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NON-INVASIVE BLADDER VOLUME MONITORIZATION THROUGH NON-GALVANIC ELECTRODES

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Abstract: In this paper we study the feasibility of bringing together the classic rheographic volume measurement approach with non-galvanic electrodes which would allow for a fatigue free holter approach in daily monitoring of bladder volume. This approach relies on a multi-frequency FPGA-based system. This viability analysis will be approached both regarding the measurement accuracy and robustness to motion artifacts. In this preliminary study focus is essentially on the accuracy.

Keywords: Capacitive electrodes, Rheography, Bladder volume, FPGA, Embedded system.

1. INTRODUCTION

Real-time volume measurement of human bladder is a novel approach in urodynamics with the potential to assess bladder functioning during daily living activities for patients complaining of symptoms arising from some sort of urinary disorders. Routine urodynamic tests may involve a simple collection of a glass of urine's sample from the patient to more complicated ones, such as the insertion of catheters through the urethra for continuous measurement of the pressure of the bladder and urethra while another catheter is filling the bladder as in multichannel cystometry, portable ultrasonic equipment to obtain a glance of the bladders volume, fluoroscopy of the bladder during voiding, uroflowmetry to assess the bladders voiding rate and electromyography of the electric activity in the bladder neck. Exams like these can take up to an hour to perform and can be very distressful for some patients.

Urinary disorders directly related with the presence of an abnormal volume of urine within the bladder, whether by voiding or filling malfunction, can be a direct result of bladder/detrusor overactivity, urinary tract infections, ureter obstruction, urethral sphincter contraction and bladder muscle weakness or even cancer. Behind the idea of designing a real-time volume measurement device lays the principle that human bladder expands depending on the volume of stored urine, which has a good biological conductivity, leading to the thinning and elongation of the detrusor muscle, up to a limit when the urination urge triggers.

The first attempts to measure bladder volume were made by Tabili *et al* (1971) in a study where the authors recorded volume-related impedance changes between potential electrodes placed on the bladder wall of dogs while stimulating electrically the bladder [1]. Baker *et al* (1975) performed the first non-invasive monitoring of the bladder volume, highlighting the potential of the approach to detect abdominal impedance changes as infusion of urine through dog catheterized bladders takes place [1]. More recently, Shida *et al* (2006) developed a non-invasive urination-sensing device based on the four-electrodes impedance measurement method for human testing [2]. The system was tested in a single subject for a couple of days and the authors were able to identify a difference in the rate of change of impedance in the recordings in the time instants immediately preceding urination urge. Gill *et al* (2008) carried out a study to test several intravesical probe designs, making use of polymer bodies with an array of electrodes attached that were introduced in an *in vitro* organ bath system consisting of excised pig bladders or bladder- like latex vessels filled with saline solutions matching the conductivity of urine [3]. The results obtained by the authors have shown a correlation between conductance and intravesical volume for all probes, with an approximately linear increase at low volumes that approaches an asymptotic value at larger volumes.

A four-electrode impedance measurement configuration is here adopted using two current-injecting and two voltage-sensing electrodes placed around the volume under investigation at specific locations: for non-invasive purposes, electrodes are attached externally to the lower abdomen region where human bladder is located. From a physiological point of view, such configuration creates a dipole-like electric field around the bladder and tissues nearby, forcing current to flow through the highly conductive urine and preventing the propagation of the electric field in the outermost regions of the abdomen. Skin impedance is not of great concern when operating with high frequency signals, although, contact electrolyte developed between skin and electrode can add series resistance and/or capacitance, causing disturbances in voltage recording.

The proposed approach aims to correctly predict the change on bladder volume by measuring potential differences developed across the sensing electrodes as a direct result of a conductance change as the bladder fills. Moreover, it can provide a setup for assessing the change of bladder conductance with volume under the influence of factors such as fluid concentration within the bladder, fluid temperature, blood extravasation and cancer [4, 5] in the recordings.

The use of capacitive electrodes as we designed reduces considerably the discomfort besides preventing the appearance of skin rashes resulting from long term recordings. This opens the possibility of through the day measurements gathering daily information.

2. MATERIALS AND METHODS

a. Hardware Design

An autonomous device for impedance (or the inverse, conductance) measurement was fully designed with no need for using external electronic devices such as commercial wave generators or data acquisition boards as shown in *fig. 1*. The device only requires a USB connection to a laptop running Matlab. The electronic device can be divided into two main modules: one responsible for generating the sinusoidal current signal with amplitude of 100 µA and user-selectable frequency in the range from 20 kHz to 100 kHz, while the other one deals with the voltage detection in the sensing electrodes, amplification and conditioning prior to signal digitalization.

Figure 1 Impedance measurement device developed. Capacitive electrodes were placed on an elastic belt.

The use of capacitive electrodes prevents the appearance of DC components automatically. This is necessary to avoid tissue electrolysis, especially important in long-term measurements [6]. The sample rate of acquisition is set to depend on the frequency of the injecting signal, that is to say that an integer number of periods must be acquired from the sensing signal so as to prevent measurements to be affected by biasing effects. A total number of 50 samples are evenly acquired during the first five periods of the injecting signal in a time frame of 1 ms. The amplification gain can be set to two user-selectable values which, together with the 18-bit resolution of the embedded ADCs, allows to sense signals as low as 5 nV and high as 5 mV.

To control the two modules and hence the entire system, a Field Programmable Gate Array (FPGA) is used which, due to its design nature, can perform all the tasks it was intended for in a concurrent way: waveform generation, data acquisition, processing of the previous samples and transmission of the results *via* USB at speeds of 1 Mbps.

b. Impedance Determination

Figure 2 depicts the final stage of waveform generation performed by a voltage-to-current converter. Current is set as the ratio between the voltage measured at the terminal inputs of the opamp and a reference resistor whose impedance value Z_R is frequency independent within the frequency interval considered. According to Kirchhoff's node law, the current that flows through the reference resistor must be the same as the one that flows through the unknown impedance Z_M, since no current flows into the input terminals of an opamp.

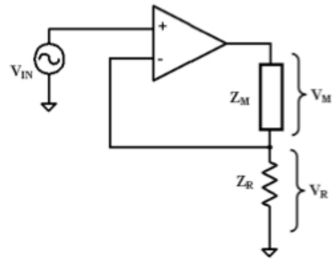

Figure 2 Schematic for the transadmitance amplifier (voltage control current source)

Previous assumption allows us to calculate the value of Z_M using the voltage drops sensed at its terminals and at resistor's ones, as stated in equation (1).

$$i = \frac{V_R}{Z_R} = \frac{V_M}{Z_M} \Leftrightarrow Z_M = Z_R \frac{V_M}{V_R} \quad (1)$$

Since the signals one is dealing with are sinusoids with a well-known frequency, a statistical measure of their magnitude can be achieved by calculating the corresponding voltage quadratic mean, so the last expression is translated into equation (2),

$$Z_{Mr.m.s.} = Z_R \frac{V_{Mr.m.s.}}{V_{Rr.m.s}} = Z_R \frac{\sqrt{\frac{\sum_{i=1}^{N} V_{Mi}}{N}}}{\sqrt{\frac{\sum_{i=1}^{N} V_{Ri}}{N}}} \quad (2)$$

where V_{Mi} and V_{Ri} are, respectively, the sensing and excitation voltage samples and N is the total number of samples acquired in one time frame.

c. Results and Discussion

First experiments performed by the electronic device were based on measuring volume-related conductance changes in solutions within a bladder geometry-like container and whose conductivities fall under the biological range. A spherical container encompassing a total volume of 80 cm^3 and made of plastic walls was used to mimic the bladder shape. Capacitive electrodes with 2 cm^2 disposed in concentric rings along the container walls established the four terminal impedance measurement configuration.

Results from filling and emptying the container with a volume of 10 cm3 in each step and using the same solution yielded the curves of *fig. 3* when the device was injecting a 50 kHz sine. A residual volume of 10 cm^3 was set as initial volume for filling procedure to allow the solution to entirely cover current-injecting and voltage-sensing electrodes. Conductance values for lower volumes were thus obtained by extrapolation.

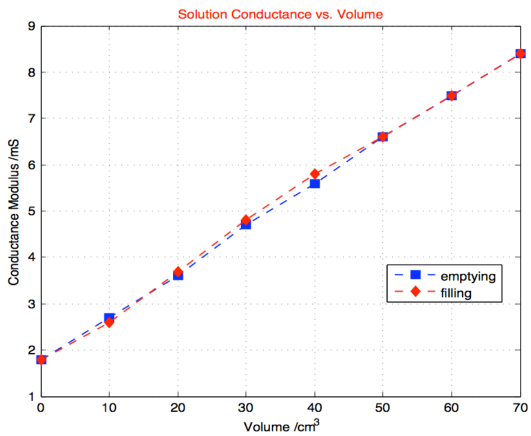

Figure 3 Volume-related conductance measurements for filling and emptying with a 0.141 Sm^{-1} solution.

The second set of experiments were performed using biological beings at 50 kHz. The prototype was tested in humans with capacitive electrodes placed on top of a cotton fabric over the skin at specific locations. Galvanic isolation was further achieved by inserting the electrodes within an elastic belt.

As bladder filling is very slow in time, thoracic conductance variations due to blood flow in the chest were firstly recorded to evaluate device performance to track physiological events that unfold in seconds. A typical conductance recording obtained by the device at the level of thorax is shown in *fig. 4*. Also shown in the figure is an ECG recording obtained independently by a commercial recording system to synchronize conductance changes with heartbeat.

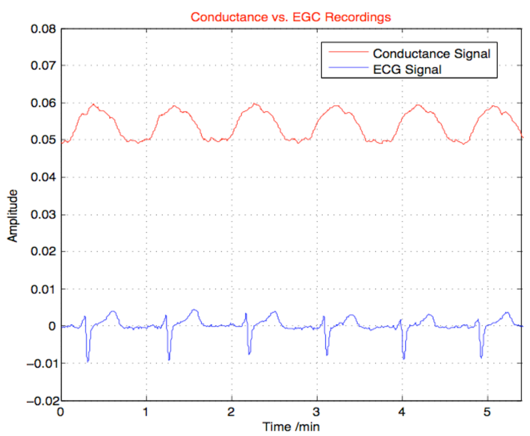

Figure 4 Thoracic conductance recording obtained by the prototype with ECG synchronisation (aVR lead).

Digital processing of ECG and conductance signals was made by means of a moving average with an equivalent low-pass filtering cut-off frequency of 10 Hz.

Finally, for monitoring conductance variations in the lower abdominal region as consequence of bladder filling and emptying, one human subject (young male in his mid-twenties with no known bladder problems) was wired to the prototype in a seated position and allowed to behave normally while drinking water in a gradual way to speed up urination urge. Current-injecting electrodes were attached at the skin near the iliac crest of hipbones in each side of human body whereas voltage- sensing electrodes were disposed along the semicircle drawn by hipbones and placed equidistantly in the lower abdomen. Again, galvanic contact was avoided by placing the electrodes within an elastic belt and over a cotton cloth.

The results of real-time monitoring are shown in *fig. 5* for the raw data given by the prototype and polynomial curve fitting.

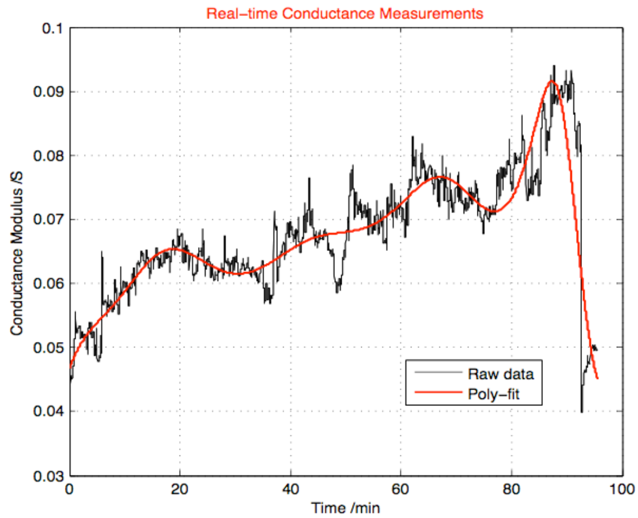

Figure 5 Conductance recording obtained during bladder filling and emptying processes with curve fitting**.**

Although histologic characterization of tissues is not yet the major aim of the electronic device, it can be developed as a future application taking advantage of the multi-frequency capability of the prototype to obtain the so-called Cole-Cole models and determine the amount of affected tissues since cancerous tissues have a higher reactance.

5. ACKNOWLEDGEMENT

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