A System Design with Mode-Switching Power Management for Closed-Loop Implantable Glucose Biosensor

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Abstract

This paper presents an integrated control circuit and system design for an implantable glucose biosensor. A microcontroller unit (MCU) is configured to work as the core control circuit for the whole system. An 8-bit MCU is chosen due to its low power characteristics. The front-end glucose sensing module utilizes a low-power wide-current-sensing-range readout circuit with a potentiostat for amperometric chemical sensors. Post layout simulations for the readout circuit show a redox current of 50 pA to 50 μA with a power consumption of 390 μW to 745 μW. A closed-loop insulin delivery system is integrated with the implanted module, which is controlled by the MCU. A power management chip which includes the closed-loop insulin delivery system is developed to meet the power requirements of implantable systems. The power management chip is fabricated in TSMC 0.35-μm 2P4M 3.3/5 V CMOS technology. It occupies an area of 2.382 × 2.374 mm² (core size = 2.200 × 1.086 mm²) and dissipates 36.3 mW of power. An in-depth analysis of the proposed system is undertaken. The experimental results match those of the circuit simulations.

Keywords: Glucose sensing, Readout circuit, Amperometric chemical sensor, Power management, Closed-loop insulin delivery

1. Introduction

Diabetes, an incurable disease, causes serious medical complications and leads to untimely death [1]. However, with tight glycemic control and early diagnosis, the number and severity of medical complications and medical costs can be greatly reduced. The major objective of glycemic control is to maintain a person’s blood glucose level within a physiologically acceptable range [2]. Frequent measurements of blood glucose levels are needed to determine how much insulin should be administered. However, patients often fail to maintain tight control due to pain caused during measurements and the high cost of glucose monitoring devices. An ideal treatment of diabetes requires a closed-loop insulin delivery system implanted in the human body. Such a system, which comprises a glucose sensing component and an insulin pump, acts as an artificial pancreas [3].

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To achieve closed-loop feedback, a clinically applicable implantable artificial pancreas requires the miniaturization and coordination of three components, namely a safe and reliable insulin infusion device, an accurate and stable glucose sensor, and a control system that regulates insulin delivery according to blood glucose variations assessed by the glucose sensor [4].

The continuous delivery of insulin based on the obtained glucose data is the key function of an artificial pancreas [3]. The glucose sensor should be able to rapidly provide highly precise blood glucose data to a microcontroller unit, whose function is to compute the proper amount of insulin based on the measured glucose level and control the administration of insulin [5].

Complementary Metal-Oxide-Semiconductor (CMOS) technology is commonly used for fabricating modern electronic systems [6]. Micro-sensors based on CMOS integrated circuit (IC) technology have received a lot of attention in the electronics industry, and especially in biomedical implant research [7-16].

The present study describes the design and implementation of an implantable bio-sensing system for closed-loop glucose monitoring that uses CMOS IC technology. The implantable
bio-sensing system provides a stable reference voltage to an electrode component for the precise measurement of blood glucose levels.

In this research, the proposed analog intellectual property (IP) and the designed reusable components, including a band-gap reference generator, an op-amp, and a comparator, were validated for application in glucose sensing control and feedback. The calibration algorithm used for programming the microcontroller unit (MCU) meets the required functional performance.

2. System features

Figure 1 shows the architecture of the proposed closed-loop glucose monitoring system, which comprises an amperometric glucose sensor readout circuit with a potentiostat, an MCU (for the control system), a 2-channel adjustable power supply, a power management IC, and a drug delivery subsystem. An Atmel Advanced Virtual RISC (AVR) MCU is utilized. The MCU digitizes the linear sensed glucose value obtained from the readout circuit and sends the corresponding drug driver signal. The readout circuit uses an amperometric method for glucose sensing. Since the target application of proposed system is implantable biomedical devices, the power management IC ensures that the system’s power level, battery charge, temperature, and related parameters are at appropriate levels. The drug driver is embedded in the power management IC to complete the closed-loop system for glucose monitoring. An LCD module is used for experimental verification, but could be used to display data to patients if necessary. The power management chip and readout chip were fabricated in TSMC 0.35-μm 2P4M 3.3/5V CMOS technology.

3. Circuit design and considerations

3.1 Amperometric glucose sensor readout circuit

3.1.1 Basic concept of current-mode readout circuit

Glucose sensor is a type of amperometric sensor [17]. Therefore, a readout circuit that operates in the current-to-current transfer mode should have very low input impedance to receive most of the generated sensor current. The resistance of electrochemical cells varies with the concentration of analyte [17]. A reliable readout circuit should thus be insensitive to the variation in sensor resistance.

3.1.2 Readout-1

The prior work of a low-power potentiostat circuit (referred to as Readout-1 hereafter) is illustrated in Fig. 2. \(V_{ox}\), a specific constant redox potential provided by a bandgap reference circuit, is independent of the temperature and supply voltage. The operational transconductance amplifier (OTA) compares \(V_{ox}\) to the potential at the reference electrode (RE), and sends a signal proportional to their difference to the gates of transistors M1 and M2. M1 serves as a pass device; it has a drain current equal to the redox current of the amperometric sensor.

![Figure 2. Readout-1 (low-power potentiostat) [18,19].](image)

When oxidation occurs near the working electrode (WE), a corresponding reduction appears near the counter electrode (CE). As a result, an electron current flows from the CE to the WE through the chemical solution and a corresponding conventional current flows from the WE to the CE. This redox current is proportional to the concentration of the chemical solution and is the drain current of M1. Since M2 is a mirror of M1, the same redox current also flows through M2 into its series resistor R1. Hence, the output \(V_o\) across R1 is proportional to the concentration of the solution.

M1 and M2 have identical sizes and both operate in the saturation region; their drain currents are thus functions of \(V_{ox}\) only. However, considering the effect of channel length modulation, \(V_{ox}\) causes a slight difference between their drain currents. In order to mitigate this mismatch, a cascode current mirror (M3 and M4) is used. The equivalent input resistance of Readout-1 \(R_{in}\) is taken from the drain of NMOS M3 where the sensor is connected.

\[
R_{in} \equiv R_{ox} \cdot r_{o3} \cdot r_{o1}
\]  

(1)

where \(r_{o1}\) and \(r_{o3}\) are the output resistances of cascode
transistors M1 and M3, respectively, and are inversely related to their drain currents. Consequently, when the sensor current, which is also the drain current of M1 and M3, goes low, \( r_{ds} \) and \( r_{g} \) increase, making \( R_{\text{out}} \) extremely high. This high \( R_{\text{out}} \) value makes Readout-1 non-ideal as a current-to-current readout circuit and insensitive to the sensor current.

The loop gain of Readout-1 \( (A_1 \beta_1(s)) \) is directly affected by changes in the sensor resistance \( (R_{\text{sensor}}, \text{see Fig. 2}) \):

\[
A_1 \beta_1(s) \equiv A_1(s) \cdot \frac{g_{m1}}{g_{m2}} \cdot R_{\text{sensor}} \tag{2}
\]

This causes the poles of the OTA and of the sensor to move with \( R_{\text{sensor}} \), which degrades the phase margin and generates oscillations in the output, making it unstable.

### 3.1.3 Readout-2

The proposed design of a wide-current-range potentiostat (referred to as Readout-2 hereafter) is shown in Fig. 3. The readout circuit adopts the structure of the potentiostat shown in Fig. 2. However, a low-input impedance is added to the circuit design to improve the current sensing range and enhance stability.

Figure 3. Readout-2 (proposed wide-current-range potentiostat).

To establish low input resistance, the source terminal of the MOS is chosen as the input point. Then, to maintain the direction of current, transistor M3 is changed to PMOS. Thus, the source terminal of PMOS M3 serves as the input of Readout-2. A very low input resistance \( (R_{\text{in}2}) \) is derived:

\[
R_{\text{in}2} \equiv \frac{1}{g_{m2} + r_{ds}} \approx \frac{1}{g_{m2}} \tag{3}
\]

The low \( R_{\text{in}2} \) makes Readout-2 suitable for a current-to-current transfer interface circuit and sensitive to sensor currents in the pico-ampere range.

The loop gain of Readout-2 \( (A_2 \beta_2(s)) \) is stable and unaffected by \( R_{\text{sensor}} \) (see Eq. (4)), especially when the input resistance of PMOS M3 \( (1/g_{m3}) \) is designed to be very small compared to \( R_{\text{sensor}} \).

\[
A_2 \beta_2(s) \equiv A_2(s) \cdot \frac{1}{1 + g_{m2} \cdot r_{ds}} \cdot \frac{R_{\text{sensor}}}{1 + R_{\text{sensor}}} \tag{4}
\]

Transistors M1 and M2 serve as current mirrors to transfer the sensor current \( (I_{\text{sensor}}) \) to the output current \( (I_o) \). Transistors M5 to M9 form the self-bias differential amplifier.

### 3.2 Power management and drug delivery system

Figure 4 shows the system architecture of the power management system, which mainly consists of a power supply system, a monitoring system, and a battery charge system. Power from an outside source is directed to the power recovery circuit and converted using an internal system power supply, \( V_{\text{data}} = 4.5 \text{ V} \).

The power recovery circuit uses a low-drop-out (LDO) voltage regulator circuit to improve power supply performance. The power recovery circuit mainly consists of an over-voltage protection circuit, voltage regulator circuits, a power detection circuit, and a power feedback circuit. When the input voltage is too high (7.5 V), the over-voltage protection circuit clips the voltage to 7.5 V, while the power detection circuit feeds back the signal to an external controller for proper power reduction. If the power for the circuit is at an appropriate level and stable, the power recovery circuit sends a corresponding signal to the data receiver that it is now ready to receive data. A power-on reset is used to ensure that the device state is in a known state.

After power is delivered to the internal system, the monitoring circuit determines whether the battery needs to be charged. If so, the power is divided into two paths using a power switch. One path is to the charger (to charge the battery) and the other path is to the voltage regulator (feedback) (with \( V_{\text{in}} = 3 \text{ V} \)). If charging is not needed, the power goes directly to the voltage regulator. The above situation of the path selection is mainly processed by the system mode exchange interface (S.M.E.I.) circuit. 3 V is used as the voltage regulator’s output because the lithium-ion battery requires a minimum operating voltage of about 3 V. The design allows a battery voltage range of 3.2–4.2 V.

Because heat generated by implanted device might cause burn injury to the subjects, the design requires either an external power supply or a battery to be used. For high-voltage applications, the voltage regulator output (3 V) can be converted by a charge pump to the required voltage (~15 V). For applications that do not need 24-hour continuous monitoring, an additional timer driver circuit is designed that allows the system to work for a predetermined time cycle to reduce power consumption.

In vivo power management system’s different blocks should be designed to meet their different requirements of a closed sensing system. The power recovery circuit of the power management system must have a high power supply rejection ratio (PSRR) because the received power affects the obtained data. Voltage ripples affect the accuracy of voltage regulation, so resistance to the fluctuations of power is required. The
voltage regulator block is composed of a second-order power supply circuit. The circuit is optimized for low quiescent current and high output accuracy, which are needed to improve battery life and provide stable power for high-precision subsystems in the absence of an external power supply. The charging system block is designed for high current efficiency. If maximum power is transferred to the battery, the charging time can be significantly reduced to prevent heat-related injuries.

Figure 5 shows the S.M.E.I. flowchart. The six switching modes are (1) charge mode, which provides power to the internal battery and internal system power supply; (2) external power mode, which directs all external power to the system (starts when the internal battery does not require charging); (3) battery power mode, which uses the battery to power the system (start when there is no external power); (4) protect mode, which starts under special circumstances, such as overheating and over-charging; (5) battery-saving mode, which is used by the designer who can determine whether it is time to start a fixed turn-off feature; and (6) sleep mode, which starts when battery is fully discharged and there is no external power source fed into the system.

The mode switching system mainly consists of two blocks, the external power supply (EPS) and the battery power supply (BPS). After the EPS is turned on, the EPS power switch turns on and triggers the wake up block. When the system is initialized for the first time, the battery is charged and the EPS provides power to the system. After the battery is fully charged, the system operations will be moved to EPS mode and stop charging. When the system goes to the wake up block, the system operation will no longer be in sleep mode, so the battery charge monitor will check whether the battery of the implanted sensing system needs to be recharged. If recharging is not needed, EPS mode starts.

When there is no external power source, the EPS power switch turns to the battery power supply of the proposed implanted sensing system. The BPS mode provides the power to the system. If over-current is detected, the sensing system immediately goes into protected mode. The battery charge monitoring system continues to monitor the battery. When a full discharge is detected, the system goes into sleep mode. The system will then wait for the transmission of signal to the external power supply to start the sensing function all over again.
As for the drug driver circuit, when the readout sensor detects excessive glucose levels, it sends a signal to the MCU, which computes the amount of drug to be delivered and sets the programmer timer to control the drug delivery driver and oscillator activities. The driver determines the waveform and strength of the electric field from the oscillator and reference voltage; the sensing system can also drive different qualities of load (like Ionic Polymer-Metal Composites, IPMC). In the proposed system, a single power supply is used, which reduces complexity, power consumption, and area compared to those of a twin power supply system. The proposed circuit is thus well suited for biomedical micro-system applications.

Figure 6 shows the core circuitry of the drug delivery driver. The feedback system with an operational amplifier (OPA) allows V+ to equal V-, and it can generate a cross loop for current and vary the electric field. Waveforms are used to control the switches and vary the electric field. Vsw can be set to change the driving capability while still supplying sufficient current to the load (IPMC membrane) throughout the membrane’s pressure variation.

3.3 Microcontroller unit

An 8-bit MCU (Atmel ATmega16) is used for data collection, and data processing. The MCU supports external memory due to large amount of data gathered for the succeeding glucose level forecast. Its instruction set supports multiplication and division instructions for floating point operations. The MCU has an internal program flash. A comparison of four commonly used 8-bit microcontrollers is given in Table 1. The MCU performs the following functions:

1. Convert the analog glucose signal into a digital signal using its ADC.
2. Compute the right amount of insulin to be administered based on the obtained glucose data.
3. Send a corresponding control signal to the driver of insulin.
4. Check whether the right amount of insulin was administered.
5. Check the temperature sensor results.
6. Check the battery charge level.

(7) Send corresponding warning signals if the above values are not in the the specified ranges.
(8) Display the results on the LCD module.

Table 1. Comparison of microcontrollers.

<table>
<thead>
<tr>
<th>Feature</th>
<th>Microcontroller</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ATMEGA 16L</td>
</tr>
<tr>
<td>Architecture</td>
<td>AVR</td>
</tr>
<tr>
<td>Supply voltage</td>
<td>2.7 V</td>
</tr>
<tr>
<td>Program memory</td>
<td>16 kB</td>
</tr>
<tr>
<td>RAM (bytes)</td>
<td>1k</td>
</tr>
<tr>
<td>I/O Pins</td>
<td>32</td>
</tr>
<tr>
<td>Clock Speed</td>
<td>16 MHz</td>
</tr>
<tr>
<td>ISP</td>
<td>Yes</td>
</tr>
<tr>
<td>MULT DIV Inst</td>
<td>Yes</td>
</tr>
<tr>
<td>Interrupts</td>
<td>11</td>
</tr>
<tr>
<td>Timers</td>
<td>2, 8-bit</td>
</tr>
<tr>
<td>UART module</td>
<td>Yes</td>
</tr>
<tr>
<td>SPI module</td>
<td>Yes</td>
</tr>
</tbody>
</table>

Figure 7 shows the READOUT routine flow chart. Following the READOUT program, the MCU waits for glucose data from the sensor. The MCU knows that there is glucose data when the input voltage (Vin) on one of the channels of its ADC goes below 2.7 V. The reference voltage is 3.067 V, so a voltage drop proportional to glucose level was sensed by the readout circuit. The MCU waits for a certain time (i.e., 7 seconds), and then calibrates the data using a linear equation obtained from experiments on the readout chip. The process continues and it just likes an artificial pancreas which would be analyzed if insulin is needed.

\[
Y = \frac{1023 \times 3.3}{Vin} - 0.002X + 3.0672
\]

Where: \( Y \) = voltage
\( X \) = glucose concentration
4. Experimental results and discussion

The readout chip and power management chip results are presented in this section to verify the functionality of the fabricated ICs. Simulations were conducted using Synopsys HSPICE. These two ICs were fabricated in TSMC 0.35-μm 2P4M 3.3/5 V CMOS technology. The whole system, which includes the MCU, was implemented as an on-board prototype.

4.1 Set-up of circuit simulation for readout circuit

The post layout performances of Readout-1 and Readout-2 were simulated in HSPICE using TSMC 0.35-μm mixed-signal 2P4M 3.3 V CMOS models. The amperometric chemical sensor was replaced with an $R_{wr}$, $R_{oss}$, and $C_s$ model, as shown in Figs. 2 and 3, for circuit simulations. The redox potential ($V_{sw}$) was set to 0.4 V.

4.2 Improvement on feedback stability

Modifying the sensor resistance ($R_{wr}$) simulated the change of concentration of the analyte. When $R_{wr}$ was set to 8 kΩ, 800 kΩ, 8 MΩ, and 8 GΩ, the corresponding generated sensor currents were 50 μA, 0.5 μA, 50 nA, and 50 pA respectively. Figure 8 shows the effect of $R_{wr}$ on the circuit loop gain ($|L|$). The loop gain of Readout-1 ($|L_1|$) varies with $R_{wr}$ whereas that of Readout-2 ($|L_2|$) is unaffected by $R_{wr}$ at low frequencies.

![Figure 8. Bode plot of loop gain $|L|$ for various sensor resistances $R_{wr}$.](image)

4.3 Widening of current sensing range

Figure 9 shows the simulation and experimental results of the current sensing range of Readout-2. The simulation used a 3.3-V power supply whereas the experiments used a 5-V supply.

Nevertheless, the results confirm that Readout-2 can detect current down to the nano-ampere range. The post-layout simulation results show a current range from 50 pA to 50 μA, whereas experimental results show a current range from 10 nA to 50 μA. 10 nA is the limit of detection for the instrument used.

Figure 9 also shows the linearity between the input sensor current ($I_{input}$) and the mirrored output current ($I_o$) for Readout-2. There was a mismatch between the sensor current and output current, which can be attributed to the load device limiting the drain source voltage of transistor M4.

![Figure 9. Post-layout transfer characteristics of Readout-2.](image)

4.4 Set-up of experiment

An on-board module of Readout-2 was built to verify its design and performance. The LMC6484AIN OTA was used due to its suitable rail-to-rail output range. CD4007UBE MOS transistors were employed. The power was set to 5 V. A semiconductor device analyzer (Agilent B1500) was used to set up the current down to the nano-ampere range (limit of detection: 10 nA).

4.5 Physical layout of the proposed readout circuit

Excluding the pads, the physical layout of the Readout-2 circuit occupies a chip area of 0.43835 mm × 0.403 mm (0.1767 mm²) using TSMC 0.35-μm mixed-signal 2P4M 3.3 V CMOS technology.

4.6 Power management and drug delivery

Figure 10 shows the results of all cases of the switching mode system, divided into 15 sections (A–O). The observations of the system operation at regions A to O can be summarized as follows:

(A) Initially, there is no input power and the battery supply is below 2.8 V (over-discharge status). $V_{dd}$ (external power) is at 0 V, which makes $V_{ddw}$ (internal power) also 0 V. $V_{out}$ (analog circuit power supply) and $V_{cmd}$ (digital circuit power supply) at 0 V, and thus initially provide no power.

(B) When the battery power is increased to 3.7 V, the start-up circuit inside the chip is not yet enabled because all the other circuits (specifically the reference circuit) are still off, so the system is in the standby status and cannot be activated.

(C) External power ($V_{dd} = 6.5$ V) is delivered, the battery power remains at 3.7 V, and $V_{ddw}$ increases steadily to 4.5 V. The battery draws 35 mA of current from the charger (not shown in the figure), and $V_{out}$ is at low state. $V_{POR}$ signal enables reference circuit and leads the voltages $V_{out}$ and $V_{cmd}$ to be pulled to 4.5 V.
(D) After the system has been activated, the circuit becomes stable and \( V_{outN} \) and \( V_{outD} \) decrease to the specified 3 V, but the charger does not activate.

(E) When each part of the circuit is in a steady state, \( V_{cout} \) is completely activated and pulled up to 4.2 V with a supply current of 35 mA (not shown in the figure).

(F) When the charger’s current is reduced to 100 \( \mu \)A, the charger turns off.

(G) \( V_{dd} \) is pulled down to 3 V (below the threshold voltage of 5 V), which turns the power recovery circuit off, resulting in \( V_{dd} \) being decreased to the battery power supply voltage of 3.7 V.

(H) To test the performance of \( V_{inner} \) the second-order voltage regulator is turned off, which pulls down voltages \( V_{outN} \) and \( V_{outD} \) to ground.

(I) The second-order voltage regulator is turned on again, resulting in \( V_{outN} \) and \( V_{outD} \) being 3 V.

(J) When battery power goes down to 2.6 V (below the threshold of 3 V), the system enters sleep mode, \( V_{outN} \) and \( V_{outD} \) are pulled down to ground, and \( V_{cout} \) is still a negative voltage.

(K) When battery power goes back to 3.7 V again, the system recovers to normal operation, \( V_{outN} \) and \( V_{outD} \) are 3 V, and \( V_{dd} \) remains at 3.7 V.

(L) When external power returns to 6.5 V (as in the initial state), \( V_{outN} \) and \( V_{outD} \) are 4.5 V and \( V_{ddv} \) increases to 4.5 V.

(M) When each part of the system circuit recovers to its stable state, because the signal, \( V_{POPG} \) has not enabled the charger, so the charging function is still not active and the draw current keeps on 100 \( \mu \)A.

(N) At this operating region, \( V_{POPG} \) enables the battery charger (\( V_{cout} \) output is at 4.2 V). The other circuits are still in their stable state.

(O) After \( V_{POPG} \) completes the enable action of the charger, then the current sensing circuit turns off the battery charger and pulls \( V_{cout} \) down to a negative voltage.

Figure 10 shows the simulation results of the switching mode system.

Figure 11 shows the post-layout simulation results of the dynamic power for the whole system. The initial surge of power was caused by the initial switching on. The circuits that generate periodic pulses, POR and POPG also consume power. The whole system power consumption in sleep mode is almost zero (0.374 pW) with an average dynamic power from 0 to 7 seconds of \( 1.95 \times 10^{-3} \) W (supply voltage: 6.5 V; dynamic current: 300 \( \mu \)A).

Figure 12 shows a die micrograph of the power management chip. The major subsystems are power recovery, LDO second-order regulator, battery charger, drug driver, temperature sensor (TS), oscillator, bandgap reference (BGR), multi-level comparator (MLC), and battery charge meter (BCM). The chip was fabricated using TSMC 0.35-\( \mu \)m 2P4M 3.3/5 V CMOS technology. It occupies an area of \( 2.382 \times 2.374 \) mm\(^2\) (core size = \( 2.200 \times 1.086 \) mm\(^2\)) and dissipates 36.3 mW of power. It is packaged in a 48 pin DIP.

4.7 Amperometric glucose sensor readout circuit

The amperometric i-t curve, shown in Fig. 13, was obtained using glucose strips (commercial strips from HMD BioMedical) for various glucose concentrations.
The results were obtained using an electrochemical analyzer connected to a computer with appropriate software. The amperometric i-t curve was used to derive the calibration curve shown in Fig. 14. Based on the data results shown in Table 2, the data of calibration curve was obtained using MS Excel by first plotting the concentrations versus their corresponding currents on the 34.5 seconds of Fig. 13 (time of reading. Treading = Ta + Tr, where Ta is the time when glucose was added and Tr is the response time of the glucose strip) Treading = 20 s + 14.5 s = 34.5 s. Some data manipulation was then done and a linear trend line was obtained using MS Excel to get a linear equation and a linearity value. This linear equation, in slope-intercept form, was used in the algorithm for the MCU.

4.8 Whole system

The whole system prototype is composed of the power management module, the AVR MCU module, the glucose readout module, and the LCD module. The major functions of the power management module are shown below: (a) the system can output a power feedback signal (V_fb) when the power received is at one of these three levels: (i) $6.4 \text{ V} > V_{fb} > 5 \text{ V}$, or (ii) $V_{fb} > 6.4 \text{ V}$, or (iii) $V_{fb} < 4.9 \text{ V}$. (b) the system may read a battery charge level ($V_{bat}$) that is at one of the following five levels: (i) $V_{bat} > 3.9 \text{ V}$, or (ii) $3.9 \text{ V} > V_{bat} > 3.5 \text{ V}$, or (iii) $3.4 \text{ V} > V_{bat} > 3.2 \text{ V}$, or (iv) $3.2 \text{ V} > V_{bat} > 2.9 \text{ V}$, or (v) $V_{bat} < 2.9 \text{ V}$.

![Figure 13. Amperometric i-t curve.](image)

**Figure 13. Amperometric i-t curve.**

**Figure 14. Calibration curve of the glucose sensor strip.**

**Table 2. Data results for the calibration curve of Fig. 14.**

<table>
<thead>
<tr>
<th>Glucose concentration (mM)</th>
<th>Total current ($10^{-7}$ A)</th>
<th>Actual glucose current ($10^{-7}$ A)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>-1.791</td>
<td>0.000</td>
</tr>
<tr>
<td>0.02</td>
<td>-1.336</td>
<td>0.455</td>
</tr>
<tr>
<td>0.06</td>
<td>-4.101</td>
<td>-2.310</td>
</tr>
<tr>
<td>0.157</td>
<td>-4.076</td>
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<tr>
<td>0.301</td>
<td>-4.742</td>
<td>-2.951</td>
</tr>
<tr>
<td>0.602</td>
<td>-5.915</td>
<td>-4.124</td>
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<td>1.205</td>
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<tr>
<td>20.04</td>
<td>-157.9</td>
<td>-156.109</td>
</tr>
</tbody>
</table>

**Figure 15 shows the input (sensor) data. The equation in the figure was used as the glucose calibration equation.**

**Figure 15. Linearity curve of glucose sensor strip.**

Table 3 shows the experimental results obtained for the whole system prototype. Various glucose concentrations were used to imitate the human glucose range. All experimental results in the table were obtained from the LCD module. From the glucose column and the statistical analysis of five trials, there was only an average of 0.299 on the % error. For the power management functions, namely the power feedback and battery charge level, there was no error (0% error) compared to...
the setting level. The drug driver results were satisfactory to control the insulin delivery. The results show that the whole system works properly.

5. Conclusions

A system design with power management for closed-loop implantable biomedical systems was proposed. The proposed readout circuit with a potentiostat for amperometric chemical sensors can operate in a wide current range of 50 pA to 50 μA with a power consumption of 390 μW to 745 μW. Unlike most existing glucose monitoring devices, which use a separate glucose sensor and a drug administration device, the proposed device in an integrated module.

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References
