SCHOOL OF COMPUTATION, INFORMATION AND TECHNOLOGY — CHAIR OF CIRCUIT DESIGN

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The Report in Research Internship

Explore ultrasound blood pressure measurement algorithm

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Abstract

This report introduces the traditional ultrasound imaging system with its circuit and three kinds of transducers for ultrasound applications. With the development of medical ultrasound imaging, there is more and more research on integrated transceivers that can be used in smaller and portable devices. There is an increasing need for low-power and portable medical devices for long-term tracking of patients. The low peak power of frequency-modulated continuous wave (FMCW) makes such medical devices possible. The report introduces how FMCW can be used in ultrasound systems and how the circuit of a low-power and portable device could be.

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1 Introduction

In this section, the basic function of ultrasound is first introduced. Ultrasound is a kind of high-frequency sound wave typically between 2 and 18 MHz, which is higher than the audible range for humans (20 Hz to 20kHz). Figure 1.1 is a table that shows the function and application of ultrasound. It can be used in various fields, such as working as a sensor for distance measurement or transferring data in an underwater environment. Non-invasive imaging in the medical field and ultrasound cleaning also benefit from ultrasound. Another table in Figure 1.2 shows the difference between ultrasound data communication with radio frequency (RF) and conventional wire. It can be seen that ultrasound is better than RF or conventional wire with its low interference and ability to cross obstacles. As it is also non-invasive and does less harm to the human body than RF, ultrasound is widely used in the medical field.

Modern ultrasonic systems are becoming increasingly smaller but more powerful

Main function of ultra sound	Application
Sensor	Distance measurement
Data transfer	Underwater communication
Cleaning	Ultrasonic cleaning
Imaging	Non-invasive imaging in the medical field

Figure 1.1: A table shows the application of ultrasound

with the development of integrated ultrasound transceivers. The highly integrated ultrasound systems make portable or even wearable devices possible. It is introduced in [McC+19] that nowadays the need for long-term disease detection is increasing, especially for cardiovascular disease (CVD). For example, blood pressure can be affected by many factors and is a quite important data for diagnosing CVD, so portable devices are needed for a long-term disease detection for better accuracy in diagnosis. A circuit design for such a portable ultrasound device , which can help measure blood pressure with transducers, is the main goal of this report. The methodology of the blood pressure measurement is introduced in the next section. Traditional ultrasound systems in imaging and a new type of system with frequency-modulated continuous wave (FMCW) with a low peak power, which helps to reach low power consumption of the device, have also been introduced.

Data transfer methods	Advantage	Disadvantage	Application
Conventional wire	Reliability, Security, High Speed	Limited mobility, Installation complexity	Local Area Networks(LAN), High-speed Internet connection
RF data	Wireless connectivity, Easier installation, Scalability	Interference, Limited range, Security concerns	Wireless networking, mobile communication, Radar systems
Ultra sound	Low interference, Ability to cross obstacles, Non-invasive	Highly directional, Limited range, Medium dependency	Medical imaging, Short-range data transfer, Underwater communication

Figure 1.2: A table shows the difference between ultrasound and other data transfer methods

2 Methodology

In this chapter, the methodology of the traditional ultrasound system for imaging is introduced. It shows how ultrasound is generated and used. Then, a newly developed system, frequency-modulated continuous wave (FMCW), is introduced. It is specialized with its low peak power, which makes a low-power and portable device possible. With such an ultrasound system, the way to measure blood pressure instead of imaging is explained.

2.1 Traditional ultrasound system

Ultrasound imaging systems are widely used in medical diagnostics to create images of the inside of the body. These systems use high-frequency sound waves to produce images, which can help in diagnosing various conditions. The process of ultrasound imaging involves several key components and steps, which are shown in Figure 2.1.

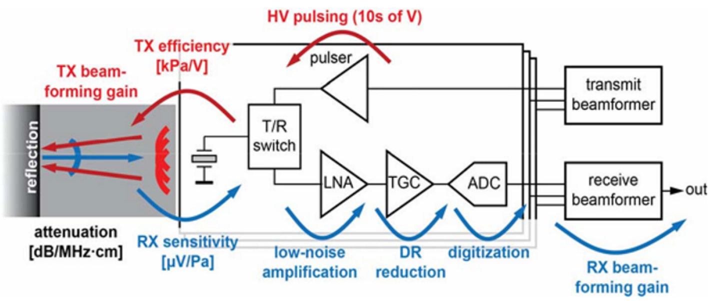

Figure 2.1: Traditional ultrasound imaging system from [CP21]

There are two main parts: Transmitter (TX) and Receiver (Rx). In the figure, the red arrows show the TX part, and the blue arrows show the receiver signal. The first component, the Tx beamformer, is responsible for generating and shaping the ultrasound pulses. The beamformer controls the direction and focus of the ultrasound beam. It adjusts the timing of the electrical pulses to the transducer elements to create focused ultrasound beams. For traditional ultrasound imaging systems, the pulse signal is used. The pulser generates high-voltage pulses that drive the transducer. These pulse signals are very brief and occur in rapid succession, creating short bursts of ultrasound waves. It is shown in Figure 2.2 what the pulse signal looks like. These pulses determine the frequency and duration of the sound waves emitted by the transducer. Transducers can turn high-voltage pulses into audio signals. The way how transducers work will be explained in the next chapter. The ultrasound waves travel through the body and reflect off different tissues and structures. The transducer receives the echoes of the ultrasound waves that bounce back from the tissues. The received echoes are converted back into electrical signals by the transducer. There is a switch controlling the mode changing of the whole system. This switch alternates the connection of the transducer between the transmitter and receiver circuits, which ensures that the high-voltage transmission pulses do not damage the sensitive receiver circuits.

After the echoes are received by the transducer, they are turned into an electrical

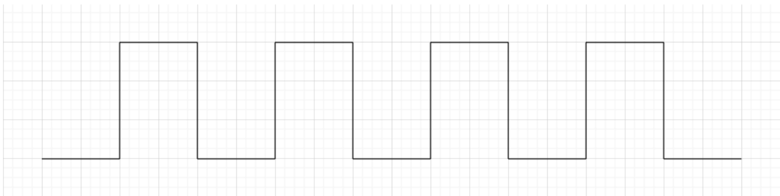

Figure 2.2: Successive pulses

signal again. The Low-Noise Amplifier (LNA) amplifies the weak echoes received from the body with minimal added noise. LNA is the most complex part of the whole system. Ultrasound echoes returning from the body are extremely weak, especially those reflected from deeper tissues. The LNA amplifies these weak signals to a level where they can be further processed without significant loss of information. High gain can also increase the noise from the reflection as well as from the LNA itself. To manage this, the LNA may need to use larger transistors and additional circuits to keep the noise levels low, contributing to increased area and power consumption. So, the design of LNA is always a hot topic, especially for portable devices. The Time Gain Control (TGC) adjusts the amplification of the received signal based on the depth of the tissue being imaged. Since echoes from deeper tissues are weaker, TGC increases the gain for these signals to ensure a consistent image quality. The final part is Analog-to-Digital Converter (ADC), which converts the analog signals from the TGC into digital signals. These digital signals can then be processed by the receiver beamformer to create an image. The whole system can then provide the desired images on the monitor for medical diagnosis.

2.2 Blood pressure measurement

In [Seo+15], the method to measure blood pressure is introduced. The example system is shown in Figure 2.3. It is specialized for its simple structure, which requires only two transducers to get all the information needed. The red tube represents the blood vessel and two "CHs" are the ultrasound transducers.

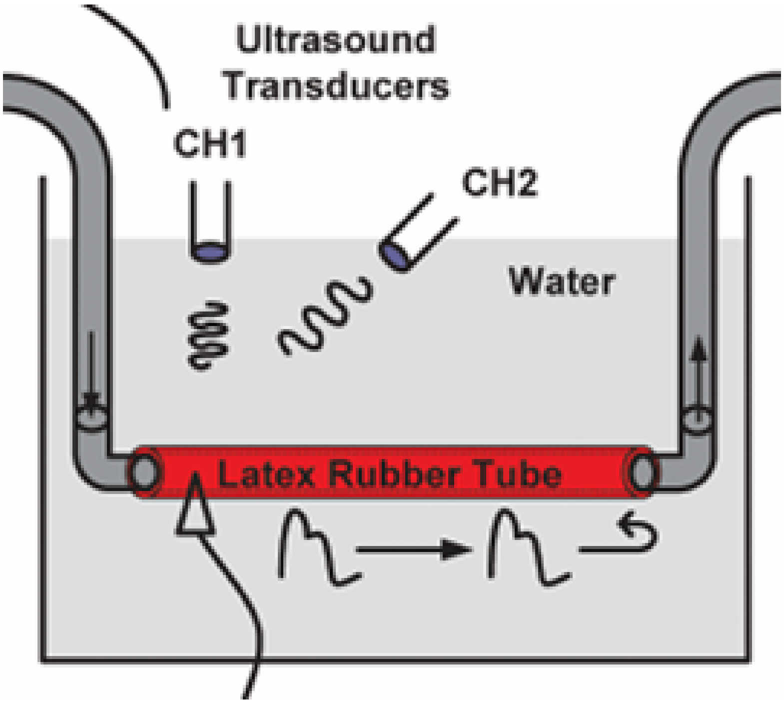

Figure 2.3: The ultrasound system for blood pressure measurement from [Seo+15]

What is needed for this method is pulse wave velocity (PWV) and cross-sectional area (A) of blood vessels. According to the Bramwell–Hill equation, PWV is directly related to the compliance (dA/dP) as

$$PWV = \sqrt{\frac{A}{\rho}\frac{dP}{dA}} \tag{2.1}$$

where A is the cross-sectional area, P is the blood pressure, and ρ is the density of blood. In [Seo+15], the formula is rewritten with a flow-area method. Local PWV is estimated using the flow-area method based on the change of flow rate with respect to the change of area during a reflection-free period in the cardiac cycle. PWV can then be expressed in terms of the slope of the flow-area plot during the reflection-free period, as shown in Figure 2.4. The reflection-free period is the line between A and B, and PWV can be seen as the slope of the line. It means that PWV can be expressed in the following formula:

$$PWV_{QA} = \frac{dQ}{dA} \tag{2.2}$$

Q(in cm^3/s) is the flow of the blood, which can be calculated through

$$Q(t) = v(t) * A(t) \tag{2.3}$$

v is the blood velocity, which can be measured through FMCW signal. The method is introduced in the next section.

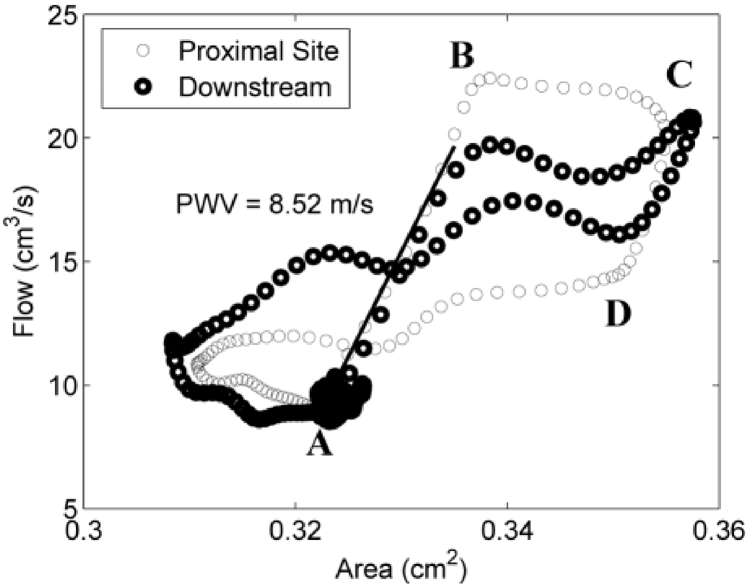

Figure 2.4: Measured flow–area plots from [Seo+15]

By reforming this formula, assuming PWV remains relatively constant with the change of area during a cardiac cycle, a pressure waveform is expressed in terms of PWV and the cross-sectional area A as

$$P_t - P_0 = \rho PWV^2 \ln(\frac{A(t)}{A_0}) \tag{2.4}$$

where P_0 is the pressure measured at the cross-sectional area A_0. These values are reference values for calculating blood pressure in reality. The area can be calculated by the diameter of blood vessels. When the transducer is vertical to the blood vessel, the diameter can be measured through the reflection effect, as usual in distance measurement with an ultrasound system. PWV can be calculated through the flow-area method. The flow is what we need and it can be estimated using the blood velocity, which can be obtained from the other transducer with the FMCW signal.

2.3 FMCW system

Frequency Modulated Continuous Wave (FMCW) is a continuous wave signal with a frequency that varies linearly over time. Compared with the pulse signal used in traditional ultrasound systems, it has lower peak power, which makes it safer as a high-voltage pulse when applied to medical devices. The power consumption of FMCW is also lower. It also has the ability to measure simultaneously the target range and its relative velocity through specific signal processing.
In [KSM08], the working process of FMCW is explained. FMCW signal is first generated and transmitted. When the signal hits a target, part of the signal is reflected back to the receiver. Then, the transmitted signal and the received reflected signal are mixed (multiplied) together in a mixer. The mixing process produces a difference frequency that corresponds to the time delay between the transmitted and received signals. This frequency is directly related to the distance of the target. The Doppler shift can be extracted from the frequency shift, providing relative velocity information. The signal processing after mixing is performed at a low-frequency range, considerably simplifying the realization of the processing circuits.

$$R = \frac{c_0|\Delta t|}{2} = \frac{c_0|\Delta f|}{2(\frac{d(f)}{d(t)})} \tag{2.5}$$

This formula from [FMCWradar] shows how FMCW radar calculates the distance between the antenna and the reflecting target. In an ultrasound system, it is similar, but c_0 should be replaced by the speed of the audio signal in the medium, e.g. blood. $\Delta t$is delay time and Δf is measured frequency difference. R is the distance that needs to be calculated. df/dt is the frequency shift per unit of time. How Δt and Δf is measured is shown in Figure 2.5.

This figure contains three plots. In the first plot, the red line is the transmitted signal and the green line is the received echo signal. The echo signal is shifted due to the running time compared to the transmission signal to the right. A Doppler frequency shifts the echo signal in height. Without a Doppler frequency, the frequency difference during the rising edge is equal to the measurement during the falling edge which means $\Delta f_1 = \Delta f_2$. In Figure 2.5, the second plot represents the beat frequency as a function of t. This plot shows how the frequency difference between the transmitted

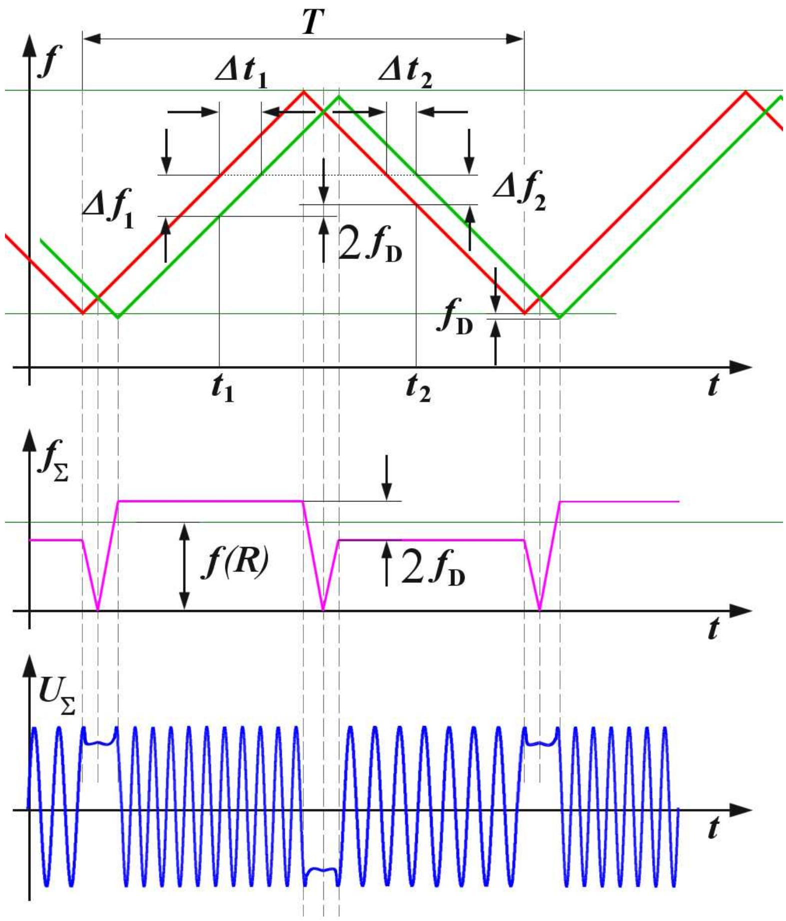

Figure 2.5: A common relationship with triangular modulation pattern from [FM-CWradar]

and received signals varies over time. The plot shows how these components add up to give the instantaneous beat frequency. For a moving target, $2f_D$ shifts the frequency up or down depending on the relative motion. The third plot represents the actual mixed signal U_Σ in the time domain. This is the signal obtained by mixing (multiplying) the transmitted and received signals, producing the beat frequency. It combines both the range and Doppler information. Performing a Fourier Transform on the mixed signal can convert the time-domain signal into the frequency domain, which is exactly the second plot. It makes it possible to extract f_D and f_R. The following formula shows how Doppler frequency and the frequency as a measure of the distance measurement is calculated.

$$f(R) = \frac{\Delta(f_1) + \Delta(f_2)}{2} \tag{2.6}$$

$$f(D) = \frac{|\Delta(f_1) - \Delta(f_2)|}{2} \tag{2.7}$$

After we get f_D, we can use the formula from [Kun+10].

$$f_D = \frac{2v}{c} f_0 \cdot cos(\theta) \tag{2.8}$$

In this formula, v is the velocity of the target, c is the velocity of ultrasound in different mediums, f_0 is the carrier frequency, and θ is the angle between the ultrasound beam and the target. v can be calculated by rewriting the formula. It is essential for our calculation of PWV and blood pressure. Such an FMCW system can help measure the speed of blood flow in blood vessels. Combined with the cross-sectional area, the blood pressure can then be calculated and made into plots. This is the methodology of the whole FMCW system, which can measure blood pressure and be designed into a portable and low-power device.

3 Implementation

In this section, the implementation of circuits of the FMCW ultrasound system is introduced. It contains transmitter and receiver parts. Before the circuit is shown, different types of transducers are also introduced.

3.1 Transducers introduction

There are three kinds of transducers that are now used widely for ultrasound systems. They are Piezoelectric Transducers (PZT), Piezoelectric Micromachined Ultrasound Transducers (PMUT) and Capacitive Micromachined Ultrasound Transducers (CMUT). The main function of transducers is to turn electrical power into sound waves that penetrate the body. The sound waves get reflected by the body tissue and are received by the transducers. The transducer then converts the echoed sound waves to electrical signals. Traditional ultrasound probes contain transducer arrays made from slabs of piezoelectric crystals or ceramics, which are the most commonly used transducers, the PZT. When hit with pulses of electricity, these slabs expand and contract and generate high-frequency ultrasound waves that bounce around within them. The structure of PZT is shown in Figure 3.1 (a).

By using a type of microelectromechanical system (MEMS), new architectures are created. It is introduced in [CP21] that the CMUT operates based on electrostatic forces within a capacitor composed of two conductive plates separated by a narrow gap. One plate, the micromachined membrane, is constructed from silicon or silicon nitride with a metal electrode. The other plate, typically a thicker and more rigid micromachined silicon wafer substrate, provides structural support. Applying a voltage induces opposite charges on the membrane and substrate, generating attractive forces that pull and flex the membrane toward the substrate. Introducing an oscillating voltage alters this force, causing the membrane to vibrate similarly to a struck drumhead. Figure 3.1 (b) shows the structure of CMUT with V_{ac} applied to it.
PMUT is similar to PZT in working theory, but it is driven by MEMS. It functions like a miniature version of a smoke alarm buzzer. These buzzers are made up of two layers: a thin metal disk fixed around its edges and a smaller, thin piezoelectric disk attached on top. When voltages are applied to the piezoelectric material, it expands and contracts

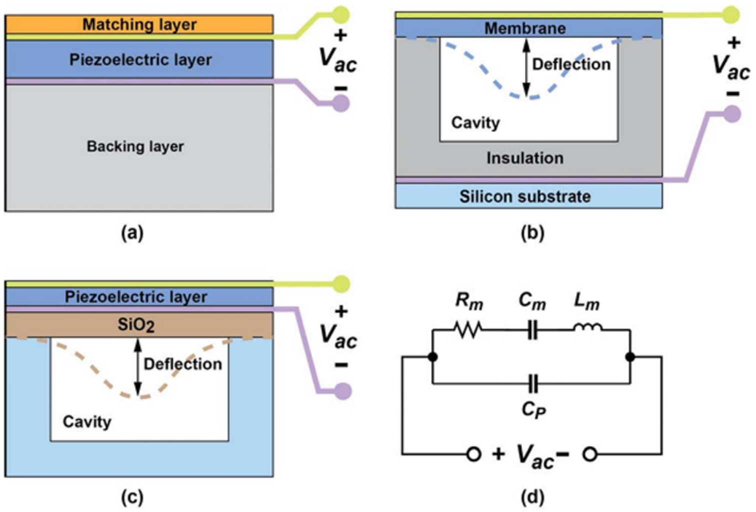

Figure 3.1: Figures from [CP21] show (a)a bulk piezoelectric transducer, (b) a Capacitive Micromachined ultrasound transducer (CMUT), (c) a piezoelectric Micromachined ultrasound transducer (PMUT), and (d) the Butterworth-Van Dyke equivalent circuit model

in both thickness and lateral dimensions. Since the lateral dimension is much larger, the piezo disk's diameter changes more significantly, causing the entire structure to bend. In Figure 3.1 (c), we can see the structure of PMUT which is similar to that of CMUT.
Figure 3.2 is a table made with the data of state-of-art devices in recent years gathered in [CP21]. It can be seen that CMUT has the widest range in frequency and bandwidth, which is suitable for high-resolution imaging. PMUT requires a lower voltage supply compared to CMUT and PZT, which makes it possible to work on portable devices. PZT usually works with conventional and stationary equipment, which can meet its high power consumption and benefit from its simple architecture.

Besides the methodology of the transducers, the electrical equivalent circuits of them are also important for further circuit design. Figure 3.1 (d) shows the Butterworth-Van Dyke equivalent circuit model that simulates the structure of a PZT transducer with electrical circuit.

In [BVD model], a detail BVD equivalent circuit model is shown as in Figure 3.3. The BVD model translates physical interactions into an equivalent electrical circuit using

Parameter	CMUT	PZT	PMUT
Operating Frequency	1 MHz - 20 MHz	1 MHz - 15 MHz	1 MHz - 10 MHz
Bandwidth	50% - 100% relative bandwidth	30% - 70% relative bandwidth	40% - 80% relative bandwidth
Max pulse Voltage	50 V - 200 V	5 V - 150 V	5 V - 100 V
Power	0.5 mW - 10 mW	1 mW - 100 mW	0.5 mW - 50 mW
Applications	IVUS, ICE, high-resolution imaging	General medical imaging	Wearable devices, high-frequency imaging

Figure 3.2: A table shows the specification of three kinds of transducers. These data come from [CP21]

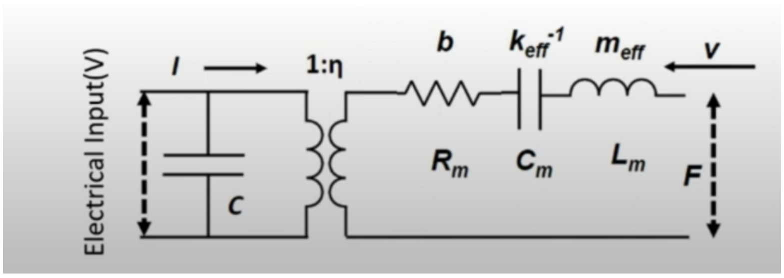

Figure 3.3: Detail BVD equivalent circuit model from [BVD model]

two formulas that connects the mechanical and electrical domains. One is a formula representing the force on the transducer. The other shows the voltage applied on it.

$$F(S) = m_{eff}\ddot{x}(s) + b\dot{x}(s) + k_{eff}x(s) \tag{3.1}$$

$$V(S) = L_m\ddot{q}(s) + R_m\dot{x}(s) + q(s)/C_m \tag{3.2}$$

These two formulas are connected with a piezoelectric transduction coefficient η, reflecting the coupling between mechanical and electrical domains in a piezoelectric material.

$$\eta = \frac{F}{V} = \frac{q}{x} \tag{3.3}$$

$$m_{eff} = \frac{L_m}{\eta^2} \tag{3.4}$$

$$b = \frac{R_m}{\eta^2} \tag{3.5}$$

$$k_{eff}^{-}1 = \frac{C_m}{\eta^2} \tag{3.6}$$

m_{eff} is effective mass, b is damping coefficient and k_{eff} is effective stiffness. R_m (Motional Resistance) represents the mechanical losses in the resonator. C_m (Motional Capacitance) corresponds to the elasticity of the resonator. L_m (Motional Inductance) represents the inertia of the resonator mass. These parameters can be transferred with coefficient η with x as displacement and q as charge in the circuit. These three components consist of the motional part in the circuit. C (Shunt Capacitance) represents the capacitance due to the electrodes and the dielectric of the resonator. Such a model is essential for the simulation of transducers in the circuit.

Another model indicates the structure of CMUT. Figure 3.4 shows a similar structure but with different parameters. Z_w (Water Load Impedance) represents the mechanical impedance of the medium (here is water) into which the CMUT is radiating ultrasonic waves. Z_{mem} (Membrane Impedance) shows the mechanical impedance of the CMUT membrane, which includes the mass, damping, and stiffness of the membrane. Z_{loss} (Loss Impedance) contains various mechanical losses in the system, including damping due to the medium and internal losses within the membrane material. CMUT differs from PMUT in that it has one more capacitance in the circuit. C_0 (Initial Capacitance) is the initial capacitance of the CMUT when there is no DC bias applied. C_p (Parasitic Capacitance) is the parasitic capacitance present due to the physical layout and connections in the system. The last component Z_{amp} (Amplifier Impedance) is the impedance of the amplifier used to process the electrical signal. This model is introduced in [Jin+01], which makes a perfect electrical equivalent model for CMUT.

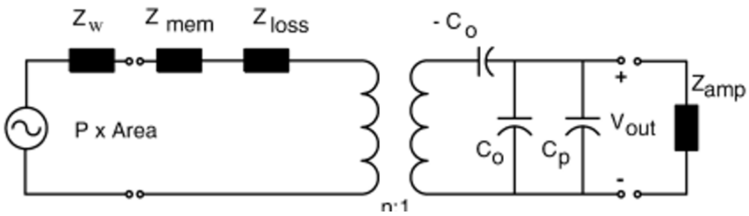

Figure 3.4: CMUT electrical equivalent model from [Jin+01]

3.2 Circuit design

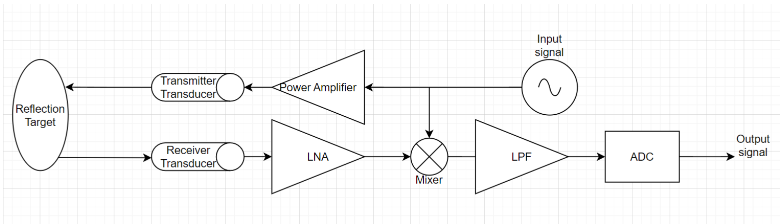

Figure 3.5: Circuit design for the ultrasound system

In this section, the whole ultrasound system is introduced. The main idea of this design comes from [LMJ23]. In Figure 3.5, all parts are shown. The upper part is the transmitter routine and the other part is the receiver path. In this figure, power amplifier is PA and LNA is the low noise amplifier. LPF after the mixer is low-pass filter and ADC is analog-to-digital converter. In this circuit, the FMCW signal as input is first sent to PA as well as the mixer. The enhanced signal is then transmitted through the transducer to the target. The reflected signal is received by the other transducer and sent to LNA. The mixer then combines the reflected signal and the original signal as reference signal. The difference signal is transferred through LPF and ADC and becomes the output we need for further signal processing. The function of these parts is introduced in the following sections.

3.2.1 Transmitter

In transmission part, there are input signal, transmitter transducer and PA. At first, there is FMCW signal designed as input signal. It performs also as reference signal which can be used in the mixer. However, the peak power doesn't satisfy the need of transducer. So, the PA is essential. PA can enhance the input signal to a level that meet the needs of transducers. After the amplify of the signal, it turns into audio signal by the transducer and is sent to the target.

3.2.2 Receiver

From the target, the reflection signal can be received with the receiver transducer. The echo signal has lower peak power with delay and noise. LNA in this system is the most important part for amplifying the weak echo signal with minimum noise. Then the mixer is used to generate the difference signal with a much lower frequency, which leads the filter after the mixer to be a low-pass filter. It allows the signal below specific frequency to pass. After all these components, the signal then converts from analog to digital signal which is the final output that needs further signal processing. To get the blood pressure, the difference signal should be separated into f_D and f_R as explained in previous sections to get speed and distance information.

4 Simulation and Results

In this part, the circuit of the whole system simulated in LTspice is introduced. How blood pressure plots can be generated from the transmitted and received signal is simulated in Python. Plots are made to show the results of signal processing.

4.1 Simulation

The simulation of the whole circuit in LTspice is shown in Figure 4.1. All parts are introduced in the previous section. In this circuit, there are PA, LNA, mixer, and LPF, the functions of which are already explained. The voltage source works as a signal generator. After PA, a BVD model is used as a transducer in the simulation. Between two transducers, the delay and loss of the signal are realized through a transmission line model. The model in LTspice can simulate a delay in the transmission process. The loss of the signal is controlled by the number of resistances. The mixer, LM1496N, in this circuit works as a multiplier, which multiplies the original signal and the received signal to get the needed output for further signal processing.

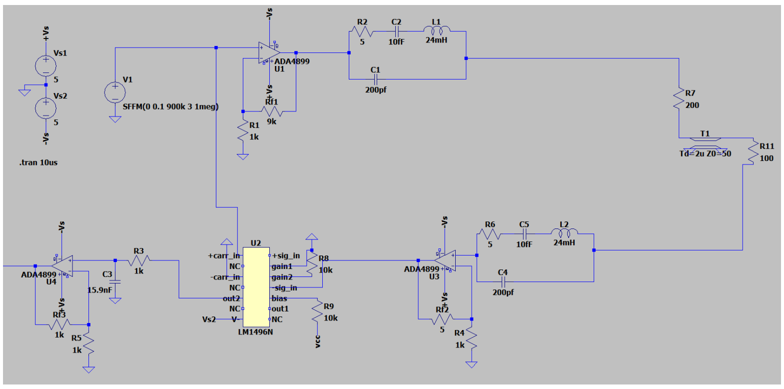

Figure 4.1: Circuit simulation in LTspice

4.2 Results

The signal processing part is simulated in Python and contains several plots. It aims to show the whole process of how mixed signals are generated from the transmitted signal and echo signal, how Doppler frequency f_D can be calculated, and what the diameter of the blood vessel and the velocity of the blood plots look like. Finally, the blood pressure is calculated with flow rate Q and the cross-sectional area A and made into plots.

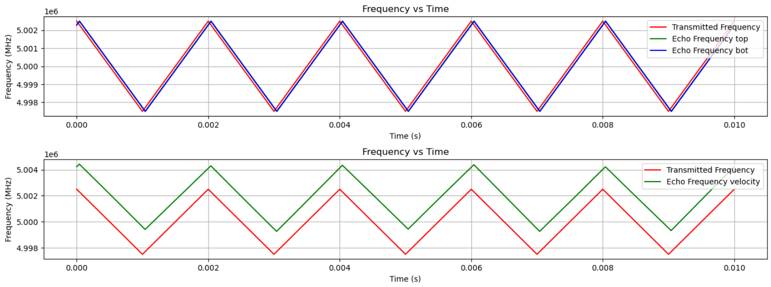

Figure 4.2: Changing frequency of transmitted and received signals

In Figure 4.2, there are two plots. The first plot on the top contains three different modulation frequencies. They are from the first transducer, which is designed for measuring blood vessel diameter. The red line belongs to the transmitted signal. The green and blue ones refer to the frequency of the echo signals from the top wall and bottom wall of the blood vessel. The translations between the red line and green and the red line and blue line show the information on time delay, which is the time the signal transmits from the transducer to the top and bottom wall of the blood vessel. The time delay can be calculated by the diameter of the blood vessel together with the speed of ultrasound in the human body. The second plot at the bottom shows the modulation frequencies from the second transducer, which aims to get Doppler frequency for velocity measurement. In this plot, the red line refers to the frequency of the transmitted signal, and the green one contains translation in both time and frequency direction, which is caused by time delay and Doppler frequency. These two plots can help us get the parameters we need.

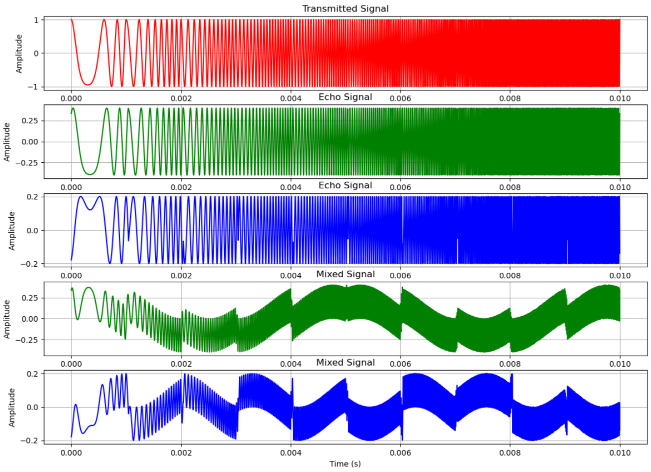

Figure 4.3: Signals from the first transducer for range measurement

Figure 4.3 shows the signals from the first transducer. There are two echo signals and one transmitted signal. The main difference between these three signals is caused not only by time delay but also by phase shift. The loss of signal also reflects on the lower amplitude of the echo signal. The mixed signals in the last two plots are generated from the multiplying of transmitted and echo signals, which can be used for further data processing in real-world experiments.

Similarly, Figure 4.4 shows the signals from the second transducer. The echo signal contains Doppler frequency and time delay, which makes it different from the transmitted signal. The amplitude of the echo signal also decreases because of the signal attenuation. The mixed signal can be used to extract Doppler frequency, which can help calculate the blood velocity.

Figure 4.5 shows the results calculating with the measured parameters: diameter and velocity. There are totally three plots showing the diameter and cross-sectional area of the blood vessel and the flow rate of the blood. All plots are made with variation functions, which make the plots of the parameters float among their mean values. In these functions, we can see that blood vessel diameter has a range among $6.26mm$, and the area of the blood vessel floats among $30.8mm$, which is close to the value from [Seo+15]. The flow rate of the blood is calculated with the velocity of the blood and the area. It has a range among $15cm^3/s$. With the flow rate and area, the next figure is made. Figure 4.6 is a plot showing the relation of flow and area, which is similar to

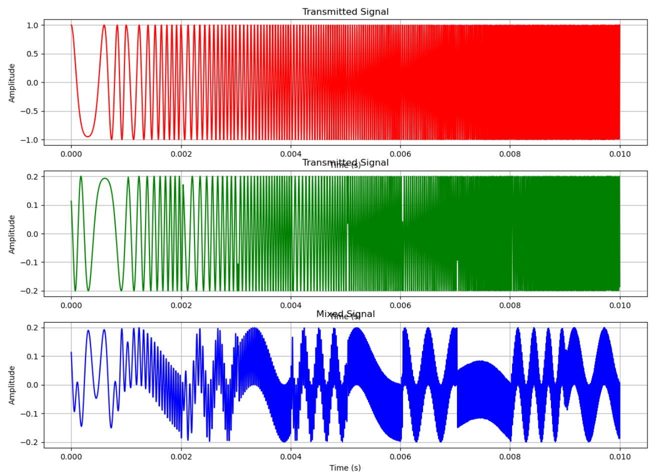

Figure 4.4: Signals from the second transducer for velocity measurement

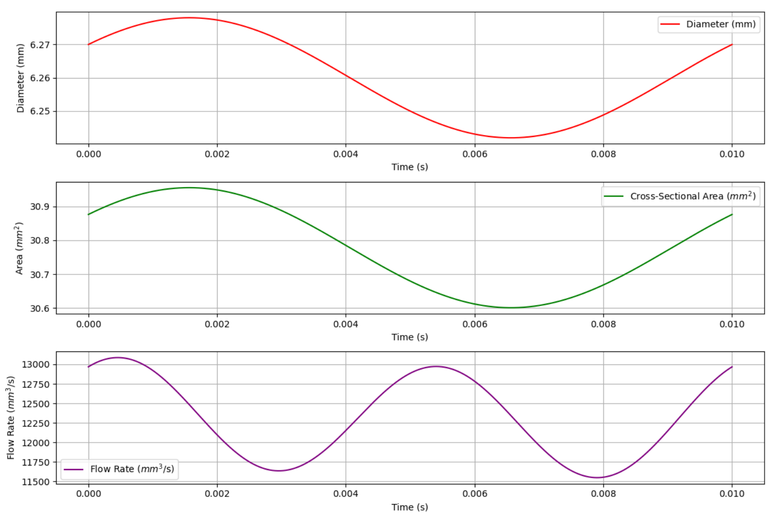

Figure 4.5: Plots of blood vessel diameter(mm), area(mm^3) and flow rate(mm^3/s)

Figure 2.4. The PWV is calculated as introduced in [Seo+15] through this plot. Figure

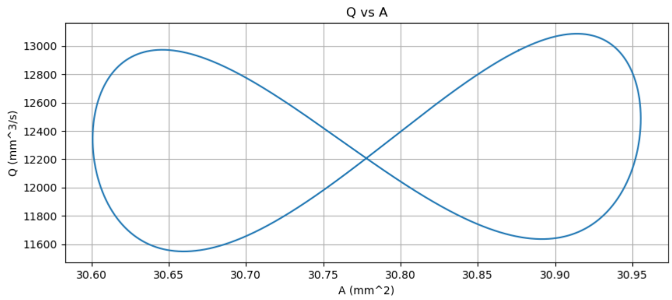

Figure 4.6: The relation of flow and area

4.7 is the plot showing PWV in this simulation. It is chosen and calculated from the linear region of Figure 4.6, which ranges from 7.7m/s to 7.9m/s. In reality, the PWV is between 5m/s to 10m/s. With all these plots above, the blood pressure can finally be measured.

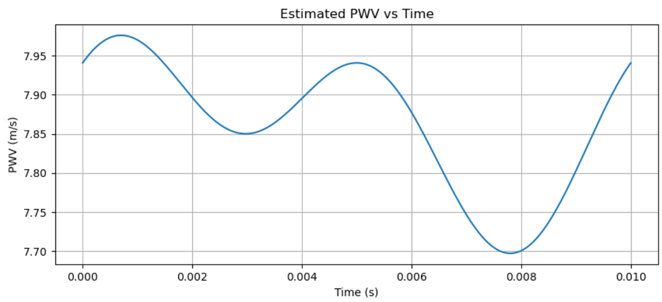

Figure 4.7: Estimated PWV in m/s

Figure 4.8 is the final result of the blood pressure, which is calculated with the formula from [Seo+15]. The blood pressure is similar to the Gauss pulse, so the variation function of the blood pressure is chosen to be Gauss pause, which aims to make the plots fit more with reality. It has a high pressure of 160 and a low pressure of about 60.

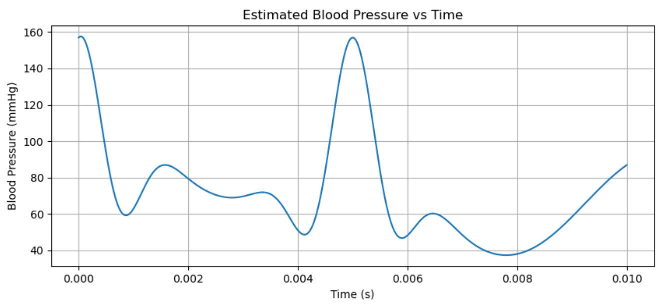

Figure 4.8: Estimated blood pressure

5 Conclusion

In this research, we introduce the method to calculate the blood pressure with PWV and the cross-sectional area of the blood vessel. A new method to estimate PWV helps us to realize the blood pressure measurement with only two ultrasound transducers. To design a more energy-efficient and portable device, we introduce the FMCW signal as our transmitted signal to measure the blood velocity and the diameter of the blood vessel. It is better than a pulse signal, which can make the device more secure and cost less energy, as the peak power of FMCW is lower. With all these reasonable methodologies, we have designed a simple circuit for the device, which is simulated in LTspice. It contains all the necessary parts that can help reduce the noise and amplify the transmitted and received signals. The data processing of such a system with the FMCW signal is also simulated in Python. The results, such as PWV and blood pressure, are made into plots, and the values are similar to those in reality.

6 Outlook

In the future, there are multiple steps that can be extended. The real circuit can be designed based on the simulation and the methodology. The data processing in the real world is a little bit different, as shown in the simulation, as the frequency of the signal can not be extracted as clearly as in the plots. The mixed signal should be dealt with FFT(Fast Fourier Transformation), and a spectrum plot should be made to get the parameters we need. More effort is still needed to optimize and put the research into practice.

The code for the simulation can be found in https://github.com/wyc991110/Explore-ultrasound-blood-pressure-measurement-algorithm

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