Vital Sign Monitoring Using Capacitive Sensing

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Abstract—Unobtrusive vital sign monitoring is essential for most wearable monitoring applications. This paper presents a new method for unobtrusive vital sign monitoring based on the change in capacitance of passive sensors in contact with the user’s body. Unlike traditional capacitive coupling that is commonly used for elimination of the DC component of bioamplifiers, we propose monitoring of heart rate using galvanically isolated capacitive sensors. This paper presents the theory of operation and the implementation of a prototype device for heart rate monitoring. A prototype capacitive sensor was implemented using parallel coplanar plates in contact with the tip of the finger. Preliminary testing allowed us to observe changes in capacitance in the range of 0.1-0.3 pF per heartbeat, or 0.3-0.9% of the measured signal. The detected heart rate was validated using a PPG sensor positioned on the same hand. During a single recording session of 772 s, the detected heart rate was within 3 BPM 42% of the time. Proposed method is sensitive to motion artefacts; however, processing of signals collected from multiple contact points may facilitate removal of motion artifacts. Since many microcontrollers directly support high quality monitoring of capacitance on multiple I/O pins, the proposed method provides a promising solution for a low-power, low-cost, multi-contact monitoring.

I. INTRODUCTION

Low power and robust vital sign monitoring is essential for ubiquitous monitoring [1], [2] and health applications in smart homes [3]–[5]. Traditional methods for wearable monitoring include bioamplifiers (ECG), optical methods (pulse oximeters in smart watches), and piezoelectric/piezoresistive sensors. Capacitive coupling is frequently used with bioamplifiers as a method of AC coupling and elimination of the DC offset of the input signal. Electrical impedance plethysmography has been proposed as a method of vital sign monitoring, but it requires 4 electrical contacts with the skin or body tissue and electrical stimulation [6]. We propose new method of monitoring of the capacitance of user interface, instead of the use of capacitance as a part of the signal conditioning circuit of the vital sign monitoring.

Present generations of microcontrollers support direct measurement of capacitance on several pins to support touch sense interface and human-computer interface. Microcontrollers, such as NXP MK20DX256VLH7 Cortex-M4 [7], support capacitive measurements using Touch Sense Input (TSI) interface supported on 12 pins that allows up to 12 capacitive sensors connected directly to the microcontroller without signal conditioning hardware.

In this paper we present theory of operation relevant for the proposed method of vital sign monitoring, sensor design, calibration procedure, and real-time processing of sensor signals. System implementation and preliminary results are presented; sensor design and evaluation will be presented in the follow-up papers.

II. METHODS

A. Capacitance Measurement

Detailed description of the TSI interface and configuration can be found in Chapter 50 of Reference Manual [8]. Principal organization of the TSI interface and measurement controller is shown in Fig. 1.

One terminal of the sensor capacitance is connected to the ground and the other terminal is connected to one of pins with TSI interface. Internal controller uses charge current source (Ic in Fig. 1) and discharge current source (Id) to charge and discharge external capacitor in each cycle Tc. Voltage on external capacitor during charging will be

\[ V_{in} = \frac{1}{c} \int_{0}^{t} i(t) dt = \frac{1}{c} \int_{0}^{t} I_c dt = \frac{Ic t}{c} \]  

(1)

Voltage on external pin during a single charge/discharge cycle will change as presented in Fig 2. \( V_{i1} \) and \( V_{i2} \) represent internal thresholds that determine start and end of charge/discharge periods, and ultimately length of the cycle time \( T_c \).

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From (1)

\[ \Delta V = V_{t2} - V_{t1} = \frac{I_c \cdot T_c}{C} \]  

(2)

For \( I_c = I_d \) cycle time is equal to

\[ T_c = \frac{2C \cdot \Delta V}{I_c} = k \cdot C \]

(3)

and fundamental frequency of the waveform is equal to

\[ F_c = \frac{1}{T_c} \]

(4)

Therefore, cycle time is proportional to measured capacitance. Moreover, if the measured capacitance is a function of the measurement frequency \( f \), we can adjust measurement frequency by changing \( I_c \) and \( \Delta V \) according to (3).

TSI uses internal oscillator as a reference to measure cycle time \( T_c \), and returns count, or number of cycles of internal oscillator proportional to measured capacitance as shown in (3). Number of scans (NSCN) and prescale factor (PS) are programmable [8]. Default TSI module sensitivity is 0.02 pF/count.

TSI module can be configured according to application requirements. We use minimum charging current \( I_c \) (2 \( \mu \)A or 4 \( \mu \)A). Longer measurements (larger value of NSCN) produces smaller noise but increases measurement time that limits applicability to real time monitoring of vital signs. Minimum recommended sampling frequency for heart rate variability analysis is 250 Hz, which limits sampling interval to 4 ms.

B. Sensor Design and Configuration

Measured capacitance (Fig. 1) depends on dielectric constant \( \varepsilon \) of material between electrodes, symbolically represented in Fig. 1. According to Gauss’ Law, charge on electrode can be found as the area integral of the electric field \( E \) over any closed surface \( A \) divided by permittivity of the medium \( \varepsilon \) as shown in (4).

\[ \oiint E \cdot dA = \frac{Q}{\varepsilon} \]

(5)

If electrodes of the sensing capacitor are separated, total capacitance will depend on the medium surrounding electrodes. We implemented sensor with two coplanar plates of length \( l \), width \( w \) and gap/separation \( s \) on a single layer of the printed circuit board (PCB), as presented in Fig. 3.

Figure 3. A sample PCB capacitive sensor with coplanar plates

Ratio of the trace width \( w \) and separation \( s \) can be represented as

\[ r = \frac{s}{s+2w} \]

(6)

Capacitance of the sensor with coplanar plates \( C_{sc} \) can be found as a function of \( r \) (6) using (7) [9]:

\[ C_{sc} = \frac{\varepsilon_r \cdot \ln\left(\frac{2}{\sqrt{1-r^2} - 1}\left(\frac{\sqrt{1-r^2} + 1}{\sqrt{1+r^2}}\right)\right)}{377 \cdot \pi \cdot \varepsilon_0}, \]  

for \( 0 < r \leq \frac{1}{\sqrt{2}} \)

(7)

where \( \varepsilon_r \) is relative dielectric constant, and \( v_0 \) is speed of light. In our implementation \( s \approx w \) and condition (5) is satisfied for all sensor configurations we tested. For \( s=w=5\text{mm} \) and \( l=15\text{mm} \) and coplanar plates in vacuum \( (\varepsilon_r=1) \) sensor will have capacitance \( C_{sc} = 0.2 \) pF. Total capacitance of the sensor will depend on the dielectric constant of the medium around the plates \( \varepsilon_r \). Typical dielectric constant of the PCB material (FR4) is 4.4, dielectric constant of Kapton tape used for galvanic isolation is very similar - 3.4 [10]. Therefore, according to (7), capacitive sensor implemented on the PCB with Kapton insulation should have capacitance of \( C_{sc} \approx 0.93 \) pF. Capacitance of the sensor measured using GwInSTEK high precision LCR meter LCR-816/CR was 4.74 pF, which means that the equivalent dielectric constant of the environment was \( \varepsilon_r \approx 5 \), that is in agreement with dielectric constants of materials used for sensor implementation.

Since the sensor capacitance depends on the dielectric constant of the surrounding medium (5), we can design capacitive sensor to measure dielectric properties of the biological tissue, as represented in Fig. 4. Permittivity of the finger tissue \( (\varepsilon_r) \) can be modeled as a parallel capacitor \( C_r \) with capacitance determined by the properties of the tissue.

Figure 4. Coplanar sensor configuration. Tissue permittivity determines change of capacitance that can be modeled as parallel capacitor \( C_r \).

Permittivity of the tissue will change in time as a result of pulsating blood through capillaries for each heart beat. Therefore, capacitance of the sensor will change with each blood pulse and can be used for the vital sign monitoring.

Since the charge on electrodes is determined by the area integral, alternative sensor configuration could use stacked electrodes, as represented in Fig. 5. In this configuration, total capacitance is dominated by the characteristic of the medium between electrodes; however, total capacitance will still have a component determined by the tissue capacitance \( C_r \). This configuration would allow true single point measurement that is not feasible for potential monitoring using bioamplifiers.
Capacitance measured by the TSI is influenced by other factors, such as parasitic capacitances, capacitances of wires and traces between microcontroller and the sensor, and capacitance of other objects in the vicinity of the sensor. To assess absolute changes of measured parameters we implemented calibration procedure as described in the next section.

**B. Calibration Procedure**

Capacitance measured by the TSI interface during calibration and monitoring is symbolically represented in Fig. 6. with the following components:

- $C_i$ – interface capacitance includes parasitic capacitance of the interface and capacitance of PCB traces and wires connecting the sensor,
- $C_s$ – sensor capacitance (5) without body contact,
- $C_v$ – variable capacitance caused by the body tissue in contact with the sensor ($C_v(t)$), and
- $C_{cal}$ – capacitance used for the calibration of the sensor.

During calibration procedure we used calibration capacitor $C_{cal} = 35 \, \text{pF} @ 2\,\text{KHz}$ (measured using Gw\textsuperscript{2}NSTEK high precision LCR meter LCR-816/CR). Capacitance of the sensor on PCB only using the same LCR meter was 4.74 \, \text{pF}.

Calibration procedure included the following measurements:

- Calibration capacitor only:
  $$C_1 = C_i + C_{cal} \quad (8)$$
- Calibration capacitor and sensor (no touch):
  $$C_2 = C_i + C_{cal} + C_s \quad (9)$$
- Calibration capacitor and sensor touched with a finger:
  $$C_3 = C_i + C_{cal} + C_s + C_x \quad (10)$$

- Sensor (no touch):
  $$C_4 = C_i + C_s \quad (11)$$
- Sensor touched with a finger:
  $$C_5 = C_i + C_s + C_x \quad (12)$$

Using data from measurements ($C_1$-$C_3$) and (8)-(12) we can find calibration constant

$$cc = \frac{C_2 - C_1}{C_3 - C_1} = 4.74 = 0.0163 \, \left[ \frac{\text{pF}}{\text{count}} \right] \quad (13)$$

Capacitance of the finger touch $C_x$ from (11), (12), and (13) is

$$C_x = cc \cdot (C_5 - C_4) = 14.87 \, \text{[pF]} \quad (14)$$

Touch capacitance depends on skin properties (e.g. wet hands will have significant higher capacitance than dry hands for the same user). There is significant difference in touch capacitance between users.

Calibration constant is also used to assess absolute change of capacity of the sensor with each heart beat.

**C. Experiment**

We implemented the prototype heart rate monitor using Teensy 3.2 [11] and simple capacitive sensor, as presented in Figs. 3 and 4. To minimize noise, the sensor was battery powered and interfaced with a Bluetooth interface to a monitoring computer. The sensor was touched by index finger of the subject. We used optical pulse sensor on the ring finger to monitor heart rate. Pulse sensor has analog output that was sampled by the same microcontroller used to measure capacitance of the touch sensor.

Samples of both signals are sent over Bluetooth with sampling frequency of 240Hz.

Change of the touch capacitance with each heart beat is presented in Fig. 7. For this figure, we used higher resolution of capacitance measurement (average TSI count value around 47,800). Signal variation in this segment of the signal is very small (0.3 – 0.4%), but sufficient to detect heart beats.

**D. Signal Processing**

Reliable monitoring of vital signs requires careful design of signal processing procedures. Presence of high frequency noise is evident in Fig. 7. Sudden motion of finger produces large abrupt changes of measured capacitance. Both issues are typical for most bioelectrical signals (e.g. ECG). Noise was first implemented by oversampling the signal at 960 Hz and downsampling and filtering signal to 240 Hz. After downsampling we implemented baseline removal and low pass filtering of the signal. Baseline removal is implemented as linear detrending in the processing window (four seconds). The resulting signal is low-pass filtered. Heart beats are processed similar to Pan and Tompkins QRS detection algorithm [12]. Inter beat intervals detected from PPG in Fig. 7 are 1103 ms and 1062 ms, and from capacitance signal (lower plot) are 1097 and 1070 ms respectively.
A working prototype has been used to measure the performance of different sensor designs and inter-person variability of the recorded signal. In this paper we present the most important findings for assessment of feasibility of the proposed sensing method. Average amplitude of heart beat changes was 5-10 counts that is equivalent to 0.1 – 0.3 pF per heart-beat, or approximately 0.3 – 0.9% of the measured signal (approximately 35 pF, see (12) and (13)). In a 772 s record, heart rate was detected within 3 BPM from the PPG reference 42% of the time.

IV. DISCUSSION AND CONCLUSION

Preliminary results from our pilot study indicate feasibility of monitoring of vital signs using capacitive sensing. The most important findings of this study are:

- Pulsating blood provides sufficient change of the dielectric constant for monitoring of heart rate.
- Commercially available processors [8] provide sufficient resolution of TSI measurement for vital sign monitoring applications.
- Commercially available processors provide multiple TSI pins (e.g. 12 pins for MK20DX256VLH7) that can be used to monitor vital signs on multiple contact points with a user. For example, a smartphone might feature multiple monitoring points on the rim of the phone. That would allow robust monitoring, making it likely that at least some segments are in good contact with the hand of the user. Some of the current mobile phones use optical sensors to monitor heart rate of the user, but they have to place finger exactly on the optical sensor and start the monitoring application.
- Proposed method is sensitive to motion artifacts; however, collective processing of signals from multiple sensors might provide robust artifact removal.
- Capacitance measurement is much more power efficient than optical measurement that is currently mostly used for pulse monitoring.
- Inexpensive and power efficient sensing of the heart rate can be easily embedded in objects of everyday use to enable continuous and reliable physiological monitoring of users. We call this concept Smart Stuff [4].
- Proposed concept can be used for in-vivo analysis of body tissues as replacement of bioelectrical impedance analysis, due to small size, galvanic isolation, and adjustable frequency. Adjustable scanning frequency might allow application and condition specific monitoring of tissue properties [13].

REFERENCES