A REAL-TIME TECHNIQUE FOR THE CORRECTION OF INVASIVE BLOOD PRESSURE MEASUREMENTS USING COUNTER PRESSURE

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A REAL-TIME TECHNIQUE FOR THE CORRECTION OF INVASIVE BLOOD PRESSURE MEASUREMENTS USING COUNTER PRESSURE

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Thesis

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ABSTRACT

In critical care, a fluid-filled catheter transducer system is commonly used to continuously monitor blood pressure. The fluid-filled catheter transducer system, as a first approximation, behaves as a second order dynamic system. The dynamic characteristics of the system are affected by variations in assembly technique and time dependent changes in the system, often resulting in distortion and inaccurate measurements. A previous simulation study employed a counter pressure source in tandem with the transducer. The counter pressure generated minimized the fluid flow in the pressure monitoring system. When the flow in the system was zero the counter pressure generated closely approximated the true blood pressure. The current study developed a real-time technique to generate accurate and dependable counter pressure. To validate this technique, one experimental model (second order dynamic system) and two simulation models (second order dynamic system and fourth order dynamic system) of the catheter transducer system were developed and tested under varying system conditions. The real-time technique successfully reproduced the true blood pressure waveforms, regardless of variations in the system characteristics and changes in the system over time.

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CHAPTER I

INTRODUCTION

The technique of invasive blood pressure measurement has become a routine clinical measurement. In critical care, a fluid-filled catheter transducer system is commonly used to continuously monitor blood pressure.

The fluid-filled catheter transducer system consists of four main units [1]-[3]:

1. Catheter - The catheter provides access to the arterial system being monitored and detects the pressure waves generated in the arterial system by cardiac contractions.
2. Fluid-filled tubing - The catheter is connected to the pressure transducer by a fluid-filled tubing. The fluid column in the tubing carries the mechanical signal created by the pressure wave to the diaphragm of the pressure transducer.
3. Transducer - The transducer converts the mechanical signal into an electrical signal.
4. Signal processing unit - The electrical signal generated by the transducer is amplified and filtered in the signal processing unit. Output of the signal processing unit is displayed as an analog waveform at the bedside monitor.

The catheter transducer system used in critical care settings, as a first approximation, behaves as a second order underdamped system. It can be expressed mathematically by a second order differential equation (eq. 1) with characteristics determined by the compliance, inertance, and resistance of the system [4], [6], [11].

$$IC\frac{d^2P_2}{d^2t}+RC\frac{dP_2}{dt}+P_2=P_1 \tag{1}$$

- The inertance, *I*, reflects the fluid mass in the system.
- The compliance, *C*, is contributed by the flexibility of the pressure transducer diaphragm and the compliance of the pressure tubing.
- The resistance, *R*, refers to the fluid viscosity as it moves with each pulsatile change in the pressure tubing.
- *P1* is the output pressure signal measured by the transducer.
- *P2* is the driving pressure at the intravascular tip of the catheter.
- *t* is time.

These parameters define the natural frequency, F_n (eq. 2) in Hertz, and damping coefficient, ζ (eq. 3) [1]-[10] of the catheter-transducer system, which indicate the adequacy or fidelity of the system.

$$F_n=\frac{1}{2\pi\sqrt{IC}} \tag{2}$$

$$\zeta=\frac{R}{2}\sqrt{\frac{C}{I}} \tag{3}$$

An inadequate system, due to its low natural frequency, may result in waveform distortion and erroneous measurements. As an underdamped system, the catheter-transducer system tends to record falsely high systolic pressures and low diastolic pressures [3], [4]. Conversely, overdamped systems exist which result in erroneous recordings as well. With an overdamped system, the waveform loses its important features, such as dicrotic notch, and appears unnaturally smooth. Over damping results in falsely low systolic pressure and high diastolic pressure readings [3], [4]. Because variable dynamic behavior exists across catheter-transducer systems, it is often questionable if the arterial waveform is a result of an inadequate system dynamic response or is an accurate reflection of the true physiological status of the patient.

There are several factors that lead to poor dynamic responses [1]-[3], [6], [7]:

- Air bubbles in the tubing system. Air bubbles contribute to the compliance of the system, causing a distorted arterial waveform and erroneous pressure readings.
- Catheter clotting is another complication in arterial monitoring that often results in an overdamped waveform.
- Several factors that may alter the natural frequency of the catheter transducer system include:
    - Long, narrow, and compliant pressure tubing.
    - Overly compliant diaphragm in the pressure transducer.
    - Presence of additional stopcocks.
    - Loose connections.

With an inadequately functioning monitoring system, not only the actively measured hemodynamic indices but also any derived variables will be erroneous, potentially invalidating the entire hemodynamic profile of the patient. Consequently, wrong clinical decisions may be made resulting in an inappropriate treatment for the patient.

Invasive blood pressure monitoring is a costly procedure and its use can give rise to inadvertent risks to the patients. One of the most common risks is infection due to contamination of catheters, pressure transducer, stopcocks, and flush solutions [1], [2], [14]. Air embolism is another risk associated with direct blood pressure monitoring. Air embolism may reduce or obstruct blood flow and may cause neurologic complications. In critical cases, air embolism may cause death [14]. Since severe consequences are associated with the invasive blood pressure monitoring, its use can be justified only if accurate and reliable data are obtained [1].

1.1 Objective of the study

A previous simulation study [1] employed a counter pressure source in tandem with the catheter transducer system. The counter pressure generated minimized the fluid flow in the pressure monitoring system. When flow in the system was zero the counter pressure generated closely approximated the true blood pressure. The current study was conceptualized as an attempt to develop a system to generate accurate and dependable counter pressure in real-time. It aimed at minimizing the errors due to variations in assembly technique and time dependent changes in the system, allowing more accurate blood pressure measurements.

1.2 Research Hypothesis

The generated counter pressure is similar to the true blood pressure; the effectiveness of this statement was validated by calculating the mean square error (*MSE*) and the root mean square error (*RMSE*) between the generated counter pressure and the true blood pressure. These errors were used as measures to assess the accuracy of the generated counter pressure with respect to the true blood pressure.

As low *MSE* (or *RMSE*) does not necessarily mean clinical acceptance, a visual comparison was necessary to ensure the effectiveness of the models in reproducing the diagnostic information present in the true blood pressure waveform.

In addition to the error analysis and visual comparisons, following null hypotheses were tested (using a paired T test, $\alpha = 0.05$):

H_{01}: There is no difference between the output systolic pressure of the catheter transducer system with counter pressure source and the true systolic pressure.

H_{02}: There is no difference between the output diastolic pressure of the catheter transducer system with counter pressure source and the true diastolic pressure.

H_{03}: There is no difference between the output systolic pressure of the catheter transducer system without counter pressure source and the true systolic pressure.

H_{04}: There is no difference between the output diastolic pressure of the catheter transducer system without counter pressure source and the true diastolic pressure.

CHAPTER II

LITERATURE REVIEW

In critical patient management, such as intensive care and anesthesia it is very important to have a continuous and accurate recording of the patient's blood pressure. This is normally achieved by inserting a saline filled catheter into the radial artery and measuring the pressure in the artery by an electronic pressure transducer (fluid-filled catheter transducer system). The errors associated with this measurement may result from the inadequate dynamic response from the catheter-transducer system. Several investigators have attempted to address this issue. Their work is discussed in the following sections.

2.1 Fast Flush Test

The catheter transducer system approximates to a second order dynamic system. Its frequency response can be defined in terms of resonant frequency and damping ratio. The response of the system would depend on the setup, particularly presence of air bubbles in the system. In clinical settings a high pressure source in the form of heparinized saline is present. There is a high impedance valve between the saline bag and the catheter transducer system near the transducer. When this valve is opened the measurement system detects high pressure.

The valve snap shuts and provides a pressure step input to the system. The dynamic response of the system is calculated from the response to the pressure step [16].

The damped natural frequency is determined by measuring the period of a full oscillation and calculating the frequency from the period. The damping coefficient ζ is calculated from the amplitude ratio *A2/A1* which is obtained by measuring two successive peak amplitudes (figure 2.1) [2], [3], [5], [6], [7].

$$\zeta = \frac{-\ln\left(\frac{A_2}{A_1}\right)}{\sqrt{\pi^2 + \left(\ln\left(\frac{A_2}{A_1}\right)\right)^2}} \tag{4}$$

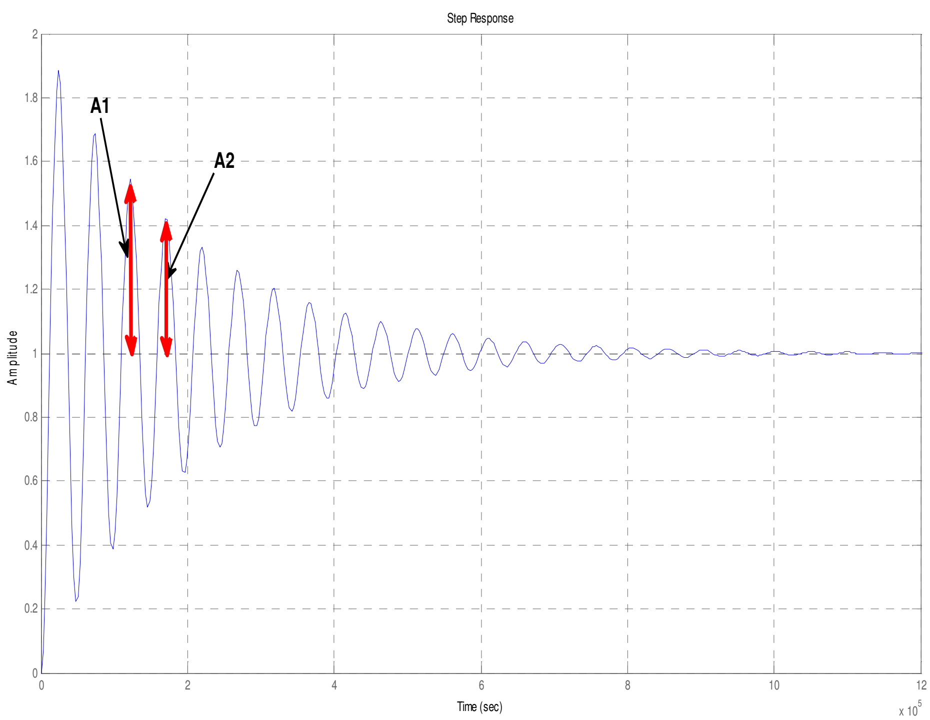

Figure 2.1: Amplitude ratio *A2/A1*

The natural frequency and the damping coefficient are mapped. If the dynamic response of the system lies in the optimal region, a faithful reproduction of the waveforms can be achieved. The fast flush test allows the verification of adequacy of the pressure monitoring system. It should be noted that the waveform distortions are only approximated from the fast flush test and documented in the clinical settings. The errors in the waveforms are not corrected, leading to the possibilities of inaccurate pressure readings [1].

2.2 Harmonic analysis for pressure correction

A Fourier based algorithm for pressure waveform correction was first used by Wellnhofer, et al. [9]. A set of correction coefficients for a catheter transducer system were obtained in vitro and they were used in correcting the pressure measurements from the same catheter transducer system.

The coefficients were obtained by simultaneously sampling and digitizing the signals from a catheter-tip transducer and a fluid filled catheter transducer system. A segmentation of both the signals was done according to the instantaneous heart cycle duration, which was determined by an autocorrelation based algorithm. The correction coefficients were determined by first taking a Fourier transform of both the segmented signals and then by a complex division of the Fourier coefficients of the reference and corresponding signals. The correction coefficients obtained for a system in vitro were valid till the system remained unchanged [9].

For clinical application, the amplitude and phase of the distorted signals were corrected by complex multiplication with the correction coefficients and taking inverse Fourier transform to get the corrected signal in time domain. Even though this method provided a way of amplitude and phase correction, it had high computational demands. A considerable lag was also introduced in real-time display. As the correction was dependent on correction coefficients for a particular system, any small change in the system such as, presence of air bubbles affected the waveform correction.

2.3 Using transfer function for pressure waveform correction

This method was developed from the second order linear differential equation that gave the relationship between the pressure obtained from the fluid-filled catheter system and the real pressure wave at the tip of the cannula [1], [8]. The transfer equation used to calculate the corrected pressure wave is given as follows,

$$\frac{d^2 P_c}{dt^2} + 2\omega_n \zeta \frac{dP_c}{dt} + \omega_n^2 P_c = CP_{pred}(t) \qquad (5)$$

Here, P_c is the pressure at the catheter-transducer system, P_{pred} is the corrected pressure, ω_n is the natural frequency, ζ is the damping ratio and t is the time [1]. Using the damped frequency ω_p the natural frequency is calculated as,

$$\omega_n = \frac{\omega_p}{(1-\zeta^2)^{1/2}} \qquad (6)$$

Comparison between the corrected waveform, P_{pred}, and the reference input pressure, P_{ref}, was performed. The corrected pressure waveform showed that the transfer function method removed the time delay in rising pressure and the overshoot in systolic pressure [1].

2.4 Counter pressure and Fourier based optimization technique

A method for approximating and reproducing the true blood pressure waveform using a numerical optimization technique was developed by Lim [1]. This method employed a counter pressure source in tandem with the transducer. The counter pressure waveform was generated by 16 Fourier coefficients, which were manipulated iteratively by the optimization algorithm to minimize the fluid flow in the pressure monitoring system. When the flow was zero, no energy was lost and the counter pressure waveform closely approximated the true blood pressure waveform. Even though this method provided a way of approximating the true blood pressure waveform, it had high computational demands. Due to the long computational delay this technique had been restricted to simulation studies [1] and had never been tested in real-time.

CHAPTER III

BLOOD PRESSURE SIGNAL CORRECTION USING COUNTER PRESSURE

The objective of the current study was to develop a real-time technique to generate counter pressure, which could be employed in tandem with the catheter transducer system to minimize the fluid flow. To develop the proposed real-time technique and to study what effect that had on the catheter transducer system, two widely accepted mathematical models of the catheter transducer system (second order dynamic system [1]-[13] and fourth order dynamic system [17], [18]) were modified to include the proposed counter pressure source.

3.1 Catheter transducer system as a second order dynamic system

Several literature sources have stated that the catheter-transducer system can be reasonably modeled by a second-order dynamic system [1]-[13]. The equivalent electrical circuit for such a system is given in Figure 3.1.

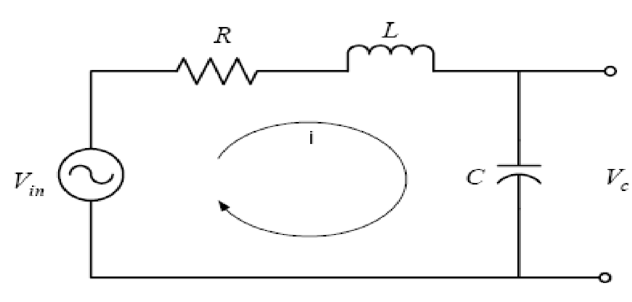

Figure 3.1: Electrical circuit (2^{nd} order system) of the catheter-transducer system

In the circuit (figure 3.1):

- The voltage source is analogous to the pressure source.
- The resistor is analogous to the fluid resistance.
- The inductor is analogous to the fluid inertance.
- The capacitor is analogous to the compliance [12]-[13].

Applying Kirchhoff's Voltage Law to the circuit shown in figure 3.1:

$$V_{in}(t) - Ri(t) - L\frac{di(t)}{dt} - V_c(t) = 0 \tag{7}$$

Suppose that the current in the circuit is zero, then

$$V_{in}(t) = V_c(t) \tag{8}$$

If the current is zero, the input voltage can be determined from the voltage across the capacitor (eq. 7-8). In the catheter-transducer system, this theory implies that the blood pressure measured by the transducer will equal to the true blood pressure if the fluid in the system is not in motion. In order to drive the current to zero, a counter voltage source, Vctr, is added in series with the capacitor as shown in Figure 3.2 [1].

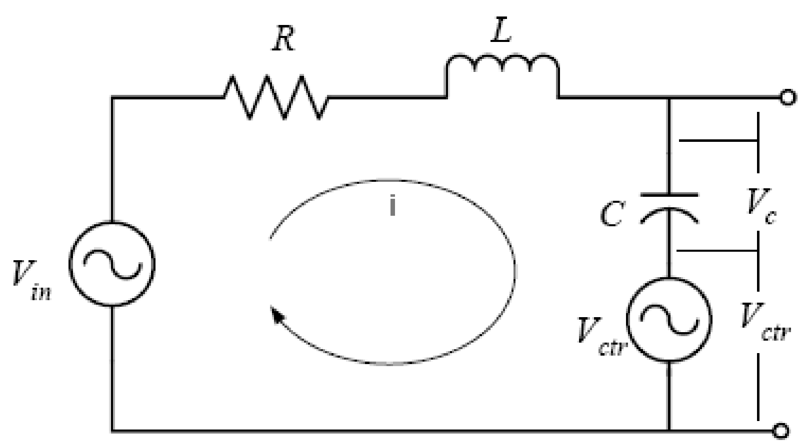

Figure 3.2: A counter voltage source added to the electrical circuit

Applying Kirchhoff's Voltage Law to the circuit shown in figure 3.2:

$$V_{in}(t) - Ri(t) - L\frac{di(t)}{dt} - \frac{1}{C}\int i(t)dt - V_{ctr}(t) = 0 \qquad (9)$$

If the counter voltage source (Vctr) is successful in driving the current to zero, then

$$V_{in}(t) = V_{ctr}(t) \qquad (10)$$

In a previous study [1], the appropriate counter voltage was generated using Fourier coefficients that were manipulated iteratively by an optimization algorithm. The biggest disadvantage of the Fourier based optimization technique was the delay in computing the required counter pressure. Due to this computational delay the Fourier based optimization technique was found unsuitable for real-time generation of the required counter pressure. As a result, a new technique involving a feedback loop was developed to generate the required counter pressure in real-time.

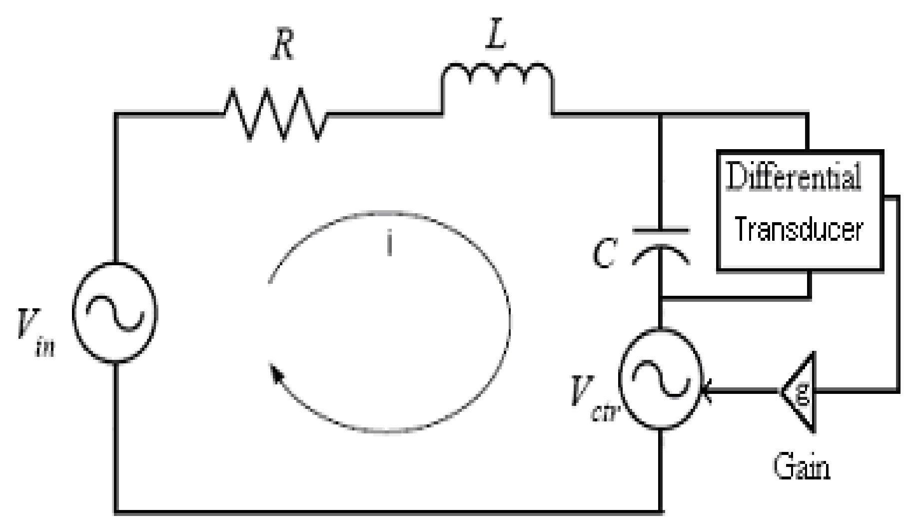

Figure 3.3: Feedback loop used to determine the required counter pressure

Transfer function of the system (in s domain) shown in figure 3.2

$$V_{out} = \frac{V_{in} + V_{ctr}(LCs^2 + RCs)}{LCs^2 + RCs + 1} \tag{11}$$

$$V_{out} = V_c + V_{ctr} \tag{12}$$

$$V_c + V_{ctr} = \frac{V_{in} + V_{ctr}(LCs^2 + RCs)}{LCs^2 + RCs + 1} \tag{13}$$

If a feedback loop (with gain g) is added to the system (figure 3.3) such that

$$V_{ctr} = g \times V_c \tag{14}$$

$$\frac{V_{ctr}}{V_{in}} = \frac{g}{LCs^2 + RCs + (g+1)} \tag{15}$$

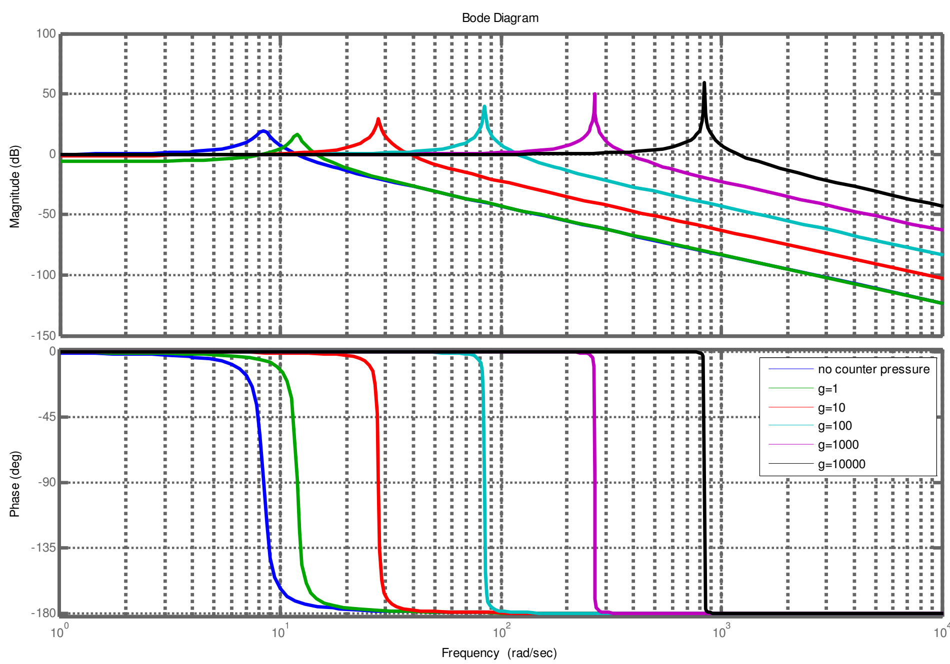

Figure 3.4: Frequency response of the system with and without counter pressure source. Parameters used (bubble location l = 0.9m and bubble size 25uL) [1]: R = 3.089×10^9 kg/m^{4}s, L = 3.523×10^8kg/m^4, C=4.017×10^{-13} m^3/Pa

Within a given frequency range, if the frequency response of the system indicates a gain of 0 dB and phase shift of 0 deg then output replicates the input ($V_{out} = V_{in}$). The frequency response of a system without counter pressure source (blue plot, figure 3.4) indicates that the gain and phase of the system start changing at lower frequencies, which result in distortion of the output. The frequency response of a system with counter pressure source (with gain g = 10 and above, figure 3.4) indicates that the gain and phase of the system remain unchanged (0dB and 0deg) for a large frequency range, which keeps on growing as the feedback loop gain (g) increases. This results in the output replicating the input. Thus selecting a feedback gain in accordance to the frequency range of interest (in this case 0 to 60Hz) results in counter voltage replicating the input voltage.

3.2 Catheter transducer system as a fourth order dynamic system

The catheter transducer system can also be reasonably modeled as a fourth order dynamic system [17], [18]. The equivalent electrical circuit for such a system is given in figure 3.5.

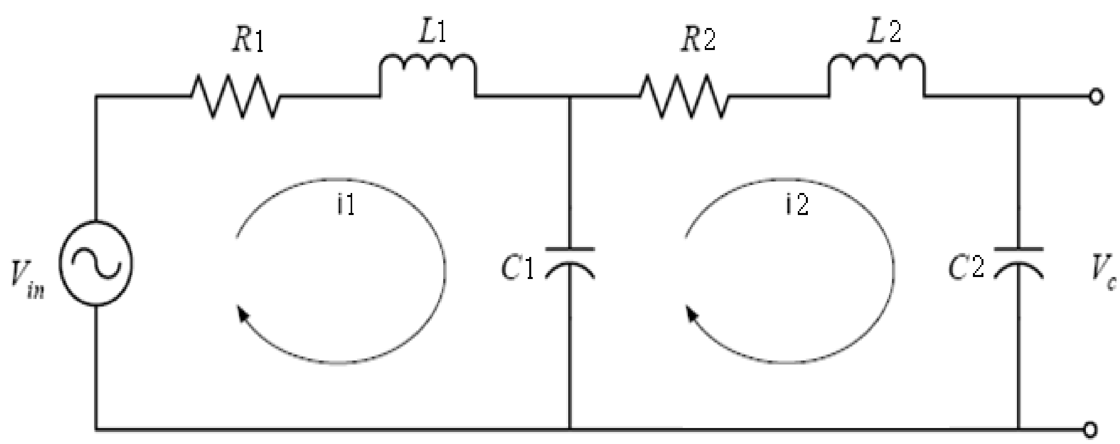

Figure 3.5: Catheter transducer system as a fourth order dynamic system

In the circuit (figure 3.5):

- The voltage source (*Vin*) is analogous to the pressure source.
- The resistors (*R1* and *R2*) are analogous to the fluid resistance.
- The inductors (*L1* and *L2*) are analogous to the fluid inertance.
- The capacitors (*C1* and *C2*) are analogous to the compliance.

Applying Kirchhoff's Voltage Law to the circuit shown in figure 3.5

$$V_{in} - R1i1 - L1\frac{di1}{dt} - \frac{1}{C1}\int i1dt = 0 \qquad (16)$$

$$Vc1 = \frac{1}{C1}\int i1dt \qquad (17)$$

$$Vc1 - R2i2 - L2\frac{di2}{dt} - \frac{1}{C2}\int i2dt = 0 \qquad (18)$$

$$Vc = \frac{1}{C2}\int i2dt \qquad (19)$$

A counter voltage source, Vctr, is added in series with the capacitor as shown in figure 3.6.

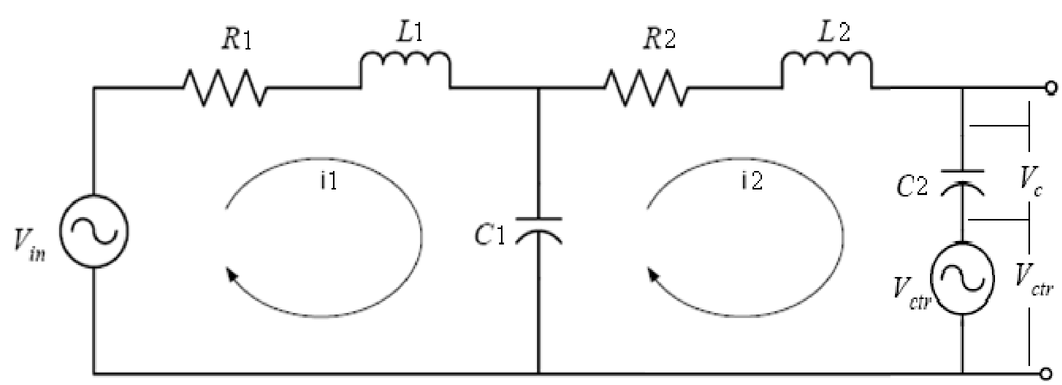

Figure 3.6: Catheter transducer system as fourth order dynamic system with counter pressure source

Applying Kirchhoff's Voltage Law to the circuit shown in figure 3.6

$$V_{in} - R1i1 - L1\frac{di1}{dt} - \frac{1}{C1}\int i1dt = 0 \tag{20}$$

$$Vc1 = \frac{1}{C1}\int i1dt \tag{21}$$

$$Vc1 - R2i2 - L2\frac{di2}{dt} - \frac{1}{C2}\int i2dt - V_{ctr} = 0 \tag{22}$$

$$Vc = \frac{1}{C2}\int i2dt \tag{23}$$

If the counter voltage source (*Vctr*) is successful in driving both the currents (*i1* and *i2*) to zero, then

$$V_{in}(t) = V_{ctr}(t) \tag{24}$$

To generate the required counter pressure, a feedback loop (with gain g) is added to the system (figure 3.7) such that

$$V_{ctr} = g \times V_c \tag{25}$$

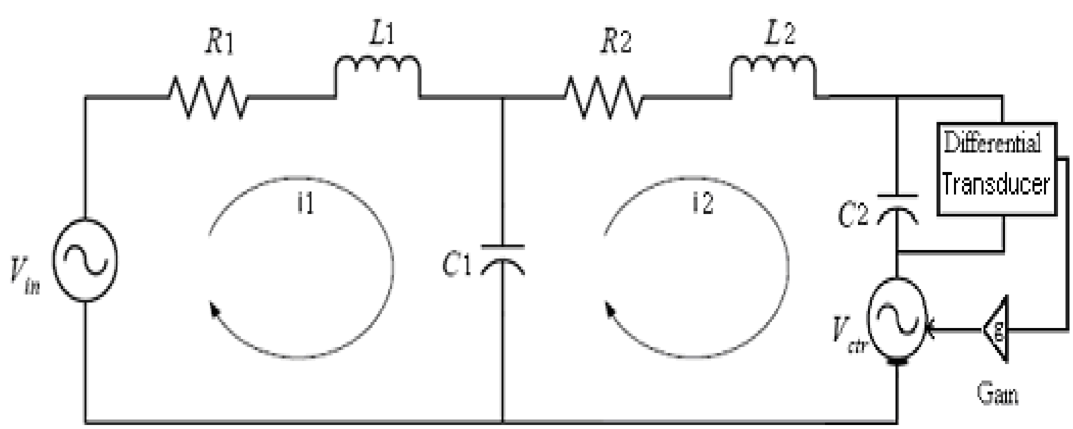

Figure 3.7 Feedback loop used to determine the required counter pressure

Transfer function of the system (in *s* domain) shown in figure 3.7 (assuming *R=R1=R2*, *C=C1=C2* and *L=L1=L2*)

$$\frac{V_{ctr}}{V_{in}} = \frac{(LCs^2 + RCs + 1) \times g}{L^2C^2s^4 + 2RLC^2s^3 + (R^2C^2 + 3LC + LCg)s^2 + (3RC + RCg)s + (1+g)} \tag{26}$$

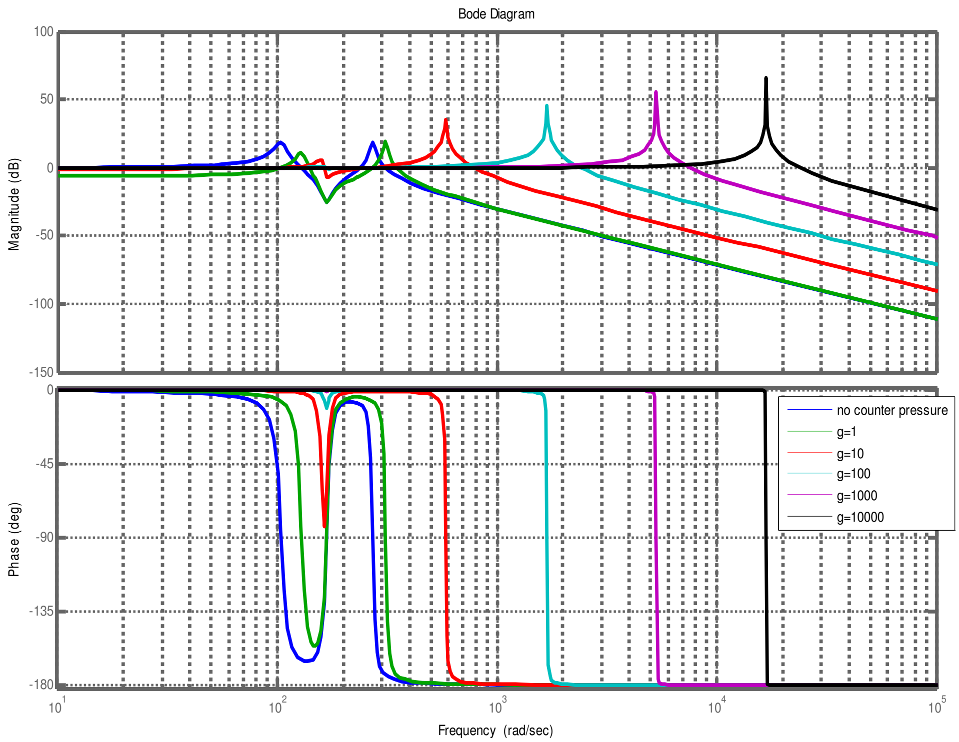

Figure 3.8: Frequency response of the system with and without counter pressure source. Parameters used (bubble location l = 0.9m and bubble size 25uL) [1]: R = 1.5445 x 10^9kg/m^{4}s, L = 1.7615 x 10^8kg/m^4, C=2.0085x10^{-13} m^3/Pa

Within a given frequency range, if the frequency response of the system indicates a gain of 0 dB and phase shift of 0 deg then output replicates the input (V_{out} = V_{in}). The frequency response of a system without counter pressure source (blue plot, figure 3.8) indicates that the gain and phase of the system start changing at lower frequencies, which result in distortion of the output. The frequency response of a system with counter pressure source (with gain g = 1000 and above, figure 3.8) indicates that the gain and phase of the system remain unchanged (0 dB and 0 deg) for a large frequency range, which keeps on growing as the feedback loop gain (g) increases. This results in the output replicating the input. Thus selecting a feedback gain in accordance to the frequency range of interest (in this case 0 to 60Hz) results in counter voltage replicating the input voltage.

CHAPTER IV

METHODOLOGY

As discussed in the previous chapter, there are two widely accepted mathematical models for the catheter transducer system:

1. Second order dynamic system.
2. Fourth order dynamic system (π model).

The effect of the feedback loop (counter pressure source) on these models was studied mathematically in sections 3.1 and 3.2. To test the proposed hypotheses, three experimental models were developed and tested individually:

Model A: Simulation model of the catheter transducer system with feedback loop as a second order dynamic system.

Model B: Experimental setup of the catheter transducer system with feedback loop as an electrical circuit (second order dynamic system).

Model C: Simulation model of the catheter transducer system with feedback loop as a fourth order dynamic system (π model).

4.1 Model A

A simulation model of the catheter transducer system (2^{nd} order dynamic system) with a counter pressure source was developed based on the electrical circuit in figure 3.3, which can be expressed by the equations:

$$V_{in}(t) - Ri(t) - L\frac{di(t)}{dt} - \frac{1}{C}\int i(t)dt - V_{ctr}(t) = 0 \tag{27}$$

$$V_c = \frac{1}{C}\int i(t)dt \tag{28}$$

$$V_{ctr} = g \times V_c \tag{29}$$

Model parameters such as Resistance *R*, Compliance *C* and Inertance *L* were selected from table 4.1:

Table 4.1: Model parameters *R* (kg/m^4s), *L* (kg/m^4) and *C* (m^3/Pa) for a catheter transducer system under different air bubble conditions (length and diameter of tubing used was 1.5m and 0.0018m) [1]. Used with permission (Appendix G).

Bubble Location	Bubble Size					
	5 μL	10 μL	15 μL	20 μL	25 μL	
l = 1.5 m	5.148×10^9 5.872×10^8 4.599×10^{-13}	5.148×10^9 5.872×10^8 4.885×10^{-13}	5.148×10^9 5.872×10^8 5.171×10^{-13}	5.148×10^9 5.872×10^8 5.457×10^{-13}	5.148×10^9 5.872×10^8 5.742×10^{-13}	R L C
l = 1.2 m	4.119×10^9 4.698×10^8 3.737×10^{-13}	4.119×10^9 4.698×10^8 4.023×10^{-13}	4.119×10^9 4.698×10^8 4.309×10^{-13}	4.119×10^9 4.698×10^8 4.595×10^{-13}	4.119×10^9 4.698×10^8 4.880×10^{-13}	R L C
l = 0.9 m	3.089×10^9 3.523×10^8 2.874×10^{-13}	3.089×10^9 3.523×10^8 3.160×10^{-13}	3.089×10^9 3.523×10^8 3.446×10^{-13}	3.089×10^9 3.523×10^8 3.732×10^{-13}	3.089×10^9 3.523×10^8 4.017×10^{-13}	R L C
l = 0.6 m	2.059×10^9 2.349×10^8 2.011×10^{-13}	2.059×10^9 2.349×10^8 2.297×10^{-13}	2.059×10^9 2.349×10^8 2.583×10^{-13}	2.059×10^9 2.349×10^8 2.869×10^{-13}	2.059×10^9 2.349×10^8 3.154×10^{-13}	R L C
l = 0.3 m	1.030×10^9 1.174×10^8 1.149×10^{-13}	1.030×10^9 1.174×10^8 1.435×10^{-13}	1.030×10^9 1.174×10^8 1.720×10^{-13}	1.030×10^9 1.174×10^8 2.007×10^{-13}	1.030×10^9 1.174×10^8 2.292×10^{-13}	R L C

Once the model parameters were defined, the model of a catheter-transducer system with a feedback loop was developed using the Simulink toolbox in MATLAB (Version R2007a).

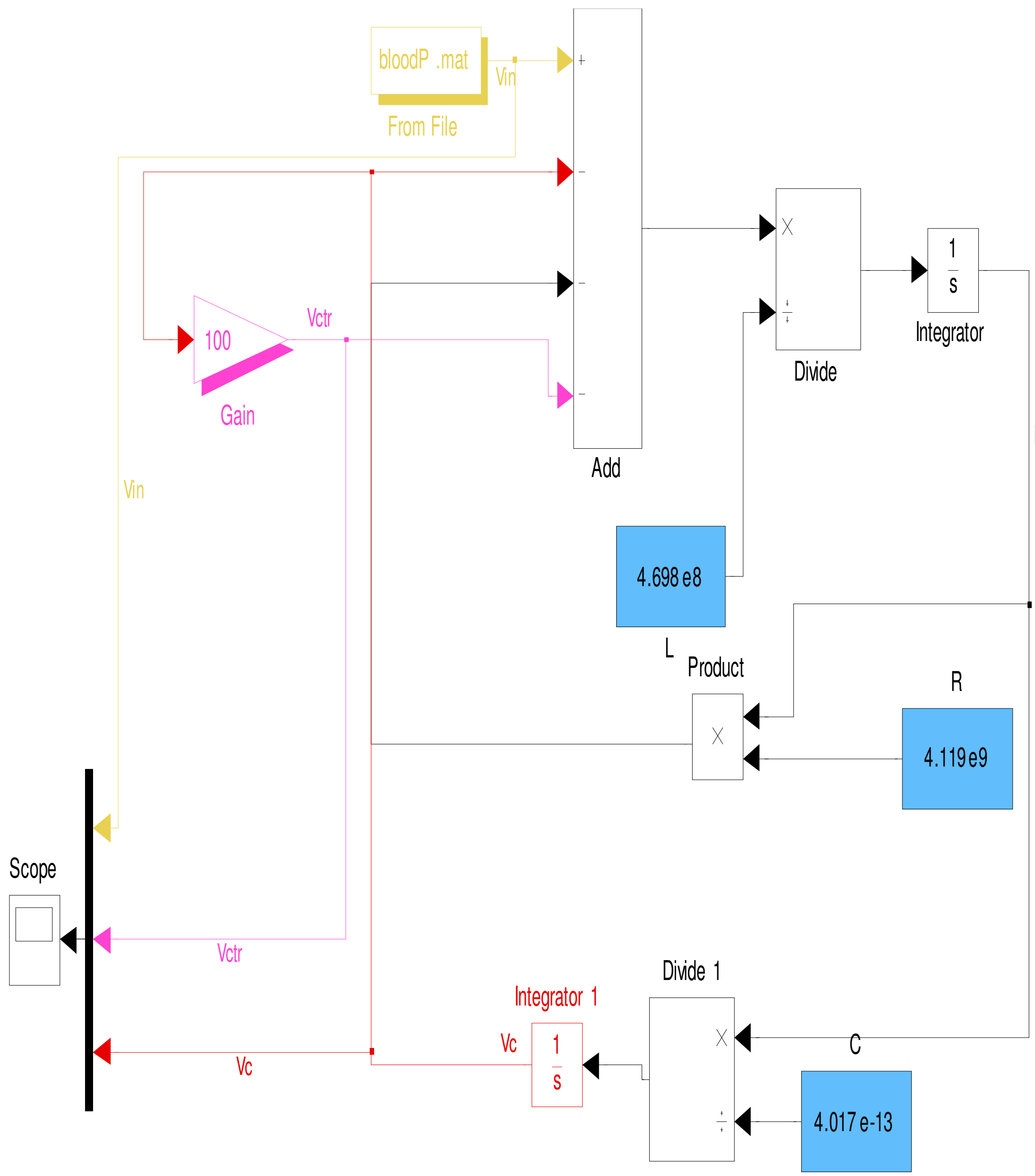

Figure 4.1: MATLAB Simulink model of catheter transducer system with feedback loop (Second order dynamic system)

For comparison with a catheter transducer system without a counter pressure source another model was developed based on the electrical circuit in figure 3.1.

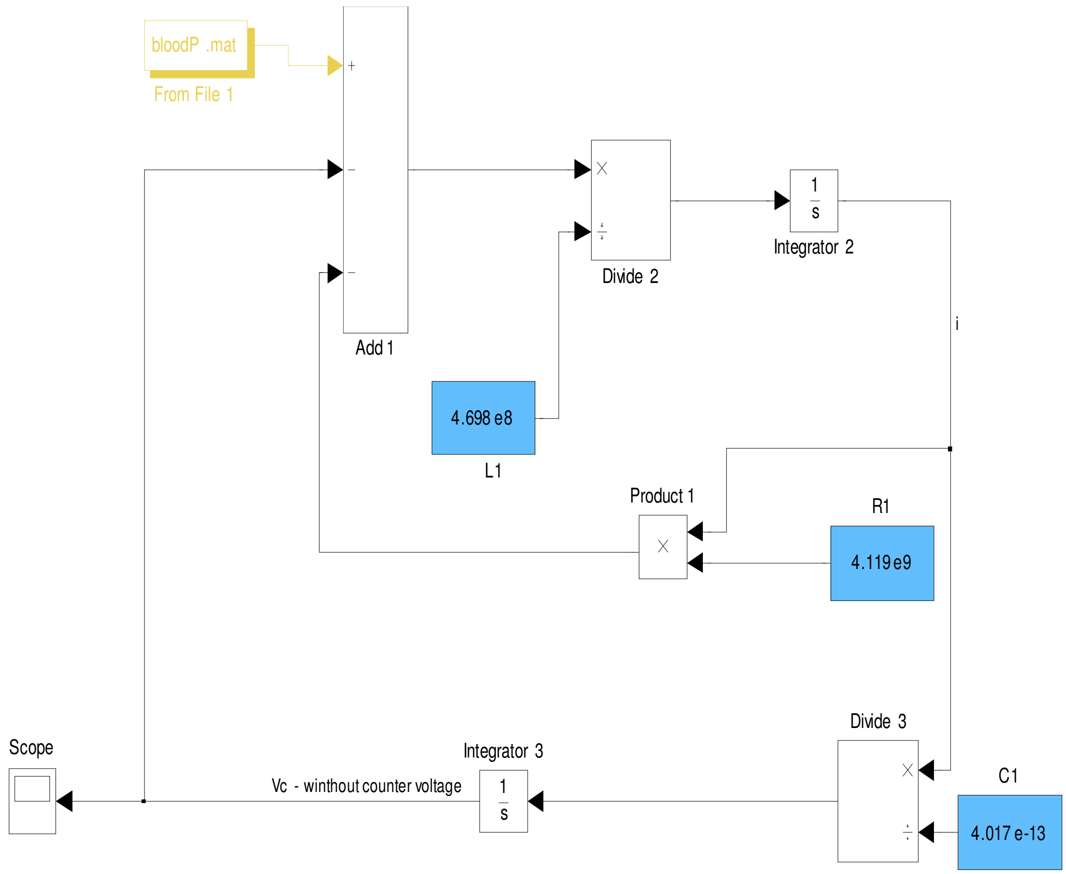

Figure 4.2: MATLAB Simulink model of catheter transducer system without the counter pressure source

These models were run for a period of 10 seconds using the fourth order Runge – Kutta solver, a fixed step size of 0.001, a blood pressure wave (randomly varying systolic pressure and diastolic pressure with a time period of 0.664 seconds) as input and a feedback gain of 1000. The simulation was repeated three times using different sizes of air bubbles, each at different locations in the system. These conditions were incorporated into the model by varying the values of R, L and C as seen in table 4.1.

4.2 Model B

An electrical circuit (2^{nd} order system) was built, equivalent to the catheter transducer system with a counter pressure source as shown in figure 4.3 and figure 4.4.

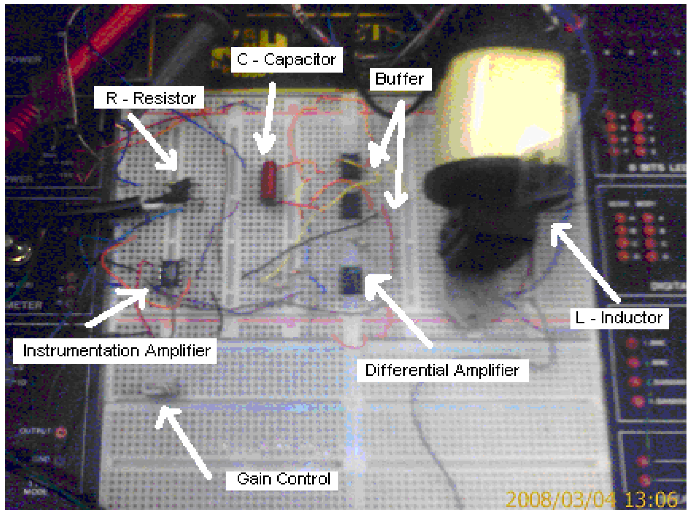

Figure 4.3: Electrical circuit (2^{nd} order system), equivalent to the catheter transducer system with a counter pressure source

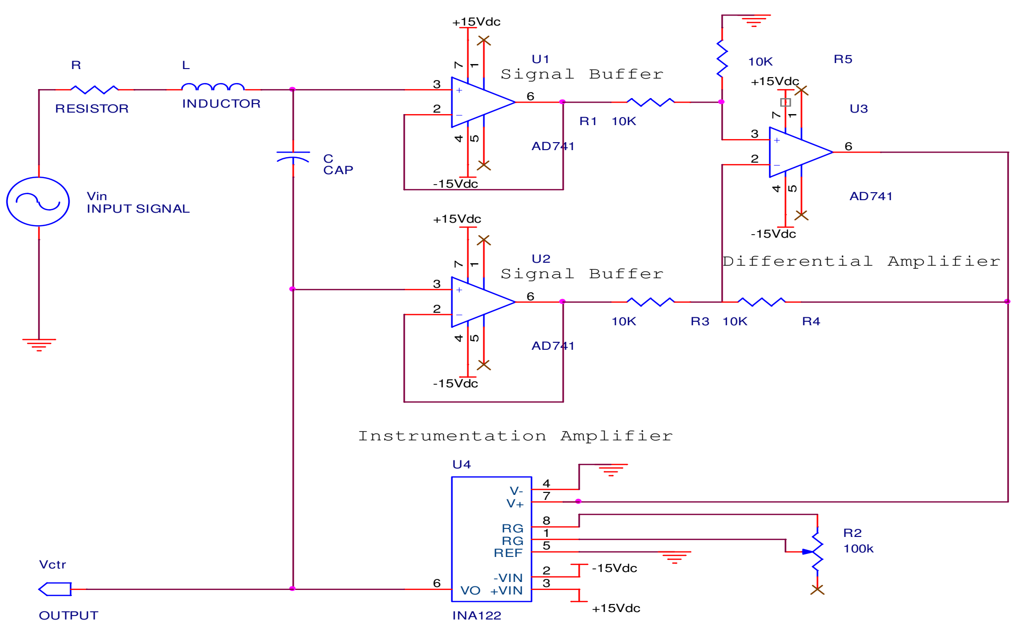

Figure 4.4: Schematic of the electrical circuit (2^{nd} order system), equivalent to the catheter transducer system with a counter pressure source

Model parameters such as Resistance R, Compliance C and Inertance L were selected according to the availability of the electrical components in the laboratory. A gain of 100 was used as the feedback loop gain (adjusted via potentiometer R2 in figure 4.4).

Blood pressure wave (randomly varying systolic pressure and diastolic pressure with a time period of 0.015 seconds), was used as the input (Vin) to the model (figure 4.3).

For generating the blood pressure wave the following setup was used:

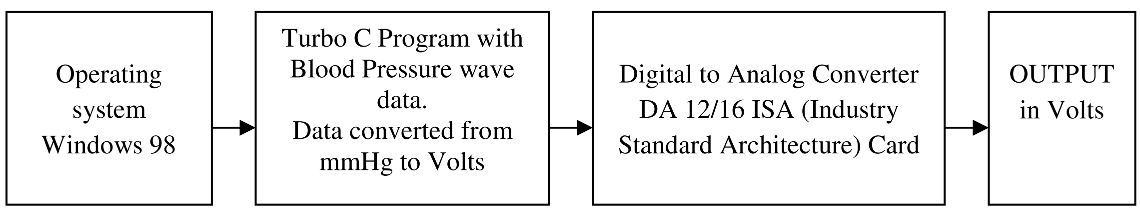

Figure 4.5: Generating blood pressure signal using a computer

Output (Vctr) data was acquired for a period of one second, using the following setup:

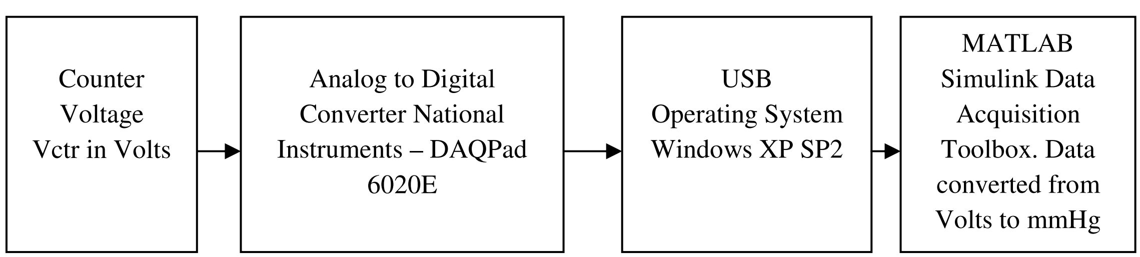

Figure 4.6: Data Acquisition setup

4.3 Model C

A simulation model of the catheter transducer system (fourth order dynamic system) with a counter pressure source was developed based on the electrical circuit in figure 3.7, which can be expressed by the following equations:

$$V_{in} - R1i1 - L1\frac{di1}{dt} - \frac{1}{C1}\int i1dt = 0 \tag{30}$$

$$Vc1 = \frac{1}{C1}\int i1dt \tag{31}$$

$$Vc1 - R2i2 - L2\frac{di2}{dt} - \frac{1}{C2}\int i2dt - V_{ctr} = 0 \tag{32}$$

$$Vc = \frac{1}{C2}\int i2dt \tag{33}$$

$$V_{ctr} = g \times V_c \tag{34}$$

Model parameters such as Resistance *R1 & R2 (R1=R2=R/2)*, Compliance *C1 & C2 (C1 = C2 = C/2)* and Inertance *L1 & L2 (L1 = L2 = L/2)* were selected from table 4.1. Once the model parameters were defined, the π model of a catheter-transducer system with a feedback loop (counter pressure source) was developed (figure 4.7) using the Simulink toolbox in MATLAB (Version R2007a).

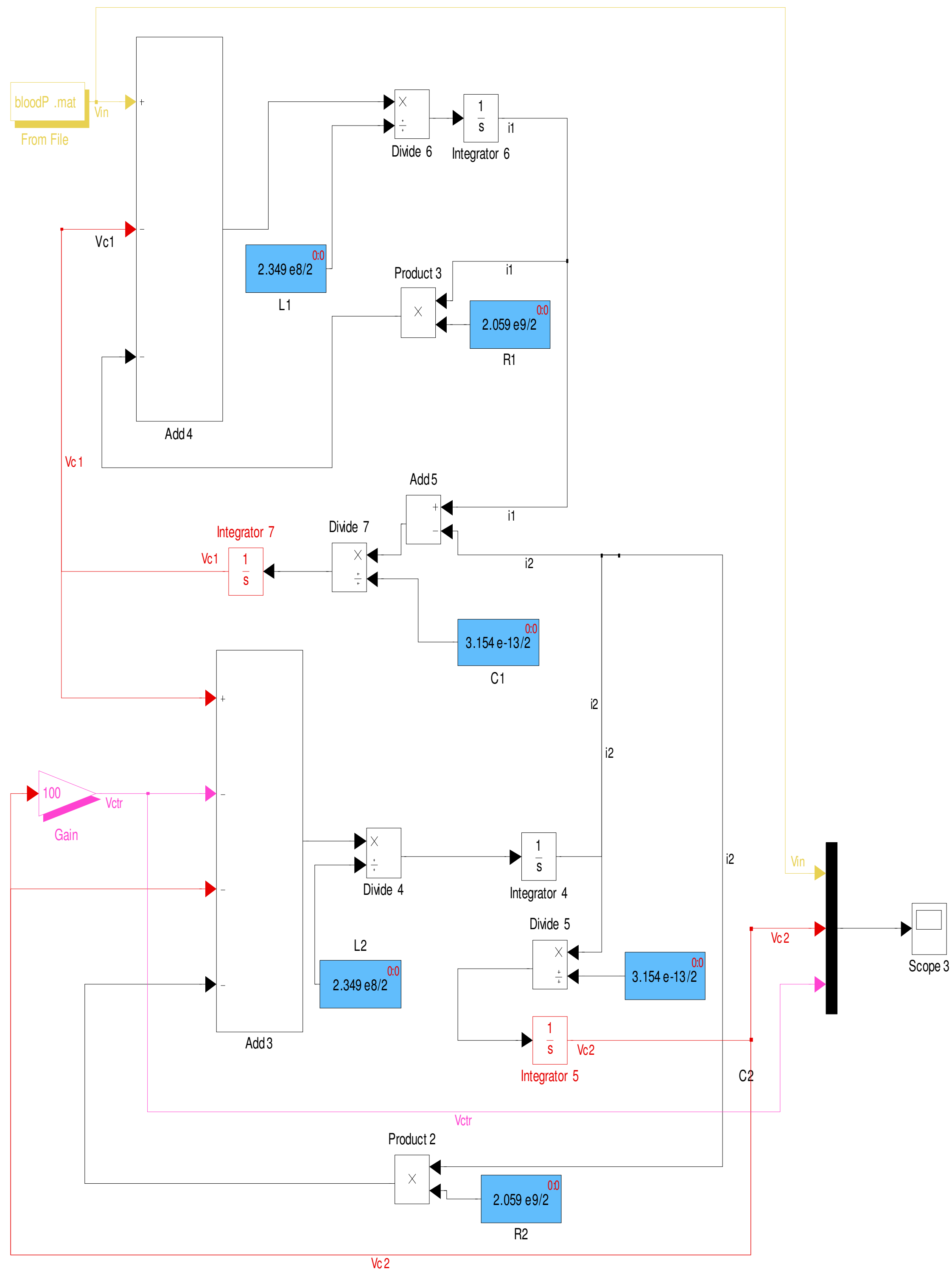

Figure 4.7: MATLAB Simulink model of catheter transducer system with feedback loop (counter pressure source)

For comparison with a catheter transducer system without a counter pressure source another model was developed based on the electrical circuit in figure 3.5.

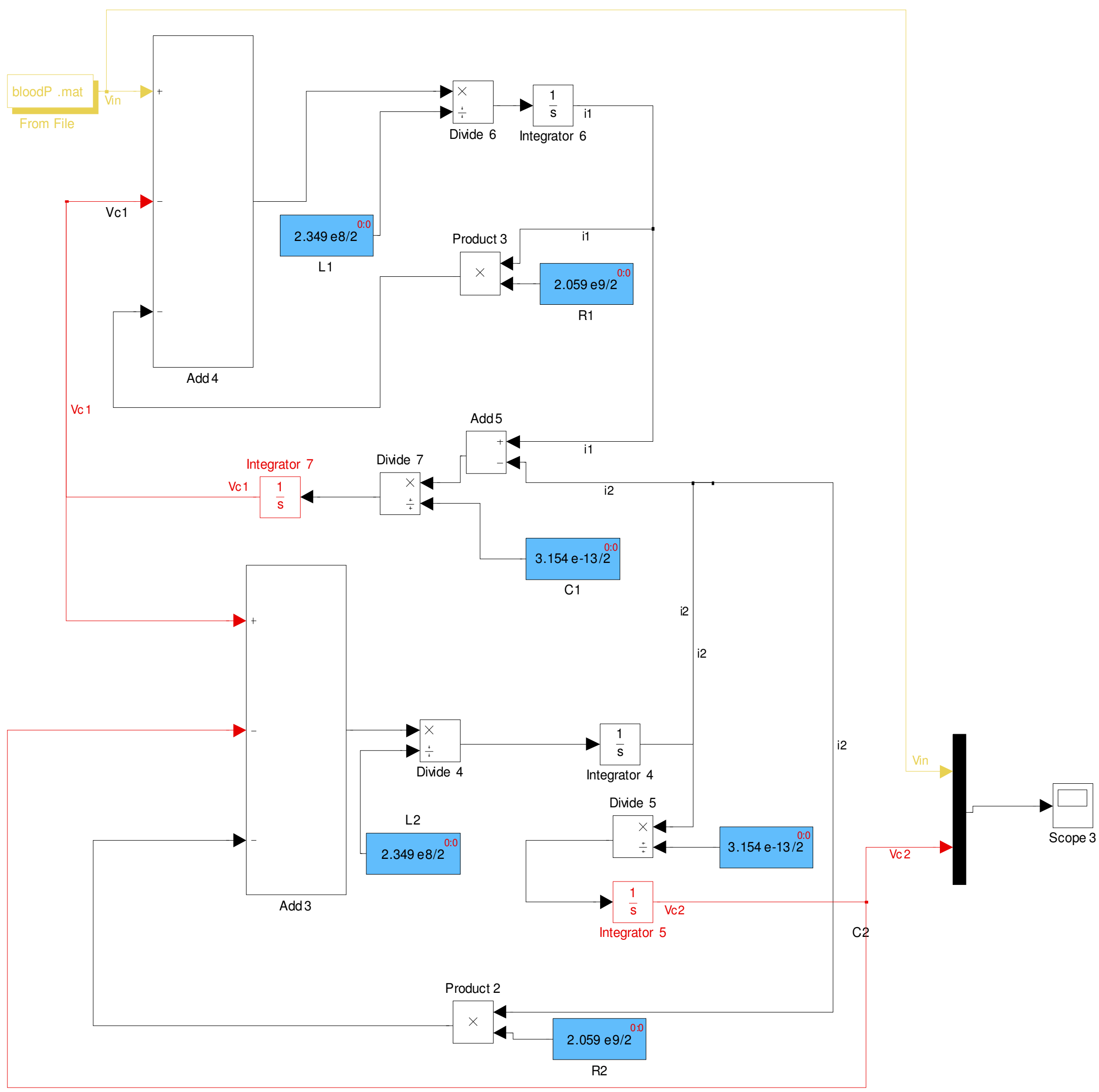

Figure 4.8: MATLAB Simulink model of catheter transducer system without the counter pressure source

These models were run for a period of 10 seconds using the fourth order Runge – Kutta solver, a fixed step size of 0.001, a blood pressure wave (randomly varying systolic pressure and diastolic pressure with a time period of 0.664 seconds) as input and a feedback gain of 1000. The simulation was repeated three times using different sizes of air bubbles, each at different locations in the system. These conditions were incorporated into the model by varying the values of R, L and C as seen in table 4.1.

4.4 Data Analysis

Error analysis was carried out on data collected from Model A, Model B and Model C. Errors were defined as follows:

Mean square error $$MSE = \frac{1}{N}\sum_{i=1}^{N}\left[V_{in} - V_{output}\right]^2 \quad (35)$$

Where, N is the number of samples, V_{in} is the true blood pressure (input) and V_{output} is V_{ctr} (for models with counter pressure source) and V_c (for models without counter pressure source).

Root mean square error $$RMSE = \sqrt{MSE} \quad (36)$$

A visual comparison was also necessary to ensure the effectiveness of the models in reproducing the diagnostic information present in the true blood pressure waveform [19].

To test the differences proposed by the hypotheses, paired t tests ($\alpha = 0.05$) were conducted using Statistics Toolbox (V5.2) in MATLAB (Only systolic and diastolic pressures were considered for the paired T tests and all the models were tested individually).

CHAPTER V

RESULTS

The results from the models discussed in the previous chapter were tabulated in spreadsheets and error analysis (as explained in section 4.4) was carried out between the true blood pressure signal (input pressure signal) and the output pressure signal (table 5.1), counter pressure signal for systems with counter pressure and output across the capacitor (V_c) for systems without counter pressure. In order to visually compare them, the output signals were plotted with the true blood pressure signal (input pressure signal).

Table 5.1: Error analysis (commonly used unit for blood pressure measurement is mmHg, SI unit conversion factor is 1mmHg = 133.322Pa)

Catheter Transducer Model	Model with counter pressure source		Model without counter pressure source	
	MSE	RMSE	MSE	RMSE
A: 2nd Order system (Simulation).	0.03 mm^2Hg^2	0.17 mmHg	90.6 mm^2Hg^2	9.51 mmHg
B: 2nd order system (Experimental Setup)	4.29 mm^2Hg^2	2.07 mmHg	64.9 mm^2Hg^2	8.05 mmHg
C: 4th order system (Simulation).	1.66 mm^2Hg^2	1.28 mmHg	84.5 mm^2Hg^2	9.19 mmHg

5.1 Graphical results for Model A

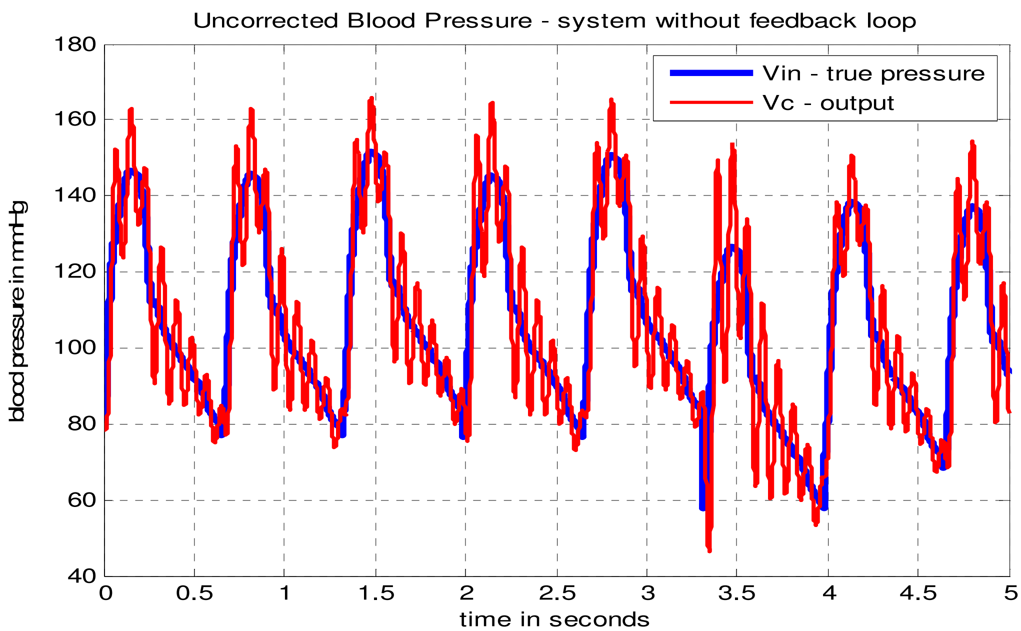

Figure 5.1: True blood pressure wave and output pressure wave for – bubble size: 5uL and bubble location l = 1.2m (refer table 4.1)

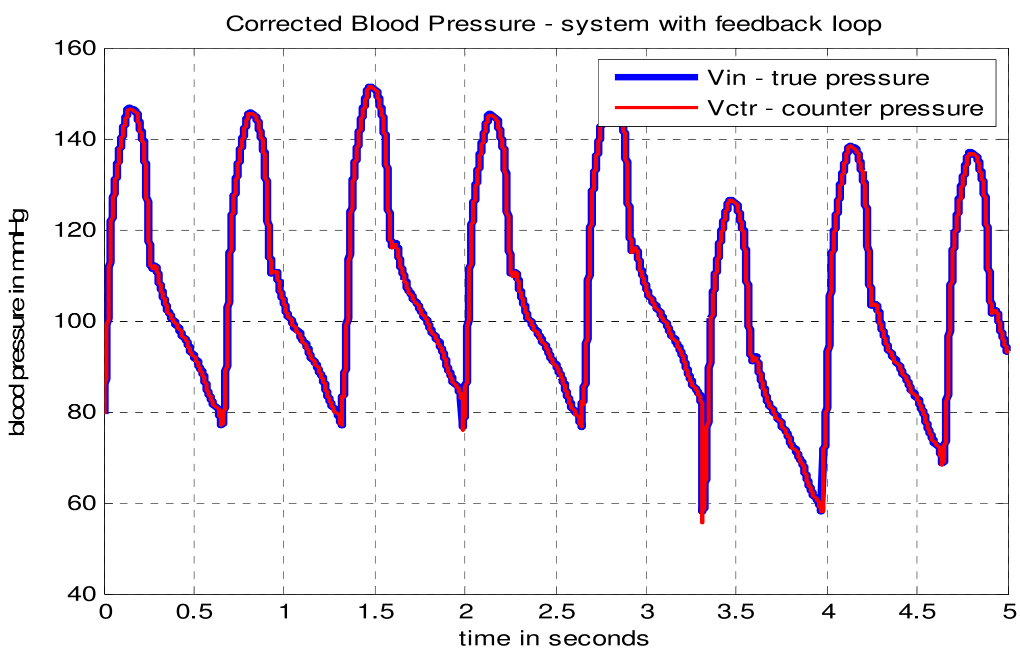

Figure 5.2: True blood pressure wave and counter pressure wave for – bubble size: 5uL and bubble location l = 1.2m (refer table 4.1)

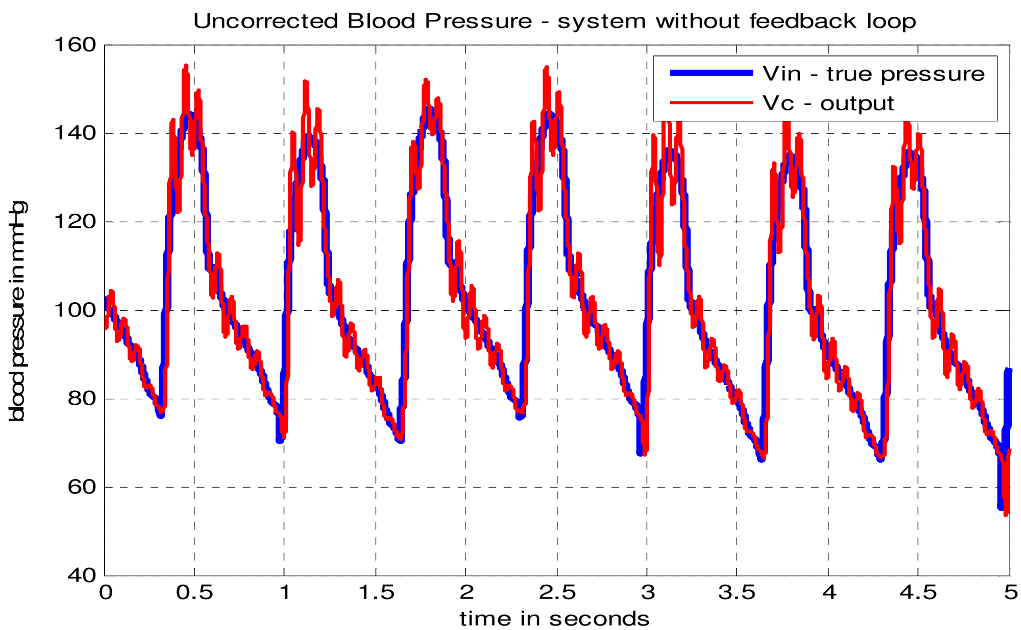

Figure 5.3: True blood pressure wave and output pressure wave for – bubble size: 15uL and bubble location l = 0.9m (refer table 4.1)

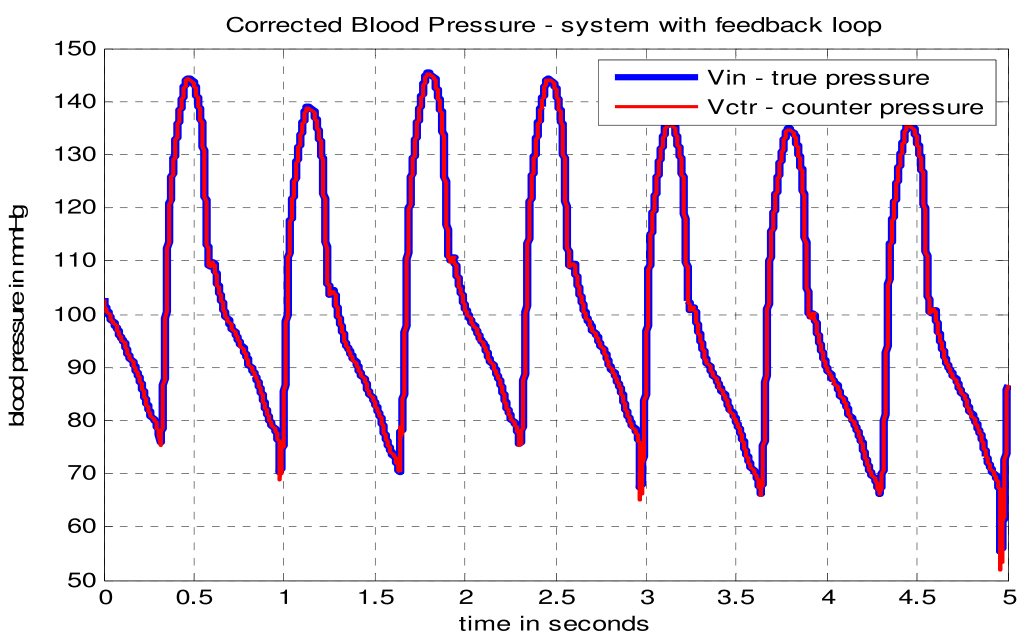

Figure 5.4: True blood pressure wave and counter pressure wave for – bubble size: 15uL and bubble location l = 0.9m (refer table 4.1)

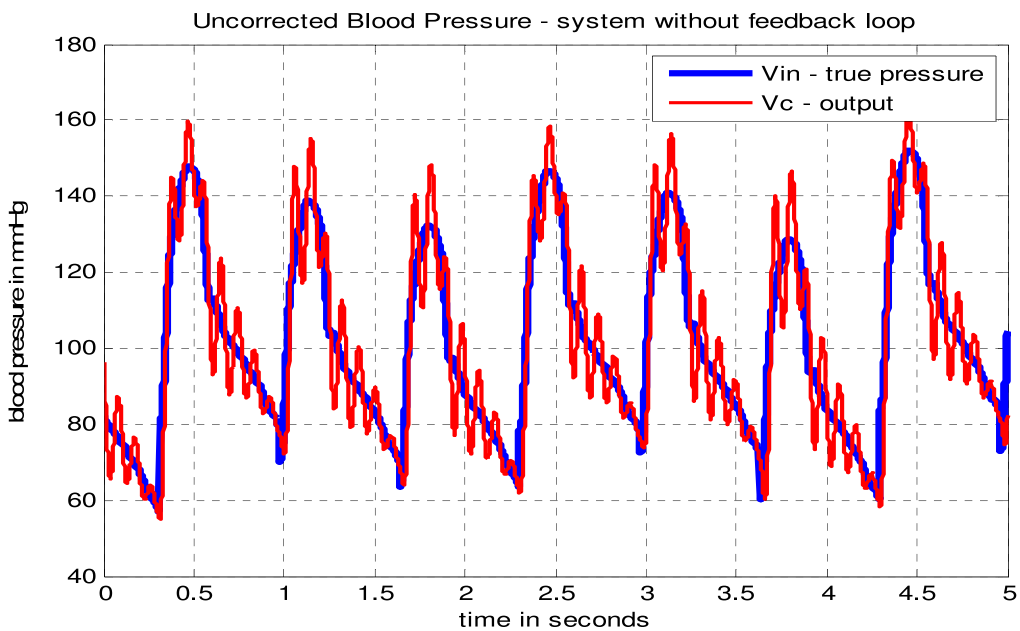

Figure 5.5: True blood pressure wave and output pressure wave for – bubble size: 20uL and bubble location l = 0.6m (refer table 4.1)

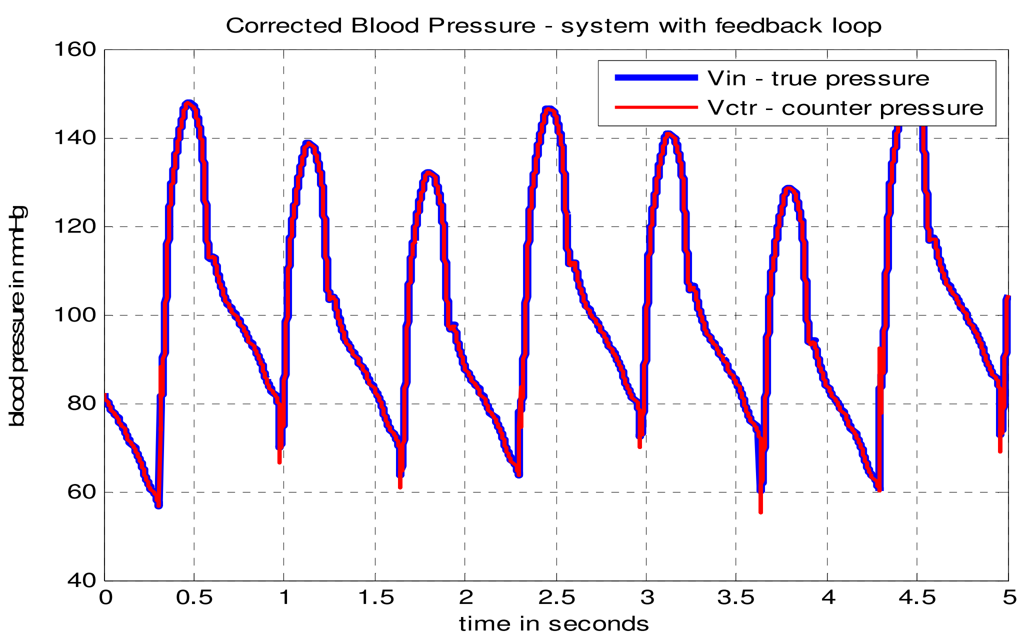

Figure 5.6: True blood pressure wave and counter pressure wave for – bubble size: 20uL and bubble location l = 0.6m (refer table 4.1)

5.2 Graphical results for Model B

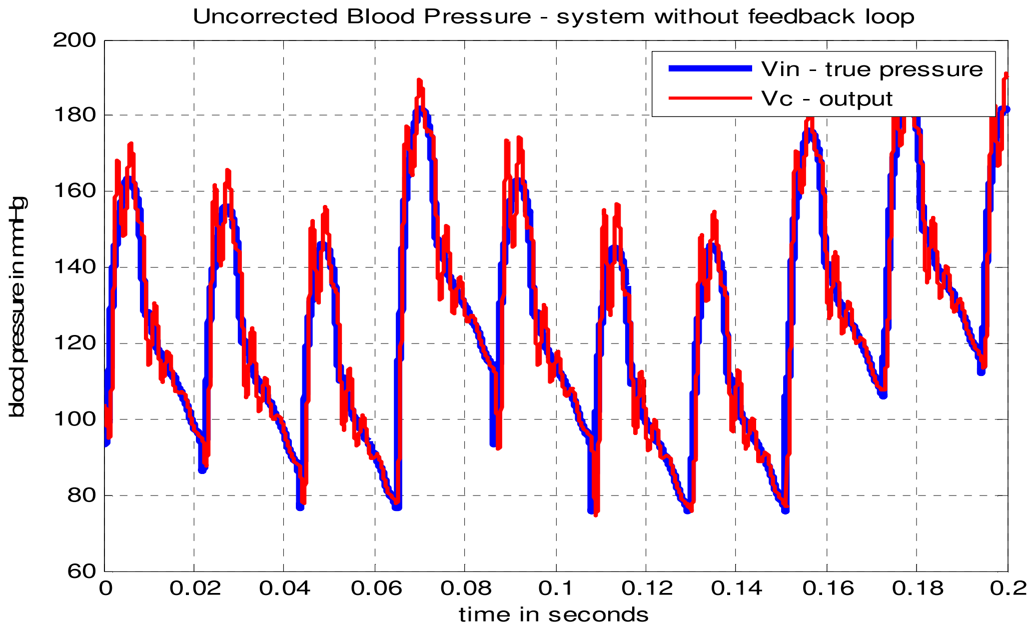

Figure 5.7: True blood pressure wave and output pressure wave for a system without counter pressure

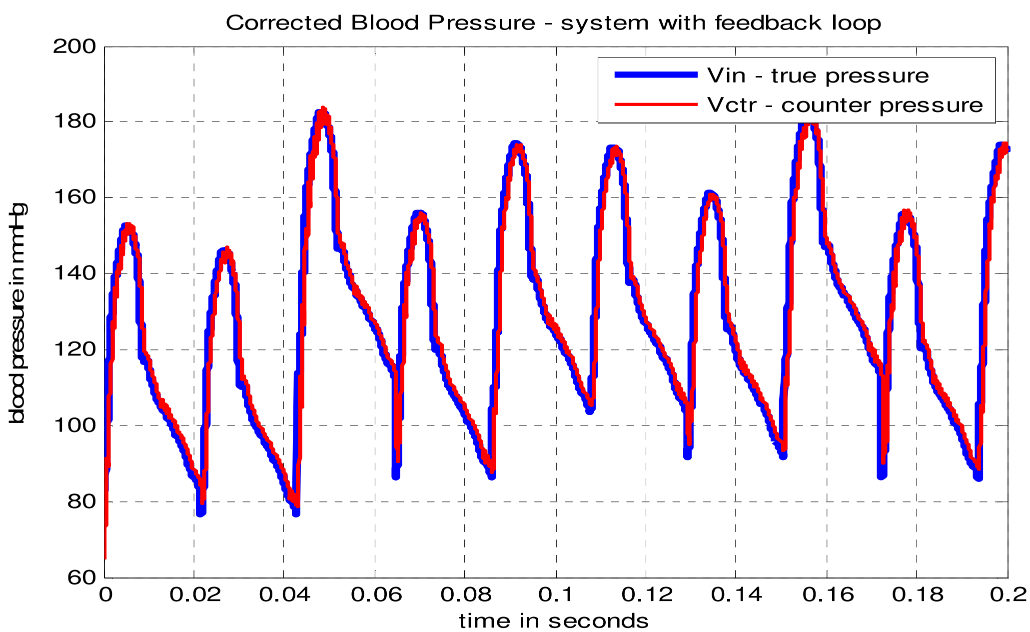

Figure 5.8: True blood pressure wave and counter pressure wave (output of the feedback loop) for a system with feedback gain of 100

5.3 Graphical results for Model C

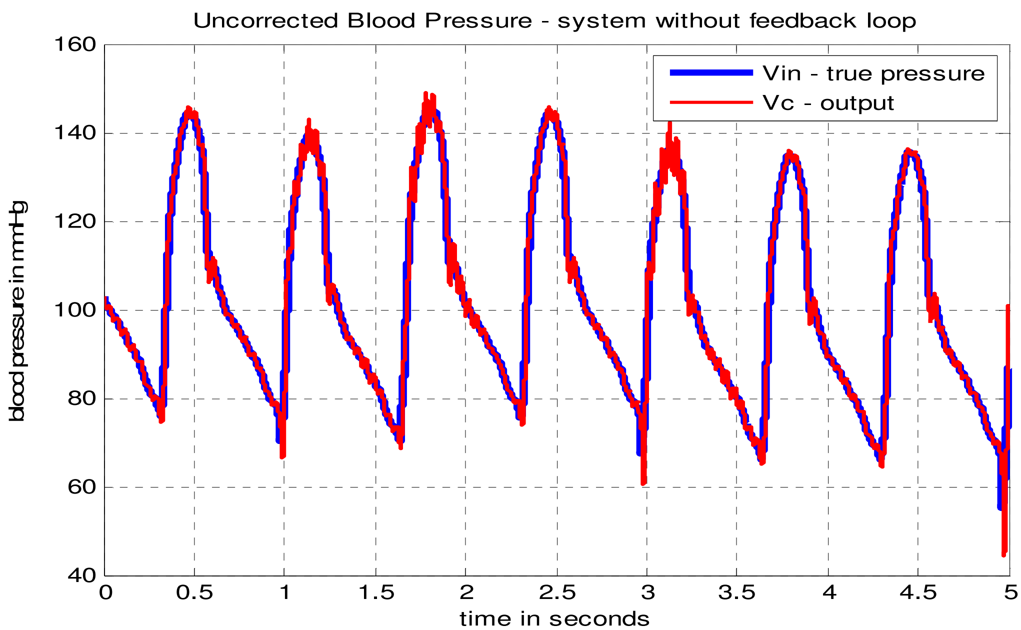

Figure 5.9: True blood pressure wave and output pressure wave for – bubble size: 20uL and bubble location l = 0.6m (refer table 4.1)

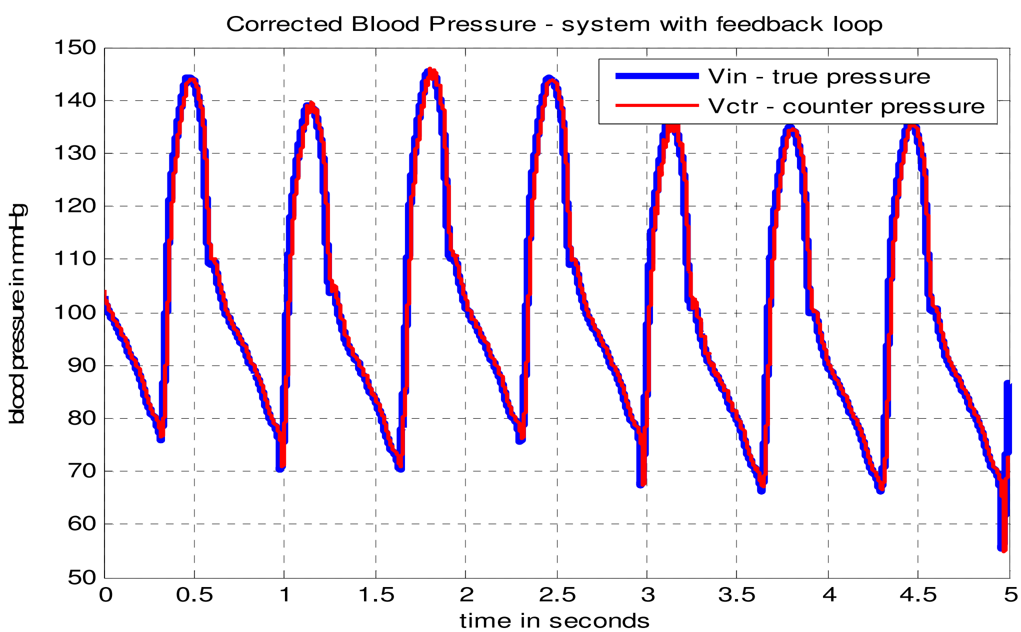

Figure 5.10: True blood pressure wave and counter pressure wave for – bubble size: 20uL and bubble location l = 0.6m (refer table 4.1)

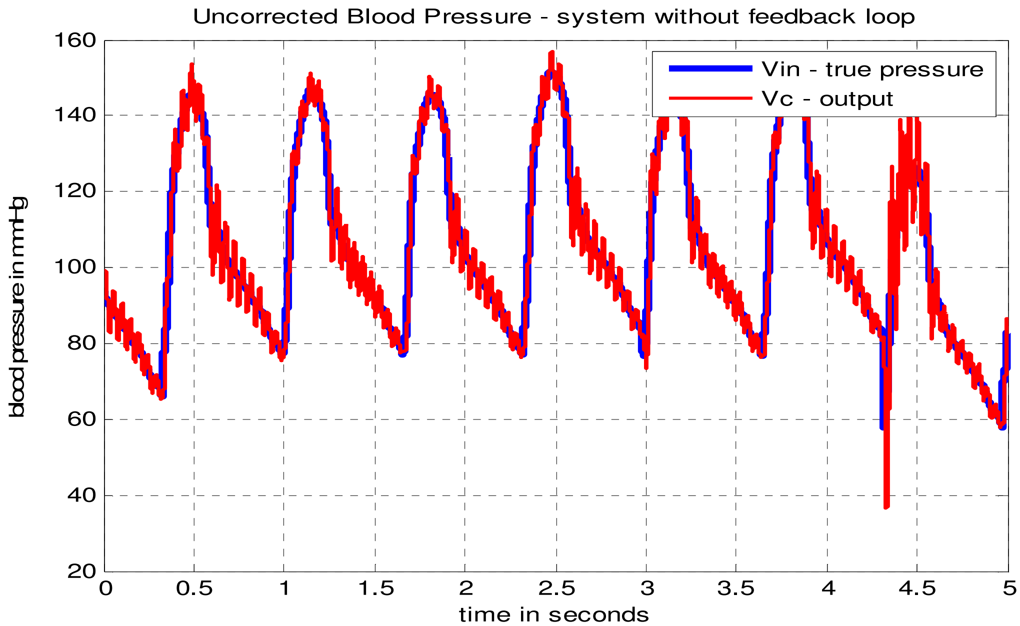

Figure 5.11: True blood pressure wave and output pressure wave for – bubble size: 5uL and bubble location l = 0.9m (refer table 4.1)

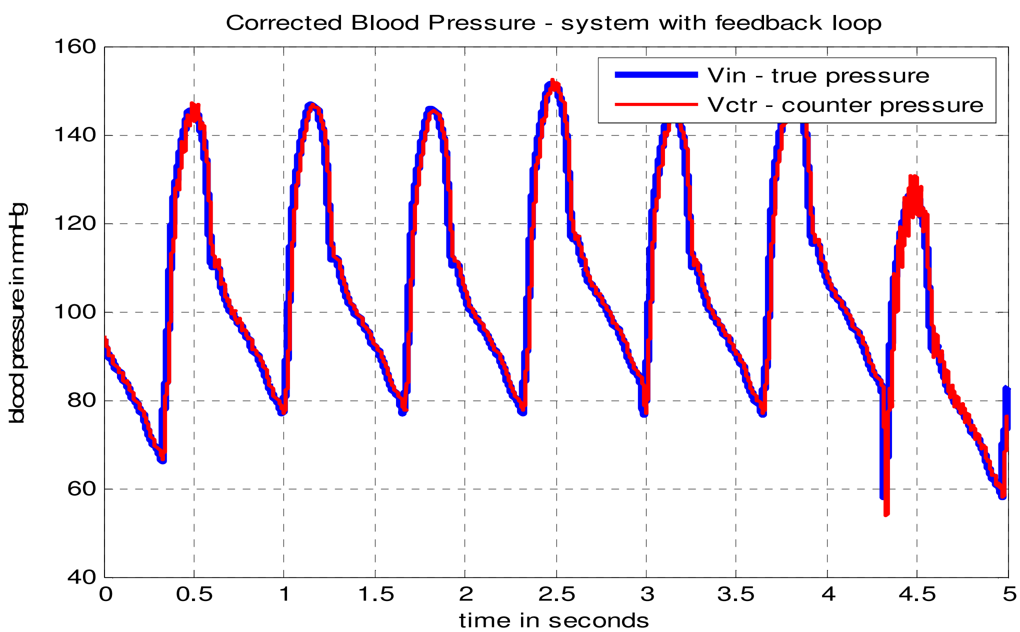

Figure 5.12: True blood pressure wave and counter pressure wave for – bubble size: 5uL and bubble location l = 0.9m (refer table 4.1)

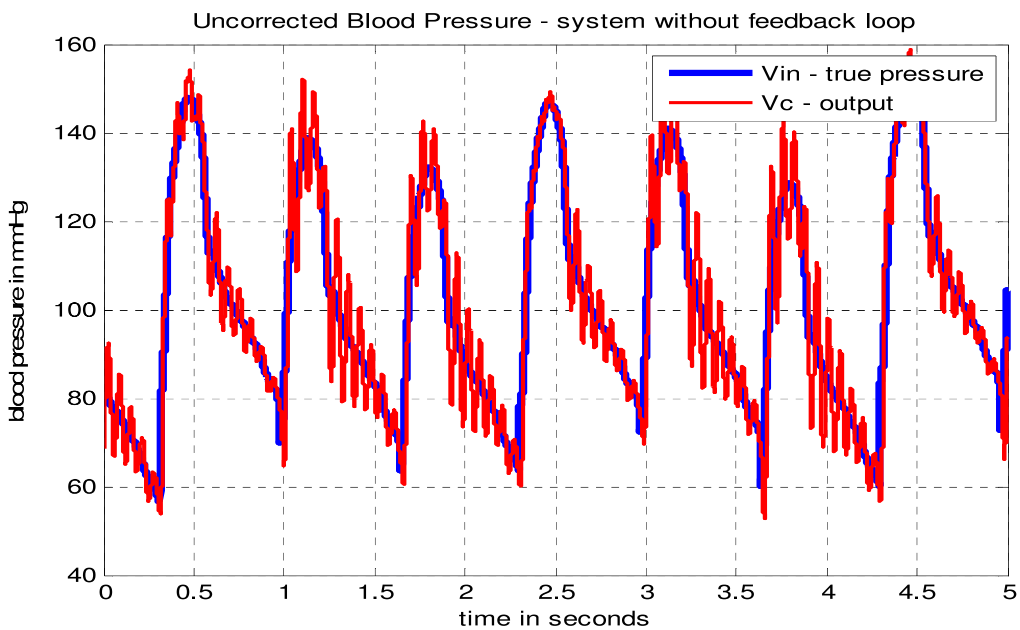

Figure 5.13: True blood pressure wave and output pressure wave for – bubble size: 5uL and bubble location l = 1.2m (refer table 4.1)

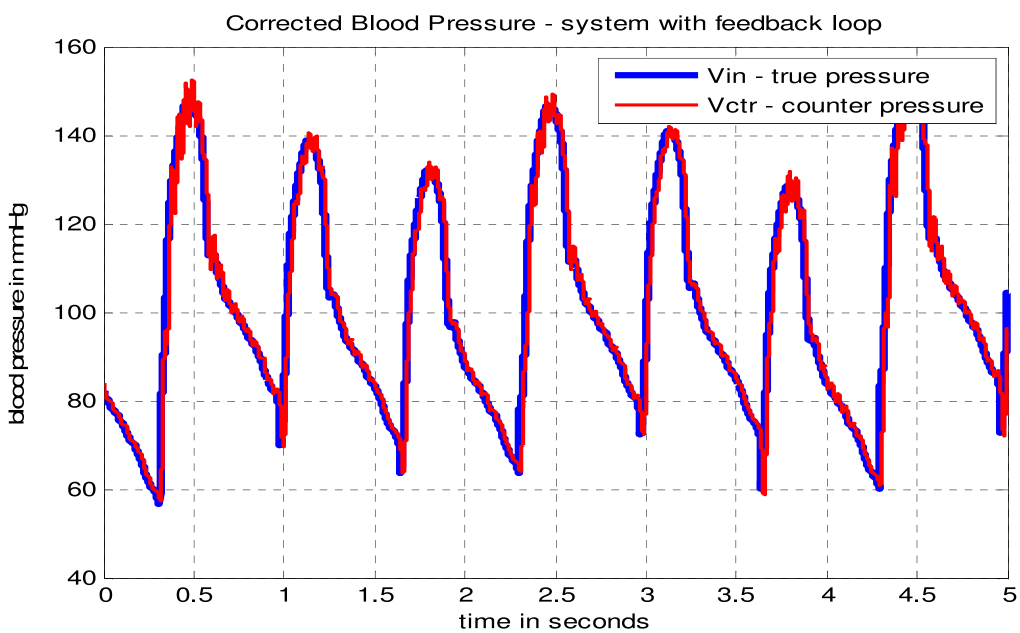

Figure 5.14: True blood pressure wave and counter pressure wave for – bubble size: 5uL and bubble location l = 1.2m (refer table 4.1)

5.4 Tabulated results for the Paired T tests ($\alpha = 0.05$)

From the graphed results (figures 5.1 to 5.14), the systolic and diastolic pressures were calculated and tabulated (Appendix D, E and F). The differences proposed by the hypotheses were then tested for every model using a paired T test with a significance level of 0.05. The results of these tests for each model are as follows (tables 5.2 to 5.4):

Table 5.2: Paired T test results for Model A

System with Counter Pressure Source		System without counter pressure source	
Output Systolic Pressure Vs True Systolic Pressure	Output Diastolic Pressure Vs True Diastolic Pressure	Output Systolic Pressure Vs True Systolic Pressure	Output Diastolic Pressure Vs True Diastolic Pressure
$p = 0.1246$	$p = 0.2054$	$p = 5.1195\text{x}10^{-21}$	$p = 4.823\text{x}10\text{-}4$
$p \rangle \alpha$	$p \rangle \alpha$	$p \langle \alpha$	$p \langle \alpha$
Fail to reject H_{01}	**Fail to reject H_{02}**	**Reject H_{03}**	**Reject H_{04}**

Table 5.3: Paired T test results for Model B

System with Counter Pressure Source		System without counter pressure source	
Output Systolic Pressure Vs True Systolic Pressure	Output Diastolic Pressure Vs True Diastolic Pressure	Output Systolic Pressure Vs True Systolic Pressure	Output Diastolic Pressure Vs True Diastolic Pressure
$p = 0.1240$	$p = 0.2826$	$p = 1.5464\text{x}10\text{-}8$	$p = 0.0023$
$p \rangle \alpha$	$p \rangle \alpha$	$p \langle \alpha$	$p \langle \alpha$
Fail to reject H_{01}	**Fail to reject H_{02}**	**Reject H_{03}**	**Reject H_{04}**

Table 5.4: Paired T test results for Model C

System with Counter Pressure Source		System without counter pressure source	
Output Systolic Pressure Vs True Systolic Pressure	Output Diastolic Pressure Vs True Diastolic Pressure	Output Systolic Pressure Vs True Systolic Pressure	Output Diastolic Pressure Vs True Diastolic Pressure
$p = 0.1856$	$p = 0.1526$	$p = 7.8253x10^{-11}$	$p = 7.0602$ x10-4
$p\rangle\alpha$	$p\rangle\alpha$	$p\langle\alpha$	$p\langle\alpha$
Fail to reject H_{01}	**Fail to reject H_{02}**	**Reject H_{03}**	**Reject H_{04}**

CHAPTER VI

DISCUSSION

As discussed in the previous chapters, the main disadvantage of the invasive blood pressure monitoring system is the inaccuracy resulting from the dynamic characteristics of the monitoring system. A previous simulation study [1] employed a counter pressure source in tandem with the transducer. The counter pressure generated minimized the fluid flow in the pressure monitoring system. When flow in the system was zero the counter pressure generated closely approximated the true blood pressure. The current study developed a real-time technique to generate accurate and dependable counter pressure. To validate this technique, one electrical circuit (Model B, second order dynamic system) and two simulation models (Model A, second order dynamic system and model C, fourth order dynamic system) of catheter transducer system were developed and tested under different system conditions (air bubbles of three sizes at three locations in the system, table 4.1).

The results from the developed models were tabulated in spreadsheets and error analysis was carried out between the true blood pressure signal (input pressure signal) and the output pressure signal.

Low values of *MSE* and *RMSE* (table 5.1) for the systems with counter pressure source indicate that counter pressure source was able to reproduce the true blood pressure accurately. As low *MSE* (or *RMSE*) does not necessarily mean clinical acceptance, a visual comparison was necessary to ensure the effectiveness of the models in reproducing the diagnostic information present in the true blood pressure waveform.

For visual comparison, output signals of the developed models (under different air bubble size and location, chosen randomly from table 4.1) were plotted with the true blood pressure signals (figures 5.1 to 5.14). Visual comparisons (checking for diagnostic information) indicate that, the systems with counter pressure source were able to reproduce the true blood pressure and its diagnostic information.

In addition to the error analysis and visual comparisons, parametric tests (paired T tests) were also conducted on the data collected from each model (tables 5.2 to 5.4). The following conclusions were drawn from the results of these tests:

- For all the models, this study failed to reject the null hypotheses H_{01} and H_{02}. Thus concluding that there is no difference between the output systolic and diastolic pressure of the catheter transducer system with counter pressure source and the true systolic and diastolic pressure.
- For all the models, this study rejected the null hypotheses H_{03} and H_{04}. Thus concluding that there is a difference between the output systolic and diastolic pressure of the catheter transducer system without counter pressure source and the true systolic and diastolic pressure.

6.1 Salient features of the developed technique

- Accurate in reproducing the true blood pressure signal.
- Independent of the system characteristics and changes over time.
- A direct feedback loop is used to generate the required counter pressure thus eliminating any computational delay.
- In the future, this technique could be implemented on an actual catheter transducer system by replacing the conventional transducer with a differential transducer and adding a counter pressure source (figure 6.1).

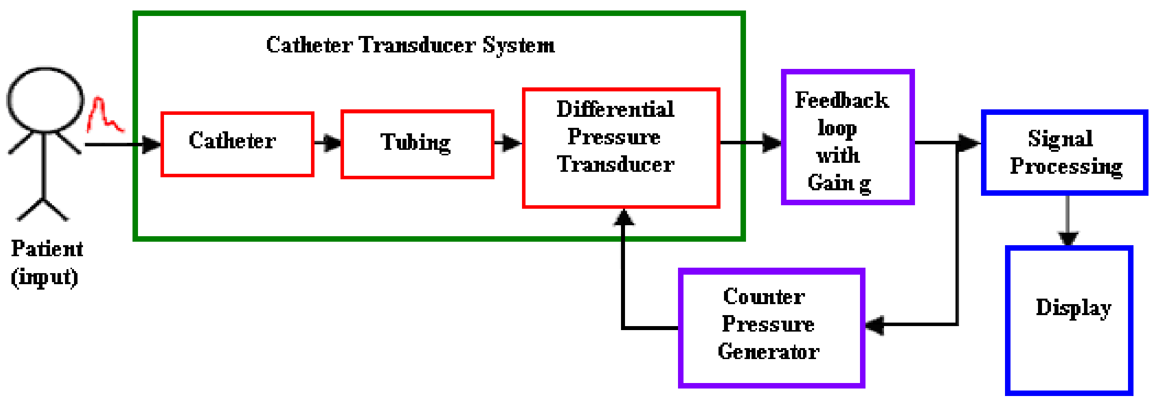

Figure 6.1: Future vision - Implementation of counter pressure on an actual catheter transducer system [1]

6.2 Conclusion

Based on the results from the error analysis, visual comparisons and parametric tests (paired T tests) it was concluded that the proposed real-time technique to generate counter pressure (via a feedback loop) was effective and accurate in approximating the true blood pressure signal.

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APPENDICES

APPENDIX A

MATLAB PROGRAM FOR RUNNING THE SIMULATION, PLOTTING RESULTS AND ERROR ANALYSIS

```
% this program is written to run the simulation, plot the output
% and to calculate Mean Square error and Root Mean square error

%runs the chosen model
sim('modelA');

%plotting the output for system with feedback
figure(1)
plot(data(:,1),data(:,2),'blue','linewidth',3);
hold;
plot(data(:,1),data(:,3),'red','linewidth',2);
grid;
xlabel('time in seconds');
ylabel('blood pressure in mmHg');
title('Corrected Blood Pressure - system with feedback loop');
legend('Vin - true pressure','Vctr - counter pressure');
hold;

%plotting the output for the system without feedback
figure(2)
plot(data(:,1),data(:,2),'blue','linewidth',3);
hold;
plot(data(:,1),data(:,4),'red','linewidth',2);
grid;
xlabel('time in seconds');
ylabel('blood pressure in mmHg');
title('Uncorrected Blood Pressure - system without feedback loop');
legend('Vin - true pressure','Vc - output')
hold;

% Error analysis
err1 = data(:,2)-data(:,3);
sqerr1 = err1(:,1).*err1(:,1);
mse1 = mean(sqerr1)
rmse1 = sqrt(mse1)
err2 = data(:,2)-data(:,4);
sqerr2 = err2(:,1).*err2(:,1);
mse2 = mean(sqerr2)
rmse2 = sqrt(mse2)
```

APPENDIX B

MATLAB PROGRAM TO GENERATE RANDOM BLOOD PRESSURE WAVE

FROM FOURIER COEFFICIENTS

```
clear all
clc;

Real = [202.037 -6.825 -12.891 -3.14 -2.381 -2.912 -0.457 -1.524 -0.65
9.61e-3 -0.573 0.335 -0.153 0.053 0.244 -0.119];
Imag = [26.858 4.722 -0.564 1.701 -0.613 -0.792 -0.137 -1.48 -0.171 -
0.602 -0.542 -2.38e-3 -0.398 0.055 -0.076];

P = 0.664; % Period
t = 0:0.001:5;

BP = 0;

for n = 1:15
    F(n)  = Real(n+1) + i*Imag(n);
    BP = BP + F(n)*exp(-i*n*2*pi/P*t);
End

BP = BP + Real(1)/2;

A = real(BP);
x = 1;
e = 1;
for i = 1:length(t)
    x = x+1;
    Arand(i) = A(i)+ e;
    if(x == 665)
    e=(10*randn(1));
    x = 1;
    end

end

RandomBP(1,:) = t;
RandomBP(2,:) = Arand;
```

APPENDIX C

C PROGRAM TO GENERATE RANDOM BLOOD PRESSURE WAVE

```
#include <stdio.h>
#include <conio.h>
#include <iostream.h>
#include <math.h>
#include <stdlib.h>
#include <stdarg.h>
#include <dos.h>

#define BASE2 0x320
#define CHOUT BASE2+0

void main(void)
{
      int j, out, k, Random = 0;

//cut paste the blood pressure data that needs to be used //the data
has to be of 664 data points

      float v[]={};

//setting to AUTOMATIC UPDATE mode

      inp(BASE2+10);
      inp(BASE2+15);

//send data to port till any key is
      while(!kbhit())
      {
            for(j=0;j<664;j++)
            {
            out = v[j]*10+1400+Random;
            outport(CHOUT,out);
            delay(1);
            }
      Random = rand()%300;
      }
}
```

APPENDIX D

DATA USED FOR PAIRED T TEST: MODEL A

	SYSTOLIC PRESSURE mmHg			DIASTOLIC PRESSURE mmHg		
	True Systolic Pressure	Output Systolic Pressure for system with counter pressure	Output Systolic Pressure for system without counter pressure	True Diastolic Pressure	Output Diastolic Pressure for system with counter pressure	Output Diastolic Pressure for system without counter pressure
Set 1	139.6	139.5	167	67.04	66.23	65.59
bubble size: 5uL and bubble location1 = 1.2m	134.5	134.5	152.1	66.03	66.04	62.83
	145.5	145.4	158	76.93	76.86	74.43
	146.7	146.6	162.8	77.15	77.01	75.12
	145.7	145.3	163	77.23	77.08	73.91
	151.5	151.3	165.6	76.75	76.06	75.33
	145.1	145.1	164.7	76.72	76.64	73.16
	150.5	150.4	165.2	59.33	59.1	46.88
	136.4	126.4	153.9	58.01	57.95	53.55
	138.4	138.2	150.9	69.84	68.47	67.36
	137.1	136.9	154.1	53.99	52.54	49.69
	122.6	122.4	145.8	53.99	53.94	49.87
	141.2	141	151.2	59.5	59.67	54.95
	127.8	127.9	150.3	59.47	59.41	55.5
	152.8	152.6	161.3	70.37	69.68	68.2
set 2	139.6	139.5	164.1	66.86	66.12	67.79
bubble size: 15uL and bubble location1 = 0.9m	138.6	138.3	149	66.86	66.79	67.31
	135.1	135	146	51.3	49.9	37.49
	149.5	149.4	152.1	51.29	51.25	54.24
	119.9	119.7	145.4	74.32	75.24	73.91
	142.9	142.8	144.9	77.34	77.02	78.42
	147.5	147.4	155.2	75.81	75.52	76.57
	145.6	145.6	156.9	70.77	70.82	71.09
	144.4	144.2	155.3	70.68	71	71.05
	139	139	151.9	75.72	75.55	76.82
	145.2	145	152.2	67.48	67.45	67.19
	144.3	144.1	155	66.26	66.12	66.64
	136.1	135.9	150.4	66.26	66.15	66.55
	134.8	134.6	145.5	71.03	71.02	68.71
	135.6	135.5	144.9	71.32	71.32	69.62
set 3	139.6	139.5	167.9	58.36	58.36	56.94
bubble size: 20uL and bubble location1 = 0.6m	139.9	139.8	152.7	58.36	58.36	56.47
	145.1	145	158.1	65.43	65.4	63.51
	126.9	126.7	146.8	63.53	65.52	58.89
	134	134	146.9	56.84	56.82	55.28
	136	136	149.5	56.84	56.82	55.09
	126.4	126.1	143.4	70.15	70.13	71.98
	125.4	125.4	140	63.58	63.58	64.13
	147.9	147.8	159.6	63.58	63.58	61.8
	138.7	138.5	155.2	72.35	72.34	73.71
	132.2	132.2	148.3	60.14	60.13	60.36
	146.7	146.7	158.4	60.14	61.12	58.41
	140.6	140.3	156.5			
	128.7	128.6	146.6			
	152	152	164			

APPENDIX E

DATA USED FOR PAIRED T TEST: MODEL B

System without counter pressure source				System with counter pressure source			
True Systolic Pressure mmHg	Output Systolic Pressure mmHg	True Diastolic Pressure mmHg	Output Diastolic Pressure mmHg	True Systolic Pressure mmHg	Output Systolic Pressure mmHg	True Diastolic Pressure mmHg	Output Diastolic Pressure mmHg
162.9	172.5	87.34	88.23	153.1	153	79.63	79.63
156.1	165.7	76.55	77.77	146.5	146.4	78.85	78.88
146.4	155.9	76.76	77.67	182.9	182.9	91.2	90.37
182	189.4	113.2	72.02	155.6	155.5	86.3	86.32
163.1	174.2	94.67	74.42	174.1	174.2	104.9	105.2
144.9	156.8	75.64	76.96	173.4	173.4	92.57	93.75
145.7	154.6	75.64	76.96	161.1	161.1	92.59	93.61
175.7	183.2	107	107.3	183.4	183.4	89.4	89.9
185.2	193.1	112.1	113.6	156	155.8	88.33	88.38
181.9	190.8	106.7	106.4	174.1	174.1	106.7	106.3
175.9	185.6	77.27	71.26	184.4	184.5	115.8	116
146.8	161.3	107.5	78.79	184.2	184.3	109.1	109.3
176.9	184.6	108	108	176.8	176.8	108.3	108.7
177.6	186.1	111.3	107.8	189.9	189.9	83.78	83.86
180.9	189.6	111.3	112	149.6	148.4	80.46	80.2
185.5	193.6	86.63	77.16	186.3	186.3	105.3	105.3
153.8	168.9	118.8	85.1	171.4	171.5	89.42	89.45
190.1	197.5	92.22	82.86	175.5	175.5	88.4	88.49
159	173.7	120.5	72.69	156.6	155.9	89.75	89.37
188.2	195.4	103.8	110.7	165.6	165.7	75.98	75.96
178.1	188.4	83.98	68.62	158.7	158.6	74.96	75.03
146	161.9	95.07	77.6	141.9	142	79.48	79.59
164.9	171.8	105.2	97.21	147.1	147.2	79.8	79.5
174.9	182.7	78.69	70.04	187.7	187.5	81.95	82.11
145.5	185.5	89.23	76.76	150.3	150	86.34	86.34

APPENDIX F

DATA USED FOR PAIRED T TEST: MODEL C

	SYSTOLIC PRESSURE mmHg			DIASTOLIC PRESSURE mmHg		
	True Systolic Pressure	Output Systolic Pressure for system with counter pressure	Output Systolic Pressure for system without counter pressure	True Diastolic Pressure	Output Diastolic Pressure for system with counter pressure	Output Diastolic Pressure for system without counter pressure
Set 1	139.6	139.5	165	70.37	71.38	63.92
bubble size: 20uL and bubble location l = 0.6m	138.6	135.6	142.6	66.86	67.54	64.92
	135.1	135	138	66.86	67	65.86
	149.5	149	158.8	51.3	51	21.41
	119.9	120.3	141.3	51.29	51.3	48.78
	142.9	142.3	159.3	74.32	74.3	73.17
	147.5	146.7	149.5	77.34	77	75.69
	145.6	145.8	147.7	75.81	75.5	74.54
	144.4	143.9	145.9	70.77	70.52	66.87
	139	138.9	143.1	70.68	70.33	68.93
	145.2	145.1	148.7	75.72	75.8	74
	144.3	143.9	145.6	67.48	67.48	60.94
	136.1	136.4	142.2	66.26	66.26	64.5
	134.8	134.7	137.9	66.26	66.26	64.46
	135.6	135.3	140.2	67.04	67.02	67.23
set 2	139.6	141	170.5	66.03	66.03	65.52
bubble size: 5uL and bubble location l = 0.9m	134.5	134.7	140.2	76.93	76.91	75.45
	145.5	145.8	153.5	77.15	77.2	77.3
	146.7	146.7	151	77.23	77.25	78.26
	145.7	157.7	150.2	76.75	76.8	74.15
	151.5	151.6	156.8	76.72	76.8	37.8
	145.1	145.2	151.3	59.33	60	58.74
	150.5	150.7	155.4	58.01	59	68.39
	136.4	137.5	147.9	69.84	69.82	42.03
	138.4	138.6	146.2	53.99	54	53.87
	137.1	137.1	141.4	53.99	54	50.42
	122.6	122.8	131.8	59.5	59.9	59.72
	141.2	141.5	153.3	59.47	59.58	43.8
	127.8	127.9	136.5	71.03	71.03	71.45
	152.8	153	168.4	71.32	71.3	67.92
set 3	139.6	139.5	163.9	58.36	58.37	46.61
bubble size: 5uL and bubble location l = 1.2m	139.9	139.4	148.1	58.36	58.37	54.25
	145.1	145.2	147.9	65.43	65.48	62.47
	126.9	131	145.9	63.53	63.2	52.6
	134	134.1	135.9	56.84	56.75	53.86
	136	135.9	141.2	56.84	56.75	54.25
	126.4	126.8	139.3	70.15	70.45	64.59
	125.4	125.3	132.1	63.58	63.58	60.56
	147.9	148.2	154.1	63.58	63.58	60.17
	138.7	138.8	152	72.35	72.34	69.59
	132.2	132.7	142.6	60.14	60.45	52.77
	146.7	146.9	149.4	60.14	60.17	56.5
	140.6	141	151.6			
	128.7	130	143.2			
	152	152.4	158.8			

APPENDIX G

AUTHORS PERMISSION

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Lim, L (2006). "Reliable Invasive Blood Pressure Measurements Using Fourier Optimization Techniques." Masters Thesis, University of Akron.

Appendix F: Values of *R*, *L*, and *C* used for simulation

Regards,
Lily Lim

PS: Congrats on finishing your thesis, Darshan. Hope all is well and best of luck!
