Non-Invasive Capacitance Based Glucometer and Controller

Yasser Abdelrahman, Alyson Fitzgerald, Delina Kambo, Sarai Velazco, Yifan Xiang

Abstract—The use of a non-invasive capacitance based glucose monitor can be implemented to eliminate the need for painful pricking and other invasive techniques involved in commercially available glucometers today. As identified by many studies and past literature, glucose concentration is linked to the capacitance of tissue and blood through a linear relationship. By going a step further than monitoring blood glucose concentration, the goal of this work is to effectively control and regulate. Through the modeling and execution of a proportional-integral-derivative (PID) controller, regulation of glucose concentration through an insulin pump can be combined with non-invasive monitoring in order to create a comprehensive blood glucose regulation feedback loop. Through the use of Simulink and Simscape, the development of promising and practical biosystems can be aided by block-diagram system models and their corresponding electrical components. To observe and further understand the response of a glucose monitoring system, we built a Simscape circuit to represent a capacitive sensor and applied a glucose-insulin model of a virtual patient through Simulink to introduce the linear relationship between capacitance and glucose concentration and how it can be used for regulation. It has been revealed that decreases in capacitance entails an increase in blood glucose concentration.

I. INTRODUCTION

Diabetes affects over 177 million individuals around the world, with this number projected to rise steadily for the next four years [1]. One of the main challenges in managing diabetes is having a continuous and effective way to monitor blood sugar levels, especially after a meal is consumed. The most commonly used glucometers today involve pricking the finger with a spring loaded lancet in order for a blood sample to be analyzed on a test strip. These methods of measuring the glucose level, although proven to be accurate, can be significantly painful for diabetic individuals, especially with repeated use due to the frequent need to measure blood sugar levels.

Additionally, the cost associated with the need to continuously purchase test strips poses a disadvantage that reinforces the need for a continuous blood glucose monitoring device. Although there has been an effort to introduce non-invasive techniques for monitoring blood sugar levels, research is still being done for their optimization and most of these products are not yet available for commercial use. Often non-invasive techniques target the use of spectroscopy or bioimpedance, where measurements of light intensity, reflectiveness and resistance of the skin can be used to produce a glucose reading. However, these methods have shown to require frequent calibration, vary from individual to individual, and are not standardized for accuracy and reliability [2]. One of the proposed noninvasive glucose meter methods is based on the concept of implementing a relationship between capacitance and blood glucose concentration. As identified by previous research work, the effective dielectric and therefore the capacitance of blood decreases as the glucose level in blood increases [3]. An individual’s finger can be inserted into a semi-cylindrical capacitor to model the finger as the effective dielectric of the capacitor. Using a low frequency operational amplifier (OP-AMP) and filtering, the value of the capacitance is obtained in terms of the output voltage and converted through a linear relationship obtained by calibration to a glucose reading. This glucose reading can be sent through a PID controller that will compare the value to that of a normal glucose level for the individual, and using the previously tuned PID parameters, it can activate an insulin pump connected to a diabetic patient in order to effectively apply a closed loop control of the glucose level.

II. PHYSIOLOGY

Diabetes is a condition that refers to problems in the functioning of insulin, the hormone produced by the pancreas that helps our body to store and use sugar from food. The food we consume is broken into glucose via the digestive system, glucose is then transferred via the bloodstream. In response to the glucose increase in blood, the pancreas releases insulin that allows glucose uptake by the cells, and its storage as glycogen in the liver eventually causes the glucose level in blood to return to normal. In the case of hypoglycemia, or low glucose levels in blood stream, glucagon is produced by the pancreas, and it converts the stored glycogen in the liver back to glucose to bring the glucose levels in blood back to normal [5].
Type I diabetes (Diabetes Mellitus) is a condition that results from the pancreas’s inability to produce insulin as a result of the immune system’s attack on the insulin producing beta cells of the pancreas. People who develop this condition experience hyperglycemia or high blood glucose levels due to the lack of insulin. As a result, patients with type I diabetes require continuous glucose level monitoring and take insulin injections when they experience hyperglycemia.

In contrast to people with type I diabetes, patients with type II diabetes produce insulin, but the insulin produced is not sufficient or the body is resistant to insulin; in either case, the insulin is not functioning properly in facilitating the glucose uptake by cells. Some patients can keep type II diabetes under control by watching their diet, weight, and exercising, though others might need glucose injections. As with people who suffer from type I diabetes, they also have a need for continuous glucose level monitoring [5].

III. Model Biosystem

In this study we focus on the design of a noninvasive capacitive glucose meter for low frequencies. In addition, we study the glucose meter in the context of a closed loop feedback-based glucose control system that includes a sensor (glucose meter), a controller, an insulin pump, and a virtual model for a diabetic patient. We focus primarily on two parts of the closed loop control system: the capacitive glucose sensor design and the PID controller.

A. Block Diagram

The system being modeled employs closed loop feedback. A capacitive sensor reading is received as an input, which is translated to a glucose level and observed as an output from a diabetic patient as a new glucose level. As shown in [6, Fig. 2], in a closed loop system, the input glucose reading measured by the sensors is run through a controller, which is represented in this study by a PID controller. The controller is preceded by an insulin pump that will ultimately regulate the biosystem in order to provide the insulin input for a case of hyperglycemia and bring the glucose level back to normal. In the present work, the biosystem components are separated and analyzed to have their results later incorporated with each other and show proof of concept, as well as simulate more simplicity in their individual designs.

![Fig. 2. Biosystem Closed Feedback Loop](image)

B. Sensor

To relate capacitance to a blood glucose level reading, the use of a semi-cylindrical capacitor can be implemented as proposed by Dutta, Abhinada, et al, shown in [7, Fig. 3] below.

![Fig. 3. Semi-Cylindrical Capacitor](image)

The curved copper plates function as a theoretical capacitor where an inserted finger becomes the effective dielectric. By reconstructing the effective capacitance of a parallel plate capacitance (Eq. 1), the theoretical capacitance can be produced as seen in Eq. 2. Where $k_s$ represents the dielectric constant of the finger, and $\varepsilon_0$ is the absolute permittivity.

\[
C = \frac{\varepsilon_s k A}{d} \tag{1}
\]

\[
C_s = \frac{\varepsilon_s k A}{\pi} \tag{2}
\]

This capacitor circuit can then be incorporated into an extended circuit in order to be amplified and processed. A comparator and differential amplifier can be combined in conjunction with a “dummy” capacitor in order to produce an output gain proportional to the capacitance of the finger. The circuit begins with an alternating voltage source in order to accumulate charge on the plate capacitor, preferably at a voltage below 50 mV to prevent saturation of the op-amps. As seen in Fig. 4, $C_s$ represents the semi-cylindrical capacitor and $C_0$ represents an identical capacitor with air as the dielectric. This combination produces a gain as shown by Eq. 4 at $V_1$. Then with the use of a unity gain differential amplifier, the signal can be isolated to simply a ratio of both capacitances (Eq. 5). After the signal has reached $V_0$, a signal rectifier and low pass filter are used in order to convert the signal from an AC to DC and filter out high frequency noise. With the use of Simscape modeling and calibrations done in previous work, blood glucose can be related to capacitance seen in $C_s$ through a linear relationship [7]. It has been proven that glucose affects the dielectric properties of blood and tissue. The effective dielectric constant of blood samples decreases linearly with the
increase of glucose concentration. The capacitance $C_s$ can be calculated with the dielectric constant of the finger $k_s$, as shown in Eq. 2. Therefore, the increases in blood glucose concentration will lead to a decrease in capacitance. Efficiencies in the proposed circuit arise from the capacitance of tissue being much larger than that of air, producing a large gain that doesn’t require further amplification. Furthermore, the use of a capacitance ratio allows for the circuit to account for changes in temperature, humidity or other environmental factors. As any changes that affect one capacitor, will be mediated by the other through their ratio.

\[
V_1 = (1 + \frac{C_s}{C_0})V_s \quad (3)
\]

\[
V_o = (V_1 - V_s) = (\frac{C_s}{C_0})V_s \quad (4)
\]

Fig. 4. Simscape Sensor Circuit Model

C. PI Controller

The delivery of insulin to the patient is modulated by a proportional-integral (PI) controller. PI and PID controllers use feedback in order to continuously reduce an error value, where the error value is the difference between a variable of interest and the desired target value. The transfer function $H(s)$ of a PI controller is given by the following equation [8]:

\[
H(s) = K_p + \frac{K_i}{s} + K_D s \quad (5)
\]

$K_p$ represents proportional gain, whose control action primarily serves to reduce rise time; $K_i$ is the integral term which decreases steady-state error “through low frequency compensation by an integrator”; and the derivative term, $K_D$, “improves transient response” at high frequencies [8]. Our electrical system model omits the derivative control because it is not necessary at the frequencies of interest; because $K_D$ does little to reduce rise time or steady-state error, it was decided that proportional and integral control by themselves were sufficient [8].

To construct a PI controller with electrical components, the role of the controller in the system must first be understood. As shown in Fig. 5, $V_s$, the voltage representing error, is obtained by summing the setpoint voltage with the “sensor” voltage that represents the current amount of glucose in the patients’ system. This is accomplished using a voltage summer, which consists of an inverting op amp that adds the two input voltages with a gain determined by the proportion of its resistor values, $R_1$ and $R_2$. Either the setpoint or sensor voltage should be inverted in order to obtain the difference between the two values. This could be accomplished by reversing the order of the positive and negative voltages of $V_{setpoint}$ with respect to ground, for example.

Fig. 5. A diagram showing the implementation of the PI controller in the complete system.

The circuitry of the PI controller consists of a further three inverting op amps: one to provide proportional gain $K_P$, an integrator to provide $K_I$, and a summer to sum the two terms together [9]. Values for the resistors were set such that all the op amps have a gain of 1; the glucose voltage obtained from the sensor is large enough in magnitude that further amplification is not necessary. The values for $K_P$ and $K_I$ in Fig. 6 are placeholders that were later tuned using a block-diagram model, since they depend on the differential equations that represent the glucose-insulin dynamics.

Fig. 6. Circuitry of the PI controller.

The following equations show how the two control parameters are determined by the passive circuit components:

\[
K_p = -\frac{R_2}{R_1} \quad (6)
\]

\[
K_i = -\frac{1}{R_2C_{fbo}} \quad (7)
\]

The sensor and insulin pump were modeled separately in order to simplify the testing of the controller.

While modeling the insulin pump with circuitry would be complex and beyond the scope of our project, we include a series of differential equations that represent the effects of an insulin dosage on the patient’s glucose concentration.
In order to determine the PID controller parameters and tune its performance, we used a model of the glucose control by insulin that is the simplest representation of a virtual patient model. The simplified model is given by the following equations for a first order low pass response with a time constant $\tau_{meas} = 1 \text{ min}$:

$$\frac{dG}{dt} = \frac{1}{\tau_m} (G(t) - G_m(t)) \quad (8)$$

$$I(t) = K_p e(t) + K_i \int_0^t e(t)dt + K_d \frac{de(t)}{dt} \quad (9)$$

$$e(t) = G_m(t) - T(t) \quad (10)$$

$$\frac{dC(t)}{dt} = aI(t) - \frac{1}{\tau_m}C(t) \quad (11)$$

$$\frac{dG(t)}{dt} = -kG(t)C(t) \quad (12)$$

In the above equations, $G_{meas}(t)$ is the glucose level measured by the sensor, or the value after the voltage output is translated into a glucose reading, and $G(t)$ is the glucose level as a function of time. $I(t)$ represents the concentration of insulin with time, $T(t)$ is the target glucose level and the error $e(t)$ is the difference between the target glucose level and the measured value. $k$ is the reaction rate constant and $\alpha$ is a constant as well [10]. This model also takes into account the changes in glucose level that occur via gluconeogenesis or the conversion of glucose to glycogen via insulin, where $G(t)$ is the glycogen concentration as a function of time. We modeled this simplified glucose regulation system in Simulink, and the schematic is shown in Fig. 7. In our system, the insulin is produced by the pancreas in a virtual patient model or pumped into the system by a carefully designed pump, as it is described by the closed loop system overview in Fig. 2.

![Fig. 7. Simulink block diagram model of glucose-insulin dynamics.](image)

In order to mimic a system that takes a glucose input that is representative of the physiological glucose ranges after a meal, we used 50 mmol of glucose per 5 L of blood, resulting in 10 mmol glucose/L concentration of glucose input as an initial condition to the Simulink model. Using these parameters for the Simulink model, we performed autotuning to get the values of $K_d$, $K_i$, and $K_p$ that would optimize the PID controller functioning for our model. This particular approach for autotuning these parameters is dependent on the model chosen to represent the virtual patient or the glucose regulation system, so for different and more complex Simulink models, the tuning parameters would change from the following values shown in Table 1 and the response shown in Fig. 8.

<table>
<thead>
<tr>
<th>PID Auto Tuning parameters</th>
<th>$K_p$</th>
<th>$K_i$</th>
<th>$K_d$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1.0333</td>
<td>0.0836</td>
<td>2.0687</td>
</tr>
</tbody>
</table>

![Fig. 8. PID Auto-tuned response](image)

Using these parameters for the PID results in a response curve shown in Figure 8. The response has a rise time of 1.25 second and a settling time of 20.9 seconds with a 9% overshoot given our choice of a virtual patient model. These parameters and the PID response are subject to change when other more complex models or data from a patient are used. This requires careful considerations when it comes to choosing a PID controller that can be applicable for glucose control measurements in a variety of patients.

IV. DISCUSSION

A. Sources of Error

Despite the well-performing design of this glucose sensor, there are a couple of considerations that we should be aware of in terms of the applicability of the design. One of the main sources of errors is the effect of stray capacitance between the two plates space and the finger, which is practically unavoidable. Factors like sweating of the finger can also cause changes in the dielectric value between the capacitor plates and therefore result in inaccurate glucose readings from the sensor. In addition, we do not have a clear knowledge on the timing between when the glucose reading is obtained by the sensor and the time it takes for glucose to change the dielectric value between the
plates. Furthermore, the device requires a one time calibration from patient to patient in order to accurately define the linear relationship between blood glucose level and the capacitance of a finger. However, this is an aspect of this design that can be tested in different experimental settings and adjusted accordingly with the sensor design. Another consideration that can also be a potential source of error is the autotuning of the PID parameters based on a chosen patient model, which need to be carefully designed for a well functioning closed loop glucose control system.

B. Advantages

The primary advantage of this design is that it relieves the user of the stress and pain associated with continuous finger pricking. The controlled delivery system also eliminates the need for the user to self-administer insulin injections, which are inconvenient and sometimes painful. The sensor, which is worn on the finger, is painless and relatively unobtrusive. Some modifications to the size and aesthetics of the sensor could also be made to further increase the practicality and appeal of the product. Advantages in the sensor circuitry include: accounting for changes in environmental conditions and containing a minimal number of components due to efficiency in the output gain.

The design is relatively simple and likely straightforward to implement thanks to the linear relationship between glucose and measured capacitance, as well as the fact that the circuitry only comprises a few op amps and a four-diode rectifier.

ACKNOWLEDGMENT

We would like to express our sincere gratitude to Dr. Cauwenberghs and Preston Fowler for their dedication to providing constant assistance and guidance.

REFERENCES