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An autonomous bladder pressure monitoring system

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Abstract

This paper reports on the development of an autonomous monitoring system, capable of continuously measuring the pressure inside the bladder. The capsule features wireless bi-directional communication and can be inductively powered anywhere inside the bladder. Short-circuiting the resonant LC tank for the data transmission maximizes the operating range of the passive telemetry. A novel clock extraction method is presented, based on lock-in of a Schmitt-trigger oscillator into the external radio-frequency field. A prototype was built using only discrete components. The system measures the pressure with a resolution of 0.04 kPa at a sampling rate of 10 Hz.
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Keywords: Pressure sensor; Inductive powering; Telemetry

1. Introduction

The diagnosis of urinary incontinence includes urodynamic tests to measure the contraction of the bladder muscle as the bladder fills and empties. This involves the use of catheters, which can be painful and is non-physiological. As a result, these tests do not always show a good correlation with the clinical findings. The aim of this project is to develop an implantable bladder pressure monitoring system that: 1. In a first phase should register and wirelessly transmit the bladder pressure, in a way unnoticeable to the patient. This would strongly enhance the comfort of the patient as well as the accuracy of ambulant urodynamics. 2. In a next phase, the real-time bladder pressure can be used to stimulate the sphincter muscles in an intelligent way [1]. This can be particularly useful for patients with neurogenic bladder dysfunction. Neurogenic bladder is the loss of normal bladder function caused by damage to part of the nervous system. A combination of the presented data logger with sacral nerve stimulation could then effectively replace the neurofysiological feedback from the bladder to the contraction muscles and thus prevent incontinence.

Past telemetric bladder pressure devices [1,2] suffer from limited accuracy and especially from short lifetimes, as they are battery powered. This can be overcome by the use of inductive powering, a concept that has become widespread in biomedical applications, in witness the systems in [3–12]. Bottleneck in these designs remains the operating range between transponder and reading unit. The design presented in this paper maximizes this range to guarantee operation over all possible positions of the telemetric capsule inside the bladder.

A general system overview is depicted in Fig. 1. The different building blocks are discussed below.

2. Inductive powering

2.1. External driver

As described above, inductive powering can provide a solution to battery related lifetime problems, on condition that the distance to an external power source can be made sufficiently small. Furthermore, the use of inductive powering

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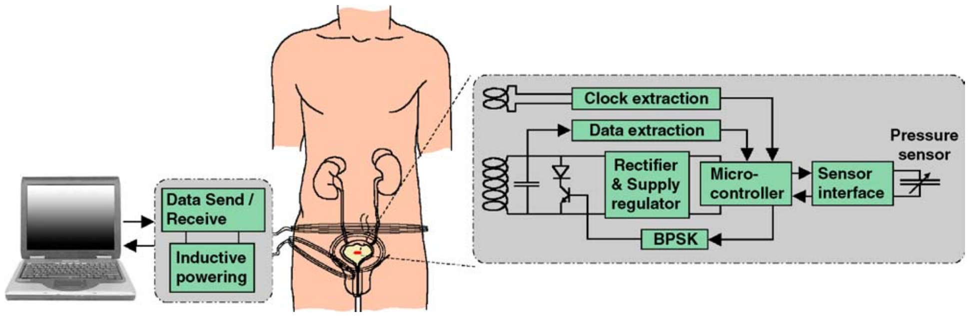

Fig. 1. General overview of the system.

allows reducing the implants dimensions, since the battery, often the largest component in biomedical implants, can be left out. This is important, since the use of the capsule as a diagnostic tool imposes strict limits on the size of the implant. The capsule diameter should be smaller than 5 mm to allow the implant to be inserted in the bladder with a cystoscope. To further minimize the volume, additional antennas are unacceptable, and the bi-directional data communication is through the same single inductive link as the inductive powering.

The efficiency of the power transfer depends strongly on the position and orientation of the secondary coil to the primary coil. The power transfer is maximal when the axes of both coils coincide, and decreases with any angular or lateral misalignment [11,13]. However, the implanted capsule can move freely inside the bladder and adopt all possible orientations. Therefore, the external driver is equipped with three orthogonal coils that cover the entire three-dimensional operating area (see Fig. 1). This solution was chosen over the dual situation with three internal coils to limit the capsules volume. Fig. 2 depicts the driver and three external coils, of which one is to be worn around the waist, one around the leg, and the third one is to be placed vertically, in front of the bladder. The coils are driven, at a carrier frequency of 132 kHz, by three parallel class-E amplifiers. This allows each driver to be matched to the corresponding coil and guarantees class-E regime [14] to obtain an optimal efficiency. The coil that couples best with the secondary coil is switched on, while the others remain off.

2.2. Internal coil

To further enhance the inductive coupling, the internal coil is equipped with a ferrite core, as this increases the induced open-circuit voltage by a factor μ_{rod} over the value for the same coil in free space [15]. μ_{rod} is the effective permeability of the ferromagnetic rod and can be found from curves in [15] as a function of the initial material permeability μ_{i} and the rod length-to-diameter ratio. The difference between μ_{i} and μ_{rod} corrects the end effects for finite length rods.

The minimal core and coil sizes are determined by the required power transfer, since the dimensions of the ferrite core have a big influence on the induced voltage through μ_{rod} and the coupling factor.

The induced power is received by a parallel resonant LC tank, rectified, and regulated to a stable supply by a low dropout voltage regulator (see Fig. 3). Besides a low dropout voltage and a low consumption to achieve high power efficiency, the power supply rejection ratio (PSSR)

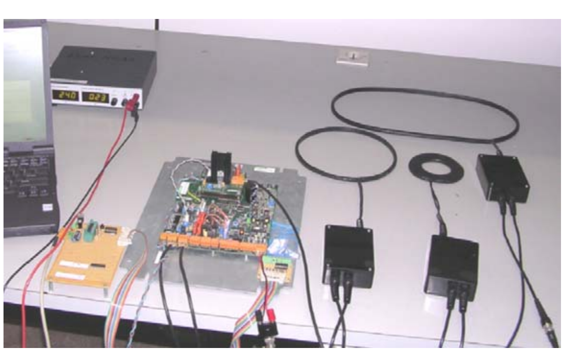

Fig. 2. Photograph of the external unit, with the Class-E drivers, and three external coils.

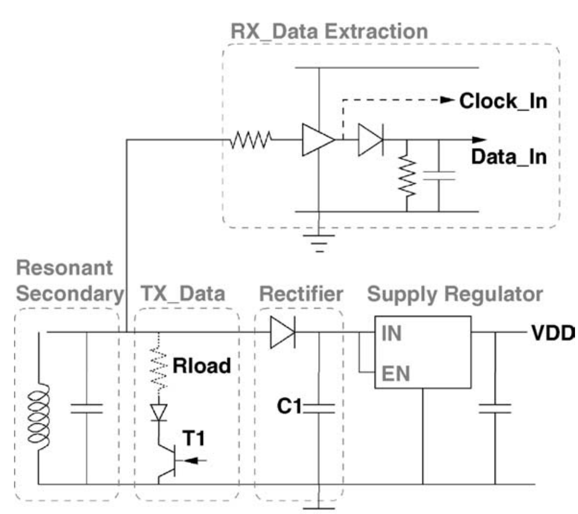

Fig. 3. Schematic of the implant, showing the building blocks for the supply generation and the data transmission.

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