RFID Technology for Continuous Monitoring of Physiological Signals in Small Animals

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***Abstract*—Telemetry systems enable researchers to continuously monitor physiological signals in unrestrained, freely moving small rodents. Drawbacks of common systems are limited operation time, the need to house the animals separately, and the necessity of a stable communication link. Furthermore, the costs of the typically proprietary telemetry systems reduce the acceptance. The aim of this paper is to introduce a low-cost telemetry system based on common radio frequency identification technology optimized for battery-independent operational time, good reusability, and flexibility. The presented implant is equipped with sensors to measure electrocardiogram, arterial blood pressure, and body temperature. The biological signals are transmitted as digital data streams. The device is able of monitoring several freely moving animals housed in groups with a single reader station. The modular concept of the system significantly reduces the costs to monitor multiple physiological functions and refining procedures in preclinical research.**

***Index Terms*—Biotelemetry, implant, *in-vivo* monitoring, radio frequency identification (RFID), sensor system.**

I. Introduction

PRECLINICAL biomedical research has the need for animal models in pharmacological studies to obtain first results on drug efficacy during drug development. In Germany alone, around 1.5 million rodents are used for medical research each year [1]. Evaluation of intended and side effects in pharmaceutical studies requires continuous monitoring of physiological signals, but intermittent manual measurements need special setups [2] and cause inconvenience and stress to the animals, which changes vital parameters like heart rate and blood pressure, consequently no longer reflect the normal physiological situation [3].

Biotelemetry systems represent an alternative to manual handling and measurements [3]–[5]. Even though the animals have to undergo an implantation procedure, healing takes only several days. During chronic long-term studies, the handling and total stress of the animals is reduced to a minimum, which will not only affect the wellbeing of the animals but also statistical variability. In using telemetric devices, the number of animals needed could be reduced, which is one of the principles of the ethical use of animals (3R concept).

However, common biotelemetry systems use either batteries [6], which limit the operation time, or additional antennas for inductive power supply [7], [8]. The communication protocols are often proprietary, and it is therefore not possible to use off-the-shelf components. Thus, even commercially available monitoring systems are costly.

In the last decades, radio frequency identification (RFID) has been introduced in many industrial fields, and low-cost devices have been developed to monitor consumer goods. Passive RFID transponders have become more and more relevant in industrial farming [9] and veterinary science [10], [11], where it is used to identify animals and is also able to store a small amount of data. Particularly, it is possible to meet tracking directives in an automated manner. The technology became widespread and widely accepted [12].

The aim of our investigations was the development of a low-cost system using RFID technology as an alternative to conventional biotelemetry systems. We developed an implant for rodents that is able to measure ECG, arterial blood pressure, and body temperature.

II. Materials and Methods

A. System Concept

The concept is focused on increasing the efficiency for laboratories with dozens of subjects as well as on cost reduction. Therefore, each animal is equipped with one telemetrical implant, which is designed for good reusability. Single animals can be housed in euro standard type III cages with a floor area of 825 cm^2, but for social animals like rats group-housing of up to five subjects in type IV cages (1815 cm^2) is preferable. Each cage is equipped with one ISO15693 [13] compatible reader device with its antenna placed below the cage.

The RFID reader inductively powers the implants and provides a digital communication link with a carrier frequency of 13.56 MHz and a data rate of approximately 26 kbps. It is bidirectional, block-oriented, secured by checksums, and able to handle multiple transponders. Using the standard RFID

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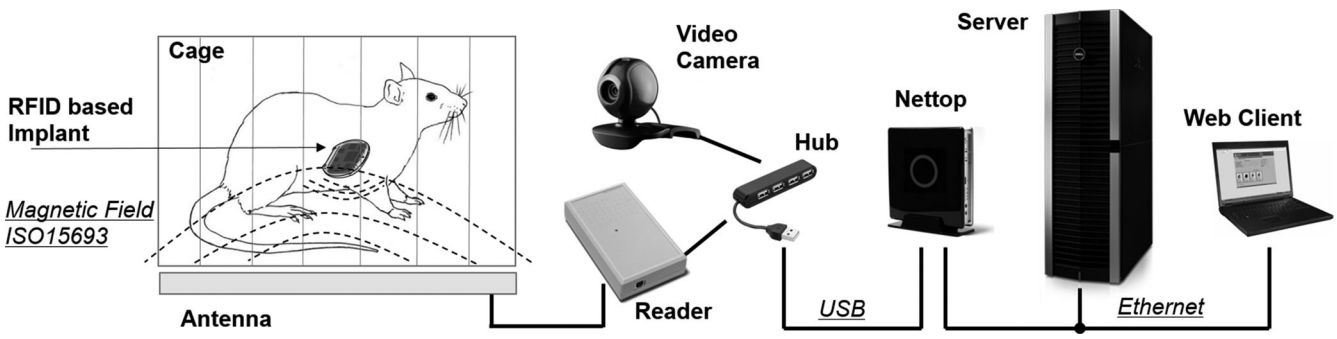

Fig. 1. System overview of a multidevice monitoring system. Up to four animals can be monitored with one RFID reader device. Several RFID reader stations and video camera(s) are connected to a local workstation (nettop). A client–server topology has been setup, which connects the local workstation to several web clients as well as to a central data and application server.

specifications, it is possible to use commercial off-the-shelf components and common interfaces.

These benefits are used to integrate implants into a modern and easily scalable client–server topology [14]. In the system depicted in Fig. 1, the reader operates like a gateway and is connected via universal serial bus (USB) to a local workstation. Its function is to control the communication of multiple readers as well as additional USB devices, e.g., a webcam to monitor the behavior of the animals. In large-scale systems, it is possible to use more than one station.

Using a local area network, each station communicates with the central data and application server, which uses a LAMP (Linux, Apache, MySQL, PHP) stack. All results are written via ODBC into a MySQL database (Oracle Corporation, USA). Commands to control the acquisition process are also held in this database, and it is possible to access the data without using a second communication protocol. Consequently, the database provides a bidirectional communication link between local workstation and server.

A customized software suite, installed on the server, is also able to access the database. The software interprets and provides the data in HTML language format, which can be accessed from most web-enabling devices using a web browser.

B. Implant

The implant to measure ECG, arterial blood pressure, and body temperature is the only proprietary element in the system. It communicates like a common RFID transponder, but implements several concepts, which allow autonomous operations and processing of physiological signals. In order to reduce the costs, the implant is designed to last for at least five consecutive uses in different animals. The implant is wirelessly powered to avoid the operation time limitations caused by power consumption. Supply interruptions, which happen if the implants leave the magnetic field or stay in an unfavorable position, are bridged by using an electricity storage device, which is charged inside the field, in a so-called semipassive operation mode [15].

To guarantee good reusability, it is possible to separate the sensors, which are directly attached to the animal's body and take physiological signals at the desired areas, from the electronics for signal processing and communication. Due to the different requirements, the so-called sensor module is described separately in Section II-C. The following section focuses on the reusable part of the implant. It presents the power supply as well as the necessary steps to process physiological signals from plug-in sensors for RFID access. Furthermore, the implementation, casting, and electronic design are shown.

1) Electronics: The electronics of the implant are built in a discrete manner. Only the RFID communication is integrated in an application-specified integrated circuit (ASIC) [16] using AMIS 0.35-μm technology (ON Semi 0.35-μm CMOS Co35M logic and mixed technology). Besides digital logic, the device includes analog circuits. It contains an envelope detector to demodulate the data signal, which uses an amplitude shift keying, from the reader. Another circuit generates a clock signal out of the carrier frequency of the magnetic field. Power transistors protect the circuit against electrical surges and handle load modulation for implant to reader communication. In addition, a power rectifier generates the necessary direct current to drive all other components on the implant.

The ASIC is placed into the implant's electronics via chip-on-board technology or is bonded in a QFN (quad flat no lead) housing with 32 pins. The antenna geometry is part of the design of the printed circuit board (PCB) and consists of two wires around the circuit on each of the four layers of the PCB. In detail, it is a 29 $\times$ 21.5 mm rectangular PCB antenna with eight turns, which provides an inductance of 4.26 μH. The rest of the electronic parts are common components. The circuit is organized as a microcontroller system, as illustrated in Fig. 2. Apart from the RFID front-end and antenna, it consists of the following:

- a common microcontroller, Atmel ATMEGA165PV, equipped with analog-to-digital-converter (ADC) inputs, a serial peripheral interface (SPI), as well as a real-time clock (RTC) circuit able to drive a tuning fork crystal oscillator;
- a power management system including an electricity storage device lasting several hours to also work outside the charging field, as well as regulation circuits;
- a storage device, Microchip 25AA1024, to buffer 50 measurements;

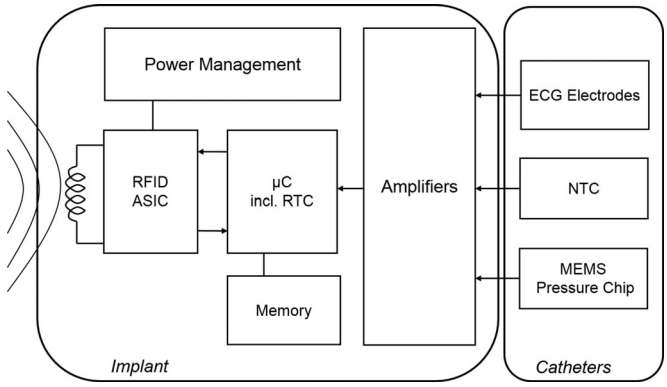

Fig. 2. Design of the hardware architecture of the implant with wireless energy supply and data transmission via RFID. The implant's electronics (left block) can be connected to versatile sensors (right block) to monitor ECG, temperature (NTC), and blood pressure.

- analog amplifiers optimized for ECG, arterial blood pressure, and temperature sensing.

Further details of the power management system are described in Section II-B2. The design of the analog amplifiers as well as the function of the remaining circuit is provided in Section II-B3.

2) Power Concept: Common RFID systems are classified into active, including a battery, or passive, without a battery [17], as well as semiactive, where active transponders operate passively inside the field to spare battery life [18]. The presented powering concept of powering the device passively inside the field and buffering energy for times without sufficient field can, therefore, be described as semipassive.

To implement the concept, the power, which is induced into the antenna, is transferred into an electricity storage device from which the implant is powered. A secondary battery, Panasonic VL1220, is used, but other technologies are suitable as well. The supply voltage of the electronics and sensors of the implant is 2.6 V and is generated by a low drop-out regulator using the voltage of the storage device as input.

To charge the storage device, the characteristics of an *LC* circuit are used, which is formed by the implant antenna together with an additional capacitor to obtain resonance at 13.56 MHz. Inside the magnetic field the voltage across the antenna rises until a power transfer is established. The voltage is converted to a rectified voltage by the internal rectifiers of the RFID front-end, which is directly connected to the storage device. A shunt regulator circuit stabilizes the voltage on the storage device to a value of 3.55 V. Therefore, a comparator, MAX9117, switches to an extra load, approximately 5 mA. An additional current limiter is not necessary in this design, because under normal conditions the power source is not able to generate critical current levels.

3) Signal Processing: To use the benefits of a semipassive power supply and to simplify the integration into the RFID system, the signal processing is organized as an autonomous data logger. A so-called profile, with the configuration of the three implemented sensor channels and measurement schedule, is programmed into the memory. Without a communication link, the implant is therefore able to measure signals independently and keep the results in the integrated storage. A profile provides the opportunity to scan, for example, the temperature channel each hour and the ECG each quarter-hour or to set the device to increase the number of samples at a specific time of interest. For a nearly continuous measurement, samples can be taken almost directly after each other. A small operation system running on the microcontroller executes the profile like a program using a RTC timer tick.

To set up the profile as well as for data exchange, only RFID is used. The communication is implemented as with common RFID transponders. Only a few extra commands to control the acquisition flow, e.g., "start" or "stop Logging" as well as "initialize," are additionally added as custom commands according to the standard. The data are transferred block-orientated using "read" or "write multiple blocks" commands. Controlled by the microcontroller, the commands directly access the memory, which is organized by a flexible file system [19], [20]. All data are kept in files together with additional calibration information and checksums.

The physical as well as parts of the data link layer of the RFID communication are implemented using the RFID front-end and the microcontroller, which are connected via a proprietary bus system. The RFID front-end modulates and demodulates signals from the magnetic field. The higher layers of the open systems interconnection (OSI) model are implemented as a program on the microcontroller. The concept follows the philosophy: real-time accesses are processed in the digital part of the front-end, whereas slower protocol parts are implemented in software.

Besides the communication, the microcontroller also controls the acquisition of the three sensor channels. According to the profile, the microcontroller powers the analog amplifiers as well as the sensors and captures the resulting signals using the integrated 10-bit ADC. It implements a single measurement for the temperature channel as well as 477-Hz synchronous sampling for the ECG and the pressure channel.

The channels transform signals provided by the sensors from low amplitude and high impedance to high amplitude and low impedance. More precise, the signals of the probes, described in Section II-C, are matched to the input voltage range of the used ADC, which is 2.6 V. For this purpose, the transponder carries four operational amplifiers, TI OPA369, as well as two instrumental amplifiers, TI INA321.

Temperature and pressure are amplified by a fixed value. The temperature channel is processed single-ended by an operational amplifier with an amplification factor of four. An instrumental amplifier processes the pressure signal with an amplification factor of 110.

The ECG channel is set to be used for rats with heart rates of approximately 300 beats/min and signals of around 0.5 mV amplitude using subcutaneous electrodes. The body potential, i.e., the right leg lead, is set to half of the analog supply voltage instead of ground. An instrumental amplifier is used as an input circuit. Using two operational amplifiers, the signal is filtered for a frequency range from 2 to 150 Hz and amplified by a factor of 770. The channel allows a resolution of 3.4 μV.

4) Biocompatible Sealing: The implant is designed to operate several weeks inside the body. Therefore, this paper presents a cost-efficient device casing, which shows less interference with

TABLE I
SENSOR CONFIGURATIONS

Configuration	ECG Cables	ECG Electrodes	Pressure
A	copper on polyimide (PCB)	electroplated gold	invasive with A. abdominalis bypass
B	copper on polyimide (PCB)	electroplated gold	noninvasive cuff-like attached to A. abdominalis wall
C	copper on polyimide (PCB)	steel 1.4435 with hooks	invasive fluid-filled catheter in A. femoralis
D	helical nickel-copper wires	steel 1.4435 with hooks	invasive fluid-filled catheter in A. femoralis

the magnetic field than metal casing. The circuit board is cast in epoxide (EPOTEK 301, Epoxy Technology Inc., USA). Only the outer shape is defined by a mold with a size of $32 \times 36 \times 5.5$ mm, which is produced with an Objet 3-D printer (Stratasys Ltd., USA). The inner space is filled with the casting material. For a good biocompatibility, the surface is finished by coating it with an additional epoxide layer as well as long-time-compatible silicon (MED-6605, NuSil Technology LLC, USA).

Only the connector for the sensor module is kept open. In a final step, the implant is equipped with a sensor module, which is plugged into the device and fixed and sealed using room temperature vulcanization silicone (MED-2000, NuSil Technology LLC, USA), which provides mechanical fixation and isolation. The whole unit is disinfected using per-acetic acid (Wolfasteril E400, Kesla Hygiene AG, Germany).

C. Sensors and Electrodes

For better reusability of the device, the ECG electrodes and sensors to measure arterial blood pressure and body temperature are separated from the major implant to make them disposable and significantly reduce the costs. Flexible wires make it possible to place the sensors according to the researchers needs. A flexible polyimide circuit board [21] combines sensors and wires and provides an interface to the implant. Calibration and testing is done separately. The concept allows the use of different sensor and electrode configurations.

1) Electrics: The electrics consist of the following:

- a resistor with negative temperature coefficient (NTC), Panasonic ERT-JZEV104H;
- a miniaturized pressure cell, BCM SE101;
- several calibration resistors;
- three ECG electrodes.

Except for the electrodes, the components are mounted on the circuit board by a reflow solder process. Gold bumps on the contacts of the pressure cell, in bare die format, make aluminum pads solderable and allow flip-chip bonding. It is possible to populate, test, as well as to do further production steps in batches. A separation is only necessary during the final steps of housing.

The NTC resistor is used as a temperature sensor. Together with a reference resistor, it is configured as a voltage divider, which is adjusted to a defined temperature of 36 °C. The signal is the voltage difference between current value and virtual ground. In the current setup, a range from 27 °C to 42 °C is covered.

The pressure is measured by a miniaturized pressure cell. The cell includes a resistive Wheatstone bridge and provides an absolute pressure value. An additional calibration resistor sets the bridge voltage to zero at 875 mmHg. (The unit mmHg is used for blood pressure in most countries. For transparency, all pressure values are given in this unit instead of SI standard unit Pascal [1 mmHg = 1 torr = 1.33 mbar = 133 Pa)]. After amplification and sampling, the sensor covers an absolute pressure range between 650 and 1100 mmHg with a resolution of 0.4 mmHg. To obtain the physiological parameter, the signal is corrected with respect to the ambient pressure, which is measured by an external device outside the cage.

For ECG, three electrodes are used, two for a bipolar lead with one reference electrode. The electrodes are metal contacts (see Table I), which are directly attached to the muscle surface. For ECG as well as pressure measurement, different configurations can be used.

2) Configurations: Based on the populated PCB four configurations, A, B, C, and D, with different ECG electrodes and cables as well as different pressure measurement methods were tested. The specifications are listed in Table I. The sensor is always located at the connecting interface.

For the ECG electrodes galvanized gold (configuration A, B; see Table I) as well as 1.4435 steel (configuration C, D; see Table I) is used. The steel electrodes are glued on the gold contacts using electrical conductive epoxide (TDS CW2400, ITW Chemtronics, USA). Additional hooks optimize the fixation to the surface. The connection between the electrodes and implant is prepared by the polyimide PCB or in the case of configuration D by helical nickel–copper wires (Osypka AG, Germany).

The pressure cell is encapsulated using a biocompatible glop top material (EPOTEK 320, Epoxy Technology Inc., USA). An additional silicone layer on the sensor membrane (MED-6605, NuSil Technology LLC, USA) keeps the membrane clear and provides chemical as well as electrical isolation.

The resulting pressure module is used to implement three different approaches. Configuration B implements a noninvasive concept using a cuff, which is formed by the isolating silicone layer. Two other approaches apply invasive methods. In configuration A, the pressure sensor is mounted over a hole in a heparized polyurethane tube (CBAS-C30, 3F, Solomon Scientific, USA), which serves as a bypass, providing direct contact between sensor and bloodstream. The configuration C and D use a paraffin-filled CBAS-C30 tube with a heparin-containing tip transferring pressure from the open end in the bloodstream to the sensor [22].

D. In-Vivo Study

Seven chronic experiments were performed with different configurations of the implant (see Table II) in compliance with the German Animal Welfare Act approved by the local authority

TABLE II
CHRONIC EXPERIMENTS

No.	No. of subjects	Sensor configuration (see Table I)	Duration
1	1	A	1 d
2	2	A & B	48 d
3	2	C	28 d
4	1	C	2 d
5	2	C	19 d
6	2	D	22 d
7	2	D	67 d

Fig. 3. Climate housing container with controlled light regime system equipped with several RFID readers for individual and group housing and a local workstation for data handling and storage.

(RP Karlsruhe AZ 35-9185.81/G-26/13). Throughout the experiment, the animals were kept in a climate cabinet under a light-dark cycle of 12:12 h. The container [Fig. 3] was equipped with a local station, two type III cages, and one type IV cage. Each type III cage is equipped with a reader with 1 W transmission power (Feig ID ISC.MR101-USB) together with a 34 × 24 cm pad antenna. The type IV cage is equipped with a self-made 55 × 35 cm antenna to match the dimension of the type IV cage as well as a long range reader (Feig ID ISC.LR2500-A), which is configured to 2 W of power. Data acquisition and implant control was done via web access over a distance of 200 km between the animal care facility (Mannheim) and the Institute for Applied Research (Offenburg). The ECG signals, including heart rate calculation, arterial blood pressure with systolic and diastolic values and body temperature were taken every 20 min throughout the experiment.

In the most experiments, two sibling male rats with an age of at least eight weeks and a body weight above 200 g were used. At the beginning the animals were kept separate in type III cages until operation wounds were healed. After one week,

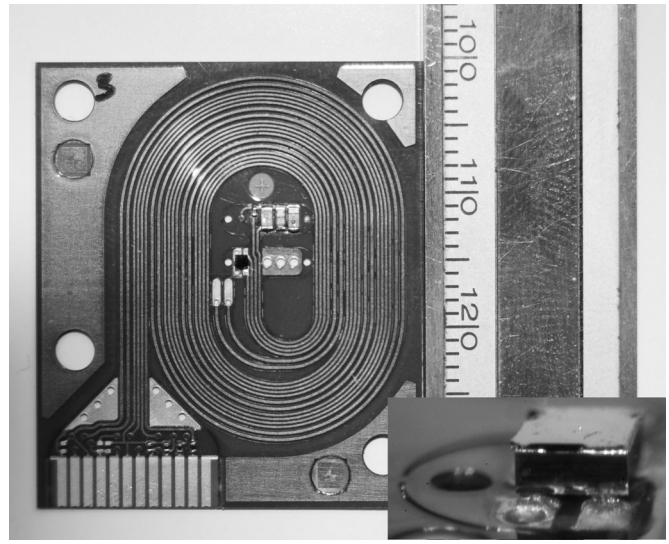

Fig. 4. Sensor part of the implant before separation. Polyimide-based PCB with connection pad array, interconnect lines, gold pads as ECG electrodes and contacts for discrete sensor elements (e.g., diodes for measurement of blood oxygen saturation). A pressure sensor is flip-chip mounted on the substrate (magnified view, lower right).

they were transferred into type IV and housed in groups of two. Implantation was done under 2–3% isoflurane anesthesia with pre- and post-analgesia using carprofen. When measuring the blood pressure in or at the aorta abdominalis an abdominal incision was made and the device was placed and secured inside the cavity. An alternative, less invasive method was also examined, whereby the device was placed in a skin pocket within the abdominal region while the pressure catheter was inserted into the aorta femoralis and advanced into the lower aorta to prevent vascular stasis and rapid clotting. The ECG electrodes were always subcutaneously attached to muscle tissue in Einthoven II positioning for all measurements.

The experiments were carried out as long the implants provided usable results. Thereafter, the animals were euthanized using a high dose of isoflurane and the implants were explanted and inspected.

III. RESULTS

A. Sensor Modules

A batch of approximately 30 circuit boards, one example shown in Fig. 4, was successfully populated and partly processed to sensor modules. Fig. 5 shows an overview of the modules as listed in Table I. Each sensor module was tested and calibrated before assembly. The yield was mainly reduced by damaged bondings during encapsulation.

The function of the electrodes was qualitatively tested using a continuity tester. The calibration resistor of the temperature sensor was set in a range between 54 and 57 kΩ. Later it was set to a fix value of 56 kΩ and calibration parameters were defined to remove the remaining variance during digital data processing. At 2.6-V supply voltage, the temperature sensor

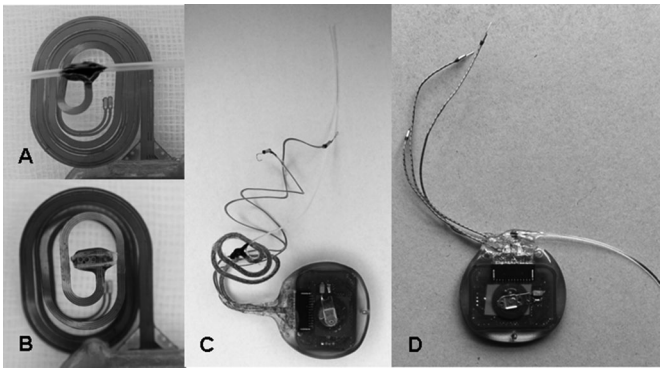

Fig. 5. Sensor configurations according to Table I.

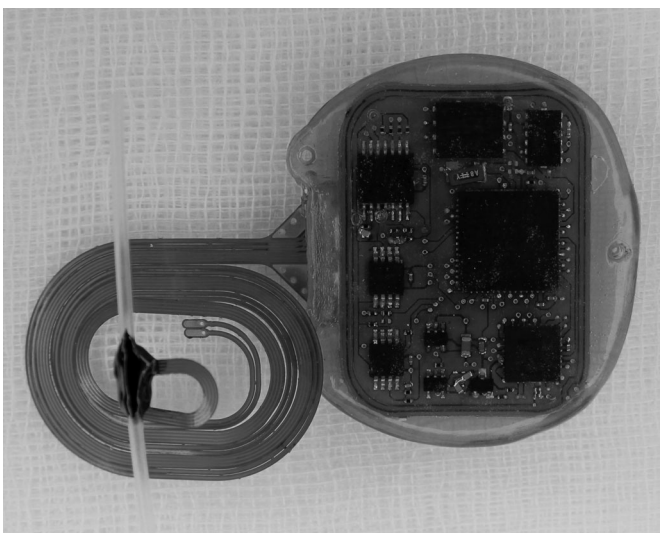

Fig. 6. Low-cost implant for small animal models based on RFID technology. Polyimide-based PCB with integrated sensors for blood oxygen saturation, blood pressure and two ECG electrodes (left side) are connected to the electronic system for signal amplification and processing, power management and data transfer via RFID. The electronic circuitry has been encapsulated in epoxy resin with an additional silicone rubber coating.

reaches a sensitivity at mean of −34 mV per K according to the calibration measurements.

To calibrate the pressure sensor, a resistor of approximately 85 kΩ was placed between positive output and ground. Therefore, the configurations A, C, and D show similar sensitivities, which are approximately 53 μV per mmHg. The cuff were only qualitatively tested using a silicone tube as replacement for the vessel. In contrast to the described method, the sensor is calibrated to zero at environmental pressure.

B. Implant

1) Manufacturing: The implants, from which one of them is depicted in Fig. 6, were produced on demand. The circuit boards were populated using a conventional reflow soldering process. Casting was done with low temperature 45 °C at the beginning and higher temperature 70 °C at the end, because the mold becomes soft at temperatures above 50 °C. Air pockets, which sometimes occurred, were opened and filled during surface finishing. The final device reaches a weight of approximately seven gram after casting and was tested via the connector before sensor modules were installed.

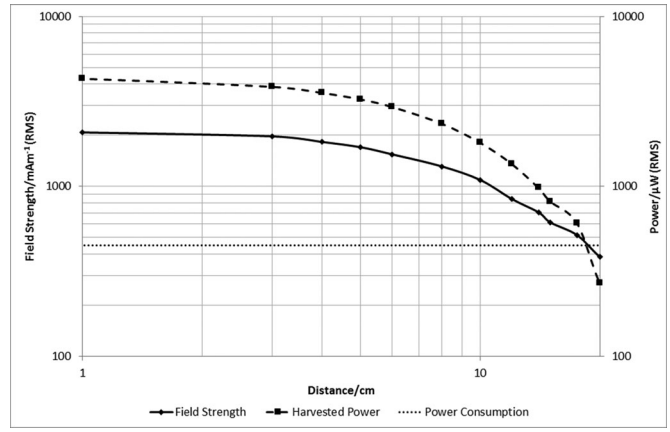

Fig. 7. Magnetic field and harvested power measured inside a type III cage. The magnetic field is generated by a RFID reader with a transmission power of 1 W. The antenna with a size of 34 × 24 cm is placed below the cage. The magnetic field strength is measured according to ISO 10373-7 [23] and is shown as a solid line (left *y*-axis). The dashed line marks the harvested energy (right *y*-axis). The power consumption of the circuit is shown by the small dotted line.

2) Power Management: The power supply was measured open circuit. Therefore, the configuration of the type III cage or more precisely a reader device with a transmission power of 1 W was used. In the middle of the cage, the magnetic field is in the range of 0.6 to 2.2 A/m (RMS) at the first 15 cm of distance [Fig. 7]. Therefore, the implant is able to harvest 4.3 mW at max. Taking the static power consumption, which is 0.5 mW inside and 0.1 mW outside the field, into account, it is possible to charge the battery or to drive the circuit passively up to a distance of 15 cm.

Due to the implemented acquisition concept, which solely takes measurement sequences and powers the whole circuit at a configured time of interest, the exact power consumption depends on the profile. The implant needs approximately 15.7 mJ to process one sequence of all sensors and is theoretically able to run outside the field for approximately six days with a fully charged battery using the standard configuration, i.e., taking one sample each 20 min.

3) Data Acquisition: One benefit of the modular concept is that the signal acquisition can be tested before mounting the sensor module using electrical signals on the sensor interface. The result of one of these measurements is depicted in Fig. 8. The graphic shows that the device is able to measure fast variations of the signal and fulfills the specifications.

After mounting of the sensors, the implants were finally checked to ensure that all device were fully operational before implantation. The mounting process did not change the signals of ECG and the temperature sensor. Only the pressure sensor shows a small offset drift, due to the contact voltage as well as the post curing process, because the calibration was close to the manufacturing. The offset drift was compensated by software,

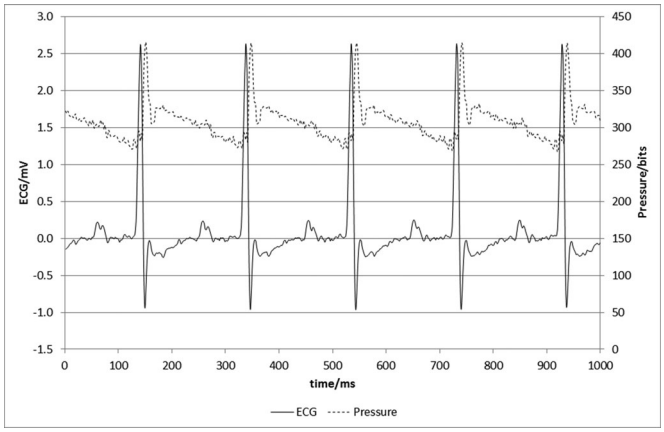

Fig. 8. Simulated ECG (lower curve) and pressure signal (upper curve) measured by the reusable part of the implant before mounting the sensor module.

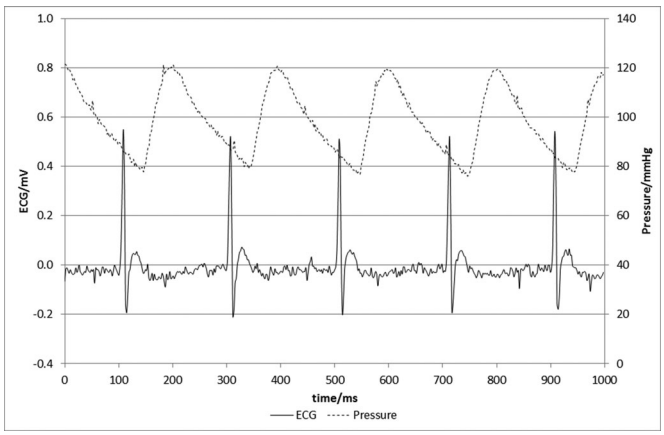

Fig. 9. Real-time blood pressure (upper curve) and ECG (lower curve) signal taken from a male rat. Blood pressure was measured with a fluid-filled catheter inserted into the Aorta femoralis. ECG was measured in Einthoven II configuration. The recordings were taken during experiment 7.

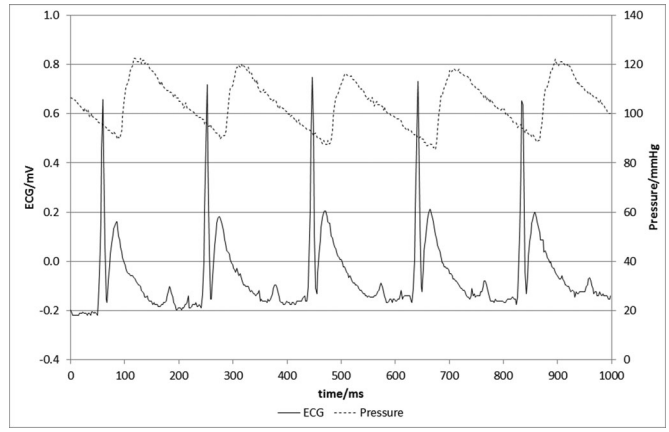

Fig. 10. Real-time blood pressure (upper curve) and ECG (lower curve) signal taken from a male rat. Blood pressure was measured by a bypass set on the Aorta abdominalis. ECG was measured in Einthoven II configuration. The recordings were taken during experiment 2.

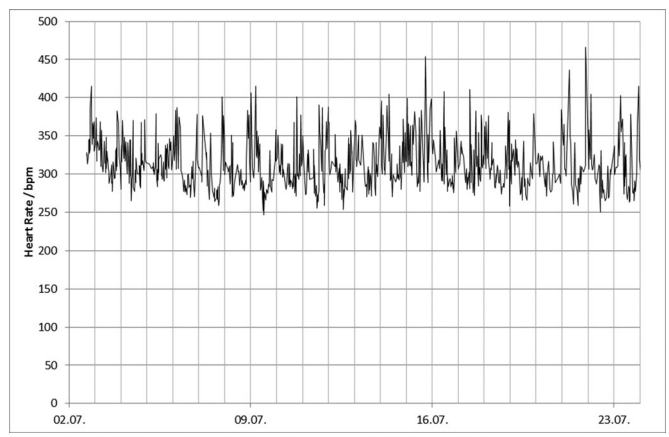

Fig. 11. Heart rate of a male rat over a period of three weeks calculated from the ECG. The graph shows diurnal variations with a minimum during the day and a maximum at night. The recordings were taken during experiment 6.

so that only calibrated and 100% tested devices were used in the following *in-vivo* study.

C. First In-Vivo Measurement

In the facility in Mannheim, the devices were successfully implanted and first results were taken. An example of an unfiltered real-time readout is shown in Figs. 9 and 10. It shows that the ECG and the pressure measurement methods used in configurations A, C, and D show a good noise to signal ratio and reliable physiological data. The noninvasive, bloodless aortic cuff measurement method (configuration B) was saturated after implantation and was not further investigated in the chronic study.

D. Chronic In-Vivo Study

Experiments were performed as long as possible to gain experience with the system. Some of the implants were used for a maximum of 67 days with proper power supply and communication.

The experiments were limited due to mechanical problems (experiments 2, 3, and 7), but also animals tried to remove their implants (experiments 4, 5, and 6) even after surgical incisions were healed, which is may be caused by the increase of activity due to group housing. The implants were, therefore, placed in the abdominal cavity instead of a skin pocket during the last experiment.

In addition, the invasive blood pressure measurements methods always resulted in a signal loss due to blood clots. Implanting the sensor directly into the blood stream (configuration A), we were able to measure the blood pressure for around half a day. Using a fluid-filled catheter (configurations C and D) enabled us to extend the measurement to around three to four days, but with increasing attenuation of the signal.

The polyimide PCB (configurations A, B, and C), especially the connector area, was vulnerable to shear and tear stress which loosens the electrical isolation of the polyimide foil and added background noise or resulted in a short circuit. Therefore, in the current design the device only provided an ECG signal up to six day using the polyimide PCB.

Longer operations were performed using configuration D. Two sample measurements are depicted in Figs. 11 and 12. The

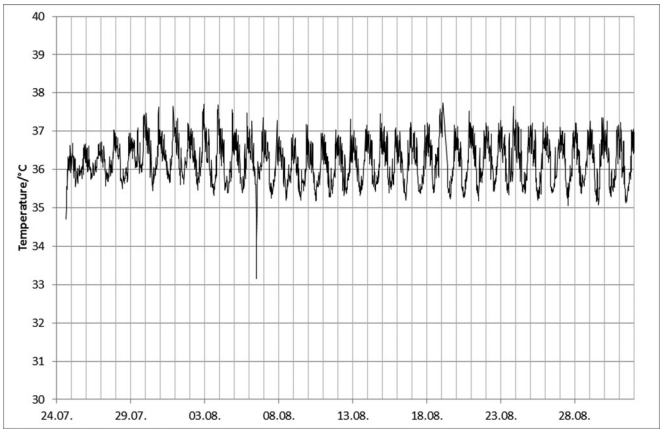

Fig. 12. Body temperature of a male rat measured inside the abdominal cavity over several weeks. The graph shows diurnal variations with a minimum at day and a maximum in the night. Two exceptional events within the graph are caused by technical incidents: a small decrease on August 6th due to a check-up as well as small interruption on August 19th due to maintenance. The recordings were taken during experiment 7.

heart rate (see Fig. 11) was calculated from the ECG signal. The body temperature was measured inside the abdominal cavity. Both measurements show reliable physiological values with the typical diurnal variation and confirm that the system was operational over several weeks.

IV. Discussion

The *in-vivo* study successfully demonstrates that the presented concept of an RFID-based modular implant is an alternative to common designs. Main advantages are the energy concept and reusability due to the modular design. Both the network as well as the reusable part of the implant could be used immediately for further analysis and applications.

The sensor module needs to be refined. Especially the mechanical stability of the sensor interconnection has to be increased. In addition, the problem with the preferable noninvasive blood pressure measurement method could be solved by filtering the offset voltages to prevent saturation and build a cuff that clings around the aorta. For the presented invasive method the material, which is in contact with blood, has to be optimized.

Nevertheless, the implant is able to measure ECG, arterial blood pressure and temperature with sufficient accuracy using the presented configurations. The signal, depicted in Figs. 9 and 10, show the required resolution as well as a good signal to noise ratio. The remaining artefacts, a common result of electrode motion, could be removed by applying high- and low-pass filters. The device is well suited to measure the moment-to-moment liability of the blood pressure and the heart function by ECG.

V. Conclusion

A concept of a low-cost RFID implant to measure physiological functions in small animals has been proposed, developed, and characterized by sampling ECG, arterial blood pressure, and body temperature data. The accuracy is within the requirements of a modern biotelemeteric device and enables research on freely moving animals, which can be kept in groups. The modular concept of the device, separating sensors and RFID implant, enables application specific changes and the use of components already available on the market. This makes the device a highly competitive product to already available telemetric devices. Especially the use of low-cost disposable sensor modules and the semipassive power supply reduce the cost of purchase and refurbishment. The main RFID implant can be reused several times. A modern, standardized, and cost-efficient IT infrastructure also allows central data management and access with web-enabled devices. Long-term stability, especially of the sensor module and blood pressure measurement concept, is subject of ongoing research.

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