WIRELESS MEMS ACCELEROMETER FOR REAL-TIME
SMALL LABORATORY ANIMAL ACTIVITY
MONITORING

by

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Wireless MEMS Accelerometer for Real-Time Small Laboratory Animal Activity Monitoring

Abstract

by

Cheng-Kuan (Bobby) Lu

Exploring genomic functions requires accurate physiological data acquisition to reveal the connection between genomic structure and medical symptoms. Most of these genetic experiments begin by changing certain genomic sequences of laboratory mice and then monitoring their physiological behavior. Physiological parameters typically include blood pressure, EKG, EEG, heart rate, body temperature, and body activity. In order to acquire accurate data with minimum distortion, a miniaturized wireless sensor without battery is highly desired.

This thesis focuses on wireless activity sensor development. A lateral-axis differential capacitive MEMS accelerometer is designed and fabricated in MUMPs process. This device occupied an area of 1 mm×1 mm and achieves a nominal capacitance value of 1.4 pF with a differential sensitivity of 4 fF/g. The low frequency activity signal sensed by the accelerometer is first modulated to a 1.8 kHz sub-carrier frequency and then transmitted by an FM Colpitts VCO transmitter operating at 30 MHz. The prototype wireless activity sensing system occupies an area of 33 mm×22 mm and weighs 5 grams including battery. The system has been characterized with an untethered laboratory mouse. The received signal shows various acceleration waveforms corresponding to different types of mouse activities. The MEMS accelerometer and interface circuit can be potentially integrated with other biological sensors to achieve a miniature implantable biosensing system.
Chapter 1

Introduction

1.1 Motivation

DNA sequencing of laboratory mice together with in vivo real time biological information, such as blood pressure, temperature, activity, and bio-potential signals, is ultimately crucial for systems biology research, genetic function discovery, and new treatments development for diseases such as hypertension, obesity, epilepsy, and cancers [1]. A miniature, and long-term, reliable bio-sensing implant network with two-way telemetry capability is highly desirable to capture a dynamic behavior of a biological system as shown in Figure 1.1. A implanted sensing microsystem records biological information and then transmits the information wirelessly through the top coil antenna to a near by computer for further data analysis. The batteryless implant unit is also powered wirelessly though the top coil to reduce the form factor. Existing commercial implant devices are inadequate to achieve the objectives due to large size and weight, causing severe post-implant trauma, data distortion, and limited functionalities. Figure 1.2 presents a general architecture of the proposed implantable bio-sensing network, which consists of an array of micro-fabricated bi-
ological sensors, low power interface and data telemetry electronics with an on-chip microprocessor control unit. This research focuses on the development of a wireless MEMS-accelerometer-based activity sensing channel for real-time small laboratory animal activity monitoring.

Figure 1.1: *In Vivo* wireless bio-sensing network

Figure 1.2: Implantable micro-system architecture

1.2 Research Summary

In this research, a MEMS capacitive accelerometer interfaced with a custom-designed integrated circuit (IC) and Colpitts oscillator transmitter form a wireless activity sensor for small laboratory animal activity monitoring. The accelerometer converts
body activities to an electrical signal, which can be transmitted through a wireless link. A nearby receiver detects and records the signal for further analysis. The prototype MEMS accelerometer exhibits a nominal capacitance of 1.4 pF with a differential sensitivity of 4 fF/g. The device achieves a resonant frequency 7.8 kHz and a noise floor of $720 \, nV/\sqrt{Hz}$ in ambient, which is equivalent to $65 \, \mu g/\sqrt{Hz}$. The prototype accelerometer and interface IC are connected by bond wires on a 10 mm $\times$ 3 mm flexible substrate, which weighs 35 milligrams with a measured sensitivity of 16 mV/g. The overall wireless activity sensing system occupies an area of 33 mm $\times$ 22 mm and weighs 5 grams including battery and other discrete components. The system was characterized with an untethered laboratory mouse. The received signal shows various acceleration waveforms corresponding to different types of mouse activities. The activity signal along with other real-time vital signals will be important for advanced medical and genetic research. The MEMS accelerometer and interface electronics can be potentially integrated with other biological sensors to achieve a miniature implantable biosensing system.

1.3 Thesis Outline

The thesis is organized as follows. Chapter 2 reviews current activity and energy expenditure measurement techniques. Also explained is how mouse activity characterization performed by using a commercial accelerometer determined the prototype accelerometer design parameters. Chapter 3 describes the prototype sensor design, analysis, simulation, and characterization results. Finally, a commercial SOI-MEMS accelerometer fabrication process flow is presented. This process will be employed to fabricate the prototype accelerometer. Chapter 4 presents the prototype wireless telemetry system design, analysis, and characterization results. Chapter 5 reports the in vitro measurement results with the prototype activity sensor attached on the
back of a laboratory mouse. Conclusion and future work of the research are given in Chapter 6.
Chapter 2

Background

2.1 Activity Characterization

2.1.1 Metabolism Measurement

There are several ways to define the activity of a laboratory mouse. The first one is measuring metabolism, which is also called energy expenditure (EE). This method measures the heat expenditure of a mouse to calculate their activity energy [2]. The energy expended by mouse can be calculated by using the Conservation of Energy Law as presented in Equation 2.1.

\[
\sum E_{\text{AirIn}} + \sum E_{\text{Mouse}} = \sum E_{\text{AirOut}}
\] (2.1)

In order to obtain the in and out air energy, air temperature, pressure, and flow rate entering and exiting the mouse cage are carefully monitored and recorded. The mouse energy expenditure is the energy difference between entering and exiting air.

The second method to measure energy expenditure is to measure oxygen con-
sumption. There is a direct relationship between body energy expended and oxygen consumption. Animals require oxygen to generate cell energy source: adenosine triphosphate (ATP). The ATP generating equation is listed in Equation 2.2. Energy expended can be found by accurately measuring oxygen and carbon dioxide concentration.

\[ C_6H_{12}O_6 + 6O_2 \rightarrow 6CO_2 + 6H_2O + ATP \] (2.2)

Both energy expended measurement methods require a highly regulated environment. Creating such an environment is difficult and expensive. Decreasing the size of a mouse cage reduces the cost of building a regulated environment. However, a limited cage space will likely restrain mice activity. The above methods are difficult to use in a geneticist’s laboratory. Some researchers, therefore, have turned to alternative methods such as implantable biological monitoring system.

Figure 2.1(a) shows a commercial implantable small laboratory animal biological monitoring system, which is capable of recording body temperature, heart rate, and activity [3]. This system includes a recording base connecting to a computer through a RS-232 port, and an implantable sensor shown in Figure 2.1(b).

![Figure 2.1: (a)Commercial small animal activity monitoring system; (b)Implantable activity sensor](image)

The base system contains several receiving coils on a printed circuit board (PCB),
which partition the cage area into a 2-dimensional grid. As a mouse moves, the position of implanted transmitter changes as well. The system is able to count how often the mouse moves from one grid to another. The information is useful only if the mouse has mobile activities such as running around in the cage. Problems, however, would occur if the mouse only exhibits stationary activities, thus resulting in an inaccurate estimation of its energy consumption.

This discrete-components-based design requires a large ferrite coil to meet the power dissipation requirements. The size and weight of the implantable device are 3 cm and 2.2 grams, respectively. Compared with the typical size and weight of a laboratory mouse (6 - 8 cm and 20 grams)[4], the device is relatively large. The bulky sensor will cause severe post-implant trauma and restrain mouse activity.

Some researchers have studied the correlation between body acceleration and energy expense [5]. They place several accelerometers on different parts of body and record acceleration waveforms. By comparing the calorimetry measurement and acceleration data researchers are able to determine the correlation between accelerations and energy expense of regulated exercise, such as walking or resting. Researchers also tried to enhance the accuracy of body energy expended by increasing the numbers of accelerometer. The configuration of placing accelerometers on human hip, back and wrist sites has been studied as the best energy expenditure calculation method [6]. The same correlation on laboratory rats has also been studied [7]. The result shows that the most significant accelerations occurring along the head-on direction. By analyzing the acceleration data, researchers can readily identify different behaviors of rodents such as walking, running, resting, and standing.

This research work will use a single-axis accelerometer as the activity sensor. The advantages of using accelerometer as activity sensor include good accuracy of energy expense, easy laboratory usage, and integrability of other sensors and electronic system, which can significantly reduce the total system size and weight to avoid post-
implant trauma and data distortion.

### 2.1.2 *In Vitro* Acceleration Characterization

In order to obtain the design specifications of the prototype accelerometer, *in vitro* characterization was first performed on a laboratory mouse with a commercial accelerometer.

An Analog Device accelerometer, ADXL320 [8], was chosen due to its small form factor and low power consumption, which allows a miniature battery to be used. The accelerometer can measure acceleration with a full range of $\pm 5$ g and exhibit sensitivity of $174 \text{ mV/g}$. In this test, the sensor bandwidth was set at 100Hz. The entire testing system occupies an area of $15 \text{ mm} \times 12 \text{ mm}$ with a total weight of 2 grams, including a coin lithium battery. The accelerometer was attached on a laboratory mouse’s back as shown in Figure 2.2. The acceleration signal was acquired by National Instrument Data Acquisition card and recorded in a computer.

![Figure 2.2: Laboratory mouse activity measurement setup](image)

The recorded acceleration signal is shown in Figure 2.3(a). Most acceleration signal is between $\pm 1$ g. Therefore, $\pm 2$ g acceleration is chosen as the maximum
Figure 2.3: Laboratory mouse activity measurement results.
design input signal range for this design. The large acceleration signals above 2 g were caused by collision between the PCB attached on the mouse back and its housing cage. Figure 2.3(b) shows the frequency content of the acceleration data. As can be seen, most body motions are below 3 Hz, mouse respiration is around 6 Hz, and heart rate is around 11 Hz. The typical heart rate and respiratory rate of a laboratory mouse are 680 beats per minute and 220 per minute, respectively [4]. The acceleration signal amplitudes due to heart rate and respiration are higher than 10 mg. Therefore, it is chosen as the minimum detectable signal amplitude for this application. The design requirement of the prototype accelerometer is therefore desired in Table 2.1.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum sensing range</td>
<td>±2g</td>
</tr>
<tr>
<td>Minimum sensing resolution</td>
<td>10mg</td>
</tr>
<tr>
<td>Information bandwidth</td>
<td>&lt; 20 Hz</td>
</tr>
<tr>
<td>Sensing direction</td>
<td>X-axis or Y-axis</td>
</tr>
</tbody>
</table>

Table 2.1: Accelerometer design specification

2.2 Summary

An overview of commercial mouse energy expenditure measurement technique is presented in this chapter. The advantage of using accelerometer for small laboratory animal activity monitoring is described. A in vitro activity characterization by using a commercial accelerometer is performed to obtain the design guideline for the prototype MEMS accelerometer design.
Chapter 3

MEMS Accelerometer Design

This chapter presents the design of the prototype MEMS accelerometer for a small laboratory animal activity sensing system. MEMS accelerometer design procedure and noise source are described. The simulation and testing results verify the prototype accelerometer design performance.

3.1 Accelerometer Architecture

This work focuses on the development of a MEMS accelerometer for small laboratory animal activity monitoring. The accelerometer converts body motion (activity) to electrical signal, which can be processed and wirelessly transmitted to a nearby receiver. A capacitive sensing architecture is chosen for its high sensitivity, low noise and low power dissipation. The proposed MEMS capacitive accelerometer architecture is shown in Figure 3.1. The devise consists a suspended structure with integrated sensing fingers between anchored fingers. An external acceleration generates a force applied to the suspended structure, causing a lateral displacement, $\Delta x$. The displacement results in a corresponding increase and decrease capacitances between the
sensing fingers and \( C_{\text{top}} \) and \( C_{\text{bottom}} \) stationary fingers. Thus, the sensor can be modeled as a set of differential sensing capacitors \( C_s^+ \) and \( C_s^- \) with a common reference node, \( C_{\text{com}} \).

![Figure 3.1: Illustration of a capacitive accelerometer architecture](image)

**3.2 Accelerometer Sensing Mechanics**

**3.2.1 Accelerometer Dynamics**

The accelerometer can also be modeled as a spring-mass-damper second-order mechanical system as shown in Figure 3.2. A force, \( F \), generated by input acceleration acting on the mass, \( m \), causes a displacement, \( x \). The differential equation describing the system response is given by Equation (3.1), where \( b \) is the damping coefficient, and \( k \) is the spring constant.

\[
F(t) = m \frac{\partial^2 x}{\partial t^2} + b \frac{\partial x}{\partial t} + kx
\]  

(3.1)

Converting Equation 3.1 to its Laplacian domain, the frequency response of this second-order system can be expressed as:
Figure 3.2: Spring-mass-damper model of an accelerometer

\[ F(s) = (ms^2 + bs + k)X(s) \]  \hspace{1cm} \text{(3.2)}

Replacing \( s \) with \( j\omega \), a system transfer function can be found as:

\[ H(j\omega) = \frac{X(j\omega)}{F(j\omega)} = \frac{\frac{1}{m}}{(j\omega^2) + \frac{b}{m}j\omega + \frac{k}{m}} \]  \hspace{1cm} \text{(3.3)}

Two key parameters of the system are the natural resonant frequency, \( \omega_{\text{res}} \), and the quality factor, \( Q \), which can be expressed as:

\[ \omega_{\text{res}} = \sqrt{\frac{k}{m}} \]  \hspace{1cm} \text{(3.4)}

\[ Q = \frac{\omega_{\text{res}} m}{b} = \frac{\sqrt{mk}}{b} \]  \hspace{1cm} \text{(3.5)}

Substituting Equations (3.4) and (3.5) into Equation (3.3) yields:

\[ H(j\omega) = \frac{\frac{1}{m}}{(j\omega)^2 + \frac{\omega_{\text{res}}^2}{Q}j\omega + (\omega_{\text{res}})^2} \]  \hspace{1cm} \text{(3.6)}
Figure 3.3 plots the magnitude of $H(j\omega)$ versus frequency. The transfer function can be expressed as $\frac{1}{k}$ for angular frequency much lower than $\omega_{res}$. At the resonant frequency, $\omega_{res}$, the transfer function has a magnitude of $\frac{Q}{k}$. Above $\omega_{res}$, the magnitude decreases as inversely proportional to the square of angular frequency.

\[ |H(j\omega)| \]

Figure 3.3: Magnitude response of second-order mass-spring-damper system vs. angular frequency

The resonant frequency of the MEMS accelerometer should be designed above the desired system bandwidth. From Equation (3.4), the displacement and acceleration can be further related as:

\[ \frac{x}{a} = \frac{1}{\omega_{res}^2} \]

(3.7)

Therefore, the minimum detectable input acceleration, $a_{min}$, will correspond to a minimum proof mass displacement by
\[ x_{\text{min}} = \frac{a_{\text{min}}}{\omega_{\text{res}}^2} \]  

(3.8)

The term of \( x_{\text{min}} \) is a key parameter to determine the minimum capacitance change of the accelerometer. The detailed capacitance calculation will be presented in Section 3.3.1.

### 3.2.2 Accelerometer Noise Characteristics

The accelerometer thermal-mechanical noise, also known as Brownian noise, is a critical design parameter. This noise signal is caused by thermally excited structural vibration. The equivalent acceleration noise power spectral density due to Brownian noise can be expressed as:

\[ a_n^2(\omega) = \frac{4k_bTb}{m^2} = \frac{4k_bT\omega_{\text{res}}}{mQ}, \]  

(3.9)

where \( k_b \) is the Boltzmann constant, \( T \) is the absolute temperature, \( b \) is the damping coefficient, and \( m \) is the structural mass. Since \( \omega_{\text{res}} \) is determined by the sensing bandwidth, it is important to maximize the structural mass in a given area to suppress the Brownian noise, thus enhancing device sensitivity. A typical resonant frequency of an accelerometer can be 1 kHz or above. An excessively low device resonance would call for low suspension compliance thus easy to break during handling. On the other hand, a large resonant frequency would result in a stiff structure, thus reduced sensitivity. As a compromise, a device resonance of 10 kHz is chosen for the prototype design. With an estimated proof mass of 14 \( \mu \)grams and \( Q \) of unity in ambient, the prototype sensor’s equivalent Brownian-noise-induced acceleration noise floor is approximately 30 \( \mu \)g/\( \sqrt{\text{Hz}} \). For an activity bandwidth of 20 Hz, the device equivalent acceleration noise is 134 \( \mu \)g, which is low enough for the proposed activity sensing application.
3.3 Capacitive MEMS Accelerometer Design

3.3.1 Accelerometer Capacitance

The interdigitated sensing fingers shown in Figure 3.4 form a set of differential sensing capacitors. The nominal capacitance of each side can be determined as:

\[ C_{so}^+ = \varepsilon_o lt \left( \frac{1}{x_1} + \frac{1}{x_2} \right) \times N = C_{s1}^+ + C_{s2}^+, \tag{3.10} \]

where \( \varepsilon_o \) is the dielectric constant in air, \( l \) is the finger overlap length, \( t \) is the finger thickness, \( x_1 \) and \( x_2 \) are the corresponding gap sizes between adjacent fingers, and \( N \) is the number of sensing finger sets.

\[ \Delta C_s^+ = \varepsilon_o lt \left( \frac{1}{x_1} \Delta x_1 - \frac{1}{x_2} \Delta x_2 \right) \times N = \left( C_{s1}^+ \frac{\Delta x_1}{x_1} - C_{s2}^+ \frac{\Delta x_2}{x_2} \right) \times N \tag{3.11} \]

Figure 3.4: Accelerometer sensing fingers structure

When the incoming acceleration pushes the shuttle, the gap space between adjacent fingers is changed by \( \Delta x \) corresponding to capacitance change of \( \Delta C_s^+ \):

Therefor, \( x_2 \) needs to be designed much larger than \( x_1 \) to minimize sensitivity degradation. However, for a given accelerometer chip area, large \( x_2 \) will result in a decreased number of finger sets and the overall capacitance, hence, the sensitivity. Calculations
were performed to determine the optimal gap spacing ratio, $\frac{x_2}{x_1}$. Figure 3.5 shows the calculation result for a device length of 1mm, finger length, $l$, of 100 μm, and device structural thickness, $t$, of 25 μm. The maximum sensitivity can be obtained with the gap spacing ratio of 2.5. In the prototype design, $x_1$ is chosen to be 2 μm, which is the minimum gap size achievable by the fabrication technology, and $x_2$ is chosen to be 8 μm, thus resulting a ratio of 4 as a compromise between the device sensitivity and performance susceptibility to the accuracy of the vertical sidewall etch profile.

![Graph showing capacitance change per unit displacement vs. gap spacing ratio for a fixed device length of 1 mm](image)

Figure 3.5: Accelerometer capacitance change per unit displacement vs. gap spacing ratio for a fixed device length of 1 mm

The prototype accelerometer is laid out in an area of 1 mm × 1 mm with 189 sets of sensing fingers on each side of proof mass. The fingers exhibit an overlap dimension, width, and thickness of 96 μm, 2 μm and 25 μm, respectively, thus achieving a 2 pF nominal capacitance. The total mass of the accelerometer is estimated as 14 μgrams. From Equation (3.8), the displacement sensitivity is depended on the sensor resonant frequency. In this prototype design, a 10 kHz resonant frequency is chosen, thus resulting in displacement sensitivity of 2.45 nm/g. With a minimum gap size of 2 μm and gap spacing ratio of 4, the capacitive sensitivity is determined as 4.9 fF/g.
3.3.2 Accelerometer Mechanical Suspension

The spring constant along the $\Delta x$ direction for a suspension structure shown in Figure 3.6 can be determined as:

$$k = 2Et\left(\frac{w}{l}\right)^3$$  \hspace{1cm} (3.12)

Figure 3.6: Detail of accelerometer spring structure

where $l$ is the beam length, $w$ is the beam width, $t$ is the beam thickness, and $E$ is the Young’s modulus of the structural material.

The structural thickness is limited by the MUMPs process device layer to be 25 $\mu$m. The beam width is chosen to be 2 $\mu$m, which is the minimum feature size of the technology. With the silicon Young’s modulus of 160 GPa, a beam length is determined to be 106 $\mu$m to achieve a spring constant of 53.7 N/m, which is equivalent to a resonant frequency of 10 kHz for a shuttle mass of 14 $\mu$grams.

3.4 Electronic Noise Floor

The electronic thermal noise from interface circuitry is the other dominant noise source in the overall system design. Figure 3.7 shows an electrical model of the prototype sensor and interface circuit system. The sensing circuitry is a continuous
time synchronous detection capacitance-to-voltage convert architecture. The MEMS accelerometer is modeled as two differential capacitors, which are driven by a stimulation clock signal with an amplitude of $V_s$. The MEMS sensor is interfaced with a differential charge amplifier converting the sensor capacitance change to an output voltage. A 1 MHz clock frequency modulates the low frequency acceleration signal away from the $1/f$ noise of the charge amplifier. An input common mode feedback circuit suppresses any offset signal due to parasitic capacitance mismatch and drift over time. The modulated signal is then mixed with the same clock frequency followed by a low pass filter to obtain the desired acceleration information [9].

![Electrical model of sensor and interface electronics](image)

Figure 3.7: Electrical model of sensor and interface electronics

The differential voltage amplitude at the output of the charge amplifier, $V_{\text{ampout}}$, for a capacitance change, $\Delta C_{\text{sense}} = \Delta C_{s+} - \Delta C_{s-}$, can be expressed as:

$$V_{\text{ampout}} = \frac{\Delta C_{\text{sense}}}{C_I} V_s$$

(3.13)

All electronic noise in the charge amplifier can be referred to a single effective noise source at its input. The differential noise voltage amplitude at the amplifier output thus can be determined as:

$$\tilde{V}_{\text{outnoise}} = \frac{C_T}{C_I} \sqrt{\frac{\tilde{V}_n^2}{Hz}} \sqrt{BW},$$

(3.14)

where $C_T$ is the total capacitance at the charge amplifier input, which includes the
nominal sensor capacitance, amplifier input capacitance, integrating capacitance, input common-mode feedback capacitance, and other parasitic capacitances, $\sqrt{V^2_{n}}$ is the input-referred electronic noise power spectral density, and $BW$ is the signal bandwidth of interest.

The signal to noise ratio (SNR) at the charge amplifier output can be determined as:

$$SNR = \frac{\Delta C_{sense}V_s}{C_T \sqrt{V^2_{n}} \sqrt{BW}}$$

(3.15)

The SNR needs to be greater than unity for practical applications. The prototype activity sensing system exhibits $C_T$ of 6.6, $V_s$ of 1.2 V, and a power spectral density of input-referred noise of $5 \text{nV}/\sqrt{Hz}$. With a signal bandwidth of 20 Hz, an SNR of 100 is expected at the amplifier output when an input acceleration signal of 10 mg is applied. The noise floor at the output of the low pass filter can be determined as $310 \text{nV}/\sqrt{Hz}$, corresponding to an equivalent acceleration sensing resolution of 25 $\text{μg}/\sqrt{Hz}$.

3.5 Simulation Results

A finite element simulation program, ANSYS, was used to verify the dynamic behavior of the MEMS accelerometer. The simulated device first-mode resonant frequency is 9632 Hz, closely matching the designed value as shown in Figure 3.8(a). Other higher order modes of resonant frequencies are all at much higher frequencies and would not affect the accelerator performance. Figure 3.8(b) shows a 1g DC acceleration applied to the proof mass, resulting in 2.6 nm displacement.

A capacitance calculation program, FastCap, was used to estimate the nominal capacitance and the capacitance sensitivity to acceleration. The total sensor nominal capacitance value is determined to be 2.6 pF with a sensitivity of 5.15 fF/g. These
values are closely matched with the designed values.

(a) Simulation of resonant mode.

(b) Simulation of displacement corresponding to an input acceleration of 1 g acting on a 14 microgram mass

Figure 3.8: ANSYS simulation of the prototype MEMS accelerometer

3.6 MUMPs Fabrication Process Flow

The prototype MEMS accelerometer sensor was fabricated by a silicon-on-insulator (SOI) multi-user process (MUMP) provided by MEMSCAP. The thickness of device layer, oxide, and substrate for SOI wafer utilizing in MUMP process are 25 μm, 1 μm, and 400 μm, respectively. The major process flow is shown in Figure 3.9. First,
a negative photoresist is deposited and patterned, and then a metal stack of 20 nm chrome and 500 nm gold is deposited for interconnect and pad metal. Etching of the photoresist leaves metal only in the patterned areas as shown in Figure 3.9(b). Next, the silicon device layer is patterned and etched by using a deep reactive-ion etch (DRIE) to define the accelerometer structure as shown in Figure 3.9(c). Then, a protective coating is applied to front side of the wafer, and the backside substrate is patterned. Another DRIE process removes the patterned substrate silicon to expose the insulating oxide. Figure 3.9(d) shows the cross-section view of the structure after HF vapor etch releasing structure from insulating oxide. The front-side protection layer is then removed, and a separate wafer is patterned and bonded to the device wafer as shown in Figure 3.9(e) as a mask for depositing a blanket metal layer of 50 nm chrome and 600 nm gold to reduce sensor series resistance. The final device structure is shown in Figure 3.9(f) after removing the shadow mask on top.
(a) SOI wafer for fabrication process

(b) Depositing and patterning pad metal

(c) Patterning and etching silicon structural layer
Figure 3.9: SOI-MUMPs fabrication process flow
3.7 Accelerometer Characterization

Figure 3.10 shows the SEM pictures of the fabricated accelerometer with close-up views of sensing fingers and mechanical suspension. The measured differential nominal capacitances are 1.53 pF and 1.34 pF.

The fabricated MEMS accelerometer is interfaced with a custom-designed low noise electronic IC [9]. The accelerometer and interface IC are bond-wired on a flexible substrate exhibiting a dimension of 3 mm × 10 mm with a mass of 35 milligrams, as shown in Figure 3.11. Accelerometer sensitivity was measured by comparing the differential DC output voltages when accelerometer was positioned flat on a test bench and positioned perpendicular to the bench. The measured acceleration sensitivity is 16 mV/g, which is slightly lower than the designed value of 20.2 mV/g due to larger finger gap size than expected.

The measured sensor noise floor in ambient is $720 \text{nV/}\sqrt{\text{Hz}}$ at 1 kHz as shown in Figure 3.12, which corresponds to an acceleration noise floor of $65 \text{μg/}\sqrt{\text{Hz}}$. The sensor was then placed in to a vacuum chamber with a pressure of 30 mtorr. Figure 3.13 presents the corresponding noise measurement. The peaking at 7.8 kHz indicates the accelerometer resonant frequency. The noise floor is reduced to the electronic noise floor of $320 \text{nV/}\sqrt{\text{Hz}}$ due to an increased quality factor at low pressure. The quality factor is estimated as 16 from -3 dB bandwidth. The measured noise floor in ambient is, therefore, dominated by the Brownian noise, which is higher than the designed value ($30 \text{μg/}\sqrt{\text{Hz}}$) due to smaller structural mass and lower quality factor.

To test the accelerometer frequency response, the sensor was attached on a type 4810 mini shaker table, which was driven by an amplified signal from a waveform generator. The acceleration signal was recorded by a dynamic signal analyzer and compared to another reference acceleration signal measured by a Laser Doppler Vi-
Figure 3.10: SEMs of the fabricated SOI accelerometer

(a) SEM of entire accelerometer

(b) SEM of sensing fingers

(c) SEM of mechanical suspension
Figure 3.11: MEMS accelerometer with CMOS electronics over a flexible substrate

Figure 3.12: Measured noise floor in ambient of the prototype sensing system

Figure 3.13: Measured noise floor in vacuum of the prototype sensing system
brometer (LDV). The testing setup is shown in Figure 3.14. Figure 3.15(a) presents the frequency response of the accelerometer normalized by the LDV-measured acceleration. At low frequencies, frequency response is close to unity, which means the acceleration measured by MEMS accelerometer is identical to the LDV measurements. At 8.2 kHz, frequency response reaches its peak point and then sharply decreases above 8.2 kHz. This characteristic represents a high \( Q \) system and is matched to previous noise floor measurement in ambient. Comparing this response to the other measurements performed with other sensors fabricated in the same run [10], the resonance peak at 8.2 kHz is likely caused by testing setup structure. Table 3.1 compares the designed and measured values for several key parameters of the prototype accelerometer.

![Figure 3.14: The prototype system dynamic response testing setup](image)
Figure 3.15: Normalized acceleration vs. frequency

<table>
<thead>
<tr>
<th>Accelerometer Parameters</th>
<th>Designed Values</th>
<th>Measured Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 Resonant Frequency</td>
<td>9968 Hz</td>
<td>7800 Hz</td>
</tr>
<tr>
<td>2 Nominal Capacitance $C_{so}$</td>
<td>2 pF</td>
<td>$\sim$ 1.5 pF</td>
</tr>
<tr>
<td>3 Displacement Sensitivity</td>
<td>2.45 nm/g</td>
<td>4.08 nm/g</td>
</tr>
<tr>
<td>4 Acceleration Sensitivity</td>
<td>20 mV/g</td>
<td>16 mV/g</td>
</tr>
</tbody>
</table>

Table 3.1: Comparison of designed and measured value of several key parameters of the prototype accelerometer
3.8 Summary

In this chapter, the design of a MEMS accelerometer for small laboratory animal activity monitoring is presented. The prototype sensor achieves the required sensing resolution and bandwidth by following the outlined design methodology. Simulation and testing were performed to verify the prototype sensor design and performance specifications.
Chapter 4

Activity Sensor Telemetry System

This chapter presents the design of the prototype wireless telemetry system for the proposed small laboratory animal activity monitoring. The major design focuses on achieving a low power and reliable voltage-controlled oscillator (VCO) transmitter for transmitting activity signal to a nearby receiver.

4.1 Proposed Wireless Activity Sensing System

The accelerometer is first interfaced with a custom-designed capacitance-to-voltage IC to convert the acceleration to an analog voltage. The low frequency bandwidth acceleration signal (< 20 Hz) is not suitable for be direct transmitting by the prototype transmitter to a WiNRADiO computer controlled receiver due to receiver’s bandwidth limitation. The WiNRADiO receiver is for audio application, which would filter out any information below 50 Hz. The acceleration signal, therefore, needs be modulated before transmission to a higher sub-carrier frequency in audio bandwidth before transmission. The computer record the modulated signal at the sub-carrier frequency from the output of the receiver. The modulated information signal can be further extracted
by a envelope detector. The proposed wireless activity sensing system architecture is shown in Figure 4.1. The low frequency activity signal (< 20 Hz) sensed by the accelerometer is first modulated to a 1.8 kHz sub-carrier frequency. The sampled signal is then FM-transmitted by a Colpitts voltage-controlled oscillator transmitter operating at 30 MHz to a nearby receiver. The 1.8 kHz sub-carrier frequency is chosen to be compatible with the bandwidth of a commercial receiver (WiNRADiO G305e) used for prototype system design and demonstration. The received signal is recorded by a computer with audio recording software. The laboratory mouse activity signal is further extracted by a Labview-based envelope detector.

![Figure 4.1: Proposed wireless activity sensing system architecture](image)

**4.2 Basic Oscillator Principle**

An oscillator is a positive feedback closed-loop system operating at a selected frequency. Figure 4.2 shows that an oscillator contains an amplifier with frequency dependent forward loop gain, $H_A(j\omega)$, and a frequency dependent feedback network, $H_F(j\omega)$.

The overall transfer function can be expressed in Equation 4.1:

$$\frac{V_{out}}{V_{in}} = \frac{H_A(j\omega)}{1 - H_F(j\omega) H_A(j\omega)}$$

(4.1)

To develop an oscillation, the loop gain, $H_F(j\omega)H_A(j\omega)$, must equal to unity, and $\angle H_F(j\omega)H_A(j\omega)$ must equal to zero. To ensure a reliable oscillation start up, the
small-signal loop gain is typically designed between 3 to 5 at the chosen oscillation frequency of $\omega_0$ with a proper loop phase shift.

### 4.3 Prototype Oscillator Configuration

![Colpitt's oscillator schematic](image)

Colpitt’s oscillator configuration is employed for the prototype oscillator design due to its simplicity. Figure 4.3 presents the schematic of the prototype Colpitt’s oscillator. The oscillator consists of a BJT, $Q_1$, functioning as a common-base amplifier, biasing resistors, $R_1$, $R_2$, and $R_E$, a resonant LC tank, and capacitive feedback network, $C_1$ and $C_2$. In the resonant LC tank, $C_V$ is a varactor whose capacitance value is varied by the 1.8 kHz-modulated acceleration signal. $C_C$ and $R_C$ are the
blocking capacitor and resistor, respectively. $C_3$ is an additional capacitor parallel to $C_V$ to tune the total tank capacitance value for desired oscillation frequency. The entire telemetry system is powered by a 3.6 V re-chargeable polymer lithium battery, which exhibits a light weight of 1.6 g and a current driving capability of more than 2 mA.

### 4.3.1 Oscillator Small-Signal Loop Gain Calculation

The small-signal oscillator model is shown in Figure 4.4. The oscillator small-signal loop gain, $A_l$, is determined by the transistor characteristics, resonator impedance, and the values of feedback capacitors.

![Small-signal model of the prototype oscillator](image)

Figure 4.4: Small-signal model of the prototype oscillator

The transistor model is based on a common gate configuration, where $R_i$ is equal to $\frac{1}{g_m}$, and $g_m$ is the small-signal transconductance of $Q_1$. $V_{tank}$ and $V_1$ are voltages at the emitter and collector of the BJT, respectively. $R_P$ represents the equivalent resistive loading of the LC resonator consisting of $L_1$ and all other tank capacitances at resonant frequency. $R_P$ can be derived at resonance as:

$$R_p = Q\omega_0 L = \frac{Q}{\omega_0 C}, \quad (4.2)$$
where $Q$ is the unloaded quality factor of the LC tank. In the prototype oscillator design, the entire LC tank $Q$ is dominated by the inductor which is approximately 25 at 30 MHz.

The transfer function of the feedback network, $H_F$, can be determined as:

$$H_F = \frac{V_1}{V_{tank}} = n = \frac{C_1}{C_1 + C_2} \quad (4.3)$$

The tapped capacitive network used in the prototype design is shown in Figure 4.5. Its input impedance, $R_{in}$, can be determined by the following equations as:

$$R_{in} \approx \frac{R_i}{n^2} = \frac{1}{g_m n^2} \quad (4.5)$$

Figure 4.5: Tapped capacitive network

Therefore, the resonant tank overall loading resistance can be determined as:

$$R_{load} = R_P \parallel R_{in} = R_P \parallel \frac{1}{g_m n^2} = \frac{R_P}{n^2 R_P g_m + 1} \quad (4.6)$$

assuming transistor, $Q_1$, provides a negligible loading. Equation 4.6 can be further simplified by using Taylor series as:
\[ R_{load} \approx R_P(1 - n) \quad (4.7) \]

with an assumption that \( g_mR_Pn \), is approximately equal to unity.

Therefore, the small-signal loop gain, \( A_l \), of the oscillator can be expressed as:

\[ A_l = g_mR_Pn(1 - n). \quad (4.8) \]

### 4.3.2 Oscillator Components Value Selection

The oscillator steady-state frequency can be determined:

\[ f = \frac{1}{2\pi \sqrt{LC_{eq}}} = \frac{1}{2\pi \sqrt{L_1 \left( \frac{C_VC_C}{C_V+C_C} + C_3 + \frac{C_1C_2}{C_1+C_2} \right)}} \quad (4.9) \]

where \( L_1 \) is the inductor, \( C_V \) is the varactor, \( C_C \) is the blocking capacitor, \( C_1 \) and \( C_2 \) are the feedback capacitors.

A nominal frequency of 30 MHz and wide-FM modulation (FMW) are chosen for the prototype design. The wide-FM bandwidth of WiNRADiO G305e receiver is 230 kHz. A surface mount inductor of 560 nH is chosen in this prototype design due to its small size and high Q (approximately 25 at 30 MHz). In this frequency band, the capacitors exhibits high Q value (> 1000). Therefore, the tank loading at resonance is dominated by the inductor loss. With the inductor value, capacitance at normal frequency and maximum capacitance change (\( \Delta C_{max} \)) in the oscillator can be calculated as:
The 230 kHz frequency deviation is chosen as the overall bandwidth. The $C_{total}$ and $\Delta C_{max}$ would be 50 pF and 770 fF, respectively. $C_{total}$ is the $C_{eq}$ in Equation 4.9 with $C_V = (C_V^+ + C_V^-)/2$.

$$\frac{1}{2\pi \sqrt{L(C_{total} - 0.5\Delta C_{max})}} = 30.115 MHz$$

$$\frac{1}{2\pi \sqrt{L(C_{total} + 0.5\Delta C_{max})}} = 29.885 MHz$$

Figure 4.6: Varactor tuning curve

However, the CV converter output voltage offset caused by nominal capacitance difference between $C_S^+$ and $C_S^-$ is 272 mV, which inherently causes in a varactor capacitance change between 32 pF and 26 pF as shown in Figure 4.6. The CV converter will also have an output signal range of ±32 mV corresponding to a maximum input acceleration signal of ±2 g. The output signal voltage range would result in an additional capacitance change of 600 fF. The total varactor capacitance range including the DC offset and maximum acceleration is between 32.3 pF and 25.7 pF. The overall 7.6 pF capacitance change needs to be reduced to less than 770 fF, corresponding to the VCO output frequency variation of 230 kHz. A capacitor, $C_C$, of 15 pF is, therefore, chosen to achieve this capacitance range reduction. As a result, the CV converter
output DC-offset-voltage causes a frequency change of 200 kHz and maximum input acceleration signal contributes 30 kHz bandwidth. The overall tank capacitance to voltage sensitivity, $\Delta C/\Delta V$, thus becomes 2.2 fF/mV. With the accelerometer sensitivity of 16 mV/g, the required minimum detectable acceleration, 10 mg, results in $\Delta C_{\text{min}}$ of 0.35 fF, which corresponds to a frequency shift of 100 Hz, and this is larger than the WiNRADiO receiver resolution of 1 Hz.

Then, $C_1$, $C_2$ and $C_3$ can also be chosen through the following steps. The $C_3 + (C_1 \parallel C_2) = C_{\text{total}} - (C_V \parallel C_C) = 40$ pF, and we chose $C_3 = 20$ pF. $C_1$ and $C_2$ values can be calculated by:

\[
\frac{C_1 C_2}{C_1 + C_2} = 20 \text{pF and} \\
\frac{C_1}{C_1 + C_2} = n = \frac{1}{5}.
\] (4.11)

Therefore, $C_1$ and $C_2$ are 25 pF and 100 pF, respectively.

![Figure 4.7: The prototype telemetry board layout](image)

A small-signal loop gain, $A_l = 4$ is chosen to ensure a reliable oscillation start up. With the values of $R_P$, $C_1$, and $C_2$, the $g_m$ of $Q_1$ can be calculated by equation (4.8) as 9.4 mS. $R_1$, $R_2$, and $R_E$ are determined to establish the collector current of $Q_1$ to
be $250 \, \mu A$, corresponding to $g_m$ of $9.6$ mS. All components values in this prototype oscillator design are listed in Table 4.1, and the telemetry board layout is shown in Figure 4.7.

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$f$</td>
<td>30 MHz</td>
</tr>
<tr>
<td>$L_1$</td>
<td>560 nH</td>
</tr>
<tr>
<td>$C_T$</td>
<td>32 pF</td>
</tr>
<tr>
<td>$C_C$</td>
<td>15 pF</td>
</tr>
<tr>
<td>$C_1$</td>
<td>25 pF</td>
</tr>
<tr>
<td>$C_2$</td>
<td>100 pF</td>
</tr>
<tr>
<td>$C_3$</td>
<td>20 pF</td>
</tr>
<tr>
<td>$C_F$</td>
<td>0.47 $\mu$F</td>
</tr>
<tr>
<td>$R_1$</td>
<td>80.6 k$\Omega$</td>
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<tr>
<td>$R_2$</td>
<td>66.5 k$\Omega$</td>
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<tr>
<td>$R_E$</td>
<td>4 $\Omega$</td>
</tr>
<tr>
<td>$R_C$</td>
<td>1 M$\Omega$</td>
</tr>
<tr>
<td>$I_{total}$</td>
<td>280 $\mu$A</td>
</tr>
<tr>
<td>$Q_{total}$</td>
<td>25</td>
</tr>
<tr>
<td>$R_P$</td>
<td>2650 $\Omega$</td>
</tr>
<tr>
<td>$A_I$</td>
<td>4</td>
</tr>
<tr>
<td>$\Delta f/10mg$</td>
<td>100 Hz</td>
</tr>
</tbody>
</table>

Table 4.1: Prototype oscillator components value

4.4 Telemetry System Testing Results

The system characterization includes the VCO testing and complete telemetry system evaluation.

4.4.1 VCO Testing Results

With the 30 MHz prototype oscillator components values listed in Table 4.1, the oscillation amplitude in the collector node can be determined as [11]:

$$A_{tank} \approx 2I_{bias}R_P$$  \hspace{1cm} (4.12)
With $I_{bias}$ of 250 \( \mu \text{A} \) and $R_P$ of 2650 $\Omega$, the amplitude can is expected to be 1.325 V. The emitter node is probed by an active probe to measure the RF power from a spectrum analyzer. The active probe has 10X attenuation driving a 50 $\Omega$ output load, thus expecting a power of -21.5 dBm to be defected by the spectrum analyzer.

The power spectrum and phase noise measurement of the prototype oscillator are shown in Figure 4.8. The measured frequency of 29.6 MHz is closed to the designed value, and the measured amplitude is larger than designed value by 1 dB likely due to variance of the inductor $Q$. The oscillator parameters comparison between designed and measured values is shown in Table 4.2.

(a) 30 MHz oscillation amplitude at emitter  
(b) Phase noise at emitter

![Figure 4.8: 30 MHz VCO measuring results](image)

<table>
<thead>
<tr>
<th>Oscillator Parameters</th>
<th>Designed Values</th>
<th>Measured Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frequency</td>
<td>30MHz</td>
<td>29.6 MHz</td>
</tr>
<tr>
<td>Amplitude (collector)</td>
<td>1.325 V</td>
<td>1.464 V</td>
</tr>
<tr>
<td>Phase Noise (@ 10 kHz)</td>
<td>—</td>
<td>-106 dBC/Hz</td>
</tr>
</tbody>
</table>

Table 4.2: Comparison of designed and measured values of several parameters of the prototype oscillator
The received power versus distance test was performed to ensure a reliable reception with respect to a laboratory mouse cage environment. The testing setup is shown in Figure 4.9. The receiver antenna is a single-ended wire and is taped on a desk. The prototype transmitter is placed above the receiver antenna with various distance. Figure 4.10 is the received power versus distance with different tilting angles. From this plot, the receiver is able to receive -81 dBm at 8 inches away from the prototype telemetry board with a 90-degree tilt, which ensures a reliable activity signal reception with a laboratory mouse bred in a cage with a similar dimension.

![Testing setup](image)

**Figure 4.9: Testing setup**

### 4.4.2 Telemetry System Characterization

The purpose of characterizing the complete telemetry system is to ensure that the computer is able to record the correct input signal applied at the VCO voltage tuning node. In this test, two different sinusoid waves (0.5 Hz / 28 mVpp and 0.5 Hz / 29 mVpp) from a function generator are applied to modulate the varactor inside the VCO transmitter. The received signal shows similar sinusoid waves with 75 unit difference corresponding to the 1 mV peak-to-peak amplitude difference as shown
Figure 4.10: Plot of received power vs. distance at various tilting angles

Figure 4.11: (a) Received signals corresponding to input signal of 0.5 Hz / 28 mVpp and 0.5 Hz / 29 mVpp; (b) 1 mVpp input signal amplitude difference corresponding to 75 unit difference.
in Figure 4.11. This means the required minimum acceleration resolution, 10 mg, which is equivalent to 12 units can be reliably detected by the WiNRADiO receiver. Therefore, the telemetry system performance achieve the design objectives.

4.5 Summary

In this chapter, the design and characterization of a prototype 30 MHz voltage-controlled-oscillator transmitter for wireless laboratory small animal activity sensing are presented. The VCO design procedure is described and followed to achieve the required bandwidth and sensing requirement. System characterization is performed to verify the designed specifications.
Chapter 5

Overall System Testing Results

This chapter presents the prototype activity sensing system, \textit{in vitro} mouse activity measurement test setup, and measurement results. Some observations are also described in this chapter.

5.1 The Prototype Wireless Activity Sensing System

The prototype system needs to be light weight to minimize loading for a small laboratory mouse. To achieve a light weight design, surface mount components are selected for implementing the transmitter design. The prototype activity sensing system exhibits an area of 33 mm $\times$ 22 mm and weighs 5 grams including battery. The system size can not be significantly reduced due to the battery diameter of 20 mm. The 3.6 V lithium re-chargeable coin battery is selected in order to support at least 2 mA of continuous discharge current. Figure 5.1(a) presents the prototype activity sensing system. Carbon fiber encapsulation is used to prevent damage during testing as shown in Figure 5.1(b). The components weight is summarized in Table 5.1.
Figure 5.1: Prototype activity sensing system

(a) Prototype activity sensing system

(b) Prototype activity sensing system with encapsulation

<table>
<thead>
<tr>
<th>Components</th>
<th>Weight</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensing Circuit</td>
<td>4 mg</td>
</tr>
<tr>
<td>Accelerometer</td>
<td>5 mg</td>
</tr>
<tr>
<td>Flexible Substrate</td>
<td>25 mg</td>
</tr>
<tr>
<td>PCB board</td>
<td>2.35 g</td>
</tr>
<tr>
<td>Battery holder</td>
<td>0.9 g</td>
</tr>
<tr>
<td>Battery</td>
<td>1.58 g</td>
</tr>
<tr>
<td>FM Transmitter</td>
<td>150 mg</td>
</tr>
<tr>
<td>Total</td>
<td>4.98 g</td>
</tr>
</tbody>
</table>

Table 5.1: Detailed components weight
5.2  *In Vitro* Wireless Mouse Activity Measurement

5.2.1 Measurement Setup

A normal laboratory mouse was chosen for the prototype system characterization. The mouse exhibits a body weight of 20 grams. The prototype sensing system was anchored on the mouse’s back by stitches and tape was added on to further secure the sensing and telemetry board as shown in Figure 5.2. The black arrow direction indicates the primary sensing axis. The computer-controlled the receiver is tuned at the prototype system transmission frequency. The receiver box detects the up activity signal through the antenna under the mouse cage. The computer then records the activity signal for further signal processing.

![Figure 5.2: In vitro wireless activity measurement setup. Prototype sensing system attached on the laboratory mouse back. The black arrow indicates the axis of activity sensing.](image_url)
5.2.2 In Vitro Mouse Activity Measurement Results

In vitro activity measurement was performed for 30 minutes. This measurement was also filmed by a camcorder to visually identify the acceleration waveform of different behaviors. According to the recorded video, from 0 to 2 minutes and 6 to 9 minutes, the mouse felt very uncomfortable of the monitoring system on its back and tried to run or walk for detaching the sensor. Between these two periods of running and 9 to 14 minutes, the mouse rest for most of time. After 14 minutes to the end of measurement, the mouse activity was alternated between running and resting. The acceleration data during a 30-minute period is presented in Figure 5.3. The baseline of -0.4 g was caused by the sensor’s tilted orientation while attached loosely on the mouse’s back.

![diagram](image)

Figure 5.3: Acceleration data obtained from the laboratory mouse

The activities were classified as three categories: resting, walking, and standing. The resting can be readily identified. During resting from 2 to 6 minutes, the laboratory mouse lay stationary yielding a near 0 g acceleration around -0.4 g due to tilting of the sensor attachment. When the mouse was walking and trying to detach the activity sensor, the acceleration data showed the waveform ranging between -0.4g to
-0.8g. Standing up was another readily identified activity. Depending on the standing angle of the mouse, the acceleration data showed a constant value around -0.8 g. Comparing the recorded acceleration waveform and video information, measurement demonstrates that the prototype wireless activity monitoring system is reliable for a practical application.

5.2.3 Additional Observation

In this in vitro activity measurement, there are some additional observations to point out. First, the prototype board was anchored by stitches and further secured by tape. Thus, the mouse was bonded to cause a limited activity. In this characterization experiment, the mouse walking behavior was more similar to jumping because the tape prevented the rear leg muscles from stretching. Second, the board size and weight also reduced activity level. The mouse rested more often than usual and spent some time trying to detach the telemetry board. The highest measured acceleration amplitude is lesser than 1g. However, it is anticipated that the acceleration amplitude of a freely moving mouse without a such external unit attached on its back should be larger than what is obtained in this experiment.

5.3 Summary

The prototype wireless activity monitoring system can detect, convert, and transmit the mouse activity signal to a nearby computer recording system. The measured data shows resting, standing, and running behaviors aided by simultaneous camcorder recording for visual identification. In the future, the mouse physical behavior profile can be identified with more measurement data, allowing the activity monitoring system to identify or estimate the activity type only through acceleration information.
Chapter 6

Conclusion and Future Work

6.1 Conclusion

In this research, a MEMS accelerometer activity sensor designed for small animal biosensing network is developed. Mouse activity was first characterized by a commercial accelerometer. The characterization results were used as a design guideline for the prototype accelerometer implementation. The prototype MEMS accelerometer was fabricated in a 25 μm SOI-MEMS process occupying a sensing area of 1 mm × 1 mm in a chip area of 2 mm × 2.4 mm, and exhibiting a nominal capacitance of 1.4 pF with a sensitivity of 4 fF/g. The accelerometer interfaced with a custom-designed IC over a flexible substrate exhibiting an area of 3 mm × 10 mm with a weight of 35 milligrams. The prototype sensor achieved a sensitivity of 16 mV/g, a sensing resolution of $65 \text{nV/} \sqrt{\text{Hz}}$ in ambient limited by the thermal mechanical noise, and a resonant frequency of 7.8 kHz. A prototype wireless link was developed to transmit the activity-induced acceleration signal to a nearby receiver. The prototype transmitter first modulated the low frequency activity signal to a sub-carrier frequency
of 1.8 kHz, followed a 30 MHz wireless FM transmission. The receiver receives the signal, which is then recorded by a computer for further data analysis. The prototype wireless activity sensing system can sense and transmit the required minimum acceleration signal of 10 mg for a least 30 cm. *In vitro* animal characterization showed that the prototype system is able to sense, transmit and record the small laboratory animal acceleration waveform due to physical motion. However, the size and weight of the prototype sensing system restrained the animal’s activity level. An improved performance can be expected by future system miniaturization.

### 6.2 Future Work

Although the sensing resolution demonstrated by the prototype system is sufficient for detecting mouse physical activity, package size and weight degraded and altered normal mouse physical behavior. For addition, it is highly desirable to reduce the package complexity for further system integration with implantable bio-sensing network. Therefore, a fabrication process capable of integrating MEMS accelerometer, sensing IC, and telemetry electronics on a same chip is highly desirable.

A post CMOS-MEMS process [12] will be considered for the proposed highly-integrated implantable bio-sensing network. The CMOS MEMS activity monitoring system is currently under developing as shown in Figure 6.1 [13]. This design integrates a pair of MEMS accelerometers at the center of the IC with sensing electronics along the peripheral area. The accelerometer is first fabricated in a commercial CMOS process followed by using a maskless post CMOS micromachining technique. The top metal layer defines the MEMS structure. A backside silicon etching is used to define the sensor structural thickness to about 50 μm as shown in Figure 6.2(a). In the next step a front side anisotropic SiO₂ etching removes the surface passivation layer and the dielectric materials uncovered by any top metal layer protection, as depicted in
Figure 6.2(b), followed by a silicon deep reactive ion etch (DRIE) to release the microstructures as shown in Figure 6.2(c). A final silicon isotropic etching is performed to undercut certain narrow isolation regions to achieve a proper electrical isolation, as illustrated in Figure 6.2(d). The overall sensing system occupies 2.2 mm × 2.2 mm area, as shown in Figure 6.1, and weighs 5 milligrams. In the future, this small size and light weight chip can greatly reduce the risk of post implant trauma and associated data distortion.

![CMOS-MEMS accelerometer die photo](image)

**Figure 6.1: CMOS-MEMS accelerometer die photo**

A Z-axis accelerometer is suggested to be included into the sensing network [7] to achieve a much accurate activity monitoring. The most significant acceleration occurs in the X-Z plan. Although the acceleration signal from x-axis accelerometer is able to identify resting, standing, and walking, a vertical axis accelerometer can provide more detailed behavioral characteristics.
Figure 6.2: Post CMOS-MEMS fabrication process flow


