Ultrasonic Doppler Blood Flow Sensor

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Abstract—With the advent of point-of-care and continuous monitoring diagnostics, noninvasiveness is a key component towards ensuring timely and health-conscious results. In the field of cardiovascular diagnostics, ultrasound medical devices have been proven as effective tools for noninvasive blood flow monitoring. In this study, an ultrasonic Doppler blood flow sensor capable of noninvasively measuring blood flow velocity dynamics of the common carotid artery (CCA) is developed and presented. By exploiting the Doppler shift phenomenon, the device characterizes the blood flow dynamics within the CCA and provides various biomarkers including blood flow velocity, volumetric flow rate, and systolic and diastolic blood pressure. To achieve these results a nonlinear frequency mixer determines the difference between the frequencies from the ultrasonic emitter and receiver to then be converted into a digital signal for further processing. The frequencies range between 2MHz and 20MHz to allow the highest resolution depending on the depth of the CCA. Sufficient sensitivity is achieved by placing the emitter and receiver 90 and 45 degrees, respectfully, from the flow of blood. These diagnostic devices can ultimately provide clinicians a clearer picture in diagnosing and treating cardiovascular diseases ranging from heart attacks to stroke.

Keywords—doppler shift, common carotid artery, nonlinear frequency mixer, ultrasound, stroke, sphygmomanometer

I. INTRODUCTION

Determining blood velocity noninvasively allows medical professionals to quickly and easily detect cardiovascular issues indicative of diseases such as stroke, heart attack, or other pathophysiology. Previous research has demonstrated a relationship between blood flow velocity in the common carotid artery (CCA) and the degree of carotid artery stenosis (the narrowing of the blood vessel), which has an impact on stroke risk[1]. The common carotid artery (CCA) is particularly useful for determining blood velocity since it directs a high volume of blood flow toward the brain and is in close proximity to the skin for easy non-invasive monitoring. In order to determine blood flow velocity, a Doppler shift can be measured using an ultrasound signal emitter and receiver where the frequency shift can then be processed and the corresponding blood velocity can be computed. Doppler Flowmeters are preferable over other common methods of blood velocity measurements such as rapid bolus injection and electromagnetic flowmeters due to its noninvasive nature, ease of use, and relative accuracy.

To process the frequencies that are coming from the emitter and receiver, they must first pass through a frequency mixer. The frequency mixer acts as a frequency multiplier, outputting a superposition of the sum and differences of the two signal inputs. The signal then passes through a low pass filter to attenuate unwanted frequencies above 10kHz, essentially functioning as an envelope detector. Additionally, a resistor of high impedance was chosen in the low pass filter to ensure that only a small current may flow near or through the skin attached circuit. The signal is then amplified since the voltages vary on the order of micro- to millivolts. Finally, an analog-to-digital converter (ADC) processes the signal and outputs an equivalent digital signal for computing the necessary variables associated with biomarkers of interest. The block diagram of the device’s functional workflow is displayed in Fig. 1.

Fig. 1: Block diagram for the design of this ultrasonic Doppler blood flow sensor.

II. BACKGROUND

Blood flow has a close correlation with a few types of strokes including Transient Ischemic Attacks (TIA) and Ischemic Strokes. TIAs are a type of stroke that occurs due to an interruption in the blood flow in the brain that can be resolved within a short amount of time. TIAs can be resolved quickly but are typically indicative of future strokes to come. These can be a warning of further Ischemic Strokes which are caused by a narrowing of the arteries due to a buildup of fatty materials that could form a clot and block blood flow to the brain and cause a stroke. Strokes are a leading cause of disability, and have also been associated with a 5-year mortality rate ranging from 50-60%[2].
As the pressure in the arteries increases without a significant increase in cardiac function, the blood flow will decrease. This decrease can be seen as one of the warning signs of increased stroke risk, and measuring this change in blood velocity easily can help identify and treat strokes before they occur.

![Diagram of blood flow](image)

**Fig. 2:** a) Location of the CCA and its bifurcation and b) location of the aortic arch [3].

Measuring blood flow velocity in the carotid artery has the advantage of proximity to the heart as opposed to alternative locations, allowing for approximations of cardiac output. The carotid artery branches directly off of the aorta leading to the common carotid artery (CCA) as seen in Fig. 2. Utilizing this secondary location may be beneficial for estimating blood pressure and flow leading directly into the brain and its relationship to strokes. If the diameter of the blood vessel is known, the blood pressure can be estimated from the flow rate using the Poiseuille’s Law for Laminar Flow, given by (1). However, this will only give a rough estimate of blood pressure due to the assumptions of a rigid vessel and newtonian fluid mechanics. In reality, the flow rate through a blood vessel is a function of the pressure gradient as well as the geometry and stiffness of the blood vessel[4]. This relationship can be utilized to estimate the stiffness of the blood given measurements of blood velocity, pressure, and vessel geometry. Arterial stiffness is another important risk factor for cardiovascular disease, and increasing stiffness occurs naturally with age, as well as conditions such as atherosclerosis.

\[
\Delta p = 8\mu LQ / \pi R^4 \tag{1}
\]

A sphygmomanometer placed on the upper arm has been the standard as a preliminary blood pressure measurement in a patient due to its ability to measure both diastolic and systolic pressure noninvasively. Abnormal blood pressure measured from a sphygmomanometer placed on the upper arm can be an indicator for patients to receive further examinations as to what may be causing the abnormal blood pressure. Abnormal blood flow patterns detected in the CCA that may not have been detected by a sphygmomanometer may provide additional markers for the onset of a stroke.

### III. METHODS

The continuous wave Doppler flowmeter uses two ultrasonic transducers, an emitter that converts an AC signal into an ultrasound wave and a receiver that converts vice versa. These transducers typically operate around piezoelectric crystals[5] that can convert applied electric energy to mechanical as well as vice versa. For the purpose of simplification, the impedance of the piezoelectric crystal as well as the diffraction patterns caused by its geometry will be ignored.

![Doppler diagram](image)

**Fig. 3:** Doppler flowmeter arrangement showing the emitter and receiver positions, the significant angles, and significant physiological measurements of the neck and artery. This schematic will be used in the calculation of Doppler shift.

The design of the emitter and receiver depends on several variables for our calculations in determining the phase of the Doppler shift. The emitter and receiver will both be placed on the surface of the neck above the CCA, a certain distance from the blood flowing in the vessel. The angle between the blood flow and the emitter and receiver respectively are represented by \( \theta_e \) and \( \theta_r \) respectively. A schematic of the transmitter orientations, the significant angles involved, and the significant distances between transmitters, from the skin to the CCA, and from the skin to the flowing blood can be seen in Fig. 3. Table 1 also includes common physiological values for the CCA that will prove useful for understanding the physiological conditions of our design and the calculations that will be performed.

To calculate the Doppler phase shift we use the frequency of the receiver and the emitter. We can then use this shift to calculate the blood flow velocity using the speed of sound in body tissue, the angle between the emitter and blood flow, the angle between the receiver and blood flow, and the frequency of the emitter.
Thus, we use equation (2) using the variables we have defined in Fig. 3 to calculate the doppler phase shift and then the blood flow velocity.

\[ \Delta f = f_k - f_e = \frac{c}{r}(\cos \theta_k + \cos \theta_e)f_e \]  

(2)

The Doppler shifted frequency as a function of blood cell velocity is given by (2), where c corresponds to the speed of sound through body tissue. A frequency mixer will be used to generate an AC signal with the Doppler shifted frequency from the emitter and receiver signals. Mixers are nonlinear circuit components that typically consist of either diodes or field effect transistors acting as switches. Ideal mixers amplify signals from input frequencies and the difference of input frequencies [9]. Given two AC signals (3) and (4), the output of an ideal mixer shown in Fig. 4 is given by (5). The driving oscillator of the emitter will be fed into the LO input while the output from the receiver will be fed into the RF input as can be seen in Fig. 5. It is also significant to note that the sensitivity, \(\Delta f/V\), is dependent on the ultrasound propagation speed, the emitter and receiver angles, and the emitter frequency.

\[ V_{RF} = A(t) \cos(\omega_0 t + \phi(t)) \]  

(3)

\[ V_{LO} = A_{LO} \cos(\omega_{LO} t) \]  

(4)

\[ V_{IF} = V_{LO} \ast V_{RF} = \ldots \]  

(5)

\[ \frac{ACT A_{LO}}{2} \left[ \cos((\omega_{LO} + \omega_0)t + \phi(t)) + \cos((\omega_{LO} - \omega_0)t) \right] \]  

The resulting output of the multiplier will be fed into a low pass filter to generate a signal with the Doppler shifted frequency, and to eliminate potential noise. This signal will then be amplified, digitally sampled, and processed with a computer to determine this frequency, which is proportional to the blood velocity. All circuit components are assumed to be ideal.

**TABLE 1**

<table>
<thead>
<tr>
<th>Values for the CCA's average cross-sectional area[6], average carotid depth from the surface of the neck[7], and blood flow velocity for healthy and diseased states[8].</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average CCA cross-sectional area (mm²)</td>
</tr>
<tr>
<td>Average carotid depth from neck surface (mm)</td>
</tr>
<tr>
<td>Healthy end-diastolic velocity (cm/s)</td>
</tr>
<tr>
<td>Diseased end-diastolic velocity (cm/s)</td>
</tr>
</tbody>
</table>

Fig. 4: Ideal frequency mixer diagram. RF represents the shifted radio frequency signal, LO represents the fixed local oscillator, and IF represents the output intermediate frequency signal.

The ideal low-pass filter is designed such that any signal with a frequency above the set cutoff frequency defined by (6) is attenuated. This value is inversely proportional to the RC time constant associated with the circuit configuration. In the context of the ultrasound parameters and the expected Doppler shift range, the cutoff frequency is tuned such that only frequencies representing the Doppler shift passes through while filtering the ultrasound frequency and other outside noise.

\[ f_c = \frac{1}{2\pi RC} \]  

(6)

A non-inverting operational amplifier functions to amplify the received signal by a modulated gain coefficient dependent on the chosen passive component values (7). The data is then processed using an n-bit analog-to-digital converter (ADC) characterized by its range and resolution (8).

\[ Gain = 1 + \frac{R_1}{R_2} \]  

(7)

\[ Resolution = Range \cdot \frac{2^n}{2} \]  

(8)

Upon receiving the signal, the necessary computations can be conducted on a computer to convert the signal into its frequency domain using Laplace transform then further computed to reveal the patient’s blood flow velocity as given in (2) and other necessary biomarkers if provided the parameters such as blood pressure as given in (1).

**IV. RESULTS**

The complete circuit diagram is shown in Fig. 5. The parameters laid out in the previous sections influenced the chosen values for the circuit’s components based on the physiological and technical needs for the device’s function and of its end users.

The emitter is powered by a sinusoidal AC voltage source with a 3.3 V amplitude (6.6 V peak-to-peak) and a variable frequency ranging from 2 MHz to 20 MHz as per conventional ultrasound ranges[10]. Ultrasound propagation speed is assumed to be \(c = 1500 \text{ m/s} \). Variable frequency allows for...
adjustable penetration depth and resolution in which higher frequencies allow for higher resolution but lower imaging depth, and lower frequencies allow for higher imaging depth but lower resolution. As referenced from Figure 2 and (2), the placement of the emitter and receiver relative to the CCA and compared and computed to more reasonable blood flow velocities and optimized to be set at 10 kHz which assumed 20 MHz ultrasound and 100 cm/s blood flow velocity. The resistor value, $R$, was chosen to be 1 MΩ to prevent any

![Fig. 5: Ultrasound Doppler shift sensor circuit.](image)

The emitter frequency influences the Doppler shift output frequency characterized in Fig. 6. Considering practicality, the ultrasound emitter ought to lose as little power as possible, so placing it such that it is flush with the skin allows for most surface area contact and the least loss in energy as the ultrasound propagates throughout the tissue. Therefore an angle of 90° relative to the blood flow was chosen for $\theta_s$, assuming that the CCA flow is mostly parallel with the surface of the skin. As dictated by (2), having the smallest $\theta_s$ allows for the highest sensitivity for the system. It is impractical to have the receiver 0° relative to the blood flow since that necessitates the receiver placement on a part of the body far from the emitter, but having a large $\theta_s$ makes for lower sensitivity and may be placed too close to the emitter. Therefore, an angle of 45° was chosen for $\theta_s$ which corresponds to a distance $x = 29.5$ mm away from the emitter along the CCA assuming a penetration depth of $y = 29.5$ mm given in Table 1. The sensitivity of the system now becomes a function of the emitted frequency as shown in (9).

$$\frac{\Delta f}{v} = \frac{\cos(45°)}{1500 \text{ m/s}} f_f$$

The passive frequency mixer works hand-in-hand with the low-pass filter to output the Doppler shift signal then attenuate frequencies higher than its set cutoff frequency. The cut-off frequency was chosen based on the worst-case, highest Doppler shift frequency which was computed to be approximately 14 kHz assuming the source emitted 20 MHz ultrasound and the blood in the CCA flowed at 150 cm/s as demonstrated in Fig. 6. The cut-off frequency was then current greater than 1 mA flowing near or into the body since the voltage supply provides upwards of 3.3 V for the devices which outputs upwards of 3.3 uA near or into the body. The capacitor value, $C$, was then calculated using (6) to be approximately 16 pF. Though the low-pass filter allows essentially all meaningful frequencies to pass through, it may not filter out noise within the same frequency band as the Doppler shift. By accounting for a wider range of blood flow velocities, the design sacrifices precision to maintain accuracy. Of course, in future design iterations, the filter can be

![Fig. 6: Different Doppler frequency shifts for different blood flow velocities at the specified emitter and receiver positions. The red line represents an unrealistic absolute worst-case velocity in the CCA, the magenta line represents the highest recorded CCA blood flow velocity[11], the cyan line represents a normal healthy CCA velocity [8], the black line represents a diseased CCA velocity [8], and the blue line represents the worst-case bare minimum of no velocity.](image)
designated to be more precise or even variable such that the end user may adjust the cutoff frequency.

A variable non-inverting operational amplifier allows for the output waveform to populate more discrete least significant bits (LSB). The amplifier gain varies from 1 to 1001 given R1 = 100 Ω and R2 is a potentiometer ranging from 0 to 1 MΩ. Given that the received ultrasound amplitude varies on the order from micro- to millivolts, the adjustable gain allows the end user to set the output waveform as high as possible within the bounds of the amplifier’s linear region [-3.3 V, 3.3 V]. A 32-bit ADC bit was chosen which has a LSB of approximately 1.54 nV and provides satisfactory resolution in all scenarios. To restate, the digitized output waveform can undergo Laplace transformation to find the Doppler shift frequency then be used to compute blood flow velocity and other necessary biomarkers determined by the end user.

V. CONCLUSIONS

Noninvasive blood flow monitoring can be a powerful tool in the context of its use; however, the design’s simplicity presents notable limitations. Firstly, the design’s components were considered to be ideal in nature. The emitter’s ultrasound signal was assumed to propagate in the intended direction while leaking negligible amounts of noise. Additionally, external noise sources were assumed to be filtered out when passing through the low-pass filter. Notably, ultrasound is known to heat the affected surrounding tissue, but this effect can be considered negligible. Additionally, resistors are assumed to be Ohmic in nature independent of the heat produced from current flowing through it since this heat’s effect is also considered to be negligible. All passive components and active components perform ideally i.e. the input impedance is infinity and the output impedance is zero for the ideal operational amplifier. The voltage supply is assumed to provide enough sufficient power to generate its whole frequency range.

There are several practical limitations to the design’s computational assumptions. The model does not take into account the diffraction pattern of the ultrasonic waves as well as possible interference between the emitted and received waves. Furthermore, the AC signal from the receiver will not be a single frequency in practical applications due to the velocity profile within the vessel, geometric properties of the waves, and possible turbulence, adding intrinsic noise to our signal. Furthermore, some of the carrier signal generated by the emitter will be leaked to the receiver due to electric field coupling. Due to these complicating factors, practical Doppler flowmeter signal processing includes more complex RF amplifiers and detectors. Typically, bands of frequencies are passed through the emitter at the same times, returning a band of frequencies to the receiver so that more information can be obtained. The design also rectifies the frequency difference so that the direction of the velocity vector cannot be determined[5].

Though the model uses many assumptions, the design serves as a solid basis for designing a more complicated Doppler flowmeter that can gauge not only flow direction, but the velocity profile as well, and can even measure blood flow at different depths from the skin.

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REFERENCES

% for BENG 186B project

v_unreal_max = 150/100; % m/s --- absolute unrealistic max velocity
v_unreal_min = 0; % m/s --- absolute unrealistic min velocity
v_healthy = 20/100; % m/s --- average (max) end-diastolic velocity for healthy subjects
v_diseased = 15/100; % cm/s --- average (min) end-diastolic velocity for diseased/at risk subjects

c = 1500; % m/s
f_s = (2:20)*1000000; % Hz
theta_s = 90; % degrees
theta_r = 45; % degrees

delta_f_unreal_max = ((v_unreal_max/c) * (cosd(theta_s) + cosd(theta_r))*f_s)'
delta_f_unreal_min = ((v_unreal_min/c) * (cosd(theta_s) + cosd(theta_r))*f_s)'
delta_f_healthy = ((v_healthy/c) * (cosd(theta_s) + cosd(theta_r))*f_s)'
delta_f_diseased = ((v_diseased/c) * (cosd(theta_s) + cosd(theta_r))*f_s)'

figure
plot(f_s, delta_f_unreal_max, 'red')
xlabel('f_s (Hz)')
ylabel('
Deltaf (Hz)')
hold on
plot(f_s, delta_f_unreal_min, 'blue')
plot(f_s, delta_f_healthy, 'cyan')
plot(f_s, delta_f_diseased, 'black')

title('
Deltaf vs. f_s for Different Blood Flow Velocities (when theta_S = 90\circ & \theta_R = 45\circ)')
legend('100 cm/s', '0 cm/s', '20 cm/s', '15 cm/s')
set(gca, 'fontsize', 15)