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EVALUATION OF A SYSTEM FOR REAL-TIME MEASUREMENT OF ANESTHETIC UPTAKE

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EVALUATION OF A SYSTEM FOR REAL-TIME MEASUREMENT OF ANESTHETIC UPTAKE

by

Mohammad Jafar Navabi

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For the Degree of

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In the Graduate College

THE UNIVERSITY OF ARIZONA

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Date

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TABLE OF CONTENTS

LIST OF ILLUSTRATIONS

LIST OF TABLES

ABSTRACT

The initial work involved the analysis of sources of errors in an anesthetic uptake measurement system. After completing the error analysis appropriate measurement instruments were selected and a computer program for real-time breath to breath analysis of anesthetic gases was implemented. The analysis system was evaluated by careful bench tests which simulted zero uptake, induction, and emergence of anesthesia. The error in the measured volumes ranged from 0.28% to 8.40%. Cumulative uptake had errors in the range of 0.18% to 8.92%. The system was later revised for on-line breath-by-breath flow-meter bias adjustment. After the revisions, the error in the measured volumes was less than 3% and cumulative uptake had a worst case difference between the precalculated value and the measured value of 4.17%.

CHAPTER 1

INTRODUCTION AND BACKGROUND

In the delivery of general anesthesia the purpose of the anesthesiologist is to raise and maintain the concentration of anesthetic gases in arterial blood high enough to put the patient in a pain free state. At the same time it must be ensured that this concentration does not exceed some limit. Knowledge of anesthetic uptake can provide the anesthesiologist with the necessary information regarding the patient's status and help him/her with decision making. A faster and more precise control of anesthetic administration and a rapid detection of abnormal conditions may also be possible.

After Eger [1] suggested that anesthetic depth is correlated with alveolar concentration of anesthetic agent, it was first assumed that knowing the end tidal concentration would help the anesthesiologist in the delivery of anesthesia. This assumption arose from the fact that end-tidal concentration is related to the alveolar concentration of the anesthetic. However it was later determined that end-tidal concentration alone is not a good descriptor of anesthetic uptake because the effects

of anesthetic may vary greatly as a function of cardiac output or the functional residual capacity.

It is believed that real time anesthetic mass uptake rate and cumulative uptake may correlate closely with anesthetic depth [1]. Anesthetic uptake is currently approximated by the anesthesiologist by relying on available monitoring devices in the operating room. These devices indicate the patient's vital signs such as blood pressure, respiratory rate, heart rate, anesthetic delivery system settings, and also the patient's response to surgical stimulus.

Anesthetic uptake can be approximated by using empirical mathematical models or electrical analogs which describe anesthetic uptake. One such model, developed by Lowe and Ernst [2] is the square root of time model. Cumulative uptake in a closed circuit anesthetic delivery system was empirically found to follow the relationship:

$$C.U. = 2 * Ca * Q * (t)^{1/2} \qquad \text{(eq. 1.1)}$$

where C.U. = Cumulative Uptake

Ca = Arterial concentration

Q = Cardiac output

t = time elapsed since the start of anesthesia

and the constant 2 has units of $[(min)^{1/2}]/(conc.\ \%)$.

Empirical models provide a reasonable estimate of anesthetic uptake provided the patient is in the same

general health category as the group of people from which the model was developed. For example the square root of time model depends on many patient parameters such as heart and lung condition and amount of fat in the patient's body. Considering the general increase in average age of our population, the number of patients who do not fit the empirical model is on the rise. Therefore the need for a system that can continuously and accurately measure the uptake of anesthetic agents is clear.

To measure mass uptake and uptake rate a continuous measurement of anesthetic gas concentration and flow is needed. One such system developed by Noe et. al [2], measured expired gas flow volume, carbon dioxide excretion and oxygen consumption, continuously for two minute periods at 15 minute intervals. CO_2 concentration was measured with an infrared CO_2 analyzer, O_2 concentration with a rapid polarographic analyzer, and gas flow rate by a pneumotachograph. A general purpose digital computer was used to process the data. Values of flow volume, carbon dioxide excretion, and oxygen consumption were found for 50 normal women during abdominal surgery. The results indicated a progressive increase in O_2 uptake and a concurrent but not necessarily simultaneous decrease in CO_2 output. The correlation between rate of decrease in CO_2 excretion and minute volume was studied. It was

concluded that with on-line computation a more comprehensive and better controlled study of this relationship would be possible.

In another study conducted by Ramanthan et. al [3], oxygen consumption and carbon dioxide production during endotracheal anesthesia was measured in real time. It was assumed that the total volume of gases leaving the system was approximately equal to the volume of fresh gas entering the system. This volume was measured and multiplied by the oxygen concentration in gases leaving the system, in order to calculate the volume of oxygen leaving the system. Oxygen consumption was calculated as the difference between the oxygen volume entering and leaving the system. Production of CO_2 was calculated by multiplying the volume of gases entering the system with the expired CO_2 concentration.

Chilcoat et. al [4] developed a control system to bring the tension of anesthetic agent in the brain to a specified value determined by the anesthetist, and maintain this value until a new value was specified. The inspired concentration required to achieve the desired brain tension was calculated from a model of the patient and set automatically on the vaporizer. During the development of this system inspired and end-tidal concentration of several gases, and alveolar and total

ventilation was measured in mechanically ventilated subjects [5]. A quadrupole mass spectrometer was used for measuring gas concentrations and the mixed carbon dioxide concentration was computed as the volume weighted average of expired carbon dioxide. The system was tested on eight Alsatian dogs. After omitting results affected by avoidable errors, the standard deviation of the measured to computed arterial tension ratio was less than 10%.

More recently, Lauria [6] designed and tested a system for real-time analysis of anesthetic gases. The outputs of a gas analyzer and flow-meter were fed into a computer for analysis. The system calculated breath by breath inspired and end-tidal concentration , the ratio of inspired to end-tidal concentration , inspired and expired volume, respiratory rate, single breath uptake of anesthetic, and the square root of time of anesthesia. Measurements were done in real-time, in laboratory simulation. Due to the computer's speed limitation a 2 seconds interval was necessary between each breath for calculating the variables and outputting the results to the screen. During this interval the signals from the gas analyzer and the flow-meter were not sampled by the computer and not included in the analysis. This produced a potential cause of errors in the measured parameters, and also put a limit of 12 BPM on the maximum breathing rate

the system could handle. Cumulative uptake measured by the system during bench tests had a worst case error of 10.6% while the other measured variables had errors of less than or equal to 5.0 % .

The objective of this thesis was to determine the feasibility of measuring the uptake of anesthetic agents in real-time using state of the art instrumentation which could be used in the operating room. A further goal of this study was to implement such a system for measurement of anesthetic uptake. These objectives were accomplished by:

1) Analyzing sources of errors in a simulated measurement system and evaluating the contribution of each error to the overall error. Possible sources of errors include gas analyzer and the flow-meter inaccuracies, analog to digital conversion, data reduction, and synchronization between flow and concentration.

2) Implementing a system for anesthetic uptake measurement and evaluating its performance by means of controlled bench tests.

CHAPTER 2

MATHEMATICAL MODEL

During the induction phase of anesthesia in a circle breathing system the patient receives anesthetic gases mixed with oxygen and nitrous oxide from an anesthesia machine. Initially there is no anesthetic in the blood and some of the gases entering the lungs diffuse into the arterial blood. Diffusion occurs rapidly and normally before expiration starts, gases in the alveoli will be in equilibrium with the gas in the blood. Compared to inspired gases, the expired gases will thenhave a lower concentration of anesthetic gas (Figure 2.1).

During emergence from anesthesia, concentration of anesthetic agent in the patient is higher than that of the gas entering the lungs. Thus anesthetic gas will diffuse from the blood to the lungs and cause the expired gas to have a higher concentration of anesthetic agent than the inspired gas (Figure 2.2).

To implement the measurement system we used a commercially available flow-meter and mass spectrometer to continuously measure gas flow into and out of the patient and the concentration of anesthetic agent in the inspired and expired gases. The outputs from these instruments are

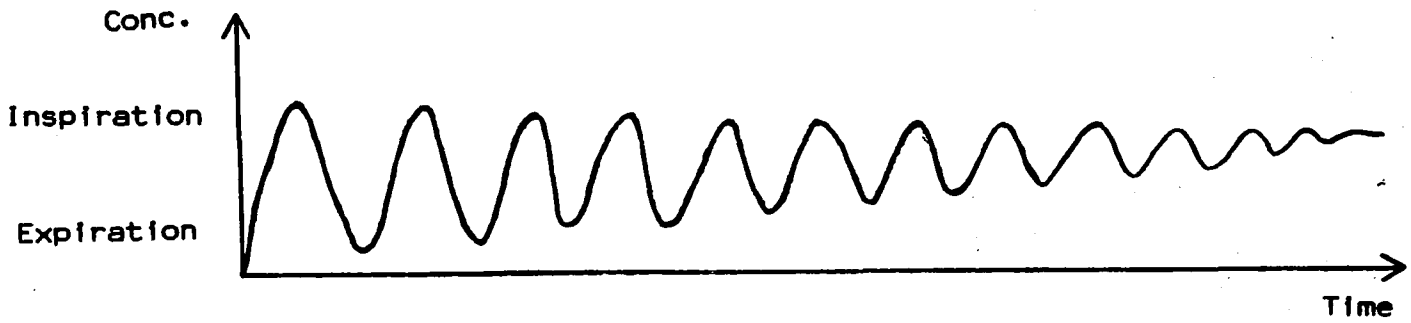

Figure 2.1. Typical changes in concentration with time during induction of anesthesia

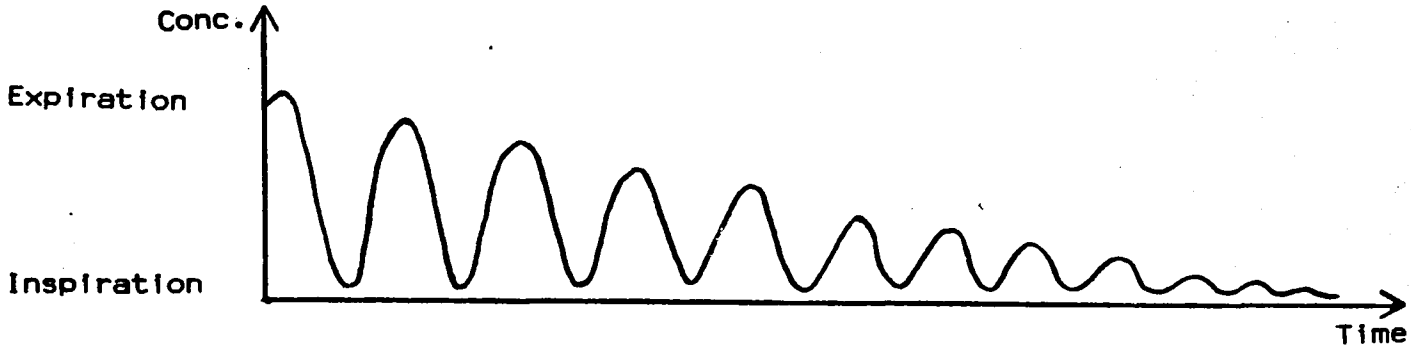

Figure 2.2. Typical changes in concentration with time during emergence from anesthesia

fed into a digital computer for analysis and calculations (Figure 2.3).

Inspired and expired volumes are measured as the time integral of the flow signal. Inspired volume is the integral of the flow wave-form from the beginning of a breath to the end of inspiration:

$$V_{INS} = \int_{INSBEG}^{INSEND} F(t)\, dt \qquad \text{(eq. 2.1)}$$

where V_{INS} = inspired volume

$F(t)$ = flow signal as a function of time

INSBEG = time corresponding to start of inspiration

INSEND = time corresponding to end of inspiration.

The expired volume is the integral of the flow over the expiration period:

$$V_{EXP} = \int_{EXPBEG}^{EXPEND} F(t)\, dt \qquad \text{(eq. 2.2)}$$

where V_{EXP} = expired volume

EXPBEG = time corresponding to start of expiration

EXPEND = time corresponding to end of expiration.

Applying the trapezoidal method of integration to equations 2.1 and 2.2 would give rise to:

$$V_{INS} = \sum_{i=INSBEG}^{INSEND} F(i) \times T \qquad \text{(eq. 2.3)}$$

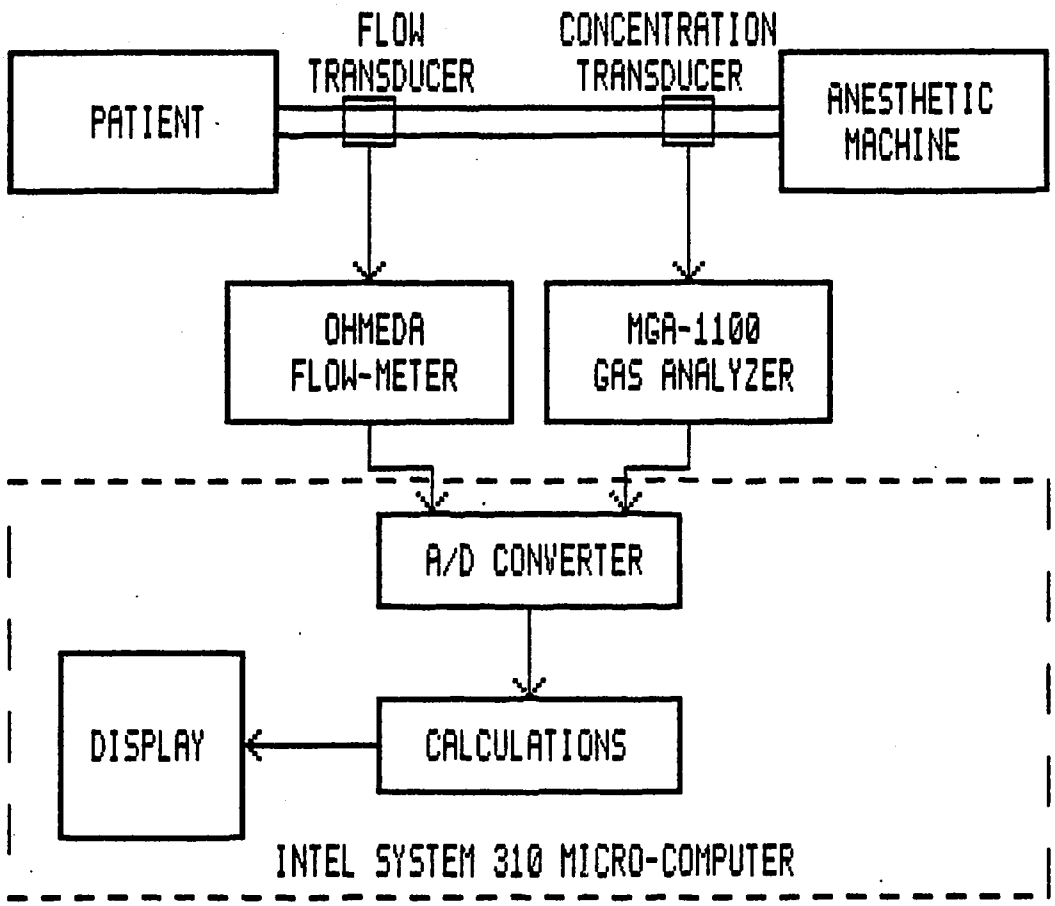

Figure 2.3. Block diagram of anesthetic monitoring system.

$$V_{EXP} = \sum_{i=EXPBEG}^{EXPEND} F(i) \times T \qquad \text{(eq. 2.4)}$$

where F(i) = the i^{th} flow sample

T = time interval between 2 consecutive samples.

Using the principle of conservation of mass, anesthetic uptake per breath is equal to the difference between inhaled and exhaled mass. Anesthetic mass is calculated by multiplying the volume of gas by the volume percent of anesthetic. Uptake per breath is the sum of inspired and expired uptakes where the expired uptake is negative.

$$U = \sum_{i=INSBEG}^{INSEND} F(i)\times C(i)\times T + \sum_{i=EXPBEG}^{EXPEND} F(i)\times C(i)\times T \qquad \text{(eq. 2.5)}$$

where U = Uptake per breath

C(i) = The i^{th} concentration sample.

Since the end of inspiration is the same as the start of expiration (INSEND=EXPBEG) and equation 2.5 can be simplified to:

$$U = \sum_{i=INSBEG}^{EXPEND} F(i) \times C(i) \times T \qquad \text{(eq. 2.6)}$$

In the above equation the flow sample must be multiplied by the corresponding concentration sample. However the gas analyzer output has an inherent time delay which must be taken into account in the multiplication as per equation 2.7:

$$U = \sum_{i=INSBEG}^{EXPEND} F(i) \times C(i-id) \times T \qquad \text{(eq. 2.7)}$$

where id = the number of samples corresponding to output delay in gas analyzer.

Figure 2.4 shows typical flow and concentration waveforms and illustrates the concept of time delay.

The cumulative anesthetic uptake is the sum of individual breath uptakes over time.

$$C.U. = \sum_{b=FIRST}^{CURRENT} U(breath) \qquad \text{(eq. 2.8)}$$

where C.U. = Cumulative Uptake

FIRST = first breath

CURRENT = current breath.

Respiratory rate (RR) is calculated as the inverse of time per breath:

$$RR = 1 / (EXPEND - INSBEG) \qquad \text{(eq. 2.9)}$$

The last inspired and expired concentrations are used to represent the inspired (C_{INS}) and end-tidal (C_{ET}) concentrations respectively:

$$C_{INS} = C(i) \text{ at } i = INSEND \qquad \text{(eq. 2.10)}$$

$$C_{ET} = C(i) \text{ at } i = EXPEND \qquad \text{(eq. 2.11)}$$

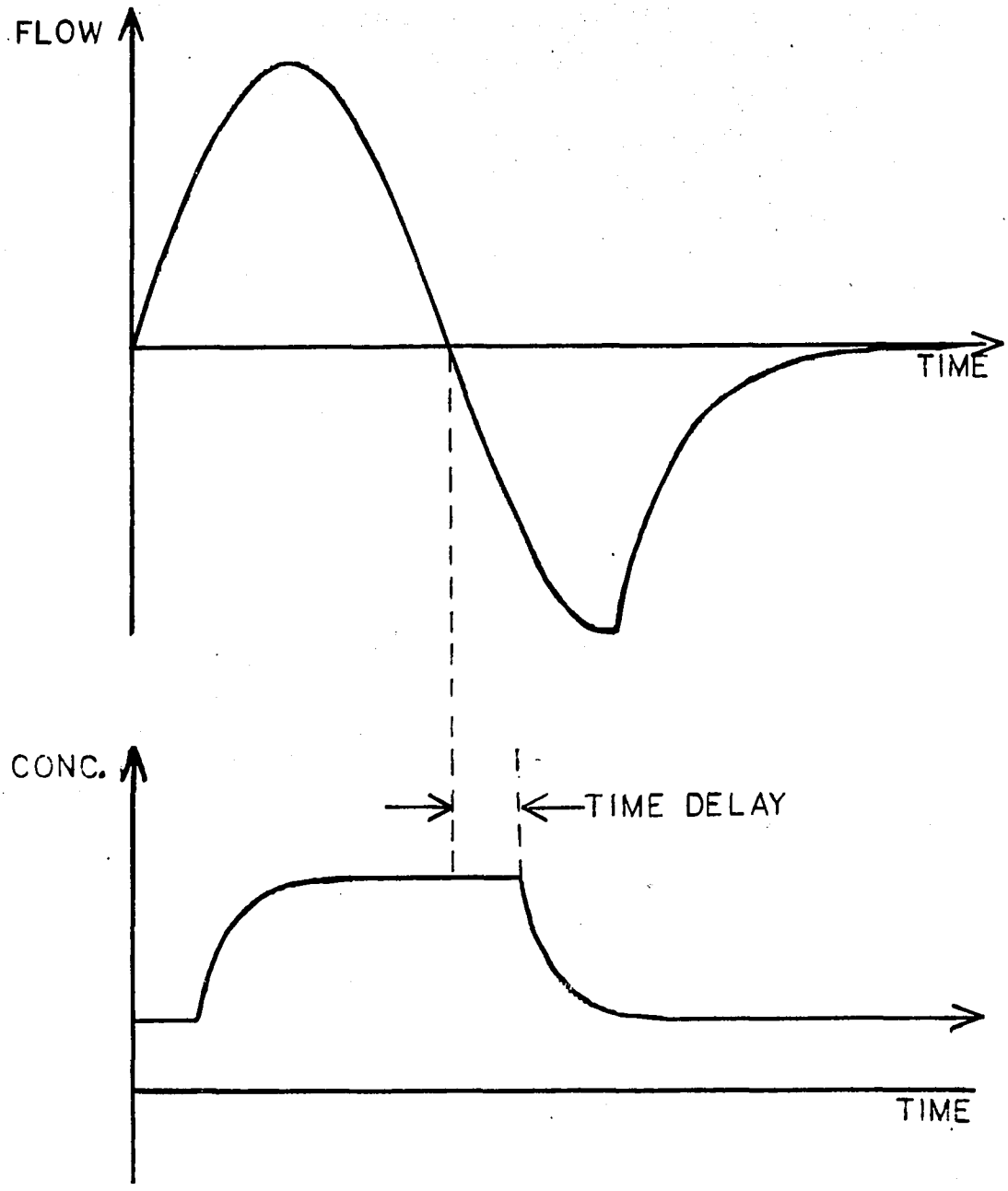

Figure 2.4. Typical concentration and flow waveforms illustrating time delay in concentration signal.

CHAPTER 3

ERROR ANALYSIS

The purpose of the error analysis was to investigate various sources of error in the measurement system and determine the potential of state of the art instrumentation to provide single breath and cumulative uptake measurements to within ±5%. An analysis of errors was conducted using computer simulation of the measurement system including the errors associated with each component of the system. Errors in the measurement of breath by breath uptake were determined by using typical flow and concentration waveforms as the input to the simulation program. In the computer simulation the following sources of error were considered in the calculation of worst case error in breath by breath uptake measurements:

- Concentration accuracy and drift.
- Flow measurement accuracy and drift.
- Timing accuracy of the computer clock.
- Digitizing flow and concentration signals.
- Selecting inspiration and expiration times.
- Digitizing integration techniques.

Estimates were made for the expected errors from the computer and instrumentation in use and the errored

single breath uptake was calculated. Error free single breath uptake was also calculated by exact integration of the product of flow and concentration waveforms over the period of a breath. Computer calculations were then conducted to determine the overall measurement error. This was calculated as the difference between the errored and the error free signal. Contribution of each of the sources of error in the overall measurement was also calculated. The estimated value for each of the sources of error was varied and the change in the calculated single breath uptake as a function of this variation was observed.

Figure 3.1 shows the flow waveform used in the error analysis. A sinusoidal shape was assumed for the inspiratory and half of the expiratory signal. A decaying exponential was used for the second half of the expiration signal. Peak flow of 524 ml/sec was produced by taking 500 mL as a typical tidal volume with an inspiration time of 1.5 seconds. The expired waveform had the same peak value, and the time constant for the exponential portion was adjusted so that the inspired and expired volumes were equal. The concentration wave-form of figure 3.2 was assumed with maximum inspiration and expiration concentration levels of 4% and 2% respectively.

The results showed that digital integration techniques introduced less than 0.02% error in the

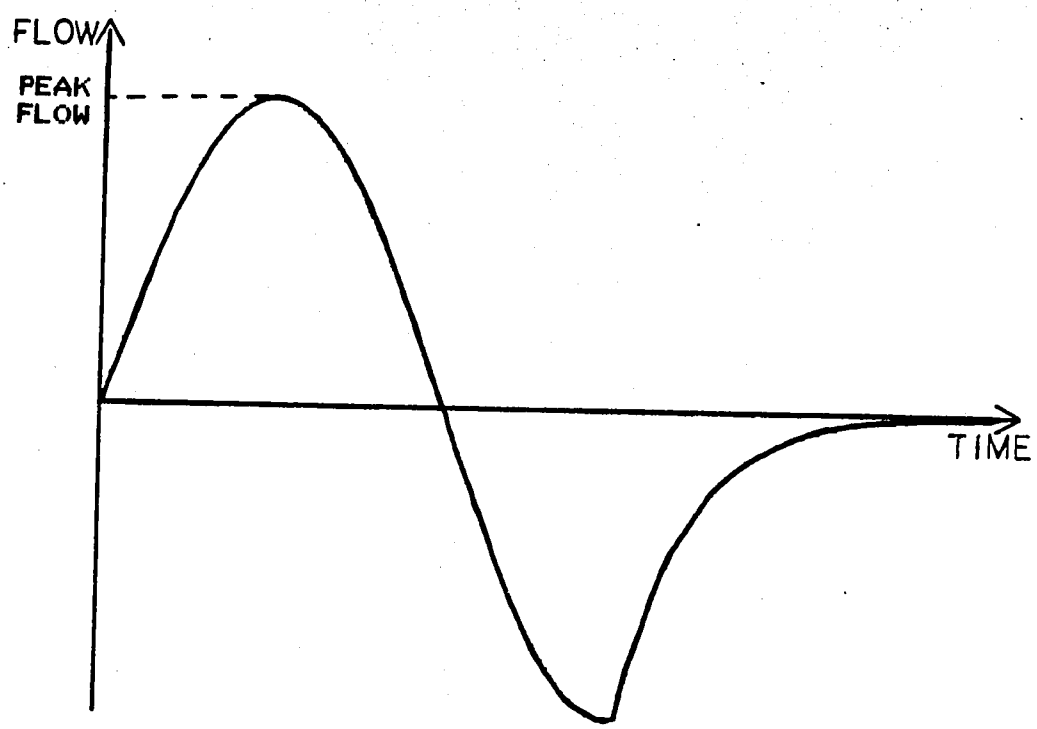

Figure 3.1. Assumed flow waveform for error analysis.

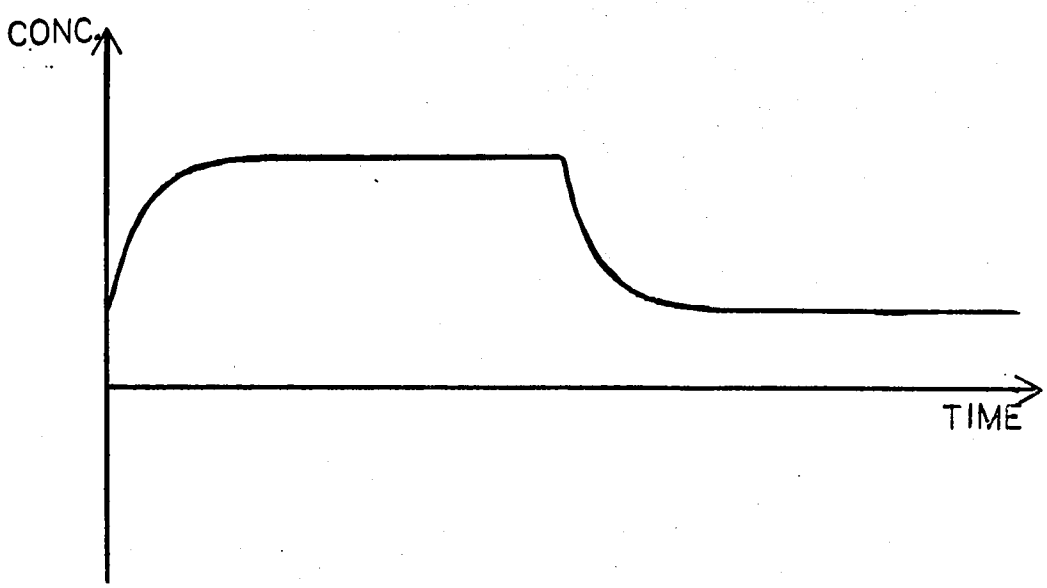

Figure 3.2. Assumed concentration waveform for error analysis.

measured single breath uptake. The error due to timing of computer system clock will contribute to the overall error directly meaning that 1% error in the timing will produce 1% error in the uptake. However since the computer timing was known to within ±0.02%, this error was insignificant. Determination of the start of inspiration and expiration together with digitizing flow and concentration signals produced 0.13% error. Concentration and flow inaccuracies were directly reflected in the single breath uptake measurement results. Expected errors in the concentration and flow are ±1% and ±5% respectively. Using all of the above assumptions, a total of almost ±7% was observed in the measurement.

Primary contributions to the overall error were from the flow and concentration measuring instruments. Flow-meter bias drift which was not considered in the above analysis, can cause considerable error. A bias variation of +1 LPM resulted in almost 13% mismatch between inspired and expired volumes and from 6.5% to 13% error in the measured single breath uptake depending on the inspired and expired concentration of anesthetic gas. Therefore in order to minimize the error in measured uptake, it is necessary to use a flow-meter and a gas analyzer with very low error and bias drift or in some way compensate for the instrument errors.

CHAPTER 4

IMPLEMENTATION

A system for on-line measurement of anesthetic uptake was implemented. The hardware and software used in this system are described here.

Hardware Description

A Perkin Elmer's medical gas analyzer (MGA 1100) was used to continuously measure concentration of anesthetic agent. The MGA 1100 has an accuracy of ±1% of full scale for nitrogen and oxygen and ±2% for for CO_2, nitrous oxide and halothane, a stability of ±2% per month, and a linearity of ±2% of full scale over concentration. The gas analyzer was calibrated as outlined in the MGA-1100 Operations And Maintenance Manual.

The gas analyzer draws 60 ml/sec of gas sample through an eight foot sampling tube which creates a fixed delay in the concentration reading. To measure this time delay, a balloon filled with pure oxygen was connected to the transducer of a HP 780-9 pressure monitor causing pressure build-up on the transducer. By breaking the balloon with a needle on the end of the sampling tube, the pressure on the transducer is released instantaneously and at the same time the gas analyzer starts sampling the gas

which was inside the balloon and after the time delay its output will indicate a change in oxygen concentration from 20% to almost 100% . The difference between the time when the pressure at the transducer drops and the output of the gas analyzer rises, is the time delay. This value was read on the oscilloscope to be 410 ±5 msec.

For flow measurement a pulsed phase measurement ultrasonic flow-meter developed by Ohmeda Corporation was used. To calibrate the flow-meter it was necessary to compare the flow reading of the Ohmeda flow-meter with flow measured by a more accurate method. A calibrated Collins 9 liter bell spirometer was used for this purpose. A constant flow of oxygen was passed through the flowhead of the Ohmeda flow-meter and into the spirometer. Flow was calculated as the rate of change of volume of gases in the spirometer with respect to time. The spirometer's chart recorder was used to keep track of the time and volume changes.

Because of drift in zero bias of the Ohmeda flow-meter, values were obtained before each calibration measurement and subtracted from the readings. This allowed accuracy and drift to be accounted for separately. 25 calibration points were obtained and average values of calibration factors were calculated for positive and negative flows. An average calibration factor of 87.92

Liters/Volt with a standard error of 0.086 LPM was found for the positive flows while those of the negative flows were found to be 94.42 Liters/Volt and 0.107 LPM correspondingly. Figure 4.1 shows the flow reading of Ohmeda flow-meter in Volts, vs. actual flow in LPM. The 25 data points are also illustrated in this figure. To account for flow-meter's zero bias the value of this bias must be known and subtracted from the flow-meter's readings. The actual flow will therefore be calculated by:

$$\text{Flow} = 87.92 \times V_{out} - \text{Bias} \quad \text{if Vout}>0 \qquad \text{(eq. 4.1a)}$$

$$\text{Flow} = 94.42 \times V_{out} - \text{Bias} \quad \text{if Vout}<0 \qquad \text{(eq. 4.1b)}$$

where V_{out} = Output voltage of Ohmeda flow-meter

Flow = Actual flow through flow-meter's flowhead.

An Intel's system 310 micro-computer with the iSBX-311 analog input multimodule board installed, was used for data acquisition and processing. The iSBX-311 is a 16 channel, 12 bits analog to digital converter, specified to have an accuracy of ±0.035% of the full scale and ±1/2 of the least significant bit. Typical conversion speed is 50 sec.

Software Description

Source code for the program was written in PL/M 86 programming language. By using multi-tasking features of the iRMX 86 operating system it was possible to assign a high priority task to sampling the two channels and a

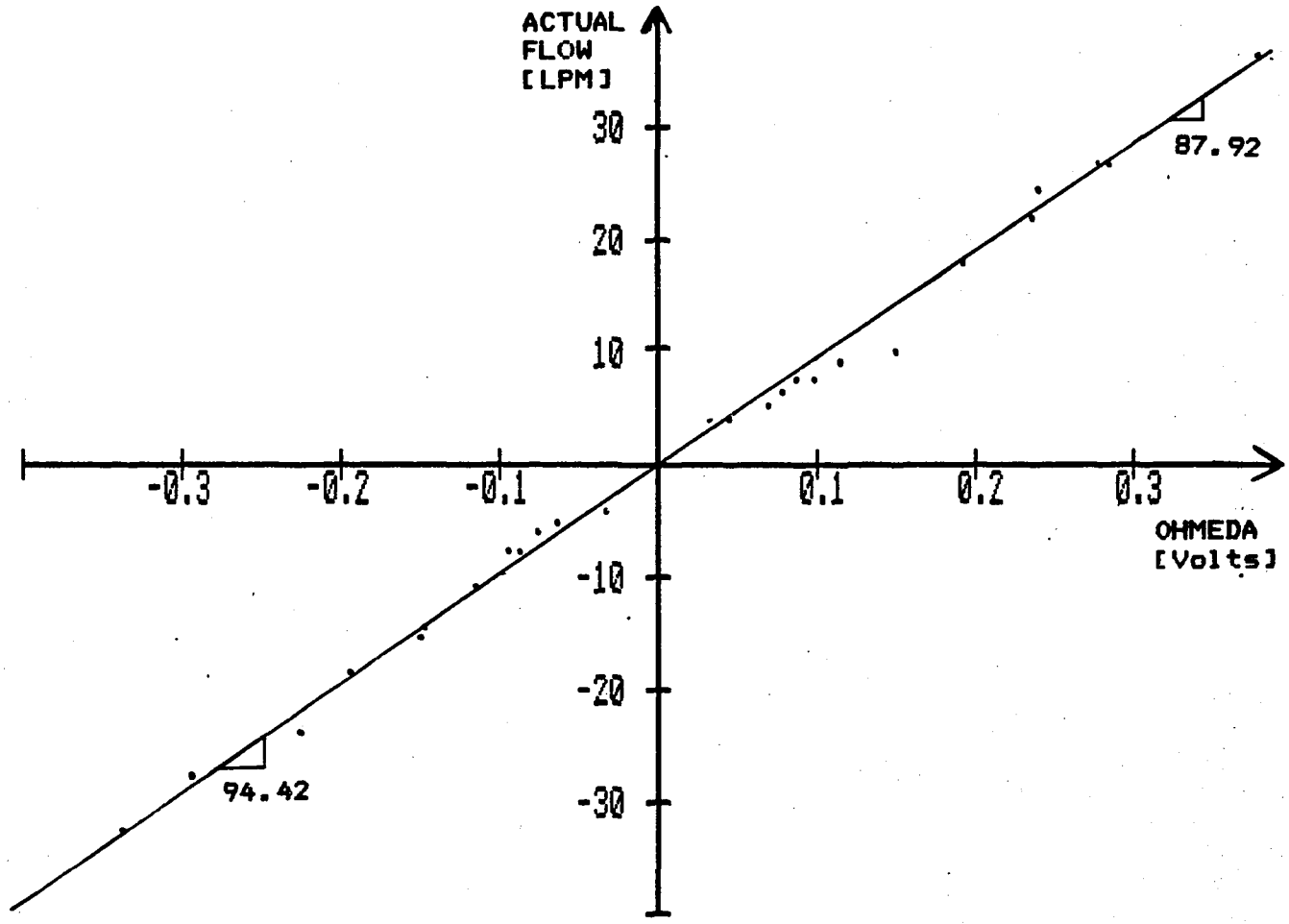

Figure 4.1. Flow-meter calibration data.

lower priority task to calculating and displaying the results. In this way continuous sampling is achieved and calculation and screen update are accomplished between samples. The program starts by sampling flow waveform until it detects the start of a breath. To avoid erroneous detection of noise as the start of a breath, a flow threshold of 5 LPM was used. In other words the flow-meter's output signal must reach +5 LPM before the program detects the start of a breath. Once a breath is detected both flow and concentration are sampled until a downward crossing of the -5 LPM flow is detected. This signifies the end of inspiration and start of expiration. Again both signals are sampled until the start of next breath. In calculating uptake, the program takes into account the time delay between the gas analyzer's and the flow-meter's output by using equation 2.7.

The computer was programmed to sample each channel at a rate of 100 Hz. A sampling rate of 100 Hz was chosen in order to satisfy the Nyquist criterion which states that the sampling rate must be at least twice the highest frequency component of the sampled signal. According to Webster [7] respiratory measurements have a frequency range from 0 to 40 Hz.

The program prompts the user for the time delay in the gas analyzer, and also for the flow-meter's

calibration information. Program outputs include breath to breath results of inspired and expired volume, inspired and end-tidal concentration of anesthetic, single breath uptake and cumulative uptake of anesthetic, elapsed time since the start of the program, and breathing rate. These are calculated as explained previously.

Flow-meter bias was calculated by using the values measured for inspired and expired volumes during the first breath. Assuming that inspired and expired volumes should be equal, the flow-meter's bias is the value which when subtracted from flow will equate the inspired and expired volumes.

CHAPTER 5

SYSTEM EVALUATION

The overall system described in the last section was evaluated by two procedures. The first procedure was designed to test the computer's ability to sample the two A/D channels correctly and perform the calculations properly. In this part the computer's timing was also tested. The second procedure was designed to test the overall system. This was accomplished by using an uptake simulation setup and comparing the expected uptake with the uptake measured by the system.

Computer's Software/Hardware Tests

A Tektronix DC504 timer was used to measure the frequency of the computer clock. The test was repeated 8 times over the period of three days and the results showed a consistent frequency of 153.66KHz ± 1%.

The analog to digital (A/D) sampling hardware and software were calibrated by applying a known D.C. voltage (measured by Tektronix 501 DMM) to the sampled channel and checking the digitized values. Readings were accurate plus or minus one least significant bit and showed a consistent bias of less than 2.0 mVolt which was taken into account in flow and concentration conversions.

A feature of the software which required careful testing was the detection of the beginning of inspiration and expiration. To test this feature a D.C. voltage was applied to the flow channel during sampling and the voltage was slowly increased. The voltage was constantly monitored using a Tektronix 502 DVM. As the voltage passed 0.05V (corresponding to 5 LPM). the program indicated detection of inspiration. When the flow was reduced the sampling of the inspiration continued until the voltage reached -0.05V (corresponding to -5 LPM). At this time the program indicated the detection of expiration.

Overall System Evaluation

The system was tested in the laboratory for three simulated situations of emergence, induction, and zero uptake of anesthetic. The system's ability to accurately determine inspired and expired volumes, breathing rate, and breath to breath and cumulative anesthetic uptake was evaluated by comparing the measurement system results with the expected values from our simulation test setup. In the case of emergence and induction the test setup consisted of a Harvard respiration pump, an anesthetic machine with a Ohio 5000 vaporizer, and a calibrated 9-liter Collins spirometer with a mixing fan mounted inside the chamber (Figure 5.1). The spirometer was filled with approximately 8.0 Liters of oxygen and forane. The forane concentration

was measured by channel 1 of the gas analyzer and arbitrarily adjusted to a level between 1 to 5%. The flow-meter and channel 2 of the gas analyzer were then connected between the spirometer and the ventilator to monitor flows and concentration of gas moving into and out of the spirometer. The ventilator pumped fresh gas (no anesthetic) into the spirometer which then pushed the same volume of mixed gas out. With each movement of gas into and out of the spirometer ("each breath"), the anesthetic concentration was reduced and eventually reached near zero.

During the induction of anesthesia the inspired concentration of anesthetic gases are higher than that of the expired gases. As the induction continues the inspiratory and expiratory concentrations approach an equilibrium. To simulate this situation in the setup of figure 5.1, the flow-meter was installed so that flow going into the spirometer (fresh gas) was considered an expiration and flow out of the spirometer (mixed with anesthetic) was considered an inspiration.

For emergence, the concentration of inspired gases is less than the anesthetic concentration of expired gases. To simulate this condition the flow-meter was reversed so that fresh gases into the spirometer

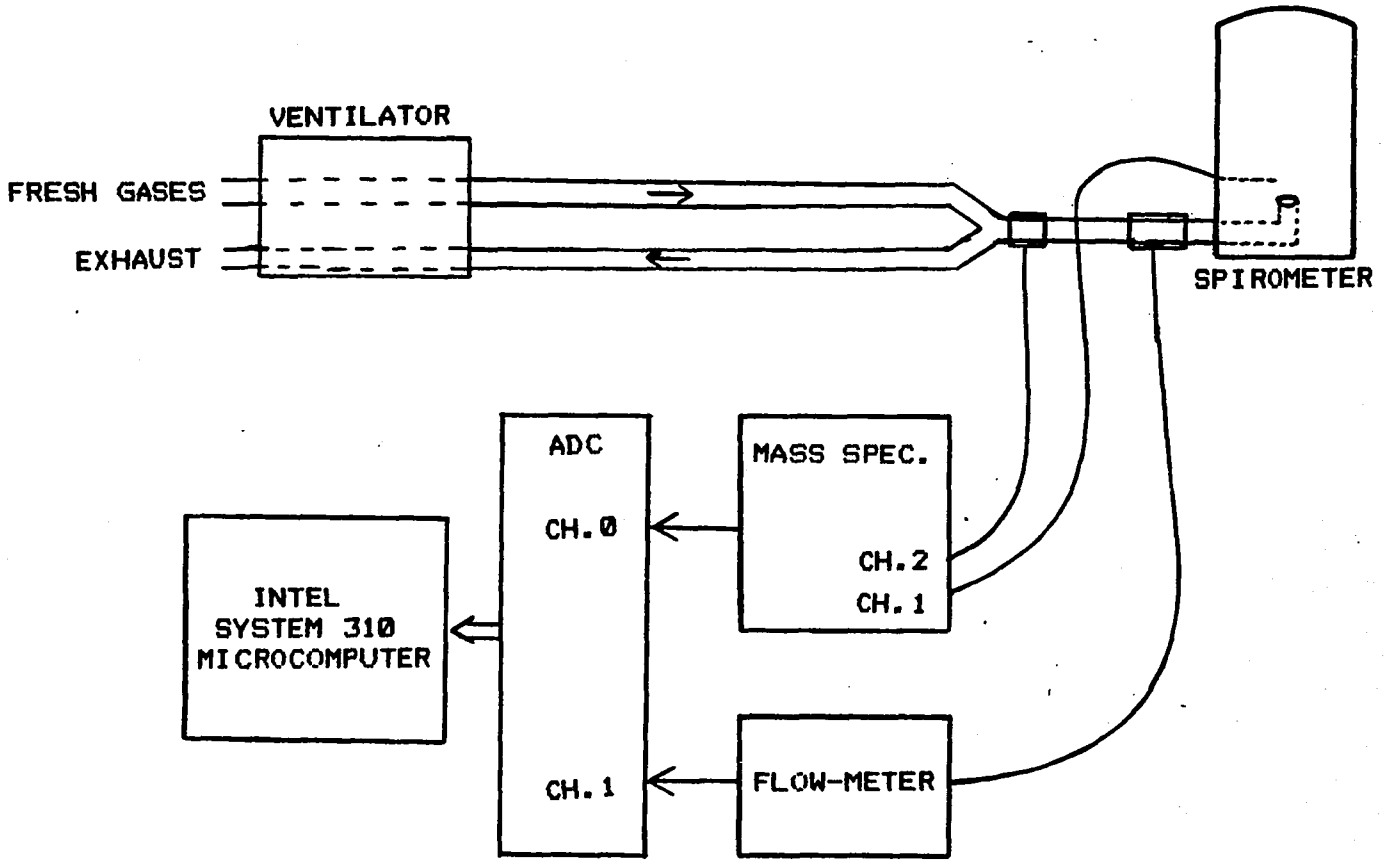

Figure 5.1. Test system used for induction and emergence simulation.

represented an inspiration and anesthetic containing gases out of the spirometer represented an expiration.

To simulate zero uptake a slightly different test set up was used. Figure 5.2 shows the set up for this test. A Fraser Harlake model 701 ventilator was used to pump oxygen with anesthetic into and out of the spirometer. The flow-meter and port 2 of the gas analyzer were used to monitor flow and concentration as in the previous tests. Since the same gases were being cycled back and forth there was no change in the concentration which simulates zero uptake.

Expected values for concentration changes and uptake were calculated from the original anesthetic concentration in the spirometer and volume changes with each breath. Dead space mixing was minimized and (as will demonstrated later) included in the calculations for expected results. Anesthetic uptake by the water in the spirometer was measured in separate tests. It was found that the absorption of the anesthetic gas in water reaches an equilibrium after about 1 minute (figure 5.3).

Expected breath by breath uptake/loss of anesthetic by the spirometer was calculated as the increase/decrease in concentration of anesthetics times the volume of gases inside the spirometer:

$$U(i) = V_{sp} \left(C_{sp}(i) - C_{sp}(i-1) \right) \qquad \text{(eq. 5.1)}$$

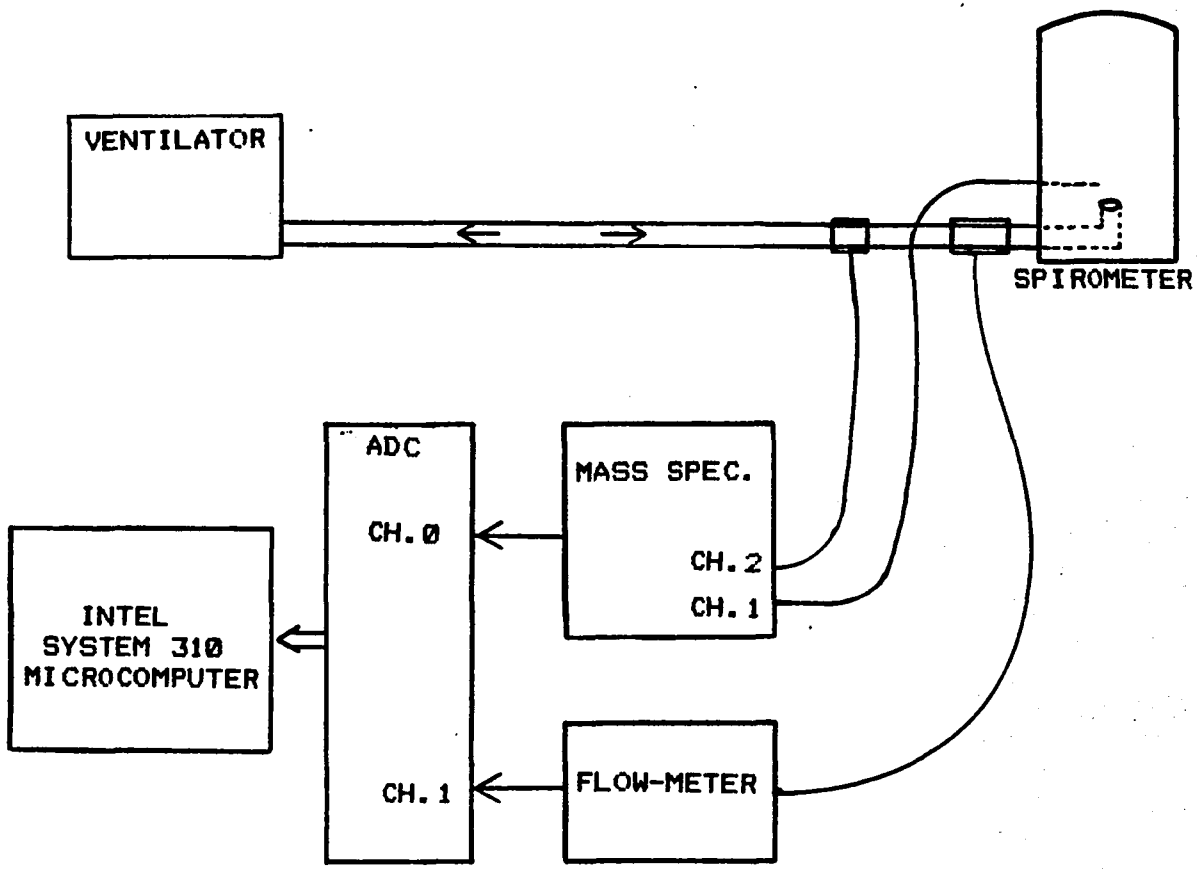

Figure 5.2. Test system used for zero-uptake simulation.

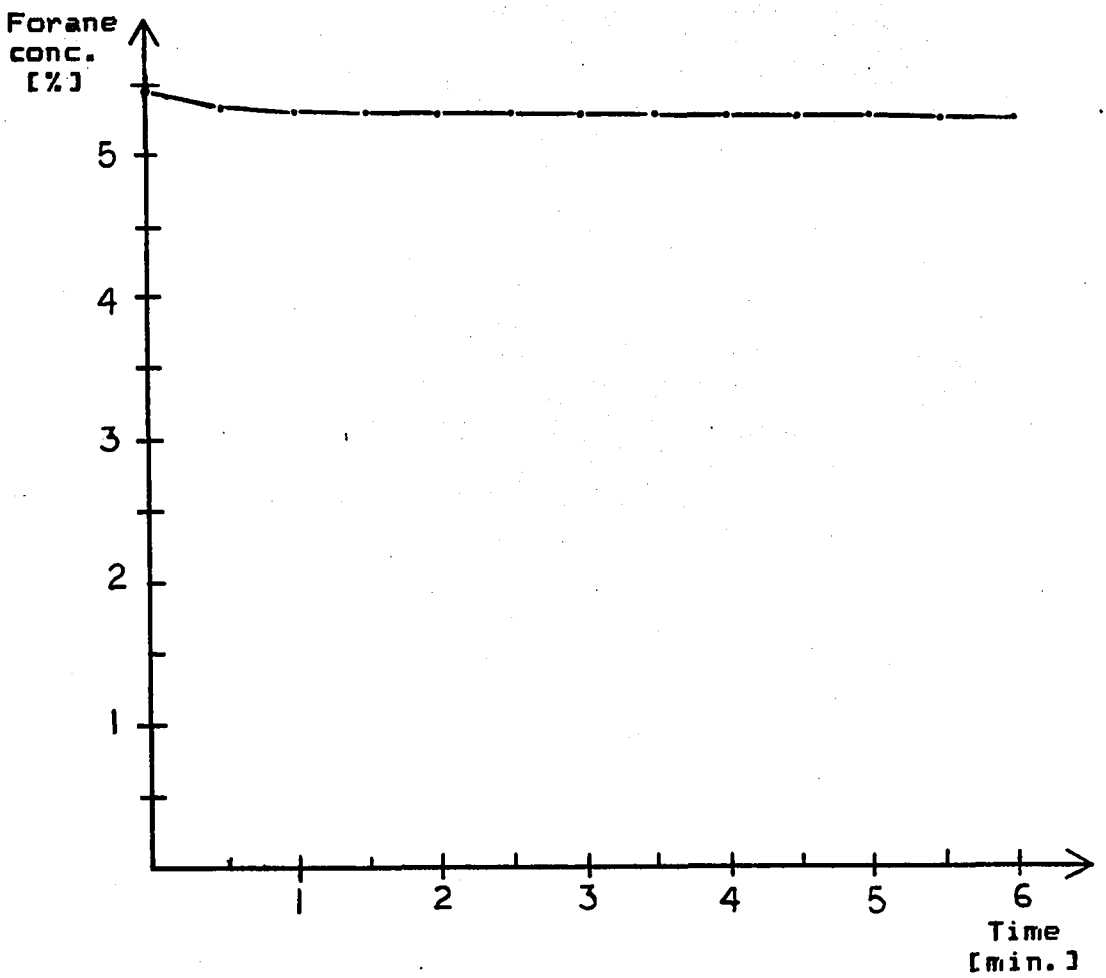

Figure 5.3. Uptake of anesthetic by the spirometer water.

where U(i) = uptake per breath

$C_{sp}(i)$ = current anesthetic concentration

$C_{sp}(i-1)$ = concentration before current breath

V_{sp} = volume of the spirometer

V_{sp} was measured prior to the test and was found to be 7785±78 c.c. $C_{sp}(i)$ can be calculated by:

$$C_{sp}(i) = \frac{(dV-V_{ds})\times C_{in} + (V_{sp}+V_{ds})\times C_{sp}(i-1)}{V_{sp} + dV} \quad \text{(eq. 5.2)}$$

where dV = volume of gases entering and leaving spirometer

C_{in} = concentration of gases entering spirometer

V_{ds} = dead space volume

The inspired and expired volumes (dV) for each test was measured by reading the volume on the spirometer's chart. This volume had been calibrated using a 3.0 liter calibrating syringe prior to the tests. Concentration of gases into the spirometer (C_{in}) is equal to zero for the induction and emergence tests. V_{ds} represents the volume of tubings where no mixing of the inspired gases occur. This is the volume between the gas analyzer sampling channel 2, and the mixing chamber of the spirometer. This volume was measured and found to be 125±2 c.c.

Notice that by using equations 5.1 and 5.2 and by knowing the initial concentration of anesthetics inside the spirometer and the concentration of gases entering the

spirometer, the expected breath by breath uptake can be calculated:

$$C_{sp}(0) = \text{initial concentration} \qquad \text{(eq. 5.3)}$$

$$C_{sp}(1) = \frac{dV \times C_{in} + V_{sp} \times C_{sp}(0)}{V_{sp} + dV} \qquad \text{(eq. 5.4)}$$

$$U(1) = V_{sp} \left(C_{sp}(1) - C_{sp}(0) \right) \qquad \text{(eq. 5.5)}$$

$$C_{sp}(2) = \frac{dV \times C_{in} + V_{sp} \times C_{sp}(1)}{V_{sp} + dV} \qquad \text{(eq. 5.6)}$$

$$U(2) = V_{sp} \left(C_{sp}(2) - C_{sp}(1) \right) \qquad \text{(eq. 5.7)}$$

During emergence tests, the spirometer simulates the patient and the anesthetic uptake is equal to the uptake by the spirometer (eq. 5.1). During induction tests, the spirometer simulates the anesthetic delivery system and uptake of anesthetic by patient is negative that of the spirometer. For the case of zero uptake, use of equations 5.1 and 5.2 is not necessary because constant concentration and zero uptake is expected.

10 runs of 40 "breaths" were performed for each of induction, emergence, and zero uptake. The tests were performed over a period of 3 weeks. Comparisons were made

between expected and computer generated results. For the cases of induction and emergence, correlation between predicted uptake and uptake measured by the system was calculated using linear regression analysis. In the case of zero uptake, standard deviation about the predicted zero uptake was calculated.

CHAPTER 6

RESULTS

Tables 6.1a and 6.1b show a typical breath by breath results of inspired and expired volumes, concentrations, breathing rate, and uptake measurements during emergence tests. Included in these tables are the corresponding errors between the measurement and its expected value. The expected value for the volumes is that read from the spirometer's chart during the particular test. The expected value of the breathing rate is calculated during each test by reading the time for a few breaths on the time scale of the spirometer (previously calibrated) and dividing the number of breaths by time. In the case of concentrations and uptake the expected values are calculated using equations 5.1 and 5.3 explained in the previous section.

Overall test results for the case of emergence are included in tables 6.2a and 6.2b. Table 6.2a includes mean measured inspired and expired volumes and their standard deviation plus the error between the mean and expected values. Table 6.2b includes the mean measured breathing rate and the cumulative uptake and the percent difference

TABLE 6.1 A- Breath to breath results for emergence trial #8.

BREATH #	Vins [L]	%Vins ERROR	Vexp [L]	%Vexp ERROR	B.R. [BPM]	%B.R. ERROR
1	0.77	0.13	-.75	2.72	7.0	3.12
2	0.76	1.43	-.73	5.32	7.0	3.71
3	0.77	0.13	-.72	6.61	7.0	3.26
4	0.77	0.13	-.73	5.32	7.0	3.41
5	0.78	1.17	-.73	5.32	7.0	3.12
6	0.79	2.46	-.73	5.32	7.2	6.82
7	0.79	2.46	-.71	7.91	6.7	0.74
8	0.79	2.46	-.72	6.61	6.9	2.97
9	0.79	2.46	-.73	5.32	6.9	2.37
10	0.79	2.46	-.75	2.72	6.9	2.52
11	0.80	3.76	-.75	2.72	6.9	2.23
12	0.80	3.76	-.78	1.17	6.9	1.93
13	0.80	3.76	-.76	1.43	6.9	1.93
14	0.79	2.46	-.73	5.32	6.9	1.93
15	0.79	2.46	-.76	1.43	6.9	1.93
16	0.80	3.76	-.77	0.13	6.8	0.89
17	0.81	5.06	-.74	4.02	6.8	1.48
18	0.81	5.06	-.76	1.43	6.8	0.89
19	0.81	5.06	-.74	4.02	6.8	1.48
20	0.80	3.76	-.72	6.61	6.8	0.89
21	0.81	5.06	-.71	7.91	6.9	1.93
22	0.82	6.36	-.72	6.61	6.7	0.00
23	0.81	5.06	-.72	6.61	6.8	0.45
24	0.81	5.06	-.75	2.72	6.8	1.48
25	0.82	6.36	-.75	2.72	6.7	0.45
26	0.82	6.36	-.74	4.02	6.7	0.00
27	0.81	5.06	-.74	4.02	6.8	0.45
28	0.81	5.06	-.76	1.43	6.8	0.45
29	0.81	5.06	-.77	0.13	6.8	0.45
30	0.81	5.06	-.76	1.43	6.7	0.00
31	0.81	5.06	-.76	1.43	6.7	0.00
32	0.81	5.06	-.77	0.13	6.7	0.89
33	0.81	5.06	-.76	1.43	6.7	0.45
34	0.81	5.06	-.77	0.13	6.7	0.45
35	0.81	5.06	-.79	2.46	6.8	1.48
36	0.83	7.65	-.76	1.43	6.6	1.78
EXPECTED	.77		0.77		6.7	

TABLE 6.1 B- Breath to breath results for emergence trial #8.

BREATH #	Cins measured [%]	Cet measured [%]	Csp calc. [%]	Uptake measured [c.c. Vapor]	Uptake calc.	%Uptake error
1	4.32	0.04	4.03	-25.80	-24.52	5.21
2	4.04	0.04	3.73	-23.00	-22.74	1.13
3	3.70	0.04	3.46	-20.80	-21.09	1.39
4	3.42	0.04	3.21	-19.90	-19.56	1.73
5	3.20	0.03	2.98	-18.10	-18.14	0.23
6	3.06	0.03	2.76	-16.60	-16.82	1.34
7	2.78	0.03	2.56	-14.70	-15.60	5.79
8	2.54	0.03	2.38	-14.00	-14.47	3.26
9	2.42	0.03	2.20	-13.00	-13.42	3.14
10	2.25	0.03	2.04	-12.50	-12.45	0.43
11	2.06	0.02	1.89	-11.50	-11.54	0.38
12	1.92	0.03	1.76	-11.20	-10.71	4.62
13	1.79	0.02	1.63	-10.00	-9.93	0.72
14	1.67	0.03	1.51	-9.00	-9.21	2.26
15	1.55	0.02	1.40	-8.60	-8.54	0.71
16	1.44	0.02	1.30	-7.90	-7.92	0.25
17	1.31	0.02	1.21	-7.10	-7.34	3.33
18	1.22	0.01	1.12	-6.60	-6.81	3.11
19	1.15	0.02	1.04	-6.10	-6.32	3.44
20	1.06	0.01	0.96	-5.40	-5.86	7.83
21	1.00	0.02	0.89	-5.00	-5.43	7.98
22	0.93	0.02	0.83	-4.60	-5.04	8.72
23	0.86	0.01	0.77	-4.30	-4.67	7.99
24	0.81	0.01	0.71	-4.20	-4.33	3.10
25	0.75	0.01	0.66	-3.80	-4.02	5.47
26	0.69	0.01	0.61	-3.60	-3.73	3.43
27	0.64	0.02	0.57	-3.40	-3.46	1.66
28	0.60	0.01	0.53	-3.20	-3.21	0.20
29	0.55	0.01	0.49	-3.00	-2.97	0.88
30	0.52	0.01	0.45	-2.70	-2.76	2.10
31	0.48	0.01	0.42	-2.60	-2.56	1.65
32	0.44	0.01	0.39	-2.40	-2.37	1.18
33	0.41	0.01	0.36	-2.20	-2.20	0.00
34	0.39	0.01	0.33	-2.10	-2.04	2.93
35	0.36	0.00	0.31	-2.00	-1.89	5.70
36	0.34	0.01	0.29	-1.80	-1.75	2.57

Table 6.2.A- Results of laboratory tests of the measurement system during simulated anesthetic emergence.

TRIAL #	EXPECTED VOLUME [L]	MEAN Vins [L]	S. D. Vins [L]	%Vins ERROR	MEAN Vexp [L]	S. D. Vexp [L]	%Vexp ERROR
1	0.77	0.82	0.0917	6.13	-.77	0.0980	0.39
2	0.77	0.75	0.1261	2.22	-.77	0.1433	0.30
3	0.77	0.75	0.1214	2.90	-.77	0.0639	0.37
4	0.75	0.76	0.0047	0.71	-.72	0.0218	3.76
5	0.75	0.75	0.0408	0.11	-.72	0.0586	4.19
6	0.75	0.73	0.0686	2.48	-.73	0.0050	2.00
7	0.77	0.82	0.0003	6.53	-.75	0.0267	3.03
8	0.77	0.80	0.0106	3.83	-.75	0.0256	3.30
9	0.75	0.71	0.0086	5.15	-.74	0.0811	1.19
10	0.75	0.74	0.0147	1.96	-.75	0.0558	0.56

Table 6.2.B- Results of laboratory tests of the measurement system during simulated anesthetic emergence.

TRIAL #	EXPECTED B.R. [BPM]	MEAN B.R. [BPM]	S.D. B.R. [BPM]	%B.R. ERROR	CALC-ULATED UPTAKE [c.c. Vapor]	MEAS-URED UPTAKE [c.c. Vapor]	CORR-ECTED UPTAKE [c.c. Vapor]	%UPTAKE ERROR
1	5.2	5.02	0.2231	2.81	-327.42	-315.70	-324.08	1.02
2	5.7	5.83	0.2617	1.72	-39.25	-39.90	-39.00	0.64
3	6.2	5.99	0.2511	2.58	-100.30	-105.20	-102.56	2.25
4	6.2	6.22	0.4350	1.06	-234.90	-224.10	-230.19	2.01
5	5.3	5.50	0.5853	3.28	-272.21	-273.60	-279.02	2.50
6	6.5	6.59	0.1750	1.31	-129.97	-122.70	-122.73	5.57
7	6.7	6.74	0.1511	0.02	-342.96	-326.60	-345.71	0.80
8	6.7	6.83	0.2386	1.36	-315.45	-312.70	-327.33	3.77
9	6.3	6.28	0.0739	0.03	-234.67	-233.30	-227.97	2.86
10	7.0	6.88	0.0292	1.09	-277.99	-270.60	-268.90	3.27

between the measured and expected values. These tables include the results of all of the performed emergence tests.

Typical breath to breath results of induction tests are shown in tables 6.3a and 6.3b and the overall results are shown in tables 6.4a and 6.4b. All the parameters and their corresponding calculations are the same as with emergence. Breath by breath uptake decreases during each breath for induction. An example of the change over one induction test is shown in figure 6.1.

Tables 6.5a and 6.5b demonstrate a typical breath to breath result of zero uptake test. Tables 6.6a and 6.6b demonstrate the overall results of the zero uptake tests.

Inspection of the data revealed that zero bias drifts in the flow-meter produced a mismatch in inspired and expired volumes. To correct the errors caused by this bias variation, a bias correction parameter was calculated for each breath and subtracted from the measured uptake for that breath. The corrected results for the cases of induction, emergence and, zero uptake are included in tables 6.2, 6.4, and 6.6 respectively.

Cumulative uptake (predicted vs. corrected) agreed closely. Regression analysis performed on all 20 induction and emergence uptake data (10 induction and 10 emergence) produced a least squares fit line with a slope of 1.003

TABLE 6.3 A- Breath to breath results for induction trial #2.

BREATH #	Vins [L]	%Vins ERROR	Vexp [L]	%Vexp ERROR	B.R. [BPM]	%B.R. ERROR
1	0.74	1.33	-.76	1.33	7.0	3.26
2	0.74	1.33	-.75	0.00	7.0	3.71
3	0.76	1.33	-.75	0.00	7.0	3.26
4	0.79	5.33	-.75	0.00	7.0	3.41
5	0.76	1.33	-.75	0.00	7.0	3.12
6	0.75	0.00	-.75	0.00	7.0	3.26
7	0.75	0.00	-.74	1.33	6.9	2.82
8	0.73	2.67	-.75	0.00	6.9	2.82
9	0.73	2.67	-.75	0.00	6.9	2.67
10	0.73	2.67	-.74	1.33	6.9	2.37
11	0.76	1.33	-.74	1.33	6.9	2.52
12	0.77	2.67	-.74	1.33	6.9	2.23
13	0.78	4.00	-.75	0.00	6.9	2.52
14	0.78	4.00	-.74	1.33	6.9	1.93
15	0.77	2.67	-.74	1.33	6.9	2.37
16	0.78	4.00	-.73	2.67	6.8	1.48
17	0.78	4.00	-.74	1.33	6.8	1.48
18	0.78	4.00	-.74	1.33	6.9	1.93
19	0.77	2.67	-.73	2.67	6.8	1.34
20	0.78	4.00	-.73	2.67	6.8	1.04
21	0.78	4.00	-.74	1.33	6.8	1.48
22	0.77	2.67	-.73	2.67	6.8	0.89
23	0.78	4.00	-.74	1.33	6.8	1.34
24	0.81	8.00	-.75	0.00	6.8	0.89
25	0.82	9.33	-.75	0.00	6.8	0.45
26	0.81	8.00	-.75	0.00	6.8	0.89
27	0.82	9.33	-.76	1.33	6.8	0.45
28	0.80	6.67	-.76	1.33	6.8	0.45
29	0.81	8.00	-.75	0.00	6.8	0.45
30	0.80	6.67	-.76	1.33	6.8	0.45
31	0.81	8.00	-.75	0.00	6.8	0.45
32	0.81	8.00	-.75	0.00	6.8	0.45
33	0.79	5.33	-.75	0.00	6.7	0.00
34	0.81	8.00	-.75	0.00	6.7	0.00
35	0.83	10.67	-.76	1.33	6.7	0.00
36	0.81	8.00	-.76	1.33	6.7	0.00
EXPECTED	.75		0.75		6.7	

TABLE 6.3 B- Breath to breath results for induction trial #2.

BREATH #	Cins measured [%]	Cet measured [%]	Csp calc. [%]	Uptake measured [c.c. Vapor]	Uptake calc.	%Uptake error
1	0.03	3.36	3.12	17.40	18.34	5.13
2	0.03	3.12	2.90	16.70	17.05	2.06
3	0.02	2.85	2.69	15.80	15.85	0.33
4	0.03	2.70	2.50	15.50	14.74	5.17
5	0.02	2.48	2.33	13.80	13.70	0.72
6	0.02	2.31	2.16	12.50	12.74	1.87
7	0.02	2.15	2.01	11.70	11.84	1.21
8	0.02	2.02	1.87	10.40	11.01	5.55
9	-.02	1.93	1.74	9.70	10.24	5.24
10	0.01	1.76	1.62	9.10	9.52	4.38
11	0.01	1.61	1.50	9.00	8.85	1.72
12	0.02	1.52	1.40	8.50	8.23	3.33
13	0.01	1.38	1.30	7.80	7.65	1.99
14	0.01	1.32	1.21	7.40	7.11	4.08
15	0.01	1.22	1.12	6.80	6.61	2.87
16	0.02	1.10	1.04	6.40	6.15	4.14
17	0.02	1.07	0.97	6.00	5.71	5.01
18	0.01	0.96	0.90	5.60	5.31	5.43
19	0.01	0.79	0.84	5.20	4.94	5.30
20	0.01	0.80	0.78	4.80	4.59	4.55
21	0.01	0.68	0.73	4.40	4.27	3.08
22	0.01	0.68	0.67	4.00	3.97	0.80
23	0.01	0.59	0.63	3.80	3.69	3.00
24	0.00	0.61	0.58	3.70	3.43	7.87
25	0.01	0.58	0.54	3.50	3.19	9.75
26	0.01	0.55	0.50	3.20	2.96	7.93
27	0.00	0.52	0.47	3.00	2.76	8.84
28	0.01	0.47	0.44	2.80	2.56	9.26
29	0.00	0.43	0.40	2.60	2.38	9.13
30	0.00	0.41	0.38	2.40	2.21	8.35
31	0.01	0.36	0.35	2.20	2.06	6.84
32	0.00	0.36	0.33	2.10	1.91	9.69
33	0.00	0.32	0.30	1.90	1.78	6.75
34	0.00	0.30	0.28	1.80	1.65	8.78
35	0.01	0.29	0.26	1.70	1.54	10.50
36	0.00	0.26	0.24	1.60	1.43	11.87

Table 6.4.A- Results of laboratory tests of the measurement system during simulated anesthetic induction.

TRIAL #	EXPECTED VOLUME [L]	MEAN Vins [L]	S. D. Vins [L]	%Vins ERROR	MEAN Vexp [L]	S. D. Vexp [L]	%Vexp ERROR
1	0.75	0.80	0.0075	7.00	-.76	0.0192	1.44
2	0.75	0.78	0.0003	4.04	-.75	0.0033	0.44
3	0.77	0.81	0.0050	4.44	-.78	0.0186	1.37
4	0.77	0.78	0.0467	1.63	-.78	0.0219	1.45
5	0.77	0.82	0.0574	6.16	-.73	0.0474	4.86
6	0.77	0.74	0.0044	4.51	-.76	0.0353	0.72
7	0.77	0.73	0.0208	5.16	-.78	0.0003	1.41
8	0.78	0.83	0.0608	6.67	-.76	0.0261	1.93
9	0.78	0.80	0.0081	2.46	-.76	0.0303	2.46
10	0.78	0.81	0.0387	3.83	-.74	0.0590	4.86

Table 6.4.B- Results of laboratory tests of the measurement system during simulated anesthetic induction.

TRIAL #	EXPECTED B.R. [BPM]	MEAN B.R. [BPM]	S.D. B.R. [BPM]	%B.R. ERROR	CALC-ULATED UPTAKE [c.c.	MEAS-URED UPTAKE Vapor	CORR-ECTED UPTAKE]	%UPTAKE ERROR
1	6.8	6.86	0.1161	0.94	271.10	277.40	274.55	1.27
2	6.7	6.85	0.1611	1.62	241.98	244.80	240.79	0.49
3	6.7	6.76	0.1586	0.32	210.86	210.40	207.82	1.44
4	6.7	6.69	0.0889	0.73	150.45	155.60	155.70	3.49
5	6.4	6.12	0.3082	4.35	282.31	288.80	279.33	1.05
6	6.4	6.37	0.0697	0.46	321.89	309.60	317.69	1.31
7	6.2	6.17	0.1475	0.37	307.54	291.90	302.24	1.73
8	6.5	6.36	0.1053	2.54	261.34	265.10	259.20	0.82
9	6.3	6.27	0.1314	0.09	213.09	212.70	211.07	0.95
10	6.4	6.46	0.1244	0.87	349.43	338.60	332.98	4.71

TABLE 6.5 A- Breath to breath results for zero-uptake trial #4.

BREATH #	Vins [L]	%Vins ERROR	Vexp [L]	%Vexp ERROR	B.R. [BPM]	%B.R. ERROR
1	0.52	2.91	-.50	1.05	9.9	2.09
2	0.52	2.91	-.50	1.05	9.9	2.09
3	0.52	2.91	-.50	1.05	10.0	2.61
4	0.52	2.91	-.50	1.05	9.9	2.30
5	0.52	2.91	-.50	1.05	9.9	2.09
6	0.52	2.91	-.50	1.05	9.9	1.99
7	0.51	0.93	-.50	1.05	9.9	2.09
8	0.51	0.93	-.49	3.03	9.9	2.09
9	0.50	1.05	-.50	1.05	10.0	2.82
10	0.51	0.93	-.50	1.05	9.9	2.09
11	0.51	0.93	-.50	1.05	9.9	2.09
12	0.50	1.05	-.50	1.05	9.9	1.99
13	0.49	3.03	-.50	1.05	9.9	2.09
14	0.49	3.03	-.50	1.05	10.0	2.82
15	0.50	1.05	-.50	1.05	9.9	2.09
16	0.50	1.05	-.50	1.05	9.9	2.09
17	0.49	3.03	-.50	1.05	9.9	2.09
18	0.49	3.03	-.51	0.93	9.9	1.99
19	0.49	3.03	-.50	1.05	10.0	2.82
20	0.49	3.03	-.51	0.93	9.9	2.09
21	0.49	3.03	-.50	1.05	9.9	2.09
22	0.49	3.03	-.51	0.93	9.9	2.09
23	0.48	5.01	-.50	1.05	9.9	1.99
24	0.48	5.01	-.51	0.93	10.0	2.82
25	0.48	5.01	-.51	0.93	9.9	2.09
26	0.48	5.01	-.51	0.93	9.9	2.09
27	0.49	3.03	-.51	0.93	9.9	2.09
28	0.48	5.01	-.50	1.05	9.9	2.09
29	0.48	5.01	-.51	0.93	10.0	2.82
30	0.49	3.03	-.51	0.93	9.9	1.99
31	0.49	3.03	-.51	0.93	9.9	2.09
32	0.48	5.01	-.51	0.93	9.9	2.09
33	0.48	5.01	-.51	0.93	10.0	2.82
34	0.49	3.03	-.51	0.93	9.9	2.09
35	0.49	3.03	-.51	0.93	9.9	2.09
36	0.49	3.03	-.51	0.93	9.9	1.99
EXPECTED	.51		-.51		9.7	

TABLE 6.5 B- Breath to breath results for zero-uptake trial #4.

BREATH #	Cins measured [%]	Cet measured [%]	Csp calc. [%]	Uptake measured [c.c. Vapor]	Uptake calc.	Uptake error
1	1.91	1.92	1.92	0.50	0.00	0.50
2	1.93	1.92	1.92	0.40	0.00	0.40
3	1.93	1.94	1.92	0.40	0.00	0.40
4	1.91	1.93	1.92	0.40	0.00	0.40
5	1.92	1.92	1.92	0.50	0.00	0.50
6	1.91	1.92	1.92	0.20	0.00	0.20
7	1.92	1.93	1.92	0.20	0.00	0.20
8	1.92	1.92	1.92	0.20	0.00	0.20
9	1.92	1.92	1.92	-0.10	0.00	0.10
10	1.92	1.94	1.92	0.10	0.00	0.10
11	1.92	1.91	1.92	0.20	0.00	0.20
12	1.92	1.92	1.92	0.10	0.00	0.10
13	1.91	1.91	1.92	-0.10	0.00	0.10
14	1.89	1.90	1.92	-0.20	0.00	0.20
15	1.89	1.89	1.92	-0.10	0.00	0.10
16	1.89	1.90	1.92	-0.20	0.00	0.20
17	1.89	1.89	1.92	-0.20	0.00	0.20
18	1.90	1.91	1.92	-0.40	0.00	0.40
19	1.88	1.89	1.92	-0.20	0.00	0.20
20	1.90	1.89	1.92	-0.30	0.00	0.30
21	1.90	1.89	1.92	-0.30	0.00	0.30
22	1.88	1.88	1.92	-0.30	0.00	0.30
23	1.90	1.89	1.92	-0.40	0.00	0.40
24	1.88	1.89	1.92	-0.50	0.00	0.50
25	1.88	1.89	1.92	-0.60	0.00	0.60
26	1.88	1.88	1.92	-0.50	0.00	0.50
27	1.87	1.87	1.92	-0.50	0.00	0.50
28	1.89	1.88	1.92	-0.40	0.00	0.40
29	1.88	1.88	1.92	-0.70	0.00	0.70
30	1.88	1.88	1.92	-0.50	0.00	0.50
31	1.88	1.88	1.92	-0.50	0.00	0.50
32	1.88	1.89	1.92	-0.50	0.00	0.50
33	1.89	1.88	1.92	-0.40	0.00	0.40
34	1.88	1.88	1.92	-0.40	0.00	0.40
35	1.88	1.88	1.92	-0.30	0.00	0.30
36	1.88	1.88	1.92	-0.40	0.00	0.40

Table 6.6.A- Results of laboratory tests of the measurement system during simulated zero anesthetic uptake.

TRIAL #	EXPECTED VOLUME [L]	MEAN Vins [L]	S. D. Vins [L]	%Vins ERROR	MEAN Vexp [L]	S. D. Vexp [L]	%Vexp ERROR
1	0.72	0.70	0.1581	2.34	-.71	0.0428	0.84
2	0.84	0.87	0.1614	3.15	-.88	0.0603	4.53
3	0.91	0.91	0.0655	1.02	-.87	0.0239	3.46
4	0.51	0.50	0.0339	1.82	-.50	0.0239	0.28
5	0.54	0.56	0.0153	3.44	-.53	0.0025	0.80
6	0.74	0.78	0.0189	5.71	-.75	0.0292	1.90
7	0.91	0.92	0.0128	1.32	-.96	0.0256	5.55
8	0.71	0.69	0.0178	2.48	-.70	0.0025	1.11
9	0.58	0.55	0.0325	4.56	-.53	0.0197	8.40
10	0.72	0.74	0.0206	3.31	-.70	0.0025	1.99

Table 6.6.B- Results of laboratory tests of the measurement system during simulated zero anesthetic uptake.

TRIAL #	EXPECTED B.R. [BPM]	MEAN B.R. [BPM]	S.D. B.R. [BPM]	%B.R. ERROR	CALC-ULATED UPTAKE [c.c. Vapor]	MEAS-URED UPTAKE	CORR-ECTED UPTAKE	ABSOLUTE UPTAKE ERROR
1	5.9	5.79	0.0169	2.35	0.00	0.00	-0.00	0.00
2	7.0	6.84	0.0317	1.70	0.00	-2.10	0.67	0.67
3	7.0	7.05	0.1361	1.29	0.00	25.50	0.68	0.68
4	9.7	9.91	0.0119	2.22	0.00	-5.80	-0.58	0.58
5	9.7	9.92	0.0719	2.28	0.00	22.00	0.05	0.05
6	10.0	9.92	0.0028	0.83	0.00	30.30	-1.54	1.54
7	8.4	8.38	0.0267	0.45	0.00	-58.00	0.26	0.26
8	8.4	8.39	0.0731	0.41	0.00	-14.50	0.63	0.63
9	10.7	11.03	0.0439	3.37	0.00	31.10	0.30	0.30
10	11.0	11.02	0.0461	0.10	0.00	47.90	-0.11	0.11

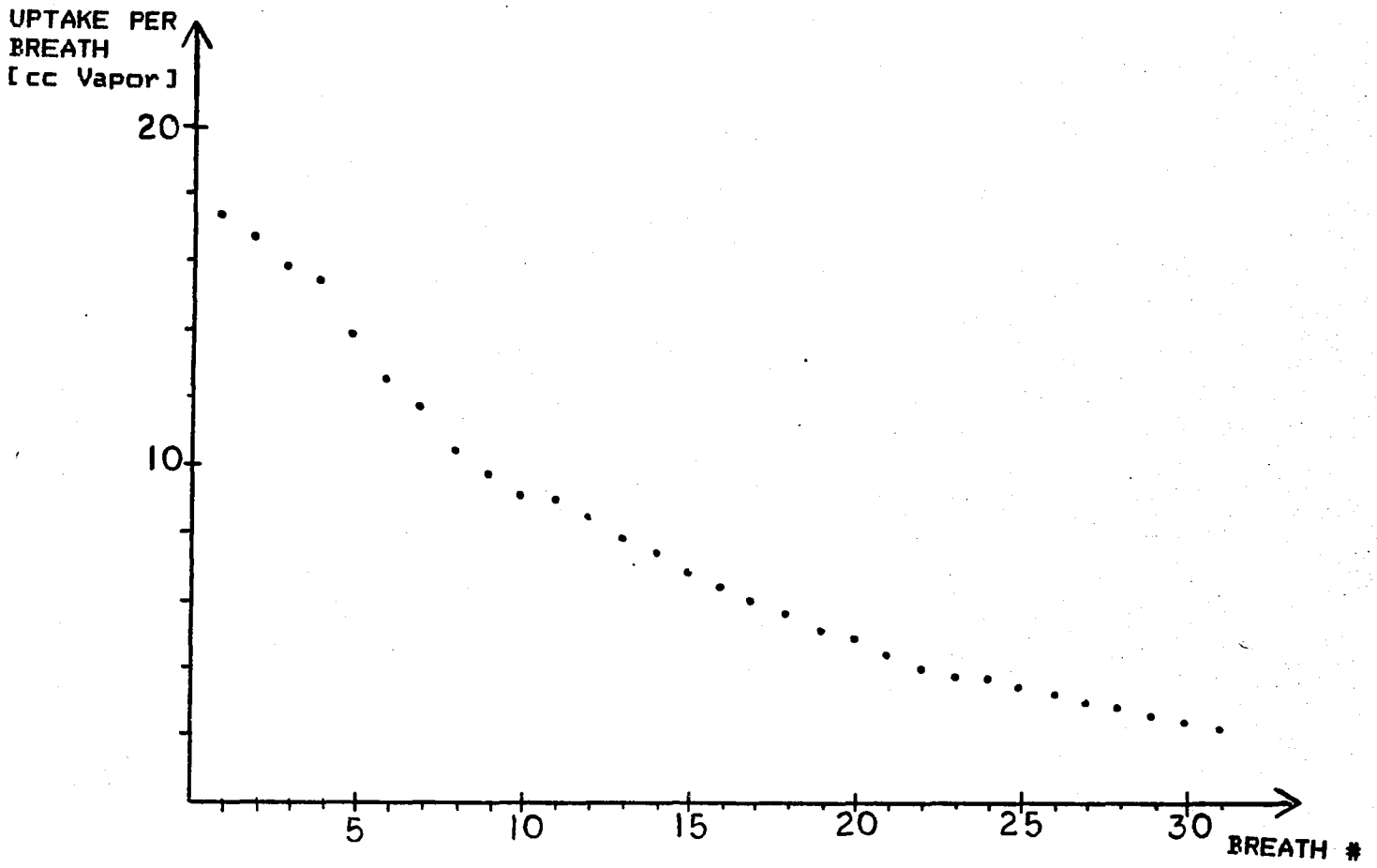

Figure 6.1. Breath by breath uptake measured during induction trial #2.

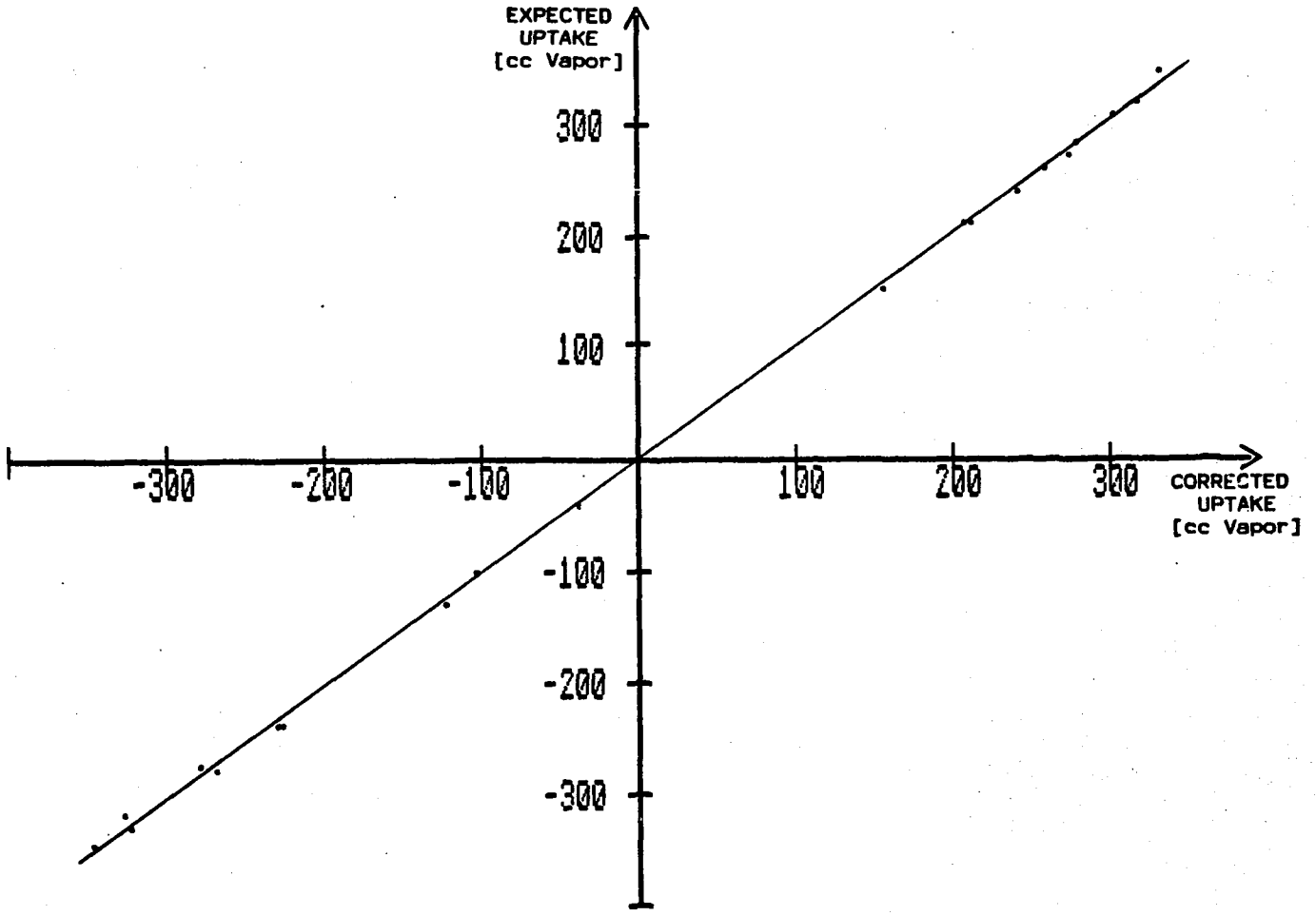

Figure 6.2. Cumulative uptake results for induction and emergence tests.

and intercept of 1.701 [cc Vapor]. The standard error of estimate was found to be 7.44 [cc Vapor]. Figure 6.2 shows the expected vs. measured cumulative uptake data and the least squares fit line. For the case of zero-uptake tests the standard deviation of corrected uptake about the predicted zero was 0.648 [cc Vapor].

As an alternative way of correcting the errors caused by the flow-meter bias drift, it was decided to reprogram the system to allow on-line breath by breath bias corrections. 10 more induction simulation tests of 40 breaths each, were conducted using the reprogrammed system. Table 6.7 shows the overall results for these tests. For the expected vs. measured cumulative uptake data a least squares fit line with a slope of 0.955 and intercept of 9.45 [cc Vapor] was found using linear regression analysis. The standard error of estimate was 5.80 [cc Vapor] and the average error in the measured cumulative uptake over the ten trials was 4.0 [cc Vapor].

Table 6.7.A- Results of induction tests with on-line breath-by-breath bias adjustment.

TRIAL #	EXPECTED VOLUME [L]	MEAN Vins [L]	S. D. Vins [L]	%Vins ERROR	MEAN Vexp [L]	S. D. Vexp [L]	%Vexp ERROR
1	0.75	0.77	0.0098	2.11	-.76	0.0144	1.82
2	0.77	0.77	0.0395	0.46	-.77	0.0036	0.05
3	0.77	0.78	0.0586	1.11	-.78	0.0032	0.80
4	0.75	0.75	0.0158	0.03	-.75	0.0121	0.41
5	0.74	0.73	0.0007	1.93	-.72	0.0168	2.19
6	0.75	0.74	0.0200	0.94	-.74	0.0228	1.44
7	0.75	0.74	0.0270	0.94	-.74	0.0048	1.44
8	0.75	0.74	0.0271	1.61	-.74	0.0158	1.85
9	0.75	0.73	0.0382	2.30	-.73	0.0715	2.87
10	0.74	0.75	0.0157	1.31	-.75	0.0019	1.09

Table 6.7.B- Results of induction tests with on-line breath-by breath bias adjustment.

TRIAL #	EXPECTED B.R. [BPM]	MEAN B.R. [BPM]	S.D. B.R. [BPM]	%B.R. ERROR	CALC-ULATED UPTAKE [c.c. Vapor]	MEAS-URED UPTAKE	%UPTAKE ERROR
1	6.4	6.42	0.1643	0.26	283.12	287.11	1.41
2	6.4	6.35	0.2168	0.79	267.08	277.28	3.82
3	6.4	6.32	0.1501	1.19	98.28	95.89	2.43
4	6.3	6.29	0.1731	0.27	260.32	269.84	3.66
5	6.4	6.32	0.1549	1.26	75.83	72.67	4.17
6	6.4	6.25	0.1857	2.40	269.67	263.64	2.24
7	6.4	6.22	0.0766	2.81	200.06	192.74	3.66
8	6.2	6.16	0.1313	0.16	188.36	181.06	3.88
9	6.2	6.18	0.1764	0.50	177.83	174.97	1.61
10	6.5	6.51	0.1983	0.27	156.32	155.45	0.56

CHAPTER 7

DISCUSSION OF RESULTS

The measurement system implemented here, seems capable of measuring cumulative uptake of volatile anesthetic agents. In spite of zero bias variations in the flow-meter used, inspired and expired volume and uptake determinations were fairly close. The zero bias variations of the flowmeter produced large values of standard deviation in some of the volume measurements, but due to randomness of these variations the mean value of volumes measured was not affected as much and therefore the value of uptake measured by the system was close to the expected.

The error in the flow-meter was specified to be ±5% or ±3LPM, and the gas analyzer's error was specified at ±1%. In the Error Analysis section it was found that for these values of error in the measurement instruments, a worst case error of ±7% could be expected. The errors for determining cumulative uptake were within this expected range except for emergence trial number 4. In this case an error of 8.9% was observed in the measured uptake. The additional error could be due the test

system's error in determining the expected cumulative uptake value.

By looking at the breath to breath results it can be concluded that the major portion of the existing errors is caused by changes in the flow-meter zero bias during a test.

Even though laboratory bench tests satisfy the requirement of low cumulative uptake error, still the implemented system has some shortcomings. The major shortcoming that needs to be overcome to make the system clinically usable is the fact that flow-meter's output bias was not consistent and had to be adjusted at the beginning of each run. The value of bias was found in the program as the value which caused the inspired and expired volumes to equal. This is a valid procedure during the tests where the inspired and expired gases are indeed equal, but during operating room activities where the system is used on an actual patient the inspired and expired volumes may not be equal. It was also observed that even during the test period sometimes the bias would drift. Since the program only corrects the bias at the beginning of the run a drift in the bias will produce erroneous results. During the testing, results were ignored if a considerable change of bias was detected during a test, a fix which is not applicable in the

operating room. Future work should first concentrate on finding another means of flow or volume measurement with zero or very low bias drift. One such alternative is the VMM2 ventilation monitor developed by the Sensor Medics corp. This instrument measures the volume of gases passing through its transducer in real time and it is specified to have zero output bias. Furthermore having an instrument which measures volumes rather than flows can save some calculation time in the program.

Another shortcoming is that the program does not automatically calculate the time delay between the gas analyzer's and the flow-meter's output. The time delay must be determined prior to the program execution and entered into the program. This however has the advantage that this time delay can be measured by a more accurate method than if it were done automatically. The error due to the time delay is therefore minimized.

The Perkin-Elmer MGA1100 gas analyzer used in this study performs satisfactorily once calibrated correctly. However the time delay must be rechecked any time the sampling capillary is replaced.

Aside from the above shortcomings it must be mentioned that by using the real-time and multi-tasking features of the operating system it was possible to sample the input channels continuously. There is no loss of data

when the program is writing the output to the screen or when calculating the output parameters.

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