

# Wireless system for monitoring Intra-abdominal pressure in patient with severe abdominal pathology

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**Abstract.** The paper discusses an experimental design of the wireless system for monitoring intra-abdominal pressure (IAP) using Bluetooth Low Energy technology. The possibility of measuring IAP via the bladder using a wireless pressure sensor with a hydrophobic bacteria filter between the liquid transmitting medium and the sensor element is grounded.

## 1. Introduction

The intra-abdominal pressure (IAP) measurement is fundamental for the evaluation of patients with severe abdominal surgical pathology at risk of intra-abdominal hypertension [1]. IAP is the pressure concealed within the abdominal cavity as defined superiorly by the diaphragm, anteriorly and posteriorly by the abdominal wall, and inferiorly by the pelvic floor. The World Society of the Abdominal Compartment Syndrome (WSACS, [www.wsacs.org](http://www.wsacs.org)) has outlined definitions and recommendations relating to the measurement of IAP [2].

Special interest deserves continuous IAP monitoring allowing to improve assessment of the effects of medical management. A continuous trend of 24-h IAP recordings can be obtained with a 3-way Foley catheter or a balloon-tipped catheter placed in the stomach [3]. During continuous IAP monitoring respiratory variations can be more easily identified which is very important for diagnostic purposes [5, 6]. Multiple assembled kits for IAP monitoring are commercially available. They are attached to the monitor unit by wires or directly [2]. Newer techniques including wireless transducers for IAP monitoring are being continuously developed [7, 8, 9]. At the same time this techniques, do not allow to use IAP monitoring for a long term, due to low battery capacity.

The goal of the project described in this paper was to design an experimental system for wireless IAP monitoring with the help of intravesical technique and to research dynamic characteristics of its catheter-sensor part.

The following problems were solved:

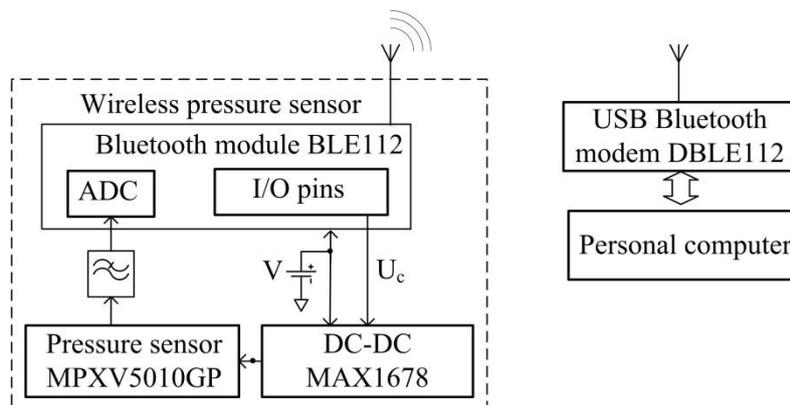
- 1) implementation of wireless transmission channel with high noise immunity and security;
- 2) elimination of sensor contamination;
- 3) registration of respiratory variations.

The design of the wireless IAP monitoring system, the theoretical model and identification results of its catheter-sensor part were described. The analysis of experimental data was presented.

## 2. The design of the wireless pressure monitoring system

The wireless pressure monitoring system consisted of wireless pressure sensor (WPS), USB Bluetooth modem and a computer with installed software. In figure 1 the system's block diagram is shown.





**Figure 1.** The block diagram of the wireless pressure monitoring system.

Wireless communication was achieved by utilizing the Bluetooth Low Energy wireless channel that is implemented on the basis of BLE112 module and BLED112 USB modem.

The MPXV5010GP transducer was used as differential pressure sensor in range 0 to 10 kPa. The WPS was powered with a pair of AAA batteries. The pressure sensor was powered with a DC-DC voltage converter made on the basis of MAX1678. Suppression of high frequency components in the sensor output signal was provided by an analog second order low-pass filter with bandwidth of 5 Hz, which is implemented using OPA376. The 12 bits analog-to-digital conversion, formation of data packets and transmission via wireless channel with a frequency of 1 Hz were made by BLE112 module. In addition the BLE112 module controlled the DC-DC converter by means of  $U_c$  signal in order to save energy.

Electronic components of WPS were placed in a small polymer casing as shown in figure 2. Measurement results were displayed on the computer using a pressure monitoring software.

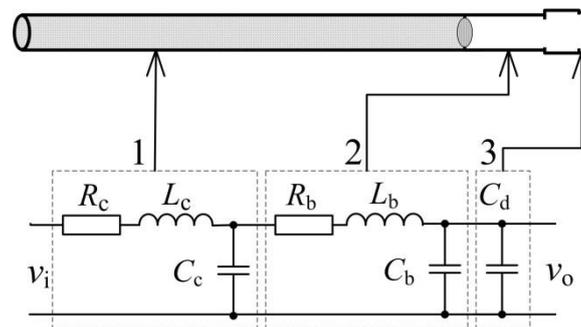


**Figure 2.** Internal placement of the WPS components.

### 3. The mathematical model of the catheter-sensor system

An air barrier with a hydrophobic anti-bacterial filter was placed between the liquid pressure transmission medium and sensor to eliminate contamination of the pressure sensor. This design required optimization of dynamic characteristics of the catheter-sensor system.

The physical model of the system consisting of the sensor and the catheter can be represented as second order model with lumped parameters [10]. Figure 3 shows an electric analog of the simplified pressure-measuring system taking into account the air barrier. The substitutes for hydraulic inertance, resistance and compliance are electric inductance  $L_c$ , resistance  $R_c$  and capacitance  $C_c$ , respectively. The substitutes for the compliance of the sensor diaphragm and the compliance of the air barrier are electric capacitance  $C_d$  and capacitance  $C_b$ . It is assumed that the compliance of the sensor diaphragm is much larger than that of the liquid-filled catheter and the catheter material is noncompliant. The inertance  $L_b$  and the resistance  $R_b$  of the air barrier are neglected.



**Figure 3.** Electric analog of the simplified pressure-measuring system: 1 - partially filled fluid catheter, 2 - air barrier, 3 - pressure sensor membrane.

The relation between the input voltage  $v_i$ , analogous to applied pressure, and the output voltage  $v_o$ , analogous to pressure at the sensor membrane is determined by the following second order differential equation

$$v_i(t) = \frac{L_c(C_d + C_b)}{dt^2} d^2 v_o(t) + \frac{R_c(C_d + C_b)}{dt} dv_o(t) + v_o(t).$$

The liquid resistance in the catheter is due to viscous friction of its layers moving along the catheter at different speeds. Assuming that the motion of the liquid with viscosity  $\eta$  in the catheter with length  $L$  and radius  $r$  is laminar, it is possible, using the Poiseuille's equation, to determine the value of resistance  $R_c$  as

$$R_c = \frac{8\eta L}{\pi r^4}.$$

The inductance  $L_c$  of the liquid with density  $\rho$  in the catheter with radius  $r$  is due to the mass of the liquid. It can be estimated by the following equation

$$L_c = \frac{\rho L}{\pi r^2}.$$

The compliance  $C_d$  of the sensor diaphragm with the volume modulus of elasticity  $E_d$  is determined by the equation

$$C_d = \frac{1}{E_d}.$$

The compliance  $C_b$  caused by the isothermal compression of the air barrier is calculated by equation

$$C_b = \frac{\Delta V}{\Delta P}.$$

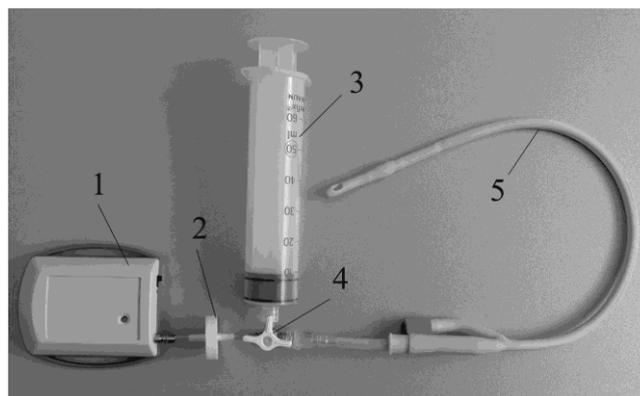
The values of the natural frequency  $\omega_n$  and damping ratio  $\xi$  can be calculated as

$$\omega_n = \frac{1}{(L_c(C_d + C_b))^{1/2}},$$

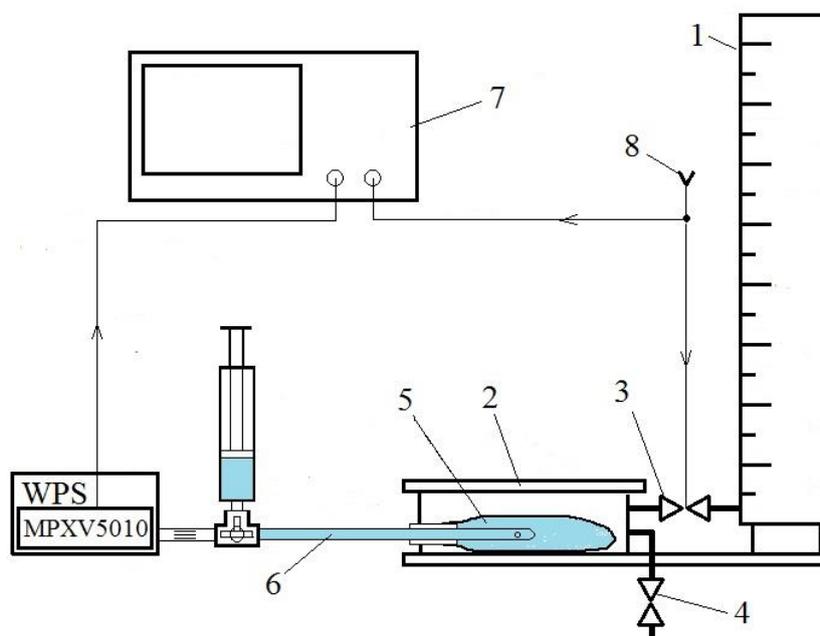
$$\xi = \frac{R_c(C_d + C_b)^{1/2}}{2L_c^{1/2}}.$$

#### 4. Experimental part

The goal of the experiment was to record transient responses of the catheter-sensor system using an experimental kit (figure 4) and a setup for the abdominal pressure simulation (figure 5).



**Figure 4.** A kit for wireless IAP measurement: 1 – WPS; 2 – hydrophobic anti-bacterial air filter; 3 – syringe with 50 ml volume; 4 – three-way tap; 5 – Foley catheter.



**Figure 5.** Block diagram of the setup for the abdominal pressure simulation: 1 – 3 l opened container; 2 – 1 l sealed container; 3 – solenoid valve; 4 – water tap; 6 – the kit for wireless IAP measurement; 7 – digital oscilloscope; 8 – synchronization input.

The abdominal model was implemented in the form of two containers which are connected to each other with a solenoid valve. A thin-walled silicone bag 200 ml volume modeling the bladder was placed at the bottom of the sealed container. The silicone bag was connected to the polymer adapter installed in its wall. The Foley catheter was placed in the silicone bag through the adapter.

Both containers were filled with 3 l of water through a water tap located at the bottom of the sealed container. The Foley catheter and the silicone bag were filled with 50 ml of water using a syringe. After filling the system the solenoid valve was closed and only then the water tap was closed.

The transition process was simulated by opening the solenoid valve. Transient responses were recorded from the output of the pressure sensor using a digital oscilloscope which is synchronized with the solenoid valve. The transitional characteristics, obtained in a series of experiments, were processed using the MATLAB System Identification Toolbox.

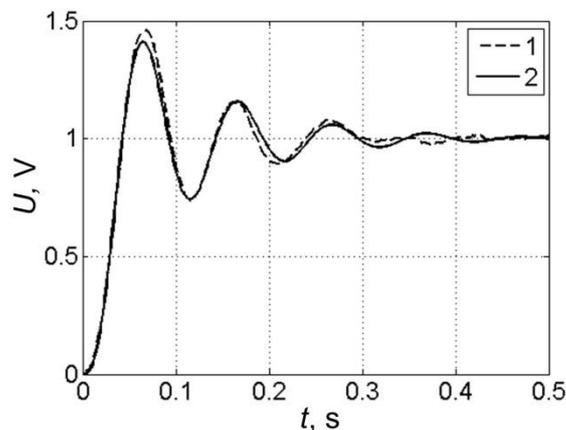
Assessment of stability of the wireless channel was made by recording cases of communication failures between WPS and the computer. The WPS was placed on the patient's thigh (without being connected to the Foley catheter). The patient was allowed arbitrary physical activity within a range of no more than 10 m from the computer.

## 5. Results and discussion

The time constant  $T_{MD}$ , the natural frequency  $f_{MD}$  and the damping ratio  $\zeta_{MD}$  of the catheter-sensor system were calculated using the mathematical model and presuming the volume of the air barrier equal to  $630 \text{ mm}^3$  (the air barrier of the kit).

Following results were obtained:  $T_{MD} = 0,0159 \text{ s}$ ;  $f_{MD} = 9,99 \text{ Hz}$ ;  $\zeta_{MD} = 0,016$ .

Figure 6 shows the normalized experimental transient response of the catheter-sensor system and the result of the identification.



**Figure 6.** Normalized transient responses: 1 - the experimental catheter-sensor system; 2 - the result of the identification.

The structure and parameters of the model, produced by the result of the identification, were defined as

$$W(p) = \frac{1}{T_{ID}^2 p^2 + 2T_{ID}\zeta_{ID}p + 1},$$

where  $T_{ID}$  is the time constant and  $\zeta_{ID}$  is the damping ratio, which amounted to:  $T_{ID} = (0,0156 \pm 0,0006) \text{ s}$ ;  $\zeta_{ID} = 0,183 \pm 0,005$ .

It was concluded that the structure of models and the values of their natural frequencies are in good agreement with each other. Damping ratios of models were different. It was due to the design features of the actual measuring system (the presence of antibacterial filter and a three-way tap).

The maximum value of the air barrier to register respiratory variations was determined on the basis of the mathematical model. This value was about  $100 \text{ mm}^3$ .

Oscillations in the natural frequency of the catheter-sensor system can serve as an indicator of the artifact caused by the patient's motor activity.

WPS demonstrated stable operation for a long term without loss of communication.

## 6. Conclusion

The results of this study demonstrate the ability of the wireless continuous IAP monitoring with the help of the intravesical technique and the Bluetooth Low Energy technology.

The possibility of applying a hydrophobic anti-bacterial filter, placed between the liquid pressure transmission medium and sensor to eliminate the contamination of the pressure sensor, was approved. Maximum air barrier volume of the measuring path was defined, to ensure the registration of

respiratory variations. Mathematical and identificational catheter-sensor models demonstrated good coherence.

Wireless technology allows free movement within the laboratory setting which simplifies experimental research activity. The developed wireless system for IAP monitoring will be used to create a non-invasive method for estimating IAP variations.

### Acknowledgments

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