Data Communication System for Sensorized Guidewire Application

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***Abstract*—Resistive and capacitive sensors are commonly used in biomedical applications for measuring the various physiological signals. A low power and high resolution sensor interface circuit is critical to achieve the required sensing range and system specifications. One of the commonly employed interface circuit for biomedical application is relaxation oscillator based sensor interface architecture, due to its low power and small area consumption characteristic. However, the resolution of the relaxation oscillator based circuit is limited by the output frequency counting scheme. In this paper, we present an improved counting scheme applied to the sensorized guidewire system that is used to manoeuvre through the arteries and blood vessels to locate areas where stenosis occurs. Measurement results shows that by using the improved counting scheme, the sensing resolution is increased from 8.1 bits to 10.55 bits.**

***Index Terms*—Biomedical, relaxation oscillator, communication system**

I. Introduction

Resistive and capacitive sensors are widely used in biomedical applications to measure various parameters such as flow, pressure, temperature, and chemical species like glucose, protein and DNA [1]-[5]. In [1], nanowire-based piezoresistive sensors are embedded into vascular prosthetic graft to monitor blood flow rate. Capacitive sensors are used in [5] to measure intracranial pressure. In this paper, the application of resistive sensors embedded in guidewire for detecting blockage in the vascular vessels is presented.

The fundamental building block for any sensorized system is a sensor interface circuit. A conventional sensor interface architecture, which consists of an analog front-end (AFE), a programmable gain amplifier (PGA) and an analog-to-digital converter (ADC), provides good performance in terms of sensing resolution and dynamic range, but has the limitation in reducing its power [1] and silicon area [2] consumption to the extreme, which is often required in biomedical applications. One well-known approach to address this constraint is converting the sensor output into a form of frequency by using an oscillator circuit [4]. An RC relaxation oscillator is commonly chosen due to its low power and small area consumption. Other advantages include constant frequency tuning gain and continuous phase readout [6].

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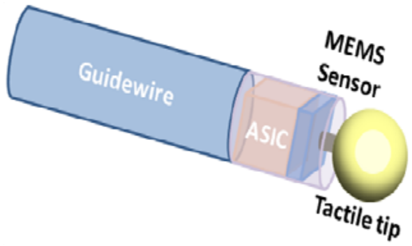
One of the major factors that limit the resolution in the relaxation oscillator based sensor interface architecture is its output frequency counting scheme. The output frequency from the sensor interface architecture is obtained by counting the number of positive-edges within a fixed period of time. The overall system resolution is limited by both the minimum least significant bit (LSB) steps of the positive-edges number and its dynamic range. Since the number of positive-edges is indirectly proportional to the output oscillation period, the minimum LSB step of the positive-triggered edges number becomes the dominant factor for limiting the system resolution. In a typical biomedical application, the sensor interface circuit is implanted inside the human body together with the sensors, and thus oscillator with low oscillating frequency is often utilised to reduce the power consumption. However, low oscillator frequency also leads to poor system resolution, as the number of positive-edges within the fixed time frame is lower.



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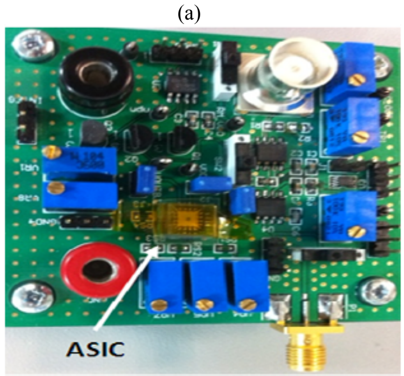

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In this paper, to mitigate the issue of low resolution relaxation oscillator sensor interface architecture, an improved output frequency counting scheme is proposed. A fixed period of timing window is generated using the sensor interface architecture while an external crystal oscillator with higher frequency operates within this time frame. The output frequency from the sensor interface architecture is obtained by counting number of positive-edges from the crystal oscillator within this timing window. Since the number of positive-edges is directly proportional to the output oscillation period, the overall system resolution can be improved since the system dynamic range is the only

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dominant limitation factor. Furthermore, in terms of biomedical application, the external counting circuit is placed outside the human body, and thus has less stringent power requirements. This paper presents the application of the improved output frequency counting scheme in a sensorized guidewire system. Section 2 provides the background information and resolution requirements for sensorized guidewire application. Section 3 describes the working principles of the improved output counting scheme. The measurement results comparing the conventional counting scheme with our improved counting scheme are presented in Section 4, followed by conclusion in Section 5.

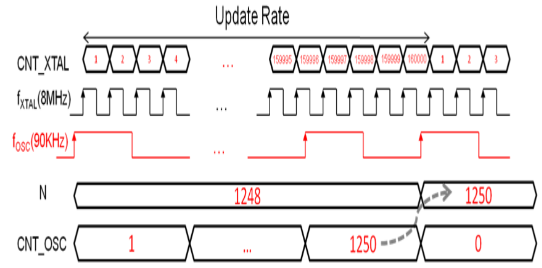


Fig. 2. Conventional output frequency counting scheme timing diagram

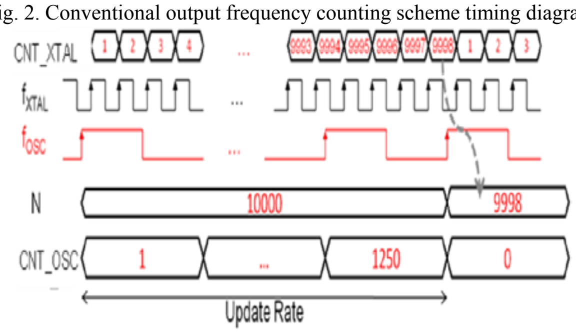


Fig. 3. Improved output frequency counting scheme timing diagram

II. Sensorized Guidewire System Overview

Guidewires are commonly used in angioplasty surgery to manoeuvre through the arteries and blood vessels to locate areas where stenosis occurs. Sensorized guidewires is realised using microelectromechanical system (MEMS) sensor to provide tactile feedback to the surgeon [2]. Fig 1a shows the sensorized guidewire assembly used for our application. The nominal resistance of our MEM sensor is 250 kΩ and the sensing range is from 0 to 40 mN. The sensor is able to achieve a resolution of 0.1 mN, which is represented by 0.05% change in the nominal resistance. The sensor resolution of 0.1 mN provides sufficient resolution to differentiate between normal blood vessel wall and stenoses (blockage). Based on the sensor specifications, the LSB that the ASIC has to detect is approximately 50 Ω and the dynamic range has to be more than 2000 steps, which is approximately 11 bits resolution. The core of the ASIC is a relaxation oscillator that converts resistance to frequency. The output frequency range for our sensorized guidewire assembly ranges from 65 kHz to 90 kHz. In our design, the ASIC uses a 20 ms sampling window to read out the resistance value, the length of the window is selected to ensure adequate refresh rate is achieved. The reader module shown in Fig 1b samples the ASIC output window length using an 8 MHz clock and transfers the information through USB connection to a laptop, where the data is translated to pressure information and displayed as a graphical image.

III. Improved Output Frequency Counting Scheme

In oscillator based sensor interface architecture, the output frequency is directly related to the sensor information. Conventionally, the output frequency from the sensor interface architecture is obtained by counting the number of positive-edges within a fixed period of timing window, which is usually, generated using an external crystal with a higher oscillation frequency shown in Fig 2. The number of oscillator edges during a fixed counting window period is given by

$$N = \frac{T_{CNT}}{T_{OSC}} \tag{1}$$

where T_{CNT} is the sampling window period and T_{OSC} is the relaxation oscillator output period

The number of oscillator edges is indirectly proportional to the output oscillation period. In order to achieve M-bit resolution in the sensor interface architecture, the counting scheme must satisfy the following requirements:

1) The Dynamic Range of the COUNTING scheme Must Be larger than 2M steps

$$N_{MAX} - N_{MIN} \geq 2^{M} \tag{2}$$

where N_{MAX} and N_{MIN} correspond to the number of oscillator edges at the shortest and longest output period respectively.

2) The Minimum LSB Step Nmin Must be Larger Than 1

$$T_{CNT}\left(\frac{1}{\left(T_{OSC,MAX} - \frac{T_{OSC,MAX} - T_{OSC,MIN}}{2^{M}}\right)} - \frac{1}{T_{OSC,MAX}}\right) \geq 1 \tag{3}$$

The second requirement is usually the dominant one in the conventional frequency counting scheme. In our design, the counting window period is fixed at 20ms. The sensor interface output period varies from 11.11 to 15.38 us and the corresponding N_{MAX} and N_{MIN} is 1800 and 1300 respectively. Equation (3) shows that the sensor interface can only achieve 8.5 bits in resolution, which fall short of the 11 bit resolution required for our system. The system resolution can be improved by extending the sampling window length or by increasing the oscillator frequency at the expense of lower refresh rate and higher power consumption.

Alternatively, an improved output frequency counting scheme that does not affect the refresh rate and power consumption is proposed to improve the resolution of this sensor interface architecture. In the improved scheme, a fixed period of timing window is generated using the sensor interface architecture while an external crystal oscillator with higher frequency operates within this time frame. The output frequency from the sensor interface architecture is obtained by counting number of positive-edges from the crystal oscillator within this timing window, as shown in Fig 3.

The number of positive-edges from the crystal oscillator is directly proportional to the sensor interface output oscillation period and is given by

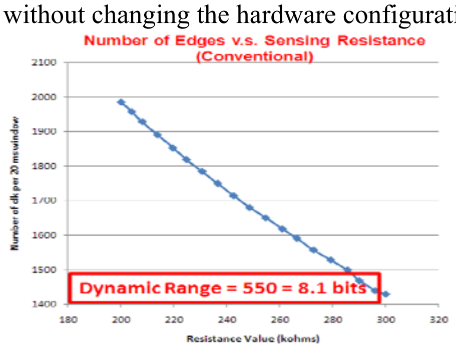
$$N = \frac{T_{OSC,MAX}}{T_{CLK}} \tag{4}$$

where T_{CLK} is the period of the external crystal oscillator. The length of the counting window becomes variable and is dependent on the maximum oscillator output period:

$$T_{CNT} \leq T_{OSC,MAX} * Sample \tag{5}$$

where Sample is the fixed number of clock edges from the sensor interface to generate the counting window

In order to achieve M-bit resolution in the sensor interface architecture, this improved counting scheme must satisfy the dynamic range requirement only. The minimum LSB of step N_{MIN} is always 1 since the relationship between the number of oscillator edges and the output oscillation period is now linear. In our design, the maximum counting window period must be less than 20 ms. The minimum and maximum output period of the sensor interface is 11.11 and 15.38 µs respectively. From equation (5), the fixed number of clock edges from the sensor interface to generate the counting window in 1250. Thus, the longest length of the counting window is 19.23 ms. By counting number of edges from the 8 MHz external crystal oscillator within these counting windows, the minimum and maximum number of edges, N_{MAX} and N_{MIN}, are 111100 and 153840 respectively. As a result, the sensor interface architecture, by deploying this improvised counting scheme, can achieve 15.5 bits in resolution, which is a significant improvement when comparing with its conventional counterpart. The improved output frequency counting scheme met the required system resolution without changing the hardware configurations.



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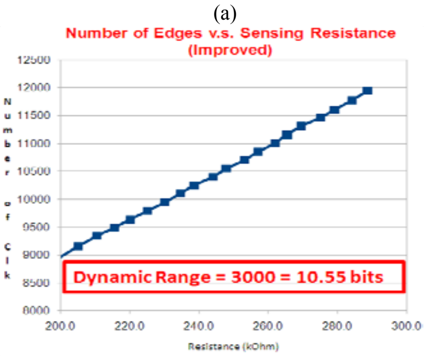

(b)

Fig. 4. Measured dynamic range of the relaxation oscillator based sensor interface with (a) the improved output counting scheme and (b) conventional output frequency counting scheme.

IV. Measurement Results

The reader module shown in Fig 1b is used to provide the measurement results for comparing our improved frequency counting scheme with conventional counting scheme. The main components on the reader module are variable resistors, ASIC and the microcontroller. The variable resistors are used to emulate the MEM sensor resistance value. The ASIC reads in the resistance value and output its equivalent frequency representation. The microcontroller running on an 8 MHz clock reads the ASIC output frequency and provides the interface for data transfers to laptop/tablet through USB connection.

For the conventional frequency counting scheme measurement setup, the microcontroller is programmed to read the oscillator edges output from the ASIC for a period of 20 msec for each resistance values. The resistance value is inversely proportional to the number of oscillator edges detected as shown in Fig 4a.

For the improved counting scheme measurement setup, the microcontroller determines the sampling period by counting 1250 oscillator edges generated by the ASIC. The sampling period varies proportionally with the oscillator period. The microcontroller calculates the length of the sampling period by counting the number of 8 MHz clock edges in the window. The resistance value is proportional to the number of clock edges in the sampling window as shown in Fig 4b.

The measured dynamic range performance of the relaxation oscillator based sensor interface with the proposed and conventional output frequency counting scheme is presented in Fig 4. The sensor interface with the conventional and proposed counting scheme achieves a dynamic range of 550 counts and 3000 counts respectively, which can be translated to 8.1 and 10.55 bits resolution. The improved counting scheme shows significant improvement over the conventional counting scheme. Jitter performance of the relaxation oscillator is the main factor limiting the sensor interface architecture using improved counting scheme from achieving better resolution. By employing the improved frequency counting scheme, the bottleneck for high resolution oscillator based sensor interface architecture is shifted from counting scheme to jitter performance of the oscillator.

V. Conclusion

An improved frequency counting scheme for oscillator based sensor interface architecture is proposed in this paper for sensorized guidewire wire application. A comparison between conventional counting scheme and our improved counting scheme shows that the system resolution can be increased from 8.5 bits to 15.5 bits resolution using the same setup without incurring additional power consumption and lowering the system refresh rate. However, based on the actual measurement results, the actual system resolution only increases from 8.1 bits for conventional counting scheme to 10.55 bits for our improved counting scheme. The discrepancy from the theoretical value is due to the jitter performance of the relaxation oscillator. By using the improved frequency counting scheme, higher resolution oscillator based sensor interface can be realised.

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