would enable us to read out the temperature data from the porcine models. In our trials, we were able to implant and measure on two pigs. The sensors were implanted on the outside of the back leg and read out with a phone (Fig. 6). The location of the implant was not indicated on the skin, but was easy to locate with the Nexus phone. The android app reported the temperature (slightly obscured due to reflections).

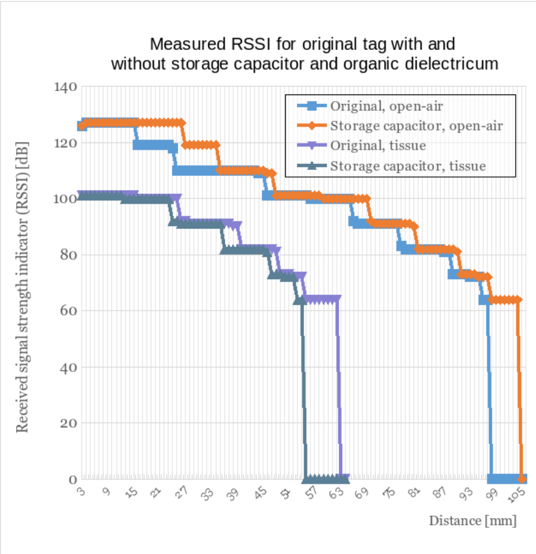

Fig. 4. Measured reception quality between commercial tag and reader as function of distance, dielectricum and decoupling.

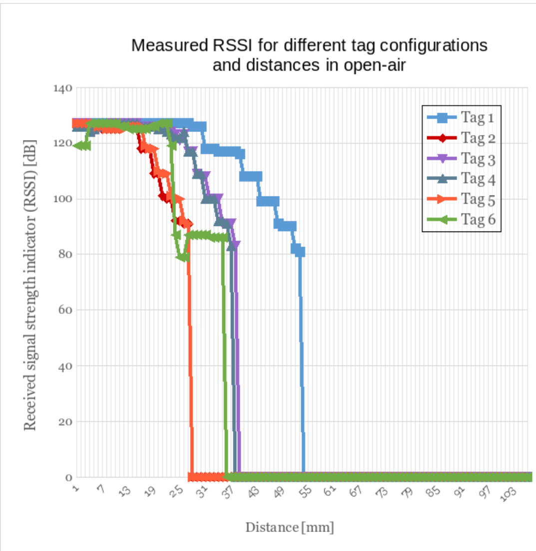

Fig. 5. Measured communication quality between developed tag and reader as function of distance.

Fig. 6. Example of the user scenario where the temperature can be read out.

V. Conclusions and Discussion

The goal of the study is to build a platform for monitoring a larger set of different biological parameters. To start this activity we have performed a theoretical study and implemented components that can be implanted, capture data and be read out from outside the body. Measurements on the flaps in the animal model were done with an android powered phone with a standard NFC interface. The system could easily identify and read the sensor hidden underneath approximately 10 mm of skin, fat and muscle fascia. This further strengthens the concept that it is possible to use ”off the shelves” measurement equipment, i.e., smart-phones for advanced biomedical monitoring. We are confident that we will be able to further improve the energy efficiency of the system and there by the penetrance of the signal in our coming studies. We are now preparing a study on rats where we will use a similar technology but with longer survival of the animals. This will give us an opportunity to evaluate the long time durability of the equipment and effect on living tissue.

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