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Wireless Pressure Sensor System For Medical Applications

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WIRELESS PRESSURE SENSOR SYSTEM FOR MEDICAL APPLICATIONS

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Submitted in Partial Fulfillment of the Requirements

For the Degree of Master of Science in

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2013

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DEDICATION

To my mother, who loves me and supports me so much.

ACKNOWLEDGEMENTS

I sincerely give my thanks to everyone listed below. To Jesus Christ, who helps and blesses me all the time. To my advisor, Dr. Guoan Wang for his leadership, teaching and helping for this long time. To Dr. Yinchao Chen, my committee member and form advisor for the knowledge and skills he taught me. To my colleague Dr. Xinchuan Liu, for the models he made and ideas he shared with me. To my group members, Farid Rahman and Yujia Peng and other office colleagues, for the discussion and communication we had. Finally, I'd like to give special thanks to my mother for her love.

ABSTRACT

Telemedicine in the modern era is a rapidly developing industry. In a telemedicine system, sensor is the key elements and play important role in any telemedicine applications, especially implantable sensors. Unlike traditional in vitro sensors, there are some special requirements for them like battery-less, compact, and bio-friendly. Therefore, many researchers in institutes as well as in industry are continually developing implantable sensor and sensor system to fulfill the requirements of medical applications. Nine in one thousand people in the United States are born with congenital heart defects and sixty percent die before one year old. As a result, it will be extremely helpful for them to get monitored all the time by certain type of sensors. Meanwhile, for this type of sensors, wireless data transmission is very important for safety and convenience concern. Heart blood pressure is an important standard for estimating the status of patients, an implantable sensor in patients' heart allows doctors to get these data real timely. This work focuses on the design of an implantable blood pressure sensor system which allows the pressure data to be transmitted wirelessly by inductive coupling between the sensor circuit and measurement device. The developed sensor system is battery-less, compact and bio-implantable. A series of the small dimensional passive sensors are introduced, analyzed and optimized in this research. Qualitative experiments are conducted, studied and validate the sensor system. The whole picture of a sensor system for a small dimensional implantable blood pressure sensor is exhibited.

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LIST OF SYMBOLS

μ_0 Permeability of free space $4 \pi * 10^{-12} \text{F/m}$

λ Wavelength of the electromagnetic wave

LIST OF ABBREVIATIONS

HFSS High Frequency Simulation System

CHAPTER 1

INTRODUCTION

1.1 Motivation and objective

With the rapid development of network technology, telemedicine has become a sunrise industry. The most significant advantage of telemedicine is it allows medical works to do the examinations and treatment to the patients without really touching them. Doctors from every corner of the world are able to help the same patient at the same time when they are sitting in their own offices.

In general, the structure of a telemedicine system is as shown in block diagram Figure 1.1 [17].

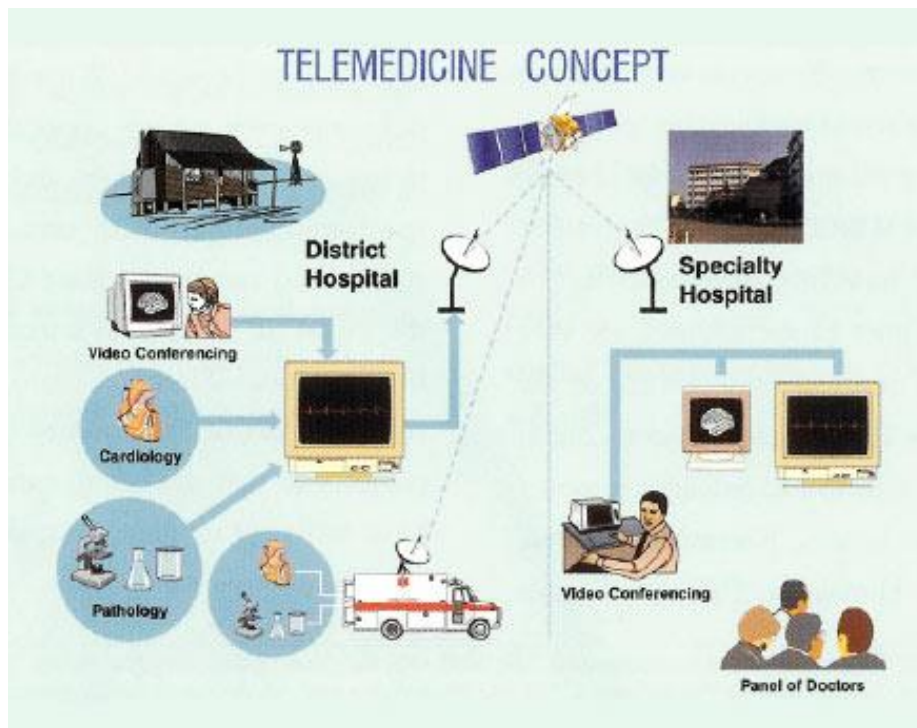


Figure 1.1 Telemedicine network concept

Telemedicine can play a role of great importance in diagnose and treatment of many kinds of disease. One among them is the congenital heart defect. Just like for other disease, sensors are a kind of very useful tools in the field of congenital heart defects for telemedicine.

According to statistic data collected by American Heart Association, nine in one thousand people are born with congenital heart defects [1]. Unfortunately, sixty percent will die before one year old if cannot be taken care of timely. As a result, it will be of great importance to diagnose and remedy the infants with these heart problems as early as possible. On the other hand, keep monitoring the infant patients is a very helpful method in the process of treatment for congenital heart defects. One significant parameter that reflects the status of the infant patient is the blood pressure within the heart which is difficult to be observed in vitro directly. To act as a solution of this problem, an implantable sensor that is able to fulfill the requirements heart blood pressure monitoring has been designed in our research.

Generally, a medical sensor is to fulfill design principles like real-time detection, accurate result providing, fast calculation rate, and convenience of using. Specially, for implantable sensors such as our design, there are still some other requirements, they must have as smaller dimensions as possible for the body cannot provide too much space for the sensors, at the same time, all the materials that applied in the sensor must be bio-friendly and harmless in order to satisfy the safety concern, last but not the least important requirement is that they are no to contain too many power components for a longer service life and extra surgeries may bring extra risk to the patients.

To summarize, this design is to come up with an implantable sensor with the characteristics as small dimensioned, bio-friendly and passive components contained only [2].

1.2 Sensor design background

A sensor is a converter that measures a physical quantity and converts it into a signal which can be read by an observer or by an electronic instrument. In other words, sensor transfers some physical quantity that cannot be observed directly into readable data by human or by compute for further analysis.

Varieties of sensors have been applied in many fields, like industry, security, education and research. One of the field in which sensors act as a series of useful tools is medicine. Modern medicine requires advanced sensors to play an important role in disease diagnose and treatment. Many institutes and industries corporations have taken part in the research, development and manufacturing of medical sensors. They are continually coming up with all kinds of those designs and products. For example, blood oximetry and pressure measurement, body fat estimation, and heart status monitoring including pulse counting, also, sensors can be used in 24-hour monitoring of newly born infants and patients in danger[2].

Generally speaking, medical sensors have two major types. They are in vitro type and in vivo type. In vitro medical sensors are being applied widely in the modern world. The medical sensors mentioned in the discussion above are all in vitro medical sensors, they acquire and transfer physiological parameters and transform it into electrical data for recording and analysis. Although the in vitro sensors do play a very important role in medical applications, there are still some limitations that cannot be overcome by them.

The most serious one is that an in vitro sensor cannot really contact the organs which generate the physiological parameters. By analyzing the cause of this problem, we come to a conclusion that it can only be solved by introducing in vivo sensors in.

Unlike the in vitro sensors, those in vivo sensors are to fulfill more requirements. They are required to come up with dimensions as small as possible, and for achieving data easier and bringing the patient less pain, they are supposed to be wireless and best to be contain passive components only. As a result, this research is focused on the design of an implantable blood pressure sensor that can fulfill all the requirements listed above.

Another design purpose to be achieved is the realization of measurement. As mentioned in the paragraph above, this measurement is to be done by wireless data transmission. An LC circuit is to be connected to a measurement device, and the sensor is to be coupled with the LC circuit wirelessly. On the screen of the measurement device, resonant frequency and the impedance phase of the equivalent circuit formed by the sensor and the LC circuit are to display. Once the sensor shape changes due to the external pressure, the inductance and capacitance will also change, and resulted in a change of resonant frequency and impedance phase. After figuring out the impedance phase change, the external pressure on the sensor can be calculated from equations provided by researchers[3] like Volpato, Ramos[4], Po Jui Chen, Damien Rodger and others[5, 6], as shown below.

$$L = 2.35\mu_0 \frac{n^2 d_{avg}}{1 + 3.55\rho}$$

In which we have,

$$d_{avg} = 0.5(d_{in} + d_{out}), \rho = (d_{out} - d_{in})/(d_{out} + d_{in}),$$

For validation purpose, coils have been made and measured. As shown in Table 1.1 below.

Table 1.1 Sample coils inductance calculation and measurement comparison

	Coil1	Coil2	Coil3
Calculation	2307nH	6657nH	179nH
Measurement	2318nH	6676nH	182nH

1.3 Review of previous designs and our design

Wireless data transmission for wireless pressure sensor can use either antenna or use inductance coupling method according to recent designs [5, 7-11]. In addition, for implantable ones, researchers tend to use inductance coupling designs for they don't require external power supply such as batteries, as a result, it is possible to make them smaller and therefore, easier to implant.

However, those recent sensor designs have a few problems with may prevent them to be putting into real medical applications. The problems are: relatively large dimensioned, difficult to fabricate, and exposed and touch the body organs directly, all the designs mentioned above have at least one of those drawbacks, some of them even have two or three.

The reasons for those characteristics to be considered as drawbacks or problems are to be explained below[12].

1. For implantable sensors, large dimensions is, and always will be a big problem, for the sensor is indeed something “should not be there” for the body, they will definitely have some influence on the patient. Therefore, the sensor must be made to be as small as possible in order to minimize the negative influence on the patient after implanted.

2. If a sensor is difficult to be fabricated, not only the cost of manufacturing will be hard to control, but also reduce the interest of the potential investors’ interest towards making those products.

3. Exposed and touching the body organs directly can bring danger to both the patient and the sensor itself. For a patient, materials peeled from the sensor circuit may be poisonous and harmful. On the other hand, tissue fluid and blood of the patient will continuously corrode the sensor and therefore, reduce its service life greatly.

Inspired by the previous designs, we came out with the idea of using inductance coupling circuit to build our sensor system. In addition, we made some improvement in order to solve the problems of recent designs.

Our design has advantages as followed:

1. Using the circuit design that reduces the dimensions.
2. Making the sensor circuit structure as simple as possible for the ease of fabrication.
3. For avoiding direct contact between the sensor circuit and the patient’s body, the entire sensor circuit is made to be wrapped by the silicone proved to be harmless for human body and widely applied in medical application in modern time.

1.4 Outline of the thesis

In Chapter 2, the theory and principle on which the sensor design is based as well as the design method and simulation results of sensor design are talked about.

Theory basis of the system design, such as measurement methods are related qualitative experiments which can prove the reliability of the measurement method are to be discussed in Chapter 3.

Chapter 4 gives a summary of the sensor design as the conclusion and also talks about future work to be done.

CHAPTER 2

SENSOR DESIGN

First things first, concepts related to the implantable medical sensor need to be went over.

2.1 Sensor

In electrical meaning, sensor is a device which can transfer nonelectrical and unreadable physical quantities into quantities that are readable or recordable for devices or observers.

In the field of electrical and computer field, that means electrical quantities like voltage, current, resistance, and inductance.

In general, sensors can be sorted in variety of ways according to all kinds of standards.

Physical quantity detected, sensors can be sorted into pressure sensors, temperature sensors, and sound sensors and so on. Requiring an external power supply or not, there are powered sensors and passive sensors. Field of applying, there are tens of types of sensors, for example, automobile sensors, industry sensors, security monitoring sensors and medical sensors. Method of signal transmission, there are wired and wireless sensors.

Medical sensors are sensors that are used to acquire biology quantities or physiological indicators that not directly detectable to the medical staff. Plenty of medical sensors have been applied in modern medicine in to order to make contributions to diagnosing as well

as treating for all kinds of body parts, and may also help to monitor the status of patients such as newly born infants.

Implantable sensors

In diagnose and treatment of certain types of disease, it is sometimes impossible or very difficult for doctors to get necessary physiological parameters in vitro. Under those circumstances, the best solution is putting a sensor inside the patient's body so that it can acquire needed physiological parameters in vivo and transmit they out to the devices held by the doctors.

The sensor discussed in this thesis, to summarize, is a passive wireless blood pressure sensor, and prosperity to emphasize, it is an implantable one and will be put into the patient's heart.

2.2 Theory basis of sensor system design

According to the discussion in Chapter 1, a resonant circuit needs to be formed as the sensor circuit structure for wireless data transmission and measurement of the sensor system.

Resonant circuit and resonant frequency

Resonant circuits have been applied in many fields. Usually, a resonant circuit is formed by inductors, capacitors and resistors. The simplest resonant circuit consists of a capacitor and an inductor. A circuit in its resonant frequency has special performance, like power radiation, minimum value of phase impedance.

Power transmission method selection

In order to build an implantable sensor, the signal which represents the physiological parameters must be able to be transmitted wirelessly out of the body. Basically, wireless signal transmitting has two main ways, by antenna radiation and by coupling.

Antenna needs much power, which means the sensor has to come with a battery or other power source. An extra power supply brings many problems: firstly, it will definitely increase the dimension of the sensor. Secondly, battery usually has a much shorter service life than the device it serves, and so it needs to be replaced at last, which means extra surgery and extra risk for the patients. Finally, for a young patient, there could be unforeseen risk for exposing in a continuous electromagnetic radiation environment.

Inductive coupling can be a solution of these problems: it doesn't need a battery so patients will only need one time surgery. And it can be made smaller for the same reason. Finally, it will only be activated and transmit signal when the patient is being examined, all other time it stays in hibernate status and do nothing, therefore, can be radiation free.

Self-resonant circuit and structure basis selection

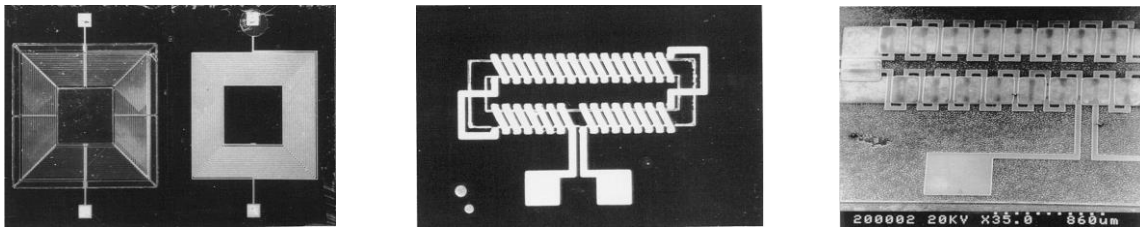
As the sensor is an implantable one for infant hearts, it should be made as small as possible. More components mean more space is required, although the LC resonant circuit is the simplest resonant circuit, there are still two components needed. Therefore, the best solution is to make the inductor and capacitor to be integrated in one component.

Transmission line theory[13] points out that, every transmission line can be transformed into an equivalent circuit consists of a resistor, a conductor, a capacitor and an inductor.

As a result, it is possible to make a component which acts both as an inductor and a capacitor, and the R and G could be ignored for they have little influence on the results needed.

If a transmission line circuit is designed to work around the frequency which is the resonant frequency determined by its inductance and capacitance, it can be named as a self-resonant circuit. A self-resonant circuit has the same characteristics as an ordinary resonant circuit. In other words, the phase impedance changes with resonant frequency, and the resonant frequency is determined by the inductance and capacitance which can be made to change with pressure.

By reviewing the recent designs mentioned in Chapter1, the part which couples with the measurement LC[14] circuit is an inductor, in other words, the most important part of the sensor circuit is an inductor. In order to minimize the dimension of the sensor in z direction, the inductor is to be a planar one. Recently planar inductor structures are shown as below in Figure 2.1, and according to the shape of planar structure, there are named as, spiral type, solenoid type, and toroidal meander type[15].



(a) Spiral type

(b) Solenoid type

(c) Toroidal meander type

Figure 2.1 Different kinds of planar inductor design

From Figure 2.1, it is obvious that among the three types, planar spiral structure is the simplest one. In addition, it can be easier connected to other external components/structures compared with the other two. Taking all those aspects into consideration, planar spiral inductor coil is selected to be the basic structure of the sensor design.

2.4 Overall construction of sensor system

Construction of sensor device/circuit

For the overall construction, there are some aspects that should be taken into consideration.

1. For safety reason and as the human body is also a conductor, the electrical part of the sensor has to be wrapped in an insulator, and the insulator should be proved harmless for human body or at least, hasn't been accused as dangerous.
2. The wrapping material must be flexible enough to transmit the pressure on the outer surface to the conductor inside.
3. Some solid structure is to be introduced into the sensor to act as a supporter for the conductor so it won't be broken due to the pressure.
4. There should be some space for the conductor to be reshaped.

Considering all the aspects mentioned above, a draft of the cross sectional view of the sensor without external pressure and with external pressure is shown as in Fig.2-2.

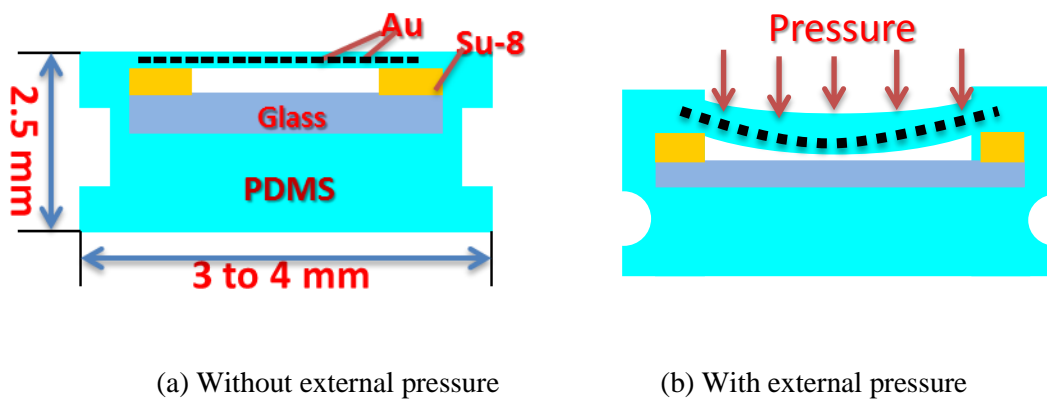


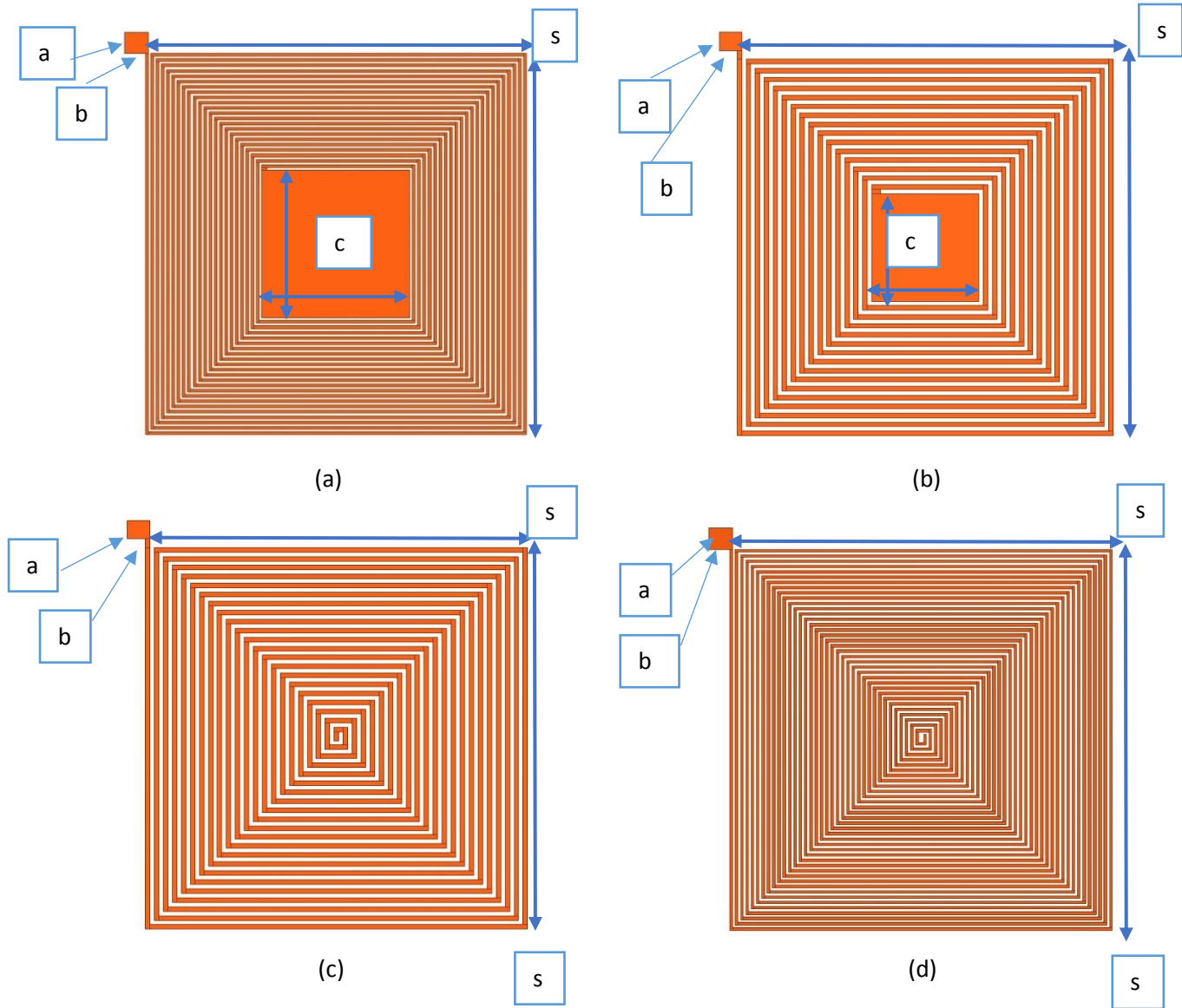
Figure 2.2 Cross sectional view of sensor with and without external pressure

Table 2.1 Calculation and measurement comparison of inductor coils

	Coil1	Coil2	Coil3
Calculation	2307nH	6657nH	4378nH

Measurement	2318nH	6676nH	4410nH
-------------	--------	--------	--------

Taking the difficulty level of fabrication into consideration, square spiral is selected to be the basic structure of the sensor's major part, the inductor. In addition, based on calculation, we designed six different structures of the sensor for simulation and optimization, as shown in Figure 2.3 (a) – (f).



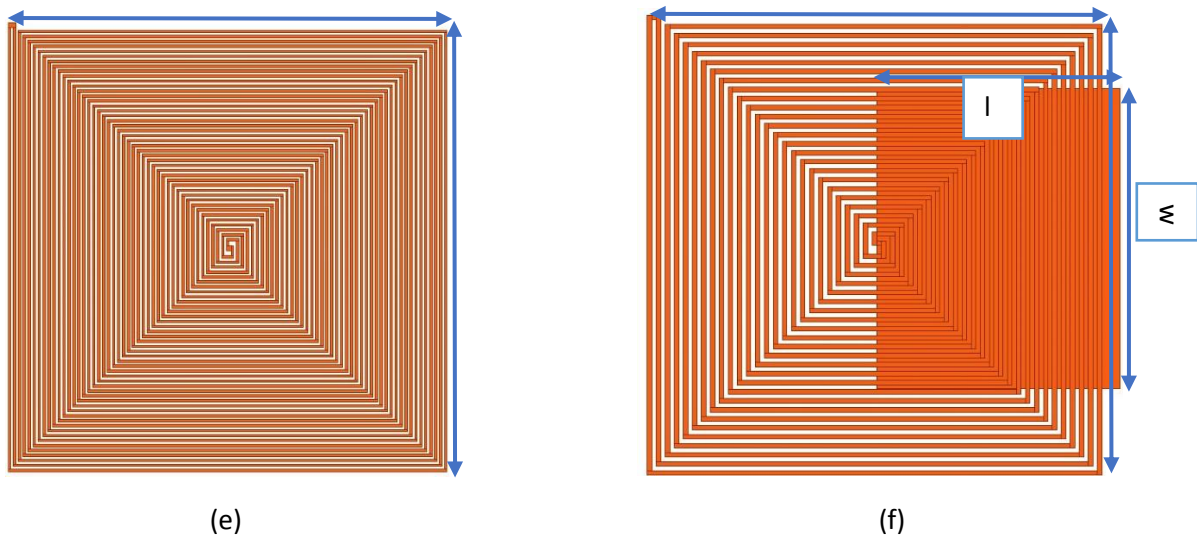


Figure 2.3 Top view of sensor circuit design

Table 2.2 Dimensions of sensor circuit design

	a	b	c	s	l	w
Design1	200um	225um	1.4mm	3.6mm	N/A	N/A
Design2	100um	125um	0.6mm	2.1mm	N/A	N/A
Design3	200um	225um	N/A	3.6mm	N/A	N/A
Design4	200um	225um	N/A	3.6mm	N/A	N/A
Design5	N/A	N/A	N/A	4.225mm	N/A	N/A
Design6	N/A	N/A	N/A	2.525mm	1350um	1075um

For the structures shown in Figure 2.3, the widths of every turn of the wires as well as distances between adjacent turns are defined to be 25um. The thickness of wires in every layer (if applicable) is 3um for the ease further work like fabrication.

2.5.2 Simulation Results, Comparison and Discussion

During the research and development process of electrical engineering, simulation plays a role of great importance.

All the simulation algorithms are designed based on different ways of solving Maxwell's equations[16].

Typically, they are several kinds of algorithms, such as Finite Element Method (FEM), Method of Moment (MoM), Finite Difference Method (FDM), Boundary Element Method (BEM), Transmission Line Matrix Method (TLM), Finite Difference Time Domain (FDTD), and Finite Integration Technology (FIT). Some of them are solutions in frequency domain, while others are in time domain.

The simulation tool selected for the sensor design in this thesis is HFSS. HFSS is a commercial FEM solver for E&M structures from Ansys. The acronym originally stood for High Frequency Structural Simulator. And HFSS has a good performance for planar structures.

For the sensor circuit simulation, parameters we concern are inductance, capacitance, Insertion loss and Q factor.

Those parameters are first grouped by the structure they came from. Then the same parameters for different structures are filled in separate tables to make comparison.

There is still something to be emphasized that those simulations used a ring around the sensor circuit as ground[17]. In addition, other simulation conducted proved that the size of those ground rings has little difference on the simulation results, and on the other hand, those rings are virtual ones and only exist in simulations. Therefore, the ground rings are not shown or considered in the following discussion.

Simulation results shown ordered by structures of design

Simulation Results of Design1:

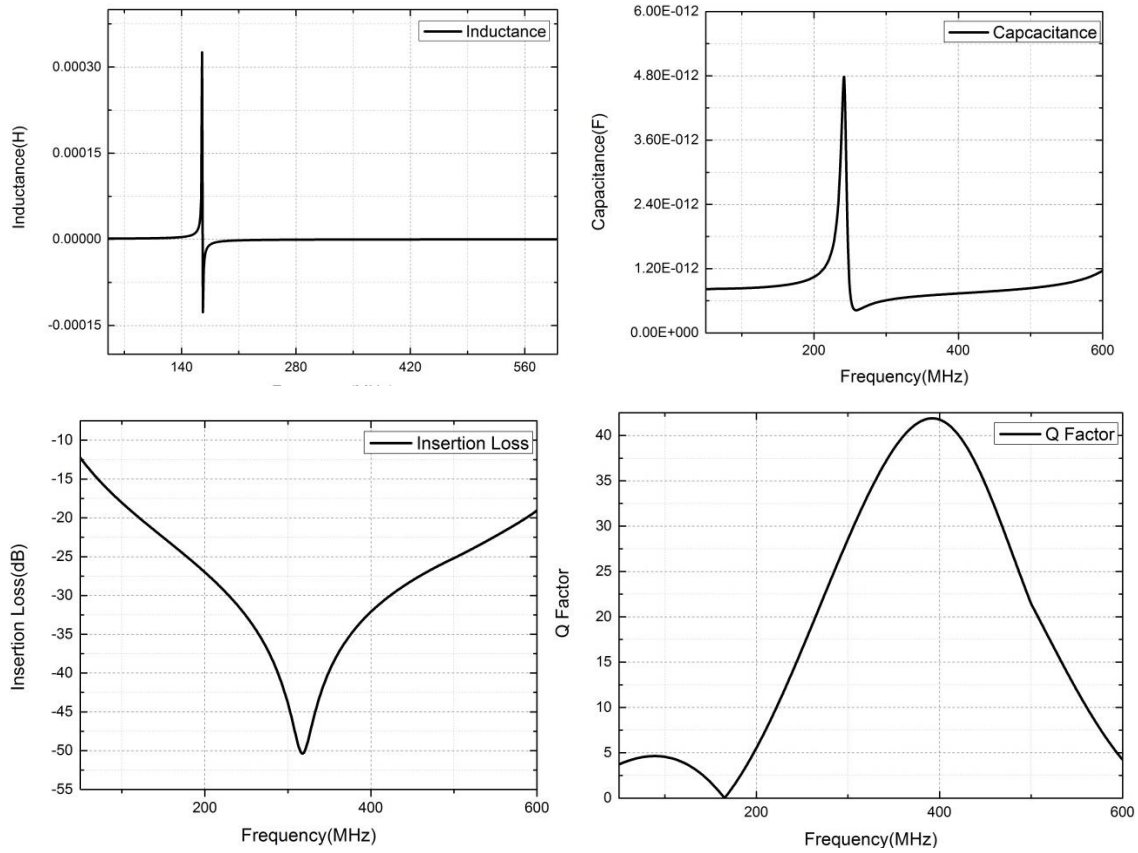


Figure 2.4 Simulation results of Design1

Insertion loss Resonant Frequency is 310MHz

Inductance Resonant Frequency is 160MHz

Capacitance Range is $7.0E-13F - 4.8E-12F$

Q Factor around Resonant Frequency is 15

Simulation Results of Design2:

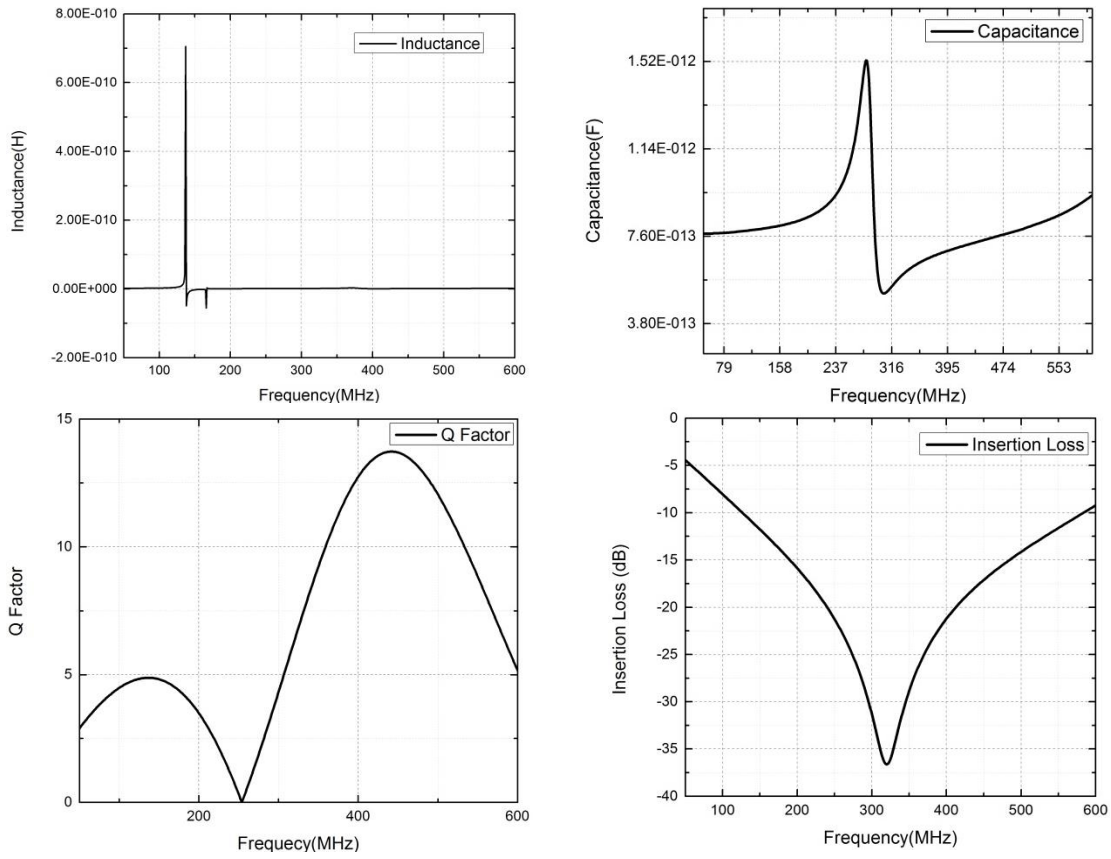


Figure 2.5 Simulation results of Design 2

Insertion loss Resonant Frequency is 320MHz

Inductance Resonant Frequency is 150MHz

Capacitance Range is 7.6×10^{-13} F – 1.5×10^{-12} F

Q Factor around Resonant Frequency is 7.0

Simulation Results of Design3:

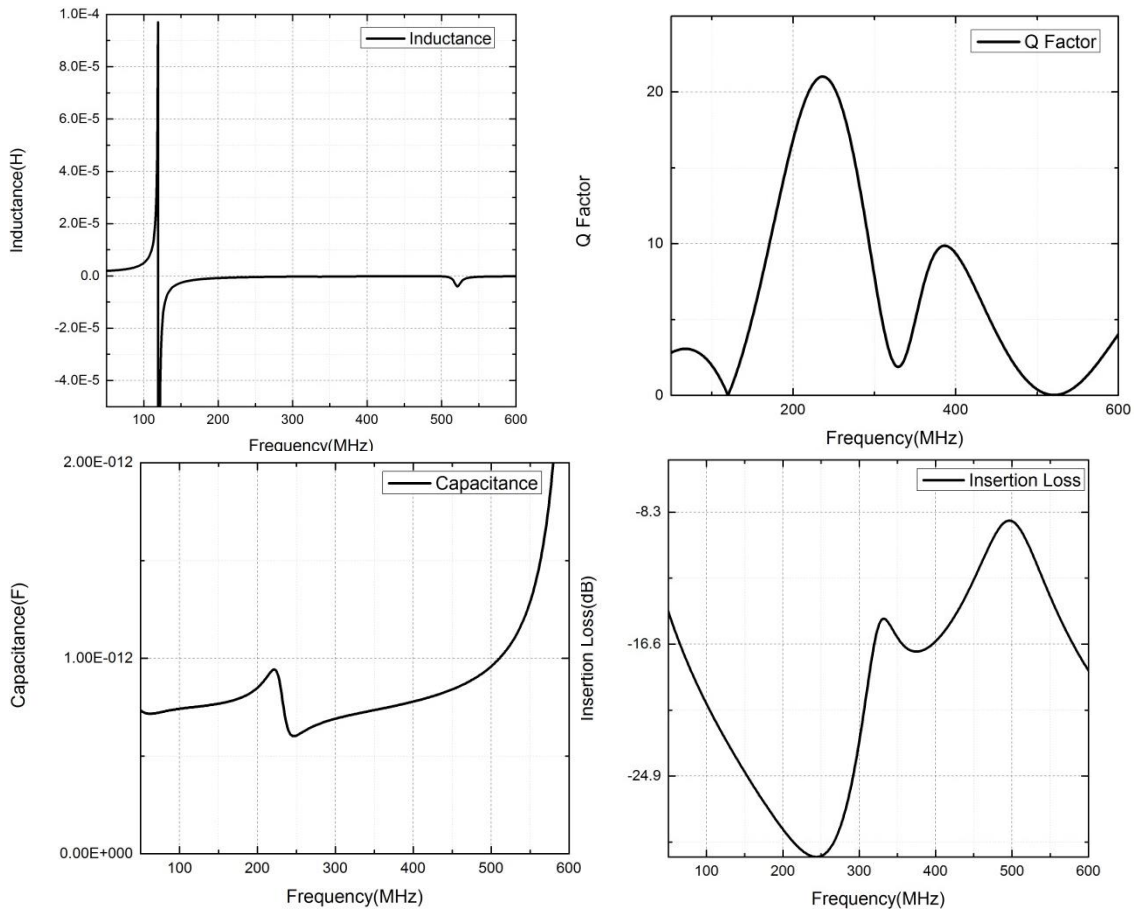


Figure 2.6 Simulation results of Design3

Insertion loss Resonant Frequency is 250MHz

Inductance Resonant Frequency is 120MHz

Capacitance Range is 7.5×10^{-13} F – 2.0×10^{-12} F

Q Factor around Resonant Frequency is 17

Simulation Results of Design4:

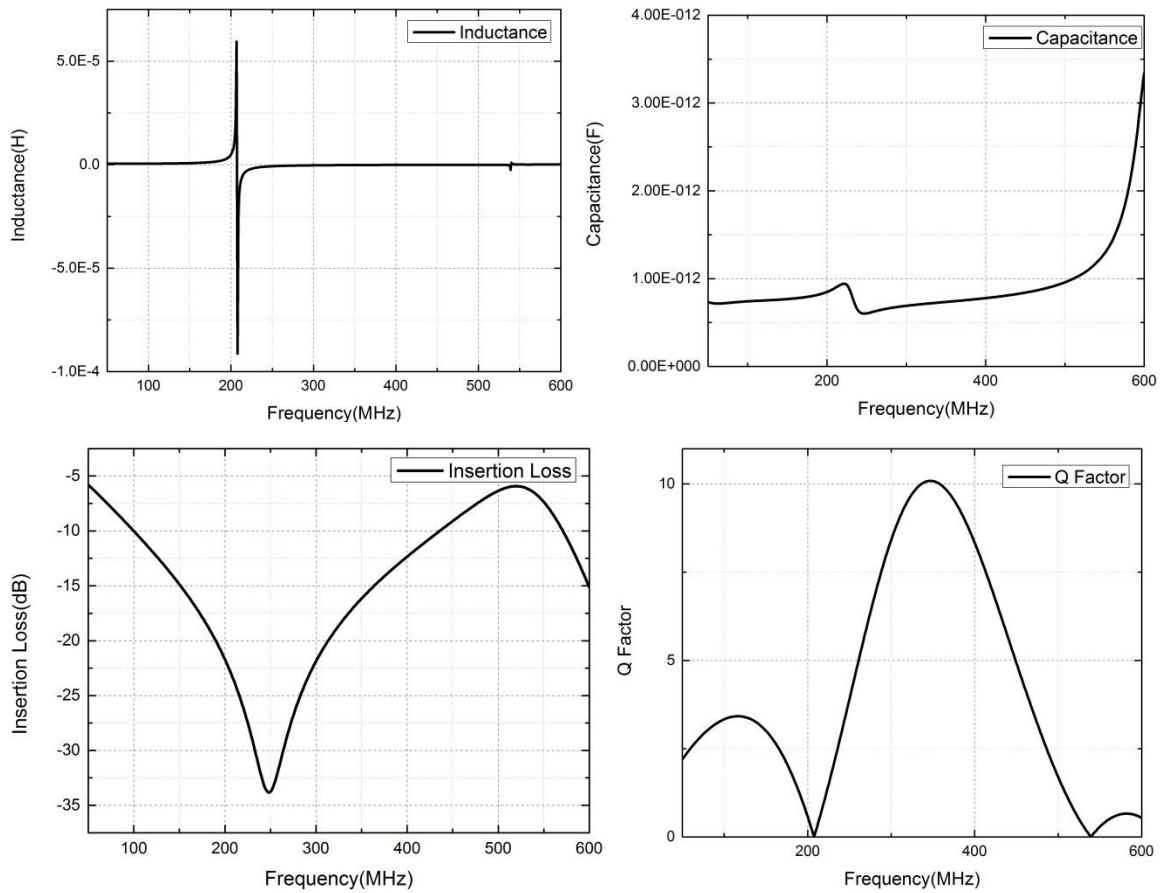


Figure 2.7 Simulation results of Design4

Insertion loss Resonant Frequency is 250MHz

Inductance Resonant Frequency is 210MHz

Capacitance Range is 9.0×10^{-13} F – 3.5×10^{-12} F

Q Factor around Resonant Frequency is 6.5

Simulation Results of Design5:

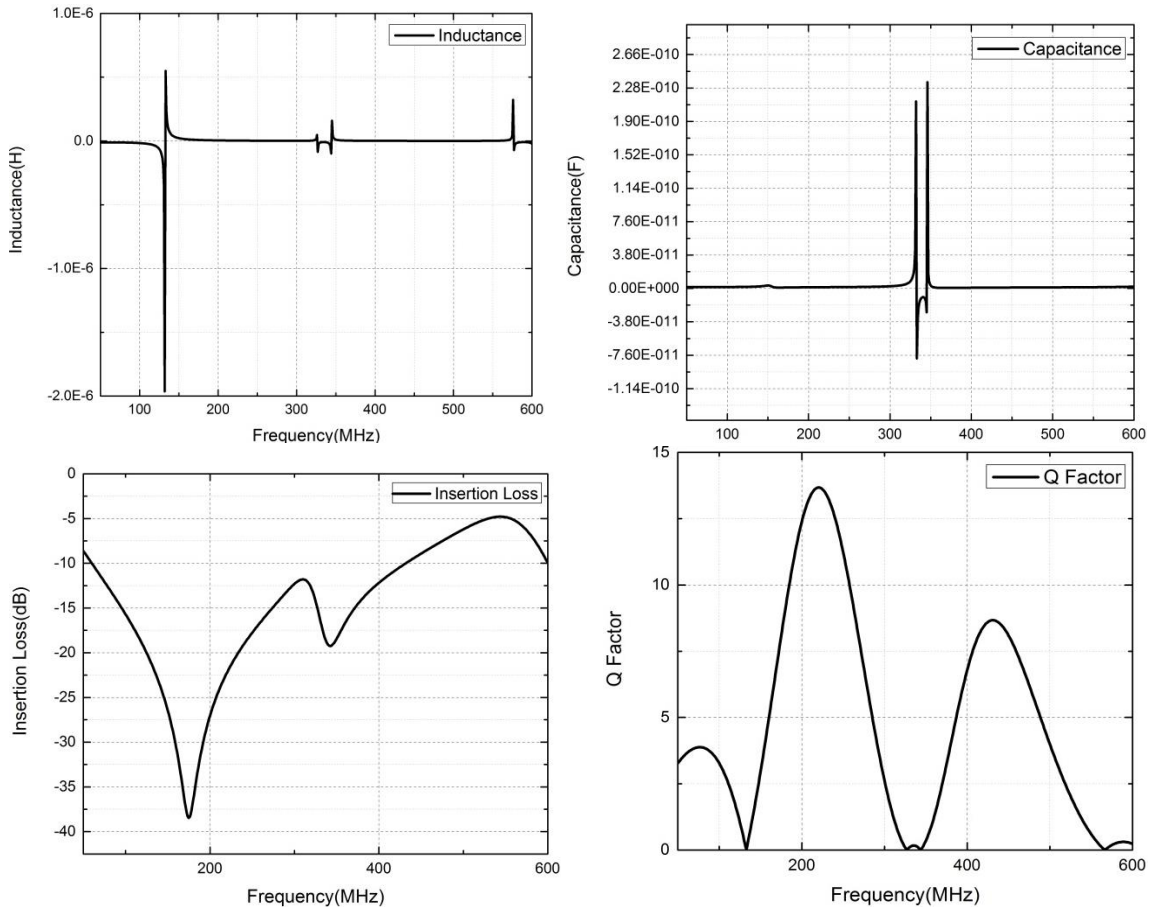


Figure 2.8 Simulation results of Design5

Insertion loss Resonant Frequency is 190MHz

Inductance Resonant Frequency is 130MHz

Capacitance Range is Negative value which is not usable – 2.4×10^{-10} F

Q Factor around Resonant Frequency is 12

Simulation Results of Design6:

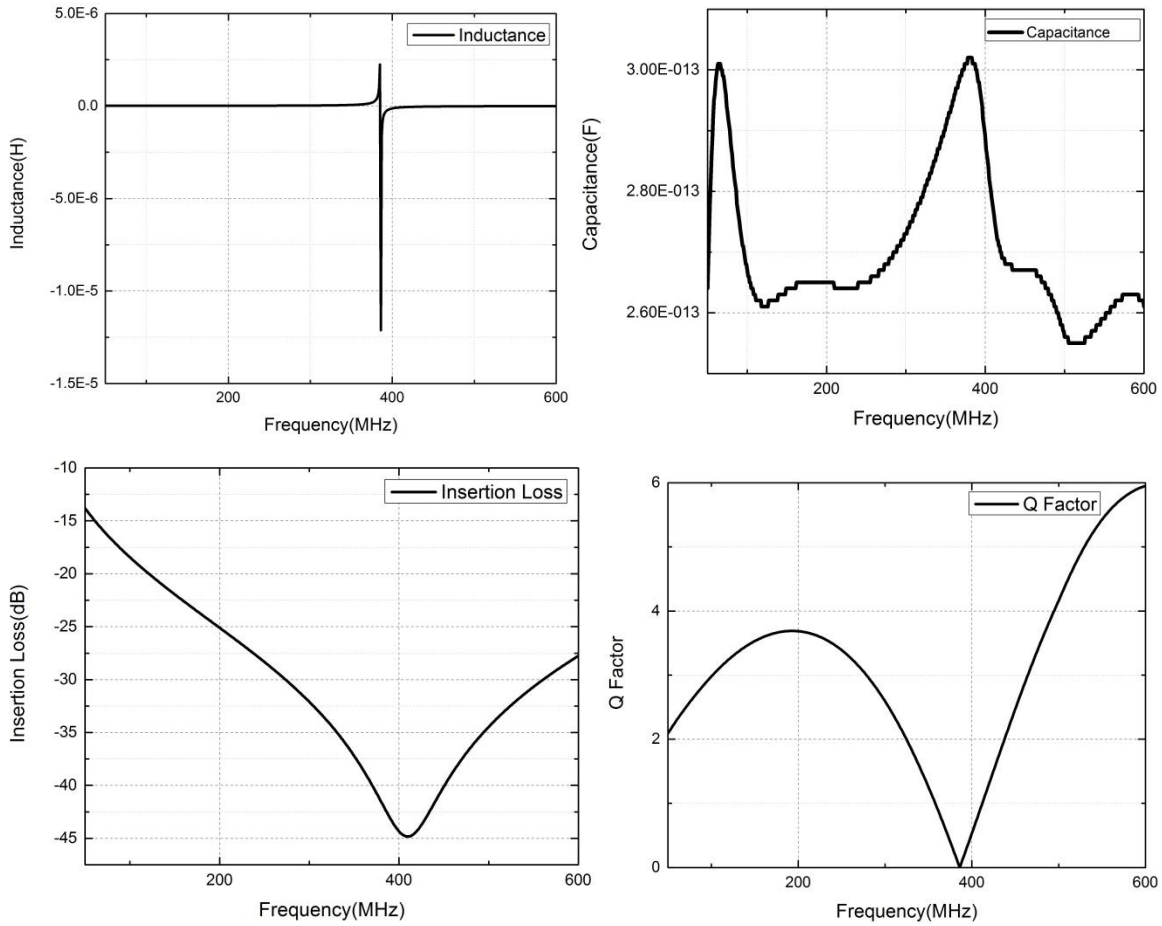


Figure 2.9 Simulation results of Design6

Insertion loss Resonant Frequency is 420MHz

Inductance Resonant Frequency is 390MHz

Capacitance Range is 2.0×10^{-13} F – 3.1×10^{-13} F

Q Factor around Resonant Frequency is 0.5

Table 2.3 Simulation values summary

	IL_RF	IRF	Capa_Range	Q Factor
Design1	310MHz	160MHz	7.0E-13F – 4.8E-12F	15
Design2	320MHz	150MHz	7.6E-13F – 1.5E-12F	7.0
Design3	250MHz	120MHz	7.5E-13F – 2.0E-12F	17
Design4	250MHz	210MHz	9.0E-13F – 3.5E-12F	6.5
Design5	190MHz	130MHz	N/A(-)–2.4E-10F	12
Design6	420MHz	390MHz	2.0E-13F – 3.1E-13F	0.5

IL_RF: Resonant frequency of Insertion loss

IRF: Resonant frequency of inductance

Capa_Range: Capacitance range

Q Factor: Q factor around the resonant frequency

Simulation results sorted by parameters

For an application based on self-resonant circuit, we can build standards to judge whether the simulation result can fulfill our requirements.

(1) Inductance resonant comparison

For the basis of our structure design is an inductance coil, it is best to make the resonant frequency of inductor higher than that of the Insertion loss, or at least near to that. Based on this standard, a table Table.2.4 is shown as following to make the comparison more clearly. All the SRFs are shown in MHz,

Table.2.4 Inductance self-resonant frequency

Design	Inductance Self Resonant Frequency(SRF)
Design1	160MHz
Design2	150MHz
Design3	120MHz
Design4	210MHz
Design5	130MHz
Design6	390MHz

(2) Capacitance Comparison:

To realize a self-resonant frequency, the other parameter of great importance is the capacitance.

According to the self-resonant frequency formula shown in Formula 8:

$$F_{SR} = \frac{1}{2\pi\sqrt{LC}}$$

For making the SRF higher, it is necessary to make capacitance smaller. The capacitances are shown as in Table.2.5.

Table.2.5 Capacitance

Design	C _L	C _H	C on Design Frequency
Design1	7.0e-13F	4.8e-12F	1.1e-12F
Design2	7.6e-13F	1.5e-2F	7.6e-13F
Design3	7.5e-13F	2.0e-12F	N/A(-)
Design4	9.0e-13F	3.5e-12F	N/A(-)
Design5	N/A(-)	2.4e-10F	N/A(-)
Design6	2.0e-13F	3.1e-13F	2.0e-13F

C_L is the lowest capacitance, C_H is the highest capacitance C on SRF is the capacitance on the self-resonant frequency of Insertion loss, C on design Frequency is the capacitance for frequency range from 402-407 MHz (design frequency), all the capacitances come in Faraday, if any resonance happens for capacitance, it is marked as N/A(-).

(3) Insertion loss comparison

Table.2.6 Insertion loss self-resonant frequency

Design	Insertion loss Self Resonant Frequency(SRF)
Design1	310MHz
Design2	320MHz
Design3	250MHz
Design4	250MHz
Design5	190MHz
Design6	420MHz

Known from FCC and FDA regulations [18] and our design propose, the resonant frequency should lie between 402MHz - 407MHz. As a result, the resonant frequency of a design should not be too far from this range, otherwise it may too difficult to optimize. Put all the Insertion loss figures together, as shown in Fig. 16.

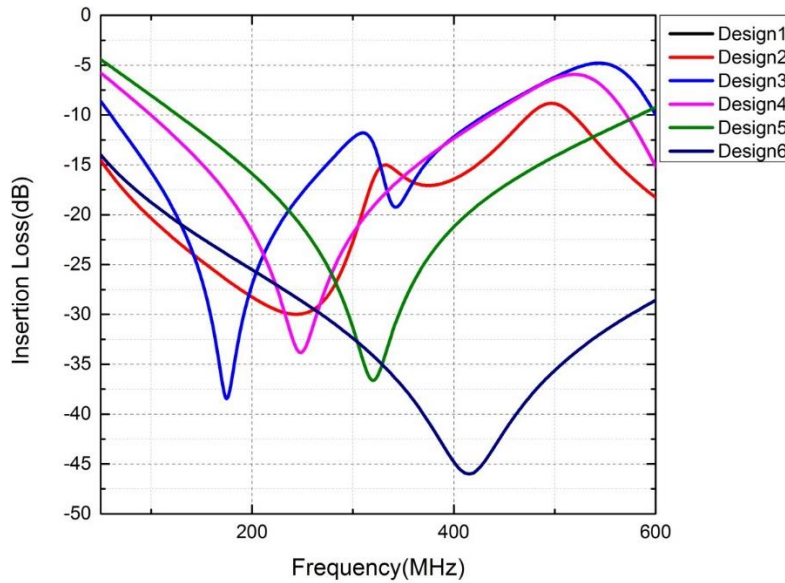


Figure 2.10 Insertion loss comparison

(4) Q-factor Comparison

Q-factor is an importance parameter represents the quality of a circuit. For the current design of an implantable sensor, Q-factor determines the farthest point at which the signal sent by the sensor can be received, in other words, Q-factor determines how far away the examine device can be placed from the patient.

The Q-factor of the chosen design is to have a Q-factor as bigger as possible around the design frequency range. As shown in Table 2.7.

Table.2.7 Q Factor comparison

Design	Q-Factor Around the Design Frequency
Design1	15.0
Design2	7.0
Design3	17.0
Design4	6.5
Design5	12.0
Design6	0.5

(5) Insertion loss SRF comparison

Table.2.8 Comparison of frequencies of different types of SRF

Design	Inductance SRF	Insertion loss SRF
Design1	160MHz	310MHz
Design2	150MHz	320MHz
Design3	120MHz	250MHz
Design4	210MHz	250MHz
Design5	130MHz	190MHz
Design6	390MHz	420MHz

From the simulation results, we can figure out that although Design6 is not one hundred percent satisfying, it is the best fitful design among all the designs.

The following part will be the discussion and optimization based on Design6.

3.2.3 Optimization of simulation

As shown in the simulation result of Design5, for some reason, the capacitance for a dual turning inductance coil, the capacitance is really unacceptable low. As a result, an extra gold pad has been introduced in order to increase the capacitance. The equivalent circuit change is adding a parallel-plate capacitor to the origin circuit.

From the capacitance formula:

$$C = \frac{\epsilon A}{d},$$

In which C is for capacitance, ϵ for permittivity, and A for the area of parallel plate.

It can be figured out that the extra capacitance is determined by the distance between the coil and the pad as well as the area of the pad. Between the two parameters that can be adjusted, it is better to make change to the area alone. The reason for us to do so is that, as shown in Chapter3, in the basic structure of the sensor, an air box is put under the coil, as a result, the extra pad, which is connect to the edge of the coil, will expose in the air and therefore, will be difficult to fabricate (no supporting material), not stable, and easy to break. So the distance between the pad and the coil, in other words, the height of the via, must be fixed.

However, by changing the area of the pad, it is impossible to make the capacitance to change along a simple curve. There are mainly two reasons.

The first one is, in the ideal equivalent circuit, we assume that adding a pad equals adding one extra parallel-plate capacitor, however, in practice, the pad forms at least one other capacitor with other parts of the structure. Meanwhile, as there are spaces between adjacent turnings, the regular pattern of the changing of equivalent area of the capacitor can be different from that of the area of the pad. For example, linear changing of the pad area doesn't certainly mean the linear change of the final capacitance.

The second reason is, adding extra pad doesn't only introduce extra capacitance, but also extra inductance in, small as it is, will still has some influence on the resonant frequency. Therefore, we can only know a trend of how the resonant frequency changes according to the area of the added pad. To know the detailed value, we still need to run some simulations with different pad areas. Eight different values of the width capacitor pad have been applied. Resonant frequencies are shown as in Figure 2.11.

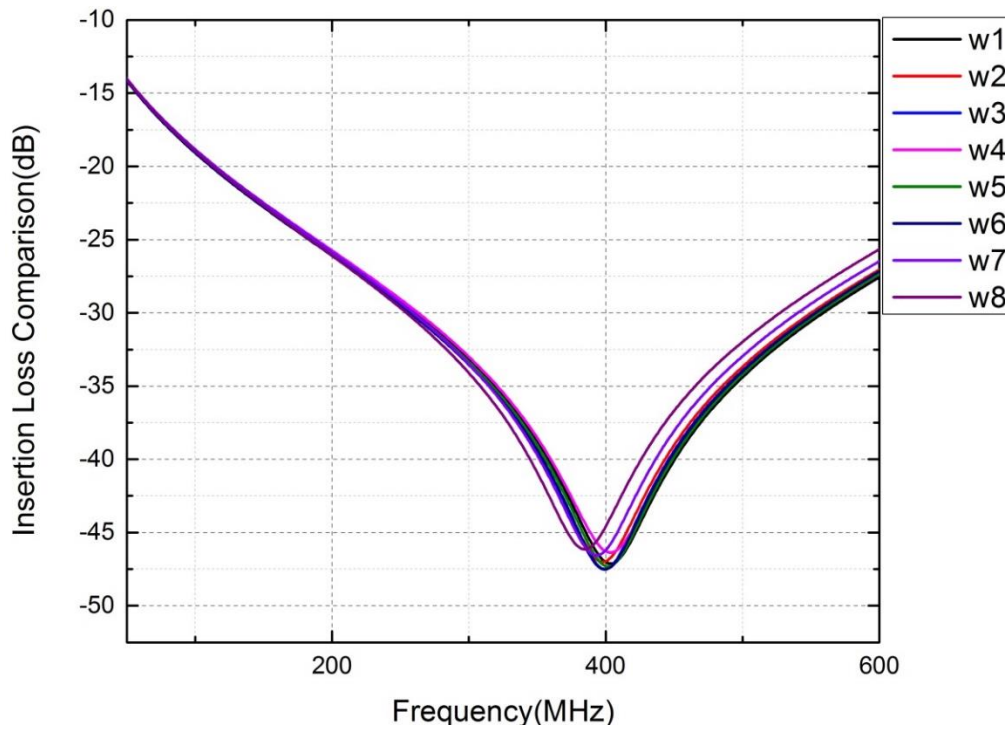


Figure 2.11 Insertion loss comparison for difference pad width of Design6

After the comparison we are now able to determine the when the area is 1665um*1350um, the structure gives the best resonant frequency of 405MHz. Taking the errors may occur in the future fabrication into consideration, it is better to make the simulation to have a resonant frequency in the middle of design frequency range. Simulation result figures are shown as in Figure 2.12.

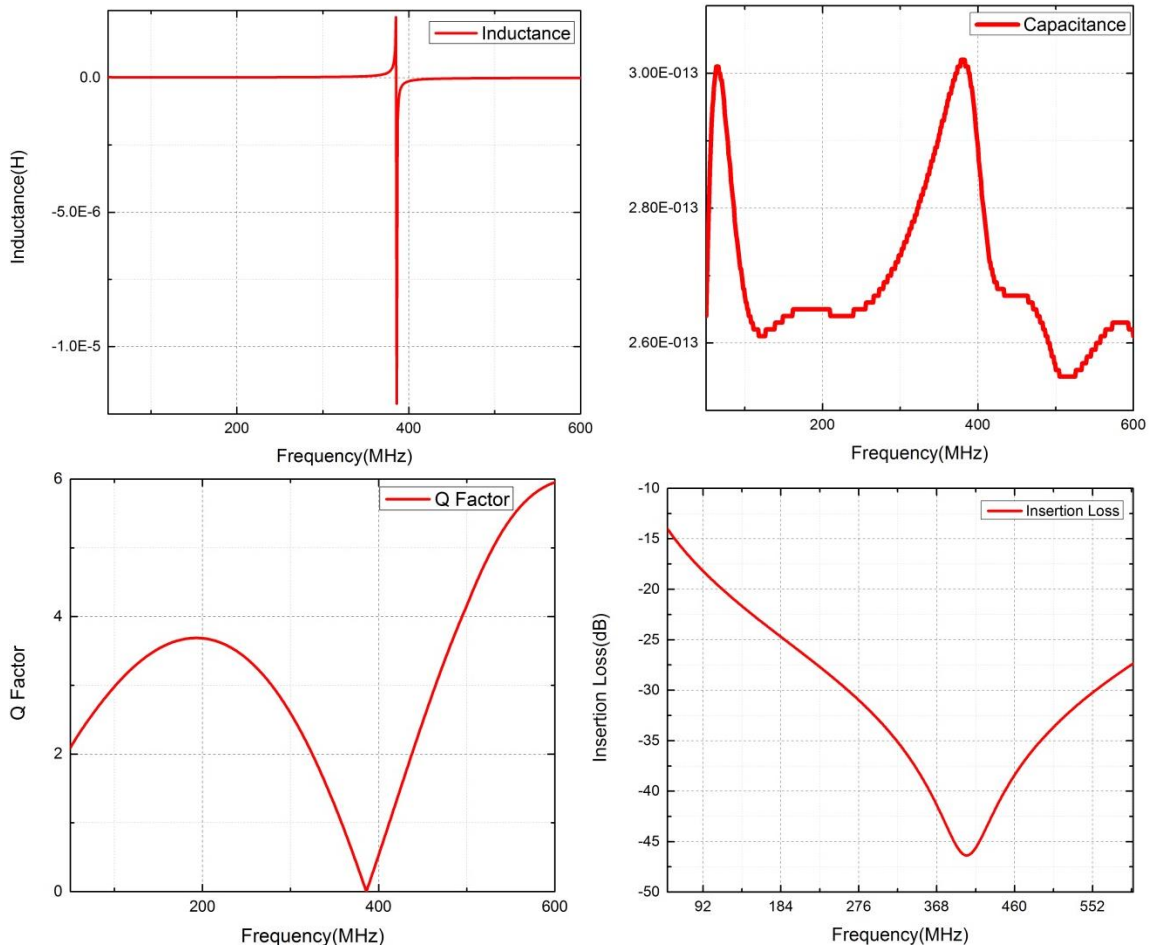


Figure 2.12 Final simulation results of optimized design6

Table 2.9 important simulation values of optimized Design6

Insertion loss Resonant Frequency	405MHz
Inductance Resonant Frequency	390MHz
Capacitance Range	1.4E-13F – 3.0E-13F
Q Factor around Resonant Frequency	0.3

2.6 Materials Selection

2.6.1 Materials of substrate

The substrate material is still changeable. Nowadays, silicon and glass are two types of most commonly use materials for substrate. So the comparison is basically between silicon and glass.

Simulation results for silicon and glass are shown as in Fig. 31(a) and (b) below. And Figure 2.13 clearly shows the difference of them by combining the two curves in one figure, and for in the design required by this thesis, we only need to see the Insertion loss curve comparison.

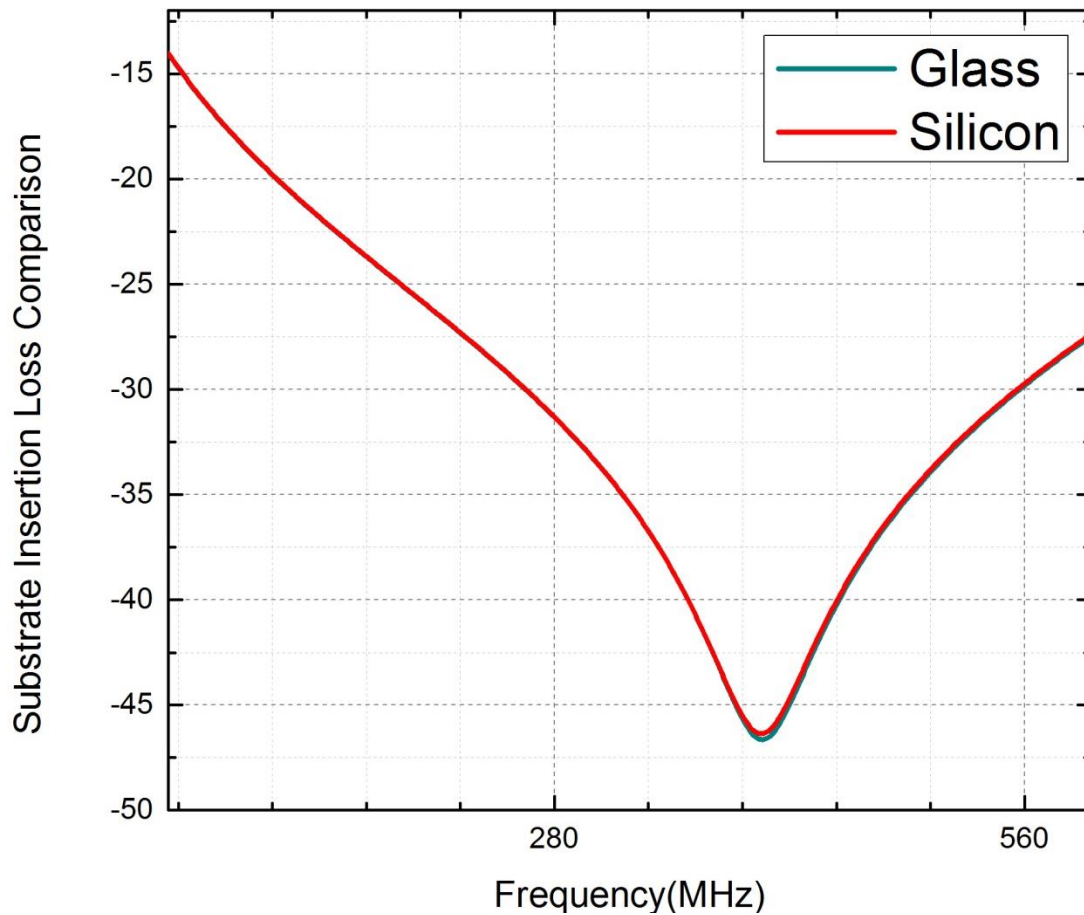


Figure 2.13 Substrate based Insertion loss Comparison

Partially zoomed in figure is shown in Figure 2.14, from which it can be known there is really little difference between glass and silicon substrate for the Insertion loss.

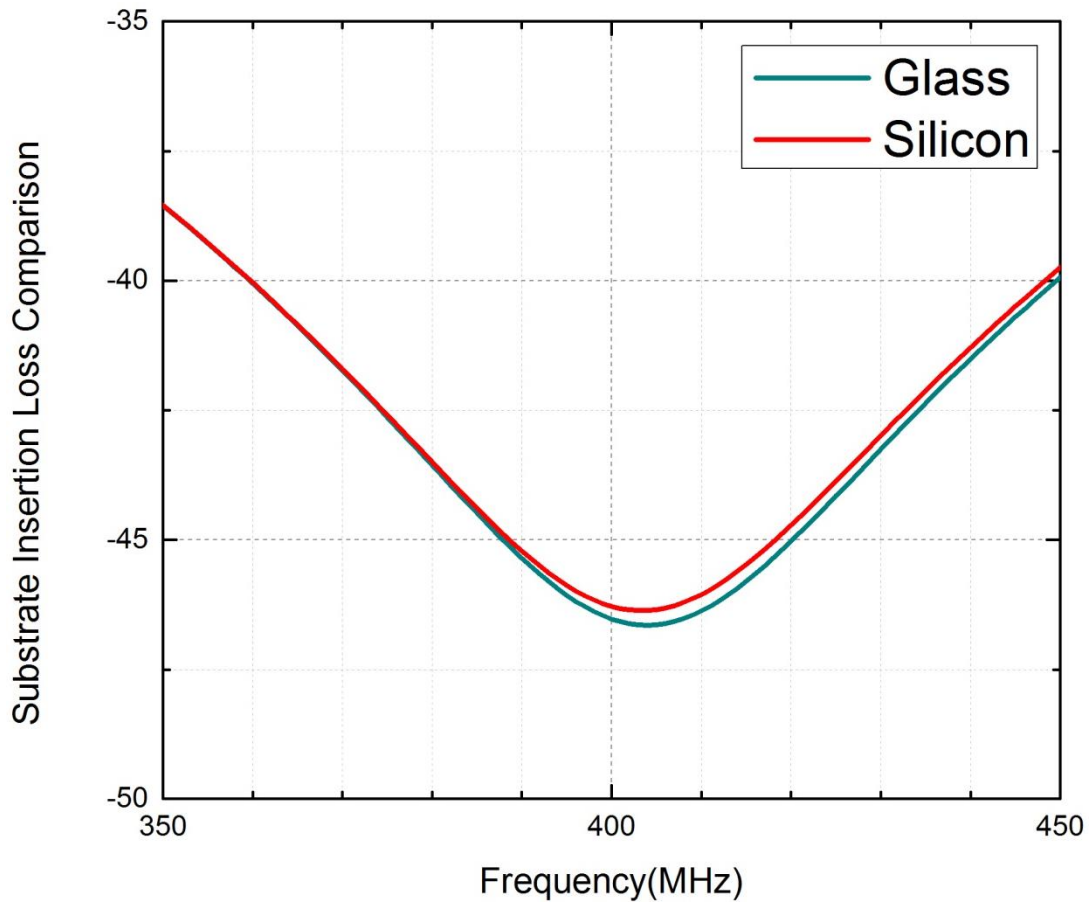


Figure 2.14 Zoomed in comparison of substrate material

Seen from the figures above, Figure 2.13 and 2.14, there are nearly no really difference between the simulation results for silicon substrate and glass substrate. The reason is that the substrate and the coil is isolated by a thick air box, therefore, no matter what substrate is applied, the real substrate will always be air, which in fact doesn't change.

As a result, the selection should be based on some other reason, for example, safety concern. Finally, glass is selected as the substrate.

4.2 Material of wire

Generally, there are three most commonly used materials for implantable sensor copper, gold and silver. All the materials are simulated based on glass substrate.

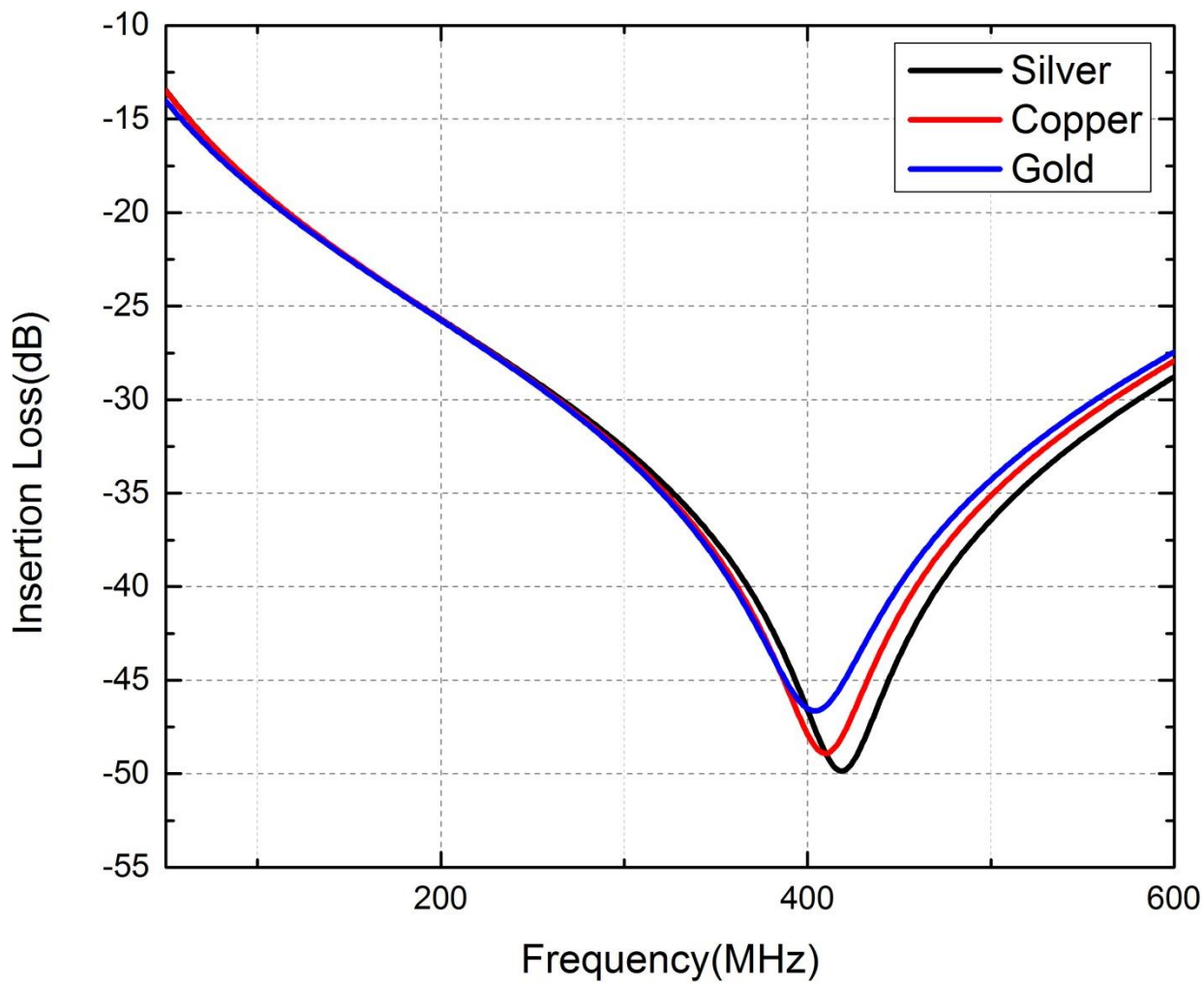


Figure 2.15 Wire material based Insertion loss comparison

Taking the simulation result and the flexibility into consideration, gold has been selected as the wire material.

CHAPTER 3

EXPERIMENT

Basic structure of the sensor system

Despite of the details of the sensor system, the equivalent circuit model can be simplified to be as shown as in Figure 3.1

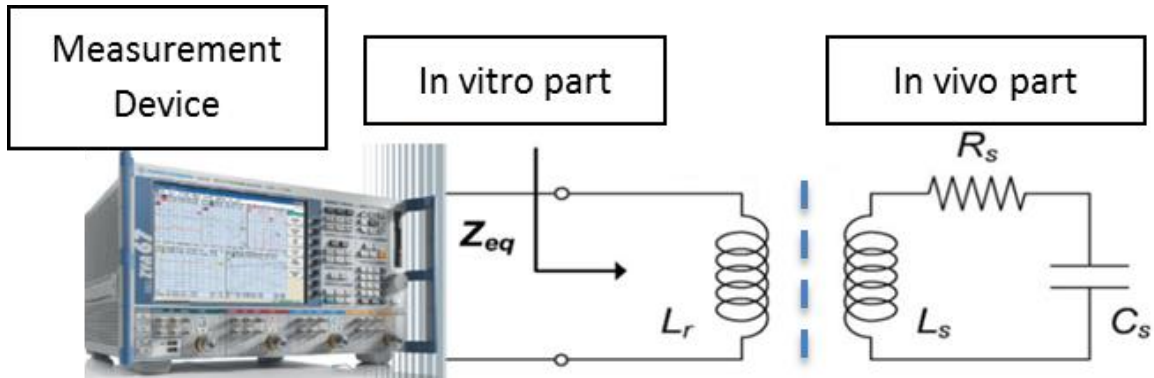


Figure 3.1 Simplified sensor system equivalent circuit model, meaning of each letter combination is shown as in Table 3.1.

Table 3.1 Meanings of letter combinations mentioned in Figure 3.1

In vitro part	The measurement part, simulating the device held by doctors
In vivo part	The circuit which simulates the function of the sensor
Z_{eq}	Equivalent impedance seeing from the measurement device
L_r	An inductance coil connected directly to the measurement device
L_s	An inductance coil, representing the equivalent inductance of the sensor
C_s	Parasitic capacitance of the coil, representing the parasitic capacitance of sensor
R_s	Resistance of the coil, representing resistance of the conductor forms the sensor

Model inductor coils

Observed from the equivalent measurement circuit, in order to validate the measurement system, a pair of inductor coils can be built and act as L_r and L_s . A series of coils has been made.

Sample coils are shown as in Figure 3.2.

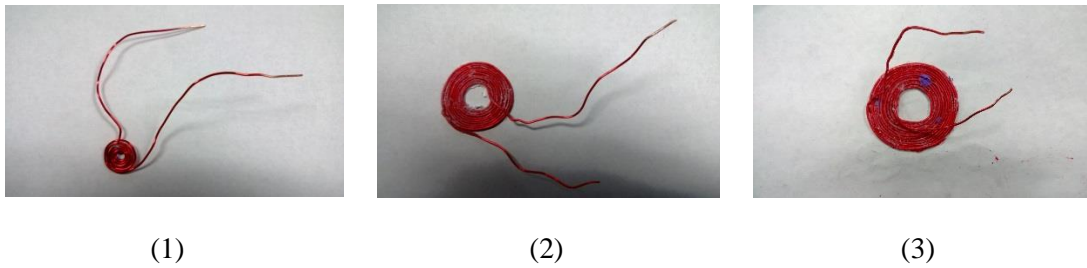


Figure 3.2 Sample coils fabricated

Dimensions and characteristics of sample coils as in Table 3.2

Table 3.2 Dimensions and characteristics of sample coils

Coil	1	2	3
D_{wire}	1mm	1mm	1mm
D_{in}	2mm	12mm	12mm
D_{out}	13mm	30mm	42mm
N_T	6	10	15
Inductance	0.18uH	2.30uH	6.60uH
F_R	22.3MHz	19.0MHz	16.4MHz

D_{wire} : Diameter of the wire used to make the coils

D_{in} : Inner diameter of the coil

D_{out} : Out diameter of the coil

N_T : Number of turns for each coil

Inductance: Inductance measured in the linear range

F_R : Measured self-resonant frequency

Measurement system construction

The overall construction of the measurement system consists of a measurement device, a measurement LC circuit, and the sensor to measure.

An R&S ZVA67 VNA[19] is applied as the measurement device. The input ports are 1.85mm type. Therefore, a cable which transforms 1.85mm port to SMA port is connected to the input port.

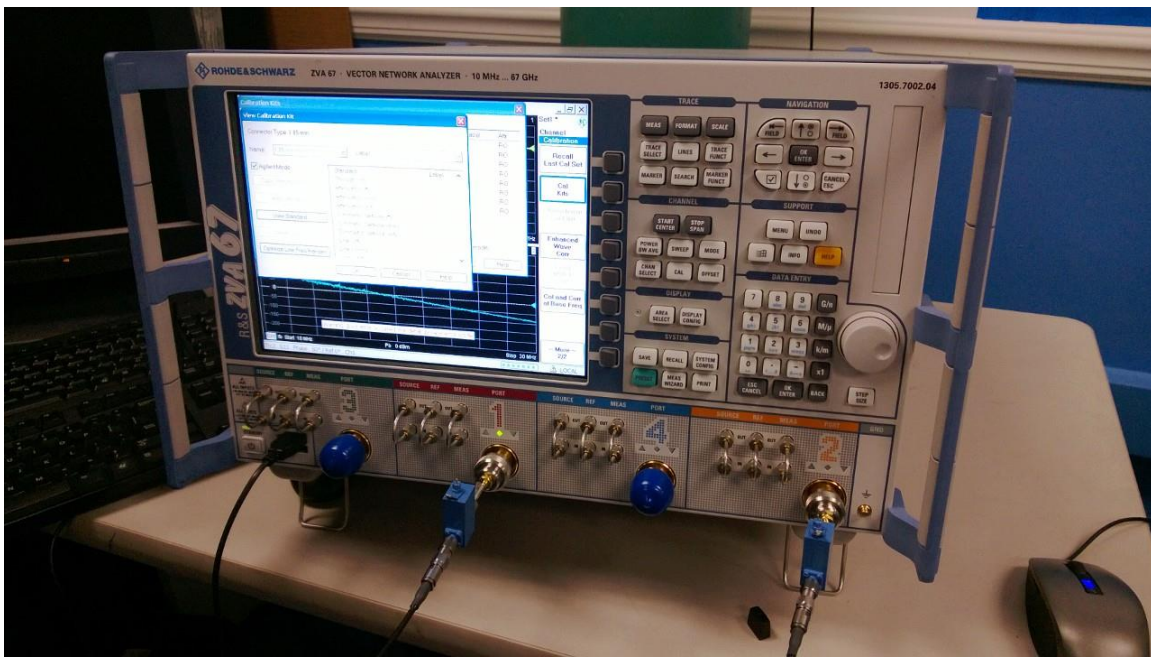


Figure 3.3 Measurement device (VNA) with connection cables (blue blocks with black cables)

In current design, an inductor coil, which will be called as the measurement coil acts as the measurement LC circuit. For the need of connection, inductor coils are connected to

SMA connectors, as shown in Figure 3.4.

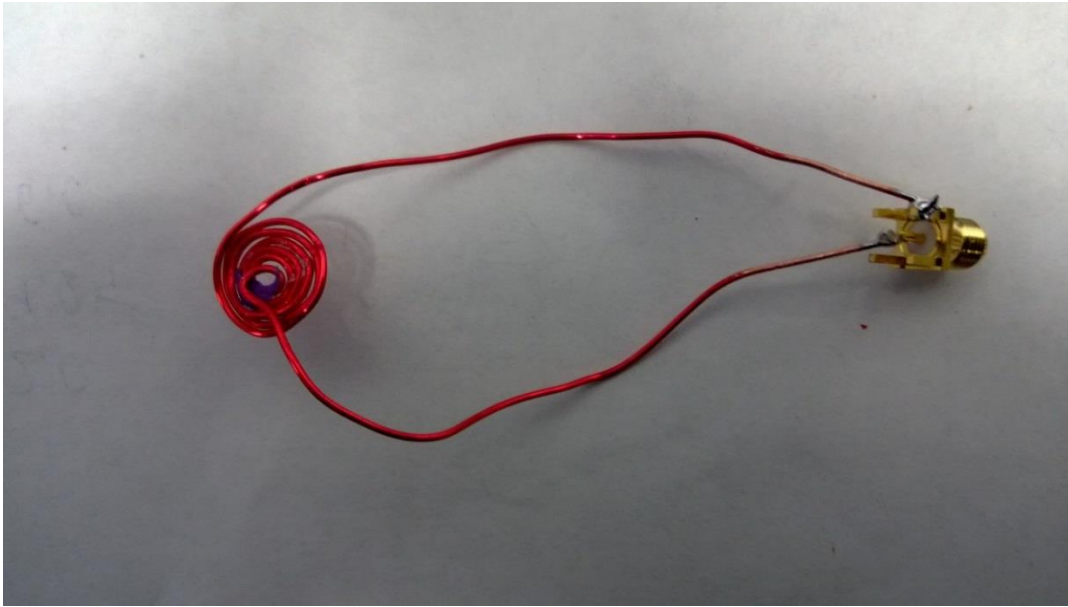


Figure 3.4 Inductor coil soldered to SMA connector

Another coil is to be couple with the measurement coil representing the sensor in shown as in Figure 3.5.

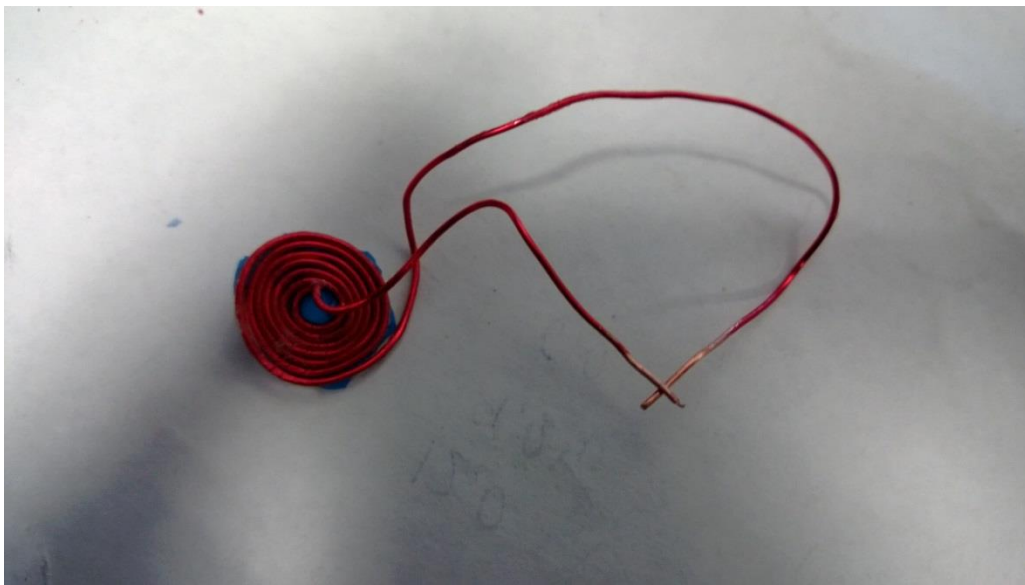


Figure 3.5 Inductor coil soldered to SMA connector

Measurement end installation is shown in Figure 3.6. The coil to be measure is connected and supported by a bread board, the coils centers are set to be in the same horizontal plane for better coupling.

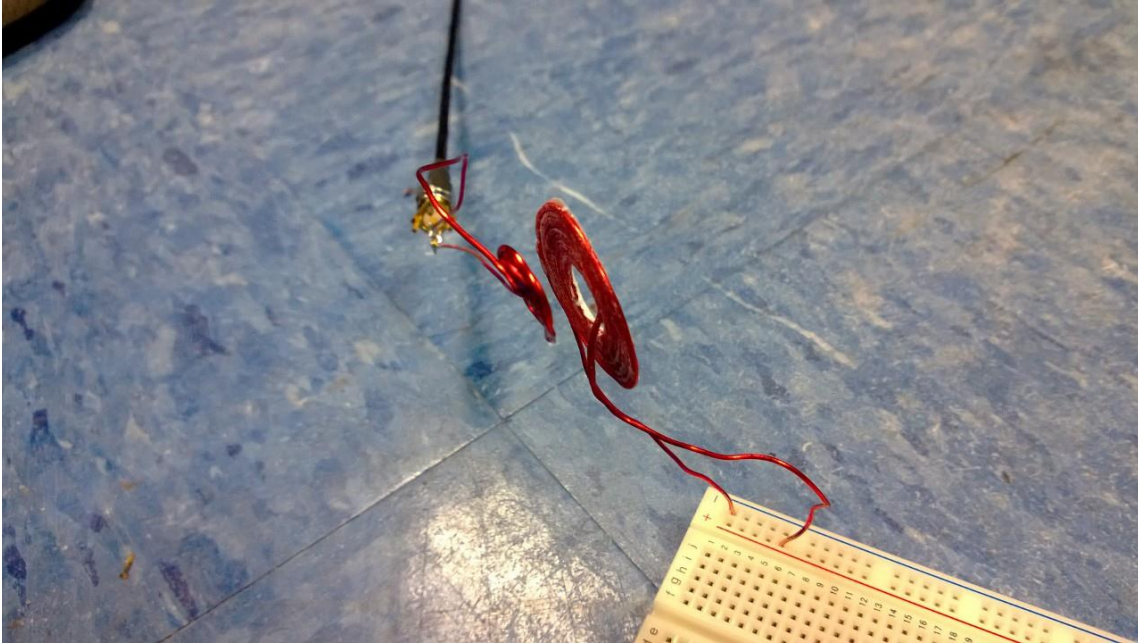


Figure 3.6 Measurement end installation

Qualitative Measurement

Calibration of the VNA is needed as first for accurate results.

A photo which shows the calibration result of VNA is shown as in Figure 3.6, in both Magnitude and phase of the impedance.

Frequency sweeping range is set from 10MHz to 100MHz at the beginning for calibration and finally set to 15MHz – 25MHz for the convenient of observation. In order to observing the resonant frequency shift, we can use either S parameter or Z parameter.

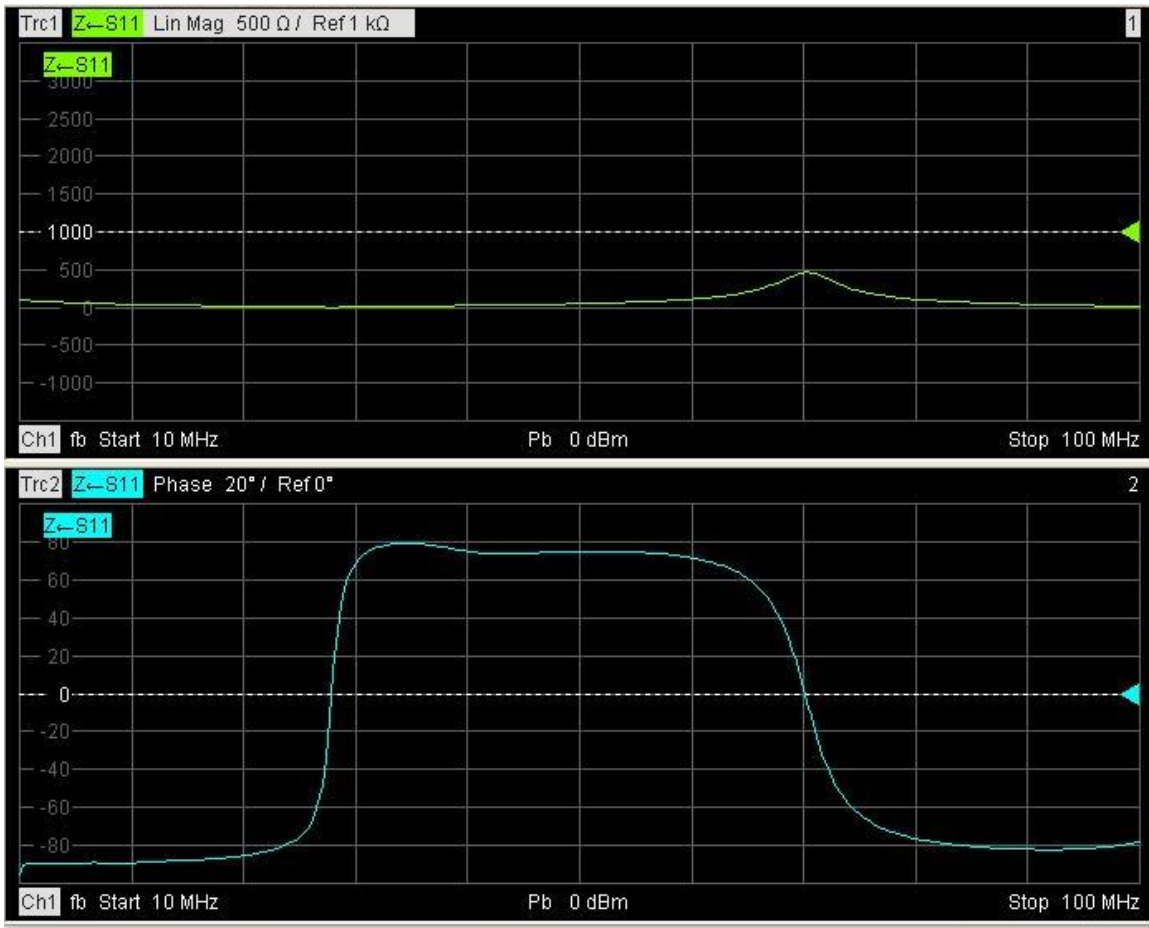


Figure 3.7 Calibration result from 10MHz to 100MHz

Selected measurement results are shown as following. The measurement coil is connected to the VNA and the measurement results are shown as in the Figure 3-6 below. In addition, as shown in the figure, the resonant frequency of the measurement coil is around 17MHz. Therefore, we redefined the frequency sweep range to be 15MHz to 25MHz for better observation.

The number of device measured shown in this chapter is 4, including a fabricated sensor, named as coil a, b, c and sensor, respectively. Comparisons of those figures are also shown. From those figures, it can be observed that although both the impedance magnitude and phase change, the variation of impedance is much more obvious and

easier to tell. As a result, the difference is mainly shown by impedance magnitude figure comparison.

In the previous paragraph, a sensor has been mentioned. That sensor is from a series of sensors fabricated by our colleagues based on the six basic designs introduced in Chapter 2 for testing purpose. The appearance of the sensor and the view under microscope are shown in the Figures 3.8 and 3.9.

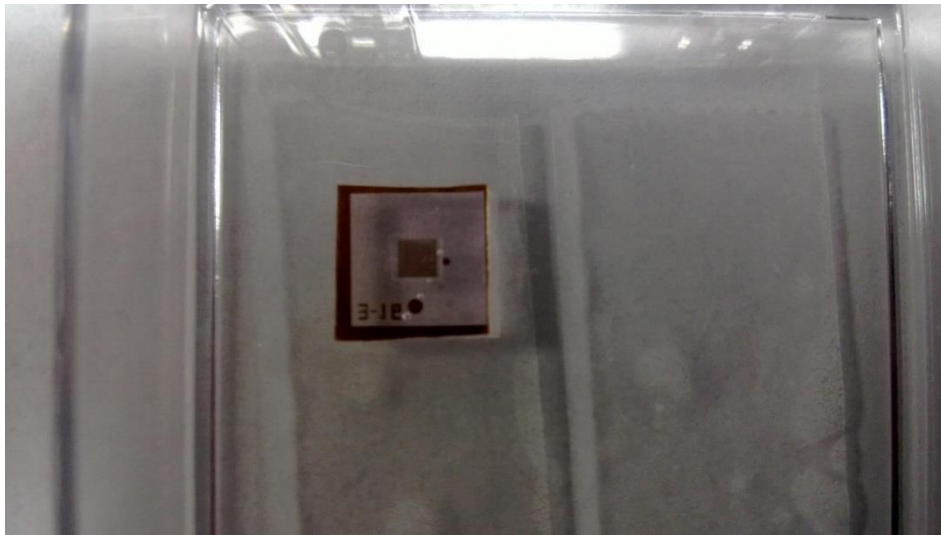


Figure 3.8 Fabricated sensor (golden block in central part)



Figure 3.9 Fabricated sensor under microscope

Figures show the measurement results as well as the comparison are listed below.

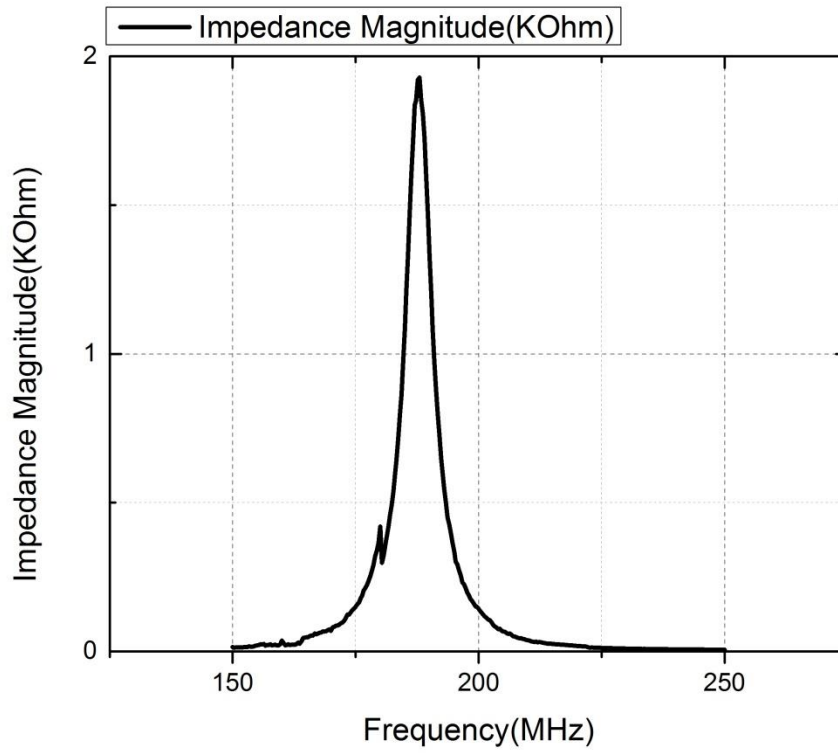


Figure 3.10 Impedance magnitude measurement result of coil a

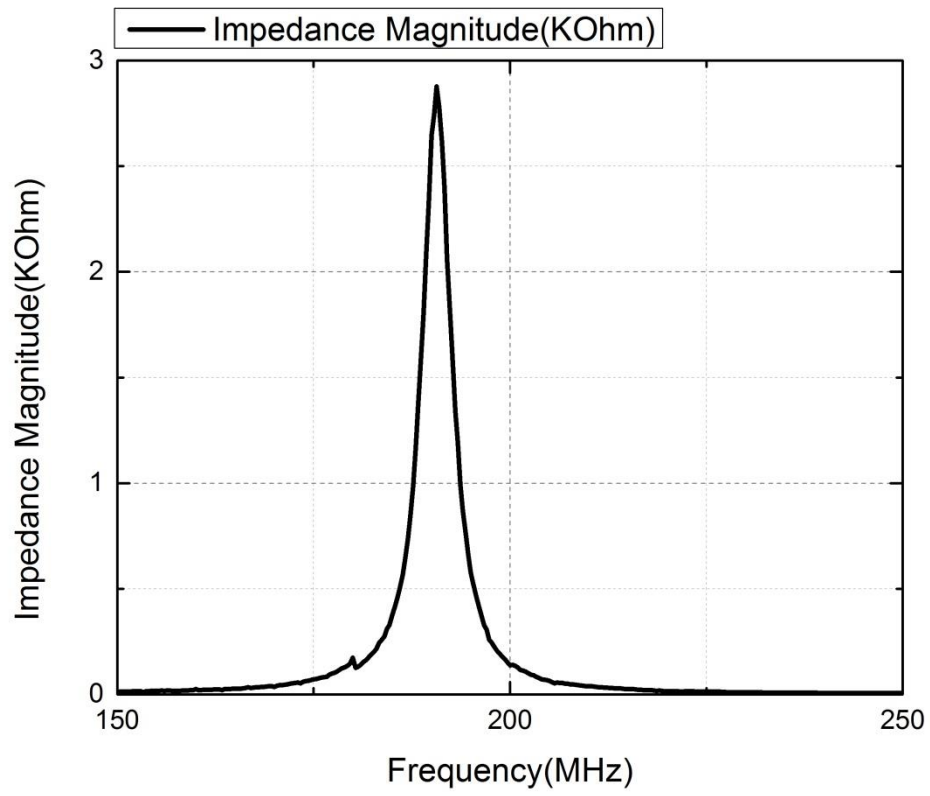


Figure 3.11 Impedance magnitude measurement result of coil b

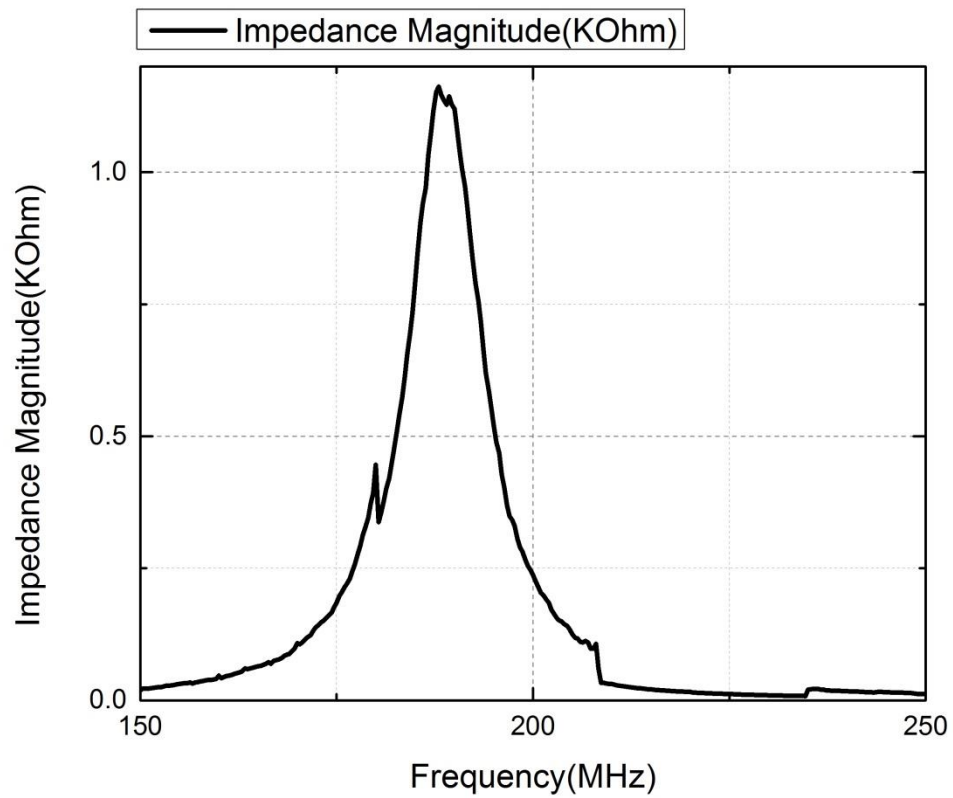


Figure 3.12 Impedance magnitude measurement result of coil c

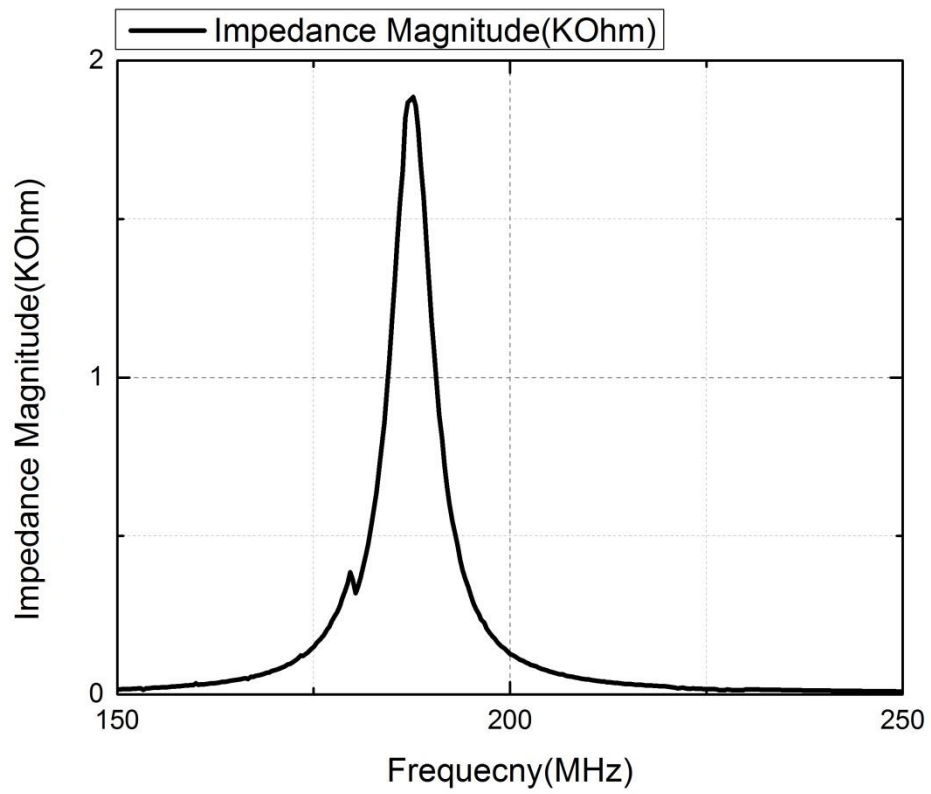


Figure 3.13 Impedance magnitude measurement result of sensor

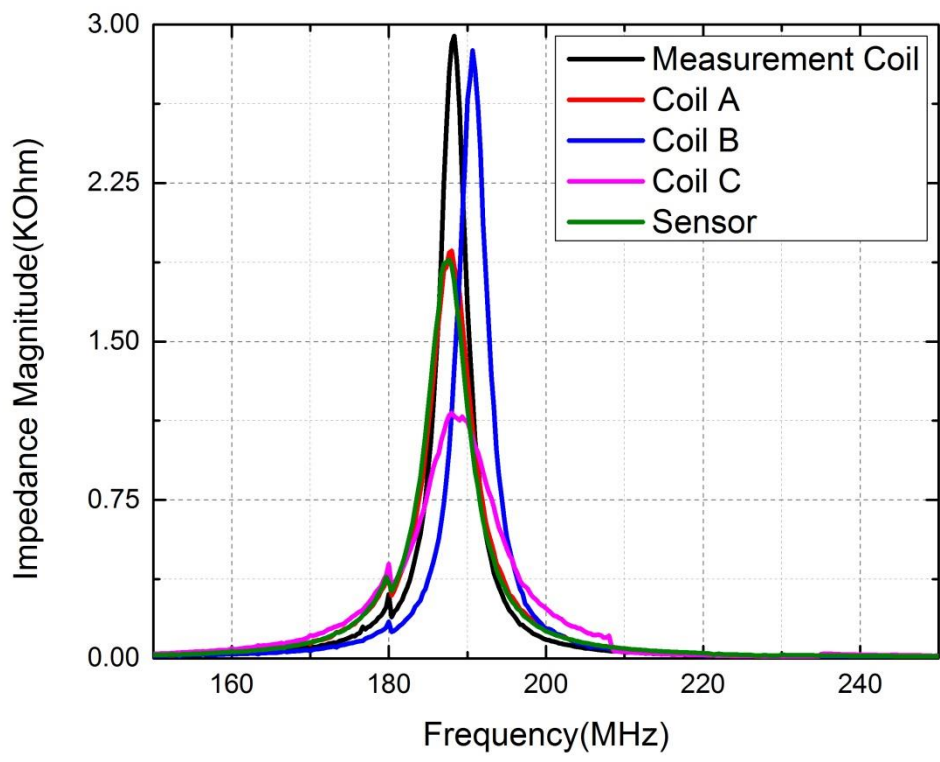


Figure 3.14 Impedance magnitude measurement result comparison

Quantitative measurement conclusion

The qualitative measurement introduced in Chapter 4 proved that when the coupling inductor coil including the sensor changes, in other words, values of inductance and capacitance changes, the variation of resonant frequency is big enough to be observed by the impedance measurement devices.

As a result, the design structure of the sensor as well as the theory basis of the sensor measurement system has been validated and probed to be practical.

Further quantitative analysis is needed to figure out the quantity relationship between the pressure and the impedance variation, which is included in the future work mentioned by Chapter 4.

CHAPTER 4

CONCLUSION AND FUTURE WORK

4.1 Conclusion

In this thesis, a novel implantable blood pressure sensor has been designed and simulated.

The theory basis has also been proved by experiment.

The whole sensor structure is wrapped in PDMS material which is flexible and easy to bend due to external pressure. The internal circuit of the sensor is made of gold and forms an inductance coil attached to a capacitance pad. The circuit is put on top of an air box which allows the coil to be reshaped and supported by a four-sided wall made of SU-8.

In details, the sensor circuit is dual 12/12 turns coil, with an extra capacitor pad attached in the middle whose size is 1665um (width)*1350um (length).

This sensor circuit has an inductance resonant frequency of 390MHz and an Insertion loss resonant frequency of 405MHz.

The sensor is a small-dimensioned one, shown as Figure1-3, the side length is only 3-4mm and the height is just 2.5mm, which makes it occupies little space in a patient's heart, this small dimension is of greater importance for infant congenital heart defects patients who are the target group of this sensor design. In addition, all the materials applied in the implantable sensor are being widely used in modern medical application and proved to be harmless based on the evidence collected up to now, and this will fulfill the safety concern.

The sensor, once implanted, will be able to transmit the patient's heart blood pressure data to the measurement device in vitro. This will help the doctors to determine the patient's status real-timely and accurately. On the other hand, since the implantable parts of the sensor circuit are passive components, no power supply such as batteries are needed, therefore, the sensor only works when it is required to, that will minimize the negative influence on patient's daily life.

4.2 Future work

Real sensors that have the same structure as the final products are to be fabricated and tested in many aspects.

All the simulations and experiments are conducted in the open air, therefore, all the data is acquired under a circumstance when the sensor is surrounded by insulators. However, in practice, as we all know, the human/animal body is a conductor. As a result, this may bring some extra electrical characteristics. All those are to be figured out by conducting a new series of experiments.

The quantity relationship [2] between the impedance phase change and pressure change is to be figured out and confirmed or optimized in order to give a more accurate diagnosing reference.

A curve fitting will be constructed based on the calculation and experiment results for further application.

Trial on animals and clinical trials are to be made after the in vitro tests can give satisfying results.

Final products are to be manufactured after enough data has been acquired and the safety and reliability are certificated.

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