An Autonomous, Capacitive Sensor Based and Battery Powered Internal Bladder Pressure Monitoring System

Philippe Jourand*, Robert Puers

Katholieke Universiteit Leuven, ESAT-MICAS, Kasteelpark Arenberg 10, B-3001 Leuven, Belgium

Abstract

A miniature battery powered internal bladder pressure monitoring system is presented, implantable through minimal invasive catheterisation. Prototype circuitry and battery fit into a flexible cylindrical pill measuring ø 5 mm x 40 mm. The device is intended for off-line monitoring of behaviour and evolution of bladder pressure under normal life circumstances. No external devices are mandatory during examination, ensuring maximal ‘unawareness’ of the system for the patient. Power is distributed from a pin-type ø 3 mm lithium cell. The sensor interface was calibrated in a dedicated test-chamber. Finally the circuitry was assembled and successfully tested on two Kapton foils measuring 4.2 x 18 mm².

Keywords: Internal bladder pressure; Battery operated; Capacitive sensor; Minimal invasive; Implantable.

1. Introduction

Analysis of bladder pressure can not only serve to prevent a great deal of urologic diseases. Possibly, urinary incontinence could be diminished or even faded out. Measurement of the bladder pressure can be done in various forms of invasiveness. Non invasive techniques focus either on the correlation between ‘urine flow’ and ‘bladder pressure’¹ or the pressure from a penile cuff, needed to interrupt urine flow¹². All of these methods however restrict patient movement and comfort. Slightly more invasive procedures focus on the pressure external to the bladder wall, as reported in³, or using catheterization⁴. A final group of devices try to measure the bladder pressure with all electronics inside the bladder⁵⁶. This research focuses on this last group, since it is the most challenging technique and probably the only way to create a truly imperceptible, minimal invasive bladder pressure monitoring device.

Key constraint in this design is the minimal invasive approach. In urology practice, F10 to F28 sized catheters are inserted through the urethra into the bladder. If the device has to be introduced through this catheter, it will hence need to fit inside the inner diameter of the largest, i.e. ~ 6 mm. To ensure a smooth insertion, the constraint was reduced even more; restricting the overall diameter to be 5 mm. Length of the device is less of an issue: it needs to be as small as possible not to hinder the bladder wall and was fixed at 40 mm. Because the device will be fully implanted inside the bladder, a protective layer must be foreseen in both ways. The electronics must be shielded...
from the harsh surroundings, but vice versa, also the bladder must be protected from any harm or infection possibly caused by the device materials. A unique feature of the presented device is that a flexible, silicone casting was chosen to house it. This leads to a pill-shaped monitoring device depicted in Figure 1 (a). The system’s block diagram is depicted in Figure 1 (b). Circuitry is divided into two parts: measurements and power. These will build up the two layers of the electronics circuitry discussed later.

![Fig. 1. (a) 3D AutoCAD representation of the pressure monitoring device, showing the relative sizes and locations of the components used. Outer dimensions of the pill, including silicone encapsulation, are ø 5 mm x 40 mm length; (b) block diagram of the device showing the two parts in circuitry: measurements and power.](image)

2. Circuitry components: measurements and power

Two major constraints influence the choice in components: size and power. Since the second will come from a battery, data will not be transmitted but instead stored on a memory module. Measurements circuitry therefore consists of: a pressure sensor, digitalisation electronics, a micro controller unit (MCU) and EEPROM.

Commercial pressure sensors are, based on a resistive, capacitive or optical measurement. The first’s major strength: a straight-forward implementation. Their weakness: power consumption. The second group’s benefit is not using actual power but introducing them into liquid environments causes parasitic effects. In the last group, ultra low power optical pressure sensors exist. However, its surrounding components are not yet small enough to introduce them into this design. A capacitive sensor was chosen for the offset can be handled and power is crucial. MicroFab’s E1.3N was selected for its miniature size, perfectly suited range and excellent protection against overpressure. Its capacitance value varies between 6.026 pF and 6.206 pF in a range of 1000 to 1300 mbar. To detect these changes, a capacitive sensor interface is needed. These interfaces can be divided into three groups: oscillators, switched capacitors and capacitive bridges. The latter is discarded for it requires an AC driver signal. In the first group, capacitance is first converted to frequency and digitised. Nowadays, capacitance-to-digital converters (CDC) can be found that fit inside a ø 5 mm cystoscope. The AD7153 from Analog Devices was chosen for its size (10-lead MSOP footprint), low power consumption (typically 100 µA), I²C-compatibility and large number of operating modes and settings.

Adding intelligence, the MCU forms the backbone of the device. The PIC10F206 from Microchip was chosen for its tiny SOT23 package. Size does however come with an important trade-off: extremely limited functionalities. It has an 8 bit counter and internal 4 MHz oscillator. I²C compatibility, asynchronous serial communication, CDC-configuration settings, time tracking and data processing routines are all manually implemented into the 512 words of flash programming memory. Of its 3 I/O pins, two are used for I²C communication and one for both serial communication and toggling the write protect of the EEPROM. This memory module (Microchip’s 24AA16) is an I²C compatible IC, which, in the smallest footprint (5 pin SOT23), can store up to 16 Kbit of information. Consuming 3 mA in a write cycle, data is written in page-write bursts of 8 data bytes, taking only 5 ms.

The power circuitry consists of: a reed switch, D-flip-flop (DFF), inverter and battery. To conserve power and ensure no accidental switching, a power switch must be implemented, excluding the most straightforward solution of a mechanical switch. Placing the inverted DFF output on the D-input, its output can be toggled by triggering the clock input. This is achieved with the reed switch, activated by a DC magnet. One of the smallest commercially
available to day is the RI-80 J-Type SMD, measuring less than 7.1 x 1.8 x 2.0 mm$^3$. The toggled output powers the measurement electronics and comes from a battery. Finding a suitable battery is not an easy task: as expected, size and capacity are inversely related. Table 1 summarizes the important features of some interesting batteries. Button cells typically have higher capacities yet the smallest diameter is still too big to fit inside the pill. Pin type batteries seem most suited. The BR316 was selected for the application for its size.

Table 1. Important features of suitable batteries summarised.

<table>
<thead>
<tr>
<th>Battery:</th>
<th>Brand:</th>
<th>Diameter (mm)</th>
<th>Length (mm)</th>
<th>Capacity (mAh)</th>
<th>Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>BR316</td>
<td>White Label</td>
<td>3.0</td>
<td>13</td>
<td>13</td>
<td>Pin</td>
</tr>
<tr>
<td>BR425</td>
<td>Panasonic</td>
<td>4.2</td>
<td>22</td>
<td>25</td>
<td>Pin</td>
</tr>
<tr>
<td>Renata 337</td>
<td>Renata Batteries</td>
<td>4.8</td>
<td>1.65</td>
<td>8</td>
<td>Button</td>
</tr>
</tbody>
</table>

3. Measurements and results

To evaluate the feasibility of the design, a dedicated test chamber was used as depicted in Figure 2. The setup uses a Druck DPI 600 both as a reference pressure and to change the pressure inside the chamber through a pump and valve. The electronics were first soldered onto a testing board and placed inside the pressure chamber. The side of the chamber houses two coaxial plugs to monitor I²C communication and a DB-9 serial connector for power supply and serial communication. The CDC was configured to give a 12 bit result in a 250 fF window at 6 pF to cover the pressure sensor range. The pressure inside the chamber was increased from atmospheric pressure to 1300 mbar in steps of ~10 mbar. Through the data points (Y-axis contains the reference pressures and X-axis the digital sensor read-out) a cubic curve was fitted. This procedure was repeated for five spread out measurements to generate an averaged fitting curve, shown in Figure 3 (a) and characterised by the following equation:

$$P = 2.28 \cdot 10^8 \cdot x^3 - 1.57 \cdot 10^4 \cdot x^2 + 0.52 \cdot x + 7.37 \cdot 10^2$$

(1)

Fig. 2. Dedicated setup used to prove the feasibility of the device and calibrating the sensor. A Druck DPI 600 (left) serves as a reference pressure and is used to change the pressure inside the chamber (middle) through a pump and valve. Power (top right) and serial communication (bottom right) pass through air tight connectors located on the side of the pressure chamber (middle). The latter houses the test board (top middle).

The pressure ‘P’ inside the chamber can than be calculated using (1) and a digital read-out ‘x’ of the sensor output. Sensor drift translates to an identically shaped curve, shifted vertically. This shift can be cancelled out by linking the first measurement to a known pressure, calculating a different constant in (1). The circuitry was then fitted onto two layers of a Kapton foil which can both be cut to a size of only 4.2 x 18 mm$^2$. These are shown in Figure 3 (b). Showing the power and measurements circuitry separately, these two foils were placed back-to-back.
and successfully tested for pressure variations of the pill in its near final form. In the next steps, the same circuitry will be transferred to a double-sided Kapton foil with vias and, after encapsulation in silicone, tested in vitro.

![Sensor Calibration Curve](image)

**Fig. 3.** (a) Sensor calibration curve. The reference pressures taken from the DPI 600 are plotted on the Y-axis, the 12 bit read-out of the sensor on the X-axis (blue crosses). Through these points, an averaged cubic equation was fitted to characterise the sensor (red full line). (b) Components soldered and tested on two pieces of flexible Kapton foil. These can be placed back to back and cut to a size of 4.2 x 18 mm². The left shows the power circuitry containing the reed switch; the right shows the measurement circuitry with the packaged sensor on the top. In future work, these two circuitries will be placed on a double sided Kapton foil with vias connecting both sides: a technology currently missing in the MICAS-labs.

### 4. Future work and conclusions

The feasibility of a device, implantable through catheterisation, measuring autonomously the internal bladder pressure was proven. Several improvements can still be made to create a truly ubiquitous device: size can be reduced even more enabling the use of smaller cystoscopes for implantation, decreasing power consumption, introducing bigger memories and adding new MCU routines such as auto calibration of the sensor offset.

### Acknowledgements

This research has been developed in the frame of BIOFLEX, an IWT funded project, contract number IWT-040101. Special thanks to Michel De Cooman for producing the flex prints used in the testing of the device.

### References