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Abstract

It is hypothesized that anaesthetic breathing systems are poorly designed, and would benefit from engineering modeling to improve function. The use of breathing systems in different environments is described. Bath fp, a software package, developed in the Department of Mechanical Engineering for mathematically modelling hydraulic systems such as breathing systems, is introduced. Bath fp was used to model the behaviour of anaesthetic circle systems at high, medium, low and minimum fresh gas flows (FGFs). Using 50% nitrous oxide in oxygen ('entonox') as a gas model for all such volatile anaesthetic agents, the behaviour of the system was observed in terms of gas concentrations, volumetric flow-rates and airway pressures. Anaesthetic circle systems studied included standard adult and paediatric systems, and a coaxial system. In the standard systems, different tubing lengths and diameters were modelled, and for the coaxial system, different diameter ratios of outer and inner tubes were modelled. Gas concentration changes at different points in the system were used to clarify system function at different FGFs. Management of exhaled carbon dioxide was examined, including changing absorber volume. A venturi was modelled and added to a valveless coaxial circle system, to see how the venturi changed system function, and under what conditions it might fail.

Clinical data were collected to validate modelling results. Nineteen adult volunteers were recruited to a trial, in which entonox was breathed through standard and coaxial systems for short periods, at different FGFs, with high flow air-breathing periods in between. Data on gas concentrations were collected from three points in each system. Results were analysed, and function was compared between different systems, and between systems and modelling results.

The simulation model clarified circle system function, and showed that function depended more on FGF than on geometry. The clinical data supported the use of simulation to predict the behaviour of breathing systems.
Notation

\(a, a', b, b'\)  Equation constants as defined in text
\(a_0\)  Area of venturi nozzle
\(a_1\)  " " " throat
\(A\)  Gas volume internal surface area, generic contact area, area of venturi throat, contextually
\(A_{Abst}\)  Total cross sectional area of absorbent
\(A_p\)  Pipe internal cross sectional area
\(A_{surf}\)  Total surface area of reservoir bag
\(ABW5A(1)\)  Bathfp model for the patient’s respiratory system
\(ABW99\)  " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " " "
$E_{\text{in}}$ Energy into a control volume
$E_{\text{out}}$ Energy out of a " "
$\Delta E_v$ Change of energy in a control volume $v$
EVA Extravehicular activity
f Respiratory rate
$f_c$ Friction factor
$F(f_c)$ Function of $f_c$
$f_{ch}$ Friction factor for choke flow
$F_{\text{damp}}$ Damping force of a valve
$F_{\text{fric}}$ Frictional force " " "
$F_{\text{elas}}$ Elastic force of reservoir bag
$F_{\text{flow}}$ Force due to gas flow of a valve
$F_{\text{press}}$ Force due to applied gas pressure of a valve
$F_{\text{spring}}$ Force due to a spring of a valve
FGF Fresh Gas Flow
$h_{\text{in}}$ Specific enthalpy of incoming gas
$h_{\text{out}}$ " " " " outflow "
$H_{\text{in}}$ Rate of change of heat transfer to the gas
$\Delta H_{\text{in}}$ Increment of heat energy applied to a gas in a control volume
ISS International Space Station
k A constant
$k_e$ Pressure limiting valve spring stiffness
$k_L$ Loss coefficient at venturi enlargement
$k_M$ Mass transfer coefficient
K Loss coefficient
$K_{LR}$ Loss coefficient (in reverse flow) at venturi contraction
KA Area enlargement coefficient in venturi
L Pipe Length
$L_{\text{eff}}$ Effective pipe length
$M_n$ Algebraic expressions, contextually defined in venturi calculations
$M_{\text{0i}}$ Molecular weight of constituent ‘i’ in a gas mixture
$M_{\text{0t}}$ " " " " gas mixture (total)
m Mass of gas, spring or valve
$m_{\text{Abst}}$ Mass flowrate of gas through the carbon dioxide absorbent cannister
$m_{\text{in}}$ Incoming mass of gas
$m_{\text{out}}$ Outflow mass of gas
m Net mass flow rate inside the gas volume
$m_{\text{co2}}$ Mass flowrate of carbon dioxide
n Polytropic index
N Number of convolutions in a corrugated pipe
P Gas pressure
$P_0$ Pressure at fresh gas nozzle of venturi
$P_1$ Pressure upstream in venturi throat
$P_2$ " " " " "
$P_3$ Pressure at venturi outlet
$P_{\text{in}}$ Pressure upstream from venturi inlet
\( \Delta P \) Pressure drop
\( \dot{P} \) Rate of change of gas pressure
\( P_{\text{aco2}} \) Arterial carbon dioxide partial pressure
\( P_c \) Downstream pressure for choke flow
\( P_d \) Downstream pressure
\( P_{\text{eco2}} \) Expired carbon dioxide partial pressure
\( P_{\text{hypst}} \) Pressure term to account for hysteresis exhibited by reservoir bag
\( P_{\text{ico2}} \) Inspired carbon dioxide partial pressure
\( P_M \) Mean pressure
\( P_{N2} \) Partial pressure of nitrogen
\( P_u \) Upstream pressure
PLSS Portable Life Support System
\( q_c \) Choke mass flowrate
\( q_f \) Mass flowrate through friction pipe
\( q_2 \) Mass flowrate at venturi throat
\( r \) Relative roughness
\( R \) Specific gas constant
\( Re \) Reynolds’ number
\( Re_{\text{ch}} \) Reynolds’ number for choke flow
\( Sch \) Schmidt number
\( St \) Stanton number
\( T \) Gas temperature
TRACH 1 Bath fp model for the trachea
\( \dot{T} \) Rate of change of temperature inside the gas volume
\( T_d \) Downstream gas temperature
\( T_f \) Gas temperature at the flow area
\( T_{\text{in}} \) Incoming gas temperature
\( T_{\text{out}} \) Outflow gas temperature
\( T_u \) Upstream gas temperature
\( u_T \) Specific internal energy within the gas volume
\( u_0 \) Gas velocity at venturi nozzle
\( u_1 \) “““ upstream in venturi throat
\( u_2 \) “““ downstream““““““
\( u_3 \) “““ venturi outlet
\( U \) Gas velocity
\( U_E \) Average gas velocity through the void space of the carbon dioxide absorber
\( v \) Gas velocity
\( v_c \) Choke flow gas velocity
\( v_f \) Gas velocity at flow area
\( V \) Gas volume
\( V \) Volume of the reservoir bag
\( V_A \) “““ limit of ““““““ when slack
\( V_B \) ““““““““““““““ stretched within elastic range
VEN01R Bath fp model of venturi
\( V_{\text{str}} \) Volume of stretched reservoir bag
\( V_{\text{void}} \) Total volume of void in carbon dioxide absorber
\( V_{\text{min}} \) Minute ventilation
\dot{V} \quad \text{Rate of change of gas volume}

w_i \quad \text{Concentration of carbon dioxide in the absorbent of the absorber}

\dot{w}_i \quad \text{Rate of absorption by absorbent absorbent of carbon dioxide}

w_{\text{Max}} \quad \text{maximum concentration of carbon dioxide in absorbent}

\dot{W}_{\text{out}} \quad \text{Work done by the gas}

\dot{W} \quad \text{Rate of change of work done by the gas}

x \quad \text{Distance}

\dot{x} \quad \text{Velocity}

\ddot{x} \quad \text{Acceleration}

\Delta W_{\text{out}} \quad \text{Increment of work done by a gas from a control volume}

\gamma \quad \text{Ratio of specific heats}

\varepsilon_{\text{Nom}} \quad \text{Nominal elastance of reservoir bag when slack}

\varepsilon_{\text{Max}} \quad \text{Maximum elastance of reservoir bag when stretched}

\mu \quad \text{Gas dynamic viscosity}

\xi \quad \text{Ratio of void volume to canister volume in carbon dioxide absorber}

\rho \quad \text{Gas density}

\rho_0 \quad \text{Gas density in venturi fresh gas flow nozzle}

\rho_1 \quad \text{“ “ upstream in venturi throat}

\rho_2 \quad \text{“ “ downstream “ “}

\rho_3 \quad \text{“ “ at venturi outlet}

\rho_B \quad \text{bulk gas density}

\rho_c \quad \text{Downstream gas density for choke flow}

\rho_M \quad \text{Mean gas density}

(1) [ ] \quad \text{Graphical lines representing oxygen, O}_2

(2) [ ] \quad \text{“ “ “ nitrogen, N}_2

(3) [ ] \quad \text{“ “ “ carbon dioxide, CO}_2

(4) [ ] \quad \text{“ “ “ nitrous oxide, N}_2\text{O}
Chapter 1.
Introduction

Artificial breathing systems to help humans survive extreme environments are used in a number of situations across a spectrum of ambient pressures, using a range of different breathing gases. These activities include, but are not limited to, anaesthesia and intensive care activity, high altitude mountaineering [1], firefighting, aerospace extravehicular space activity and underwater diving operations.

Breathing systems in these areas of activity deliver a fresh gas flow (FGF) to the human user from a stored gas source. This can be at a FGF rate high enough to provide both adequate breathing gas and excretion of waste gases through an overflow valve. However, in low FGF, closed or semi-closed systems the FGF only forms a minor part of the breathing gas. The majority of the breathing gas comes from the recirculated, exhaled gas from which the waste gas (mostly metabolic carbon dioxide) has been physically or chemically absorbed. The lower the FGF, the more economic the system is in exchange for the additional complexity of structure and function. Such complexity requires high quality technology to monitor gas flow, volume and pressure and gas concentrations, and to ensure minimisation of leaks. In anaesthesia, the expense of breathable volatile anaesthetic agents supports the economics of such a system. Likewise a limited gas supply in portable life support systems (PLSS) mitigates in favour of low flow closed systems; in diving operations the FGF is delivered at a flowrate dependent on ambient pressure, and in extravehicular space activity there are additional constraints of equipment mass and gas pressure magnitudes. In the latter two applications, the recirculation process is often aided by the inclusion of a venturi or a nozzle, either in the fresh gas supply line, or elsewhere in the breathing system. In all applications, the design of the fresh gas delivery system, unidirectional valves and the CO₂ absorption system will affect safety and efficiency. The time taken for the system to respond to a step change in gas concentrations is directly proportional to the system volume and inversely proportional to the FGF. It therefore behoves the designer to minimise the system volume in the presence of a low FGF, providing adequate gas monitoring is available.

The activity areas under discussion have problems in common, for which a study of a generic anaesthesia breathing system will have mutual design benefits. The anaesthesia circle system, in some ways, has the least demands placed on it of all such low flow systems; it uses oxygen with either air or nitrous oxide, together with a volatile anaesthetic agent at normal (sea level) ambient pressure of 101.3 kPa, although anaesthetics are occasionally given at altitude or in hyperbaric chambers. It is therefore the least well-engineered system and probably needs some improvement. Some designs
of anaesthetic circle system have utilised a venturi [43], but little is written on their efficacy. CO$_2$ absorbers are designed to avoid ‘channelling’, but much more is written about the chemical interaction of Soda Lime or Baralyme crystals with the anaesthetic agent to produce carbon monoxide or other toxic agents, than about the mechanical design of such absorbers [3,4]. The CO$_2$ absorption process is usefully exothermic and hydrating to the patient. Apart from the early history of their development, there is not much work to date on the design of other components in the anaesthetic circle system, or on the system construction.

The astronaut’s PLSS [5] provides a pure oxygen environment at approximately one third of a sea level atmosphere in order to pressurise and ventilate the suit, as well as directing adequate breathing gas into the helmet. Historically the range of spacesuit pressures is from 23 to 40 kPa. A venturi device helps to circulate the gas and CO$_2$ is absorbed by Lithium Hydroxide crystals, more efficient (defined as mass of CO$_2$ absorbed per unit mass of hydroxide consumed) and lighter than Soda Lime. This pressure is the minimum compatible with atmospheric and metabolic safety, yet low enough to allow suit mobility [6]. Such an atmosphere however, requires a careful prebreathing protocol to avoid decompression sickness and represents a fire risk to the astronaut [5]. Currently the oxygen is stored as pressurised gas in a cylinder, but a useful outlet for waste body heat and equipment heat would be to vaporise liquid oxygen to its gas, since a unit volume of liquid oxygen provides over 800 volumes of oxygen gas at 15°C. This is being investigated for use on initial human visits to Mars [7]. On the international space station, CO$_2$ is adsorbed under ambient pressure onto a zeolite, and desorbed to the vacuum of space [5,6]. An unfulfilled chemical engineering challenge would be to miniaturise the zeolite to adsorb CO$_2$ in a PLSS, as a regenerable CO$_2$ management system. Such a design could be directly applied to an anaesthetic circle system CO$_2$ adsorption device, since a binary pressure source is also available in the form of ambient pressure and the vacuum source available in all operating rooms. A further chemical engineering challenge in a space context would be to design a miniature Sabatier processor to produce oxygen directly from the Martian atmosphere consisting of 95% CO$_2$ [8,9].

The particular challenges for closed breathing systems in diving operations relate to the high gas density and mass flowrate of compressed gas [10]. The breathing gas is pressurised and stored at several tens of atmospheres and reduced to a level compatible with the ambient pressure at depth; it is sometimes compressed air, sometimes a different mix of air and nitrogen (‘Nitrox’), sometimes a helium-oxygen mix (‘Heliox’). The gas mixture is required to vary in composition with depth to minimise the chances of nitrogen ‘narcosis’, hypoxia or oxygen toxicity. At high partial pressures nitrogen is not only the cause of decompression sickness, but it also has anaesthetic properties, which is why helium is used. Helium can cause a nervous disorder called ‘high pressure nervous syndrome’[72] and is also a gas with a high thermal conductivity, which has adverse effects on the diver’s heat balance. Therefore there are times when a mix of all three gases (‘Trimix’) is used. Such a range of gas densities and pressures challenges
both the performance of the venturi, the gas mixer and the flow through the CO₂ absorber in the system.

While it is clear that there is a significant engineering input to breathing systems for space and underwater use, the hypothesis of this research is that anaesthetic breathing systems are poorly designed, with little or no engineering input. It is further postulated that such engineering input can improve efficacy and safety of such systems. The University of Bath Department of Mechanical Engineering initially developed software (‘Bath fp’) to analyse divers’ breathing systems and anaesthetic breathing systems [11]. The purpose of this study is to use the Bath fp package to analyse low flow closed systems more closely, including venturi performance in such a system. These results will be utilised to analyse and suggest possible improvements in the performance of an anaesthetic circle breathing system, and to determine the value or otherwise of a venturi in such a system, the effect of changing system volume and geometry, and of changing CO₂ absorber volume. Improvements would be based on adequate oxygenation from fresh gas delivery to the patient, adequate ventilation in the form of efficient CO₂ clearance, and adequate thermal equilibrium for the lowest possible FGF and system volume.

Current clinically used CO₂ absorbents, such as Soda Lime (mostly Ca(OH)₂ with some NaOH and KOH), Baralyme (Ca(OH)₂ with BaOH.8H₂O, and KOH) and Amsorb (Ca(OH)₂ only) [3,12], can lead to the formation of toxic agents such as carbon monoxide and compound A (a metabolic product of the anaesthetic agent Sevoflurane with reactive fluoride ions which are potentially renally toxic in high doses) in reacting with different anaesthetic agents. Using results from Bath fp modelling, an attempt will be made to optimise CO₂ absorber design. The engineering approach will contribute in a way that is different from, and advantageous to the usual clinical approaches to the understanding of the function of absorbent systems for clinical situations.

Zeolites have been used as an oxygen concentration device in areas of the world, which cannot receive reliable medical gas supplies. They have also been used as regenerable CO₂ adsorption devices in a number of underwater and space applications, including submarines and the international space station. It has also been tested, but not adopted for anaesthetic use [13], when it was found to be useful in preventing a rise in toxic compound A when using sevoflurane.

The project work will draw on previous studies on diving breathing systems and high flow anaesthetic breathing systems in the Department of Mechanical Engineering at University of Bath. It will make suggestions to improve the design of a commonly used anaesthetic breathing system, which is long overdue and has commercial potential for UK medical engineering enterprise.

The work covered in this thesis will cover the following areas. Chapters 2, 3 and 4 introduce an historical description of breathing systems used in the three areas of
activity under consideration, namely anaesthesia, space activity and diving operations. Chapter 5 introduces Bath fp as a technique for analysing breathing systems. Chapter 6 is the Bath fp modelling analysis of the standard anaesthetic circle system, in which the function of the system is analysed with standard system geometry, and with a change of tubing dimensions, and CO₂ absorber dimensions, using high and low FGFs. Chapter 7 will present a similar set of results for the coaxial circle system. Chapter 8 will introduce the functional and mathematical concepts of the venturi, and will present modelling results for the venturi system using Bath fp. Chapter 9 will show the results collected in clinical anaesthetic settings using volunteers making similar sets of measurements as the Bath fp model, and for the results acquired at the Bristol Medical Simulation Centre. Chapter 10 will analyse and discuss the results from the breathing trial, using these results to compare different breathing systems with each other, and different populations of volunteers with each other; it will also compare the collated results from the breathing trial with the results obtained from Bath fp modelling. Chapter 11 will briefly summarise the results obtained in relation to the functional efficacy of the circle system, with some suggestions for improvement in its design and use.
Chapter 2.

Anaesthetic Breathing Systems.

2.1 Introduction.
An anaesthetic breathing system is a means of transferring the breathing gas mixture from the anaesthetic machine common gas outlet to the patient. It is also the means of transferring the exhaled gas from the patient to the outside world, often via a scavenging system using high gas flowrate. Alternatively, after the carbon dioxide is absorbed from the exhaled gas, the unused fresh gas components of the exhaled gas are recirculated back to the patient, and this allows a more economic use of fresh gas flow. In general, a breathing system consists of a fresh gas limb, an inspiratory and expiratory limb, an expiratory valve, a reservoir bag, and may also consist of one or more unidirectional valves and a CO$_2$ absorber. The simpler devices have fewer components and usually involve some rebreathing of expiratory CO$_2$ containing gas, depending on the level of fresh gas flow. The ability to minimise rebreathing at as economical a fresh gas flow as possible is a measure of the breathing system’s efficiency. Depending on the precise design of the breathing system, such efficiency will vary depending on whether the patient is breathing spontaneously or is undergoing controlled artificial ventilation. The more complex systems ensure minimum rebreathing by the use of unidirectional valves and CO$_2$ absorption systems; in this way, the additional complexity allows more economical use of fresh gas and volatile agent.

The systems which use higher fresh gas flows (FGF) and involve some rebreathing were classified in 1954 by Professor Mapleson, according to their behaviour in terms of the FGF requirement to prevent CO$_2$ rebreathing [14]. At the time and for three decades beyond, they were the most popular breathing systems in UK anaesthetic practice. The Mapleson Classification of rebreathing systems is shown in Figure 2.1. Their design lends their structure and function to mathematical analysis [15].

2.2. The Mapleson Classification of semi-closed rebreathing systems
2.2.1. Mapleson A systems
The Magill breathing system was invented by the anaesthetist Sir Ivan Whiteside Magill in the early twentieth century. As shown in figures 2.1A and 2.2, the system is characterised by having the expiratory valve close to the patient and the fresh gas inflow at a distance from the patient, close to the reservoir bag. Because of this particular configuration, the system is economical in spontaneous breathing. Mapleson [14] mathematically analysed this system (and others), making some simplifying assumptions; these included constant FGF, constant patient ventilation, ‘plug’ gas flow, and ‘digital’ valve behaviour; he derives a curve of percentage of rebreathing plotted...
against ratio of FGF: patient ventilation for both theoretical calculations and assuming some gas mixing. On inspiration, fresh gas, both from the supply and previously stored in the reservoir bag, travels down the tubing towards the patient. The expiratory valve is closed at this point of the respiratory cycle and the deadspace of the system contains a small amount gas originating from the alveolae of the lungs (alveolar gas) from the previous expiration, containing CO₂. The system

![Figure 2.1: The Mapleson classification of semi-closed rebreathing systems.](from reference 2, with permission of OUP]

deadspace lies between the patient’s mouth and the expiratory valve; gas contained here is the first gas to be inhaled on inspiration. The last gas to be inhaled on inspiration is fresh gas, which therefore resides in the patient’s own deadspace, i.e. the trachea and other structures of the respiratory tract associated with transport of gas rather than gas exchange; fresh gas residing here remains fresh gas and becomes the first gas to be exhaled on the next expiration. On early expiration the pressure in the breathing system has not risen adequately for the expiratory valve to open and the dead space fresh gas from the patient’s deadspace is stored in the tubing. As expiration proceeds, the expiratory valve opens and alveolar gas is preferentially exhaled through it. In the
meantime, fresh gas from the gas supply is being stored in the tube. This means that relatively low fresh gas flows can be used while keeping rebreathing of alveolar gas to a minimum, which also reduces the likelihood of spilling fresh gas from the valve. In fact, no significant rebreathing occurs when the FGF falls as low as 70% of the patient’s own gas requirement and this efficiency is unrelated to the patient’s respiratory pattern [16, 17, 18]. If the FGF falls any lower, then more of the alveolar gas is stored in the tubing for subsequent rebreathing. Figure 2.2 shows the distribution of fresh gas, deadspace gas and alveolar gas in early and late expiration, at low and high FGFs.

![Distribution of gas in a Mapleson A (Magill) breathing system in early and late expiration at FGFs (V_F) at least equal to alveolar ventilation (V_A) and at less than V_A; white is fresh gas; hatched is deadspace gas; black is alveolar gas. [From reference 2, with permission of OUP]](image)

A modern variant of the Magill system is shown in Figure 2.3, and is known as the Lack breathing system (after Dr Alistair Lack). It consists of two concentric tubes, inspiration occurring down the outer, expiration down the inner. The expiratory inner tube must therefore be of sufficient diameter in order not to create significant resistance to expiration and it must be accommodated within the inspiratory outer tube. Therefore the concentric Lack tubing is of wider diameter than standard 22 mm tubing [19]. If the
diagram is studied carefully and compared to the Magill system, it can be seen that the Lack system is functionally identical to the Magill, and is probably as efficient as the Magill [20, 21]. The expiratory outlet from the common part of the system is close to the patient, although the valve itself is at some distance, and the fresh gas inlet and reservoir bag are distant from the patient. There is also a version called the parallel Lack system, where the tubes lie side by side rather than

![Figure 2.3.](image)

*Figure 2.3. The coaxial Mapleson A (Lack) breathing system. [from reference 2, with permission, of OUP]*

concentrically and may superficially resemble a circle system [22]. The system also imposes minimum respiratory work on the patient [23]. If such Mapleson A systems are used for controlled ventilation, the efficiency of the gas usage is relatively poor. The design, which preferentially excretes alveolar gas and stores fresh gas in spontaneous breathing, does the opposite when positive pressure breathing is applied. To facilitate inspiration, if the reservoir bag is squeezed, or if a bellows is applied to the reservoir limb, the pressure in the system increases, a partially closed expiratory valve near the patient opens and the fresh gas stored in the tube is preferentially excreted, although some will go to the patient. On expiration, the system preferentially fills with alveolar gas, partly because of the partially closed valve, and is ready to be rebreathed on the next inspiration [24]. Mapleson A systems are the least appropriate systems for controlled ventilation and a FGF of four times the patient’s gas requirement, with an expiratory time of four times inspiratory time is required to prevent rebreathing [25].

The efficiency of the Mapleson A system in controlled ventilation could be improved if the escape of fresh gas during inspiration could be prevented. Figure 2.4 shows such a modification, known as the Miller modification [26, 27]. One study has demonstrated a greater efficiency of the enclosed Mapleson A in controlled ventilation than the Mapleson D [28]. The generic term for the Miller modification is the enclosed afferent reservoir system (EAR) and has been shown to minimize rebreathing at a FGF of at least 80 ml kg\(^{-1}\) min\(^{-1}\) [29].
2.2.2. Mapleson B and C systems
These are shown in Figure 2.1B and 2.1C for historical interest only, since they were once commonly used in UK anaesthetic rooms and variations on them are currently in use as systems for use in resuscitation. Mapleson [14] estimated a required FGF of twice patient minute ventilation to prevent rebreathing.

2.2.3. Mapleson D, E and F systems – T pieces
These are shown in Figure 2.1D, 2.1E and 2.1F, otherwise known as T pieces, because the fresh gas is delivered to the system at a T junction close to the patient. This feature is common to all T pieces, whether the rest of the system has merely tubing or an open tailed reservoir bag and an expiratory valve. Functionally T pieces have a common mode of performance, due to the close proximity of the fresh gas delivery tube to the patient end of the system, which ensures preferential delivery of fresh gas. Figure 2.5 shows a sinusoidal inspiratory waveform and a constant FGF. In the early part of inspiration, FGF exceeds the inspiratory requirement. Area ‘A’ represents fresh gas that could be stored in the system for later use; area ‘B’ represents a time when the inspiratory requirement exceeds FGF; if $B \leq A$, then the gas rebreathed from the system consists solely of stored fresh gas, providing it has remained unmixed with alveolar gas. T pieces need a higher FGF than Mapleson A systems to minimise rebreathing in spontaneous rebreathing, but are more efficient than Mapleson A in controlled ventilation. A FGF of about twice the patient’s gas requirement is needed, but the exact FGF required depends on the inspiratory flow waveform, dead space volume ($V_D$) to tidal volume ($V_T$) ratios, and inspiratory (I) to expiratory (E) time ratios [14, 15, 30, 31]. The efficiency of these systems is highly dependent on a long expiratory pause [16, 17].

2.2.3.1. Mapleson D
This system has a reservoir bag and expiratory valve distant from the patient. In spontaneous breathing, the patient inspires fresh gas from the T junction and some alveolar gas from the tubing and reservoir bag. On expiration, the anatomical deadspace fresh gas is lost to the system, alveolar gas follows and is eventually excreted from the distant expiratory valve. As fresh gas continues to be supplied during expiration, it too is stored in the tubing. The higher the fresh gas flow, the greater will be the ratio of fresh gas to alveolar gas stored in the tubing for the next inspiration and the more quickly will
alveolar gas be excreted. In controlled ventilation, squeezing the reservoir bag in inspiration uses the column of stored gas (alveolar and fresh) in the tubing to push a bolus of relatively pure fresh gas into the patient. Expiration follows the same pattern as in spontaneous breathing. It can be seen from figures 2.1D and 2.5, that factors, which exacerbate rebreathing in this system are a low FGF and an inadequate length of tubing, which should generally be at least 1 m long.

Figure 2.5: The functional gas storage property of T-pieces; by storing the fresh gas and deadspace gas in the reservoir of a T-piece, rebreathing can be avoided at FGFs less than peak inspiratory flowrate. [from reference 2, with permission of OUP]

Figure 2.6 shows a coaxial variant of the Mapleson D, known as the Bain system, [32]. It is a system of concentric tubes and superficially resembles the Lack. However in contrast to the Lack, FGF is delivered down the relatively narrow bore inner tube, which delivers the gas with significant velocity close to the patient, as a T piece. Expiration occurs down the standard diameter outer tube towards a distant reservoir bag and expiratory valve.

Figure 2.6: The coaxial Mapleson D (Bain) breathing system. [from reference 2, with permission, of OUP]
In spontaneous breathing, a FGF of between 2.5 and three times the patient’s gas requirement [33] or 200 ml kg\(^{-1}\) min\(^{-1}\) [34] is required to prevent rebreathing. Any reduction in this value leads to a sudden increase in rebreathing or ventilation or both. The coaxial T piece may not be as efficient as the orthodox Mapleson D, because the coaxial arrangement at the patient end of the system may encourage gas mixing.

Figure 2.7 shows graphs of FGF delivery plotted against patient minute volume required for controlled ventilation from T pieces in adults for different values of \(P_{ECO2}\). Generally a FGF of between 70 and 100 ml kg\(^{-1}\) min\(^{-1}\) is sufficient for normocapnia, providing the ventilation delivered to the patient per minute (tidal volume\(\times\)rate) is kept sufficiently high at between 120 and 150 ml kg\(^{-1}\) min\(^{-1}\) [35, 36].

With the advent of the need to ventilate patients from a distance in a MRI scanner, the long Bain breathing system has been introduced, consisting of tubing of several metres in length. The additional length increases the compliance of the system such that smaller tidal volumes and higher end expiratory pressures result [37].

![Graph](image)

Figure 2.7: A graphical aid to estimating alveolar \(P_{CO2} (P_{ACO2})\) in kPa, from a given combination of FGF \((V_F)\) and minute ventilation \((V_E)\). [from reference 2, with permission of OUP]
Table 2.1: Fresh gas flow requirements in ml and ml.kg\(^{-1}\) body weight for children using a T piece under controlled ventilation. [from reference 2, with permission of OUP]

<table>
<thead>
<tr>
<th>Author</th>
<th>Patient Wt, kg</th>
<th>(\frac{V_E}{V_F})</th>
<th>FGFs predicted</th>
</tr>
</thead>
<tbody>
<tr>
<td>Rose &amp; Froese</td>
<td>10.0-30.0</td>
<td>2</td>
<td>(\frac{4.9}{kPa}) 1000 ml+100 ml.kg(^{-1}) 1600ml+100ml.kg(^{-1})</td>
</tr>
<tr>
<td></td>
<td>&gt;30</td>
<td>2</td>
<td>2000 ml+50 ml.kg(^{-1}) 3200 ml+50 ml.kg(^{-1})</td>
</tr>
<tr>
<td>Hatch</td>
<td>3.0 - 20.0</td>
<td>0.67</td>
<td>Measured (P_{ECO2}=4.5kPa). (P_{iCO2}&lt;0.5kPa) 1000ml+200ml.kg(^{-1})</td>
</tr>
</tbody>
</table>

\(V_{min} = V_T \times f\) where \(V_T\) is tidal volume and \(f\) is respiratory rate

\[= 7 \text{ ml kg}^{-1} \text{ body weight } \times (15 - 30 \text{ breaths min}^{-1})\]

Table 2.1 shows some suggestions for FGF for this particular T piece for controlled ventilation in children and these techniques have been confirmed in clinical use [40].

2.2.3.2. Mapleson E and F

These systems, otherwise known respectively as the Ayre’s T piece, and with its Jackson Rees modification, are shown in figure 2.1E and 2.1F. The Ayre’s T piece was invented by Phillip Ayre in 1937 for use in cleft lip and palate surgery in children. In deciding on the length of the reservoir tube, too long a tube would add unnecessary deadspace, which might permit rebreathing; too short a tube would allow dilution of fresh gas with air. Mapleson [14] showed that a FGF of somewhat more than twice patient minute ventilation is required to prevent rebreathing, but could be less than this if there were an expiratory pause and the reservoir volume were small enough. A tube of volume of one third of the patient’s tidal volume is the optimal length [38]. The valveless system is simple and provides low resistance to spontaneous breathing for small children [39]. Jackson Rees’ modification is the addition of an open tailed reservoir bag to allow visible monitoring of breathing, as well as a means of manual controlled ventilation. The FGF required to avoid rebreathing is twice the child’s gas requirement (minute ventilation), \(V_{min}\), where:

\[V_{min} = V_T \times f\] \(V_T\) is tidal volume and \(f\) is respiratory rate

\[= 7 \text{ ml kg}^{-1} \text{ body weight } \times (15 - 30 \text{ breaths min}^{-1})\]
2.3. Humphrey ADE system
This system was designed in so that, at the turn of a single lever, a system with Mapleson A characteristics can be used for spontaneous ventilation, and a system with D and E characteristics is available for controlled ventilation [41] and can be used for paediatric patients as well [42].

2.4. Venturi systems
A venturi can be incorporated into a Bain system used for adults to deliver fresh gas and entrain expiratory gas, which has had its CO₂ absorbed by a Soda lime canister. Jorgensen et al 1985 [43] used such a system to obtain some interesting results; he showed that the venturi can be too large and entrain inadequately to be of value, or too narrow and cause excess back pressure (in the fresh gas supply); the venturi entrained gas and back pressure changed to an extent which depended on the gas mixture being used (air, O₂, N₂O, helium); each 1 L.min⁻¹ FGF entrained an additional 1.0 to 1.5 L.min⁻¹; he found that FGF could be reduced to 30 ml.min⁻¹.kg⁻¹ body weight (from a value of 70 ml.min⁻¹.kg⁻¹ body weight for a normal Bain’s system, suggesting that 4/7 of CO₂ is absorbed, the rest being excreted through the expiratory valve) for no rebreathing; the absorber is exposed to a varying flowrates up to 25 L.min⁻¹ and the venturi draws gas through the absorber at about 6 to 7 L.min⁻¹; the obstruction caused by the venturi was up to 30 kPa at a FGF of 5 L.min⁻¹. It has been found that care must be taken in the design of a T-piece with a venturi if adequate inspired oxygen concentrations are to be preserved [44]. A similar device has been incorporated into a Magill system to reduce FGF requirements to 40 ml kg⁻¹ min⁻¹ from 100 ml.kg⁻¹.min⁻¹ [45]. It was found that it was difficult to incorporate into different types of breathing system for different sizes of patient. Clinicians fear complexity in technology if there is a risk of malfunction adversely affecting patient safety. The advantages of using a venturi in a breathing system to reduce FGF may have been outweighed by a lack of understanding of function.

Dorrington [15], in his book of engineering analyses of breathing systems, did so for venturi systems. He first analysed the high-air-flow-oxygen-enrichment mask, a device in which a high velocity jet of oxygen flowing to a mask entrains a larger volume of air into it. He showed how the ratio of entrained airflow to oxygen flow varies with the ratio of jet size to entrainment tube size. He notes too that stalling of the mask can occur if the exit pressure from the device increases due to an obstruction, in which case the performance of the mask is no longer predictable. He develops the analysis further by showing that the chance of stalling can be minimised by having a diffuser (a gradual change in diameter) in the entrainment system. Further details of venturis will be discussed in chapter 8.
2.5. The circle system
This is altogether a more complex system in which exhaled gas is recirculated. It incorporates unidirectional valves and a means of absorbing carbon dioxide. Its advantage is the ability to use low, very low or minimal FGF [46], which is appropriate with the use of expensive modern volatile agents. Its disadvantage is its complexity; there are many components and connections and the inspired gas concentrations are not closely related to the FGF concentrations dialled up on Rotameters or the vaporiser, particularly at low FGFs. Therefore in the modern era, sophisticated gas monitors are needed and have become integrated with the anaesthetic machine as standard, although historically low flow anaesthesia was used without the benefit of such monitoring.

In deciding where to place the various components of a circle system, shown in Figure 2.8, Eger [47] suggested three rules:
• there must be a unidirectional valve between the reservoir bag and the patient, on both the inspiratory and expiratory sides;
• fresh gas must not enter the system between the patient and the expiratory valve.
• the overflow valve must not be placed between the patient and the inspiratory unidirectional valve.

Figure 2.9 shows three configurations of the system, A, C and H, which obey these rules and which otherwise have slightly different characteristics [48]. Different configurations of circle systems have different efficiencies, as defined by fresh gas utilization [49]. In evolving circle system designs, it was found that the greatest efficiency in spontaneous breathing occurred where the overflow valve is close to the patient (c.f. the Magill system); this arrangement is least efficient in controlled ventilation, except when both unidirectional valves are also close to the patient (system H). The differences between systems are not great, but become accentuated as FGF is reduced.

Referring again to Figure 2.8, in deciding where to put the FGF supply, the best compromise is achieved in position 1, just downstream of the CO₂ absorber. Fresh gas can collect retrogradely in the absorber and reservoir bag, pushing expired gas back towards the overflow valve. At low flows, the vented alveolar gas will not yet have passed through the absorber, thus economising on its use. As the FGF increases, so does the likelihood of venting expired gas, which has already passed through the absorber, as well as venting fresh gas itself. Placing the FGF inlet at position 2, upstream of the absorber, could help humidify the fresh gas, but may also cause unnecessary venting of fresh gas and venting of absorbent dust into the inspiratory limb, as well as preventing retrograde flow of recently exhaled alveolar gas towards the overflow valve.
During controlled ventilation, gas spills out of the overflow valve during inspiration. In order to minimise the chance of this being fresh gas, the best position for the overflow valve is at 6 or 7 as shown in Figure 2.8, which will preferentially allow for venting of alveolar gas before it has reached the absorber.

Equal efficiency can be obtained during spontaneous and controlled ventilation with the reservoir bag in positions 8 or 9. If the bag is downstream of the absorber at position 9, then in controlled ventilation a squeeze of the bag forces retrograde flow of gases already cleared of CO$_2$ back through the absorber. Placing the bag upstream of the absorber in position 8 allows retrograde filling of the bag with fresh gas and this is the most common position of the reservoir bag. A modern development of circle system structure is characterised by a coaxial inspiratory limb [50]; such a modification improves the response time at low FGFs to step changes in fresh gas concentrations.

A most important part of the circle system is the CO$_2$ absorber, which has to be designed to allow low resistance to flow and to avoid channelling, where gas flows through the absorbent crystals along paths of least resistance, thus avoiding contact with the absorbent in other parts of the absorber. Although absorbent canisters have traditionally been large and arranged in two halves to allow change of half the absorbent when it is used up, modern canisters are smaller and require more frequent exchange of crystals [51].
Two absorbents in common use are Soda Lime and Baralyne. Soda Lime consists of 4% sodium hydroxide, 1% potassium hydroxide, 14-19% water and the rest as calcium hydroxide [52]. Additionally there are small amounts of silica for drying, kieselguhr for hardening and dye indicators to show when the crystals are spent. The water is essential for the reaction, which occurs as:

$$\text{CO}_2 + \text{H}_2\text{O} \Leftrightarrow \text{H}_2\text{CO}_3$$

$$2\text{NaOH} + 2\text{H}_2\text{CO}_3 + \text{Ca(OH)}_2 \Rightarrow \text{CaCO}_3 + \text{Na}_2\text{CO}_3 + 4\text{H}_2\text{O}.$$  

These reactions occur with the liberation of heat and water [53], both of which are useful by-products in the context of heat and humidity preservation of the patient.

*Figure 2.9: Three arrangements of the circle system with different behavioural characteristics. [From reference 2, with permission of OUP]*
Baralyme is a mixture of 20% barium hydroxide, 80% calcium hydroxide and some potassium hydroxide. The water for the reaction is present as the octahydrate of barium hydroxide.

In all cases the monovalent hydroxides are more reactive than the divalent Ca(OH)$_2$. Where crystals have been allowed to dry out, such as leaving the gas flowing through the canister overnight, these highly reactive monovalent hydroxides can also produce significant amounts of carbon monoxide and even formaldehyde [54, 55]. An absorbent consisting of only Ca(OH)$_2$ (‘Amsorb’, Belfast) has been produced to counteract this [56].

As well as absorbing CO$_2$, the absorbents also tend to absorb volatile agents [57]; this is generally acceptable, though wasteful, with a major exception: when the volatile agent trichloroethylene reacts with Soda Lime, it produces dichloroacetylene gas, otherwise known as phosgene, a neurotoxic agent used in the trenches of World War I. When Sevoflurane was being developed, it was thought to react with Soda Lime to produce a renally toxic agent, Compound A; this was not found to be of clinical significance [58], although in the USA, the Federal Drug Administration recommends keeping the FGF at more than 3 L min$^{-1}$ under these conditions. Indicators added to the absorbent show when it needs replacing. Ethyl violet is a commonly used example, although its usefulness can be impaired by photo deactivation by fluorescent lighting [59].

The purpose of using a circle system is to use low FGF in order to economise on the consumption of increasingly expensive volatile agents and minimise pollution. An agent concentration dialled up on the vaporiser of 1% at a low FGF consumes much less agent than 1% at a high FGF. At the start of an anaesthetic procedure, none of the body tissues contains anaesthetic agent. To enhance anaesthetic uptake and excretion of nitrogen, high FGF is generally used at the start [60]. When clinically desirable levels are reached in relevant tissues, these FGFs can be reduced, theoretically to metabolic O$_2$ consumption rate, say 250 ml min$^{-1}$, assuming no gas leakage and no further uptake by tissues or tubing. In reality both exist and some nitrous oxide and volatile agent is also switched on. The response time of the system is directly proportional to system volume and inversely proportional to the difference between gas inflow and tissue uptake [61]. At high FGF, gas concentrations delivered to the circle system from the anaesthetic machine are nearly equal to those received by the patient. As FGF becomes a significantly smaller part of what the patient receives, the majority being recirculated gas, the gas concentrations received by the patient bear significantly less resemblance to that delivered by the anaesthetic machine [62]. At the beginning of an anaesthetic, when anaesthetic gas uptake by the patient from the alveoli to the tissues is high (oxygen uptake being relatively constant), end expiratory O$_2$ concentrations appear higher than expected and this effect becomes very marked as FGF is reduced. As the anaesthetic proceeds and tissues become saturated, less anaesthetic gas is absorbed from the system, end expired O$_2$ concentrations are lower than expected and this effect also becomes more marked as FGF is reduced [63]. This degree of uncertainty means that O$_2$, volatile
agent and N₂O monitoring, with gases sampled from the patient end of the system, are highly desirable, possibly mandatory. The capital cost of such monitoring is offset within a short time against the revenue savings on volatile agent. Furthermore, the oxygen consumption of patients can vary, as can the rate of uptake of anaesthetic agents. The economic argument in favour of the use of low FGF circle systems is less strong for short surgical operations than for longer ones.

An early theoretical functional study of the circle system, using the volatile agent Halothane with O₂ and N₂O, was carried out by Mapleson in 1960 [62]. He demonstrated the fundamentally different nature of such circle systems, particularly at low FGF, compared to the open systems classified by him as A to E. Certain assumptions were made in the analysis, including no leakage, no fresh gas excreted from the overflow valve, and constant FGF and vaporiser setting. Both vaporiser in circuit (VIC) and vaporiser outside circuit (VOC, most common configuration) were studied, and major differences in inspired Halothane at low FGF (0.2 L.min⁻¹) were demonstrated; VIC has an inherent degree of hazard in that inspired anaesthetic agent levels may become inadvertently excessive, whereas the VOC configuration has to have an efficient vaporiser to maintain adequate levels of anaesthesia. For both variants, higher FGF brought the behaviour of each system closer to that of an open system. There was some dependence of behaviour on patient ventilation.

In a study by Galloon [65], a laboratory investigation was carried out using different VIC configurations at variable patient ventilations and FGF’s from 0.3 to 3.0 L.min⁻¹, although CO₂ production was modelled, patient uptake of anaesthetic agent was not. Considerable variation in rate of anaesthetic rise was shown depending on exact system geometry, although there was always a more rapid rise with higher FGF and patient minute ventilation. It was noted however, that this result was modified by gas escaping from the overflow valve when higher FGF’s were used. A clinical trial was also done, and in the single variant of VOC used, the increasing rate of rise of alveolar concentration of anaesthetic agent with higher FGF was demonstrated; however, lower final values of alveolar anaesthetic concentration at higher FGF’s were also demonstrated due to excess gas escaping from the overflow valve at high FGF, and greater amounts of anaesthetic agent collecting in the system at low FGF’s.

In the final of these three papers from the same department, in a largely clinical study, Mushin and Galloon [66] noted the factors governing gas wastage in closed systems:

\[ \text{Loss of vapour} = \text{circuit vapour concentration} \times (\text{volume Fresh Gas supplied} - \text{vol O}_2 \text{removed}). \]

Such wastage was thought to be only really relevant above a FGF of 1000 ml.min⁻¹. It was noted that in the VIC configuration, reduction of FGF always leads to an increase in inspired anaesthetic concentration and reduced wastage, whereas in VOC, the vaporiser has to be able to produce a high output at low FGF. The possibility of hazardously low
inspired oxygen concentrations in the presence of high concentrations of N₂O was noted, as was the fact that depression of the respiratory centre by anaesthetic agents in spontaneous breathing can act as a rate limiting process to anaesthetic overdose, while artificial ventilation may lead to an overdose, especially in the VIC configuration. A study by Forbes [67] noted that there was an increasing discrepancy between inspired oxygen concentrations and dialled oxygen concentrations at low FGF’s.

In a study by Snowden et al (1975) [68], using a circle system without CO₂ absorption, using two points of entry of the FGF (one upstream and one downstream of the inspiratory unidirectional valve), it was found that better CO₂ management was obtained with the FGF entering downstream.

Schonbee and Conway (1981) [69] used a model of a circle system without CO₂ absorber to examine the extent to which gas mixing in the breathing system affected efficiency of CO₂ management, starting with the hypothesis that increasing gas mixing decreased efficiency. Lower values of CO₂ were found with increasing ventilation (lower with FGF entering the system downstream of the inspiratory unidirectional valve) and with increasing FGF’s. It is deduced that the system with FGF entry downstream of the inspiratory unidirectional valve results in lower CO₂ because it allows FG to collect near the patient during the expiratory phase against a closed inspiratory valve; the system with FG entry upstream of the inspiratory unidirectional valve does not achieve this, and the results also suggested that there is greater mixing with this configuration, which confirms the hypothesis. Greater mixing also occurs with higher ventilation frequencies due to more frequent directional changes in flow.

Conway (1981) [63] did an extensive mathematical analysis of the function of circle systems with CO₂ absorber. He starts by recognising that a circle system can have a number of configurations and there can be different extents of gas mixing; using air as the gas he deduces algebraic expressions for alveolar oxygen concentration (FₐO₂) for total gas mixing and for selective fresh gas inspiration. For total gas mixing he shows that the FₐO₂ reduces to the well-known alveolar gas equation. For selective fresh gas inspiration he deduces separate equations for FₐO₂, for alveolar ventilation greater than FGF and less than FGF (although an error is noted in that equation). He recognises that his equations are necessarily simplifications, especially if account is not taken of the rate of nitrogen excretion, nor of the true degree of gas mixing in the system. A series of graphical relationships between FₐO₂ and FₐCO₂ is deduced for different amounts of rebreathing or CO₂ absorption. He then introduces the concept of rate of anaesthetic uptake and, assuming a FGF of 1.0 L.min⁻¹ of a gas mixture of 50% O₂ in N₂O, produces a further set of graphical relationships between FₐO₂ and FₐCO₂; the final result is shown in Figure 2.10; note that at high values of N₂O uptake alveolar oxygen concentration may actually be higher than inspired oxygen concentration, especially at low FGF; dangerously low values of alveolar oxygen concentration can occur at low values of anaesthetic uptake and low values of FGF; when N₂O uptake equals oxygen uptake, alveolar oxygen concentration is independent of FGF. Although modern anaesthetic
agents do not have a saturated vapour pressure high enough to make hazardous low alveolar oxygen concentrations likely, marked changes in alveolar anaesthetic concentration are likely to occur with changes in FGF and rate of anaesthetic uptake. Figure 2.11 shows this range of variation in relation to Halothane, taken from Conway’s analysis, showing particularly marked changes at low FGF’s. Conway also shows that there are relatively small differences in vented oxygen concentrations with a step change in FGF from 0.5 to 2.0 L.min\(^{-1}\). He introduces the concept of time constant of the circle system as:

\[
\text{Time constant} = \frac{\text{System volume}}{(\text{FGF} - \text{rate } O_2 \text{ uptake} - \text{rate anaesthetic uptake})}
\]

As FGF falls toward the sum of rate of gas uptakes the time constant becomes infinite. He recognises the limitation of such models in the absence of precise information about rate of oxygen and anaesthetic uptake (especially in the presence of changing ventilation/perfusion ratios in the lung), different system geometries, varying degrees of gas mixing, the presence of nitrogen (if N\(_2\)O is being used). He recommends twenty minutes at high FGF at the beginning of an anaesthetic to denitrogenate and to keep the time constant small, and tells us that low FGF’s buffer unwanted changes. He reminds us that high-level gas monitoring during clinical anaesthesia is mandatory for patient safety.

Circle systems can be used for paediatric anaesthesia, using a 1 L reservoir bag and smaller bore tubing. Despite the presence of valves, the work of breathing has been found to be acceptable [64].

### 2.6. Closure

This chapter has given an historical account of all anaesthetic breathing systems for the reader’s benefit, in a sense to show the evolution of such systems. However, due to the introduction, over the last ten years or so, of relatively expensive volatile anaesthetic agents and the subsequent requirement for economically low fresh gas flows, as well as the routine availability of gas monitoring, the circle system has taken precedence over others. For this reason, as well as for the wider application of low flow circle systems, the circle system was chosen as the object for analysis by Bath fp in this thesis.
Figure 2.10: Alveolar oxygen fraction ($F_{A02}$) plotted against FGF ($V_F$) for different values of $N_2O$ uptake ($V_{N2O}$). FGF consists of 50% $N_2O$ in $O_2$; $O_2$ uptake is 200 ml.min$^{-1}$, $N_2O$ uptake is 160 ml.min$^{-1}$. [Reproduced from reference 63, with permission]

Figure 2.11. Alveolar halothane concentration plotted against FGF ($V_F$) for different values of halothane uptake ($V_{HAL}$). FGF consists of 4% halothane in oxygen. All other assumptions and permissions are as in Figure 2.10.
Chapter 3
Introduction to Space Life Support Systems

The engineering effort and inventiveness which went into the design of space portable life support systems (PLSS) appears to be in contrast to that which went into the design of breathing systems for medical use. Space PLSSs are required to provide breathing gas and pressurization in the extremes of the space environment. By reviewing space PLSSs it may be that something can be learned from these designs for the benefit of anaesthetic breathing system design, for example by the use of venturis, which informs the subsequent research carried out by the author.

3.1. History
Suits and portable life support systems for use in space have been developed over the last forty five years mainly by Russian and American scientists, biomedical engineers and physicians in response to the needs of their respective manned space programmes.

Initially the life support systems (LSS) were not portable, since astronauts did not leave the confines of their spacecraft, although mass and volume were still major design constraints; figure 3.1 shows that even the earliest spacecraft LSS was a type of circle system, immediately recognizable to anaesthetists. Early ‘spacewalks’ had the astronaut tethered to the craft’s LSS by an umbilical cord (Figure 3.2). A self contained PLSS became necessary as the astronauts’ activities increased in duration and distance from the spacecraft, either in Earth orbit or on the lunar surface (Figure 3.3). In early years, the entire suit was constructed from flexible material, made as an overall with a removable helmet, and with the PLSS as a removable pack usually attached to the back. During the early space station years of Salyut, Skylab and Mir, the suits became semi-rigid with a hard upper torso to improve reliability and reduce suit gas volume (Figure 3.4). PLSS development has continued to allow astronauts to undertake periods of up to eight hours of work activity to construct the International Space Station (ISS) (Figure 3.5). In order to reduce the need to replenish supplies, PLSS development is now looking at regenerable systems of oxygen supply and carbon dioxide removal. In all cases, the (P)LSS is likely to be a closed, circle breathing and ventilation system, where good design and economy of use is likely to remain of paramount importance.
Figure 3.1: The earliest intravehicular suit worn by Yuri Gagarin (L) and its associated life support system after separation from the ejection seat (R). 1: oxygen bottle; 2: starter; 3: mechanism for switching over oxygen flow from the helmet ventilation line to the injector at low altitudes; 4: valve automatically opens to admit ambient air at low altitudes; 5: exhaust valve on neck dam; 6: injector; 7: CO₂ absorber. [from reference 70, with permission]

Figure 3.2: The first EVA’s in 1965 using spacecraft life support systems connected to the astronaut by umbilical. Alexei Leonov (L, USSR). Edward White (R, USA). [from reference 70, with permission]
Figure 3.3: The use of portable life support systems (PLSS) for EVA in Earth orbit (L) and on the lunar surface (R). [from reference 71]

Figure 3.4: The hard upper torso of the modern EVA suit. [from reference 71]
3.2. Technical Introduction
An intravehicular (IV) LSS is used to provide life support in the event of vehicular failure and the LSS, including the suit ventilation, is usually supplied from the vehicle. The early designs of such LSSs resembled those of high performance aircraft suits, since they had the common design features of providing protection against loss of cabin pressure and of incorporating G-trousers to prevent blood pooling in the peripheries in the presence of +G\(_z\) acceleration, which may be experienced during launch and re-entry. The IV LSS and suit also allows an astronaut to eject from the craft during launch or re-entry, and to survive a subsequent surface or water landing. The earliest extravehicular activities (EVAs) in 1965 by Leonov and White were carried out using these IV suits with the addition of a thermal insulating shell, a helmet visor filter and a PLSS (Figure 3.2). Since then, the different requirements of a spacesuit EVA compared to an IV LSS and suit have led to increasingly diverging design requirements. The IV suit became a lightweight multi-layer suit with a detachable helmet.

Figure 3.5: EVA to construct the ISS.
[from reference 71]

Shown in Figure 3.6, the Russian Soyuz TM IV suit has air supplied to it from the onboard supply in the pressurised cabin, and is fed to the suit through an umbilical, thence via separate ducts to helmet and limbs. If the cabin pressure falls below 53.3 kPa, an electro-pneumatic valve is automatically activated from the vehicle to supply oxygen to the helmet at 1.5 kg.hr\(^{-1}\). If the cabin depressurises completely, a suit pressure regulator automatically pressurises the suit to either 41.0 kPa or 26.3 kPa, depending on selection. The same suit pressure regulator also acts as a relief valve, opening at 46.7 kPa in the event of suit overpressure, and allows atmospheric air to be sucked into the suit at ambient pressures close to that at sea level in the event of cabin ventilation or
oxygen failure [6]. The IV suit of the defunct Soviet space shuttle ‘Buran’ was to have an independent unit controlling these functions, supplied by cabin air; in the event of cabin depressurisation, the suit would automatically change to a closed loop mode, purging the system with oxygen, thence using a low flow closed system; there was also the opportunity to manually change to the closed loop mode.

The US shuttle IV suit, (the launch and entry suit [LES]) consists of an integrated helmet and a counter-pressure and anti-exposure suit, designed to allow its wearer to survive for 40 minutes at 30 km altitude. On manual activation of the ‘green apple’, the associated emergency oxygen source supplies a suit mounted pressure regulator to give a minimum of 10 minutes oxygen at 37 L.min\(^{-1}\) at sea level from two oxygen cylinders each of capacity 0.98 L, pressurised to 20.7 MPa, reduced through a regulator to 0.48 MPa. The helmet is supported by a wire frame system on the shoulders, thereby reducing additional mass to the head and allowing free movement of the head within the helmet. To prevent fogging, oxygen is directed onto the visor by a spray bar. The pressure control system within the helmet delivers pure oxygen at a regulated pressure of 0.25 kPa above suit pressure while connected to the shuttle orbiter oxygen supply system or the emergency system. The suit pressure is regulated by an aneroid regulator to maintain

![Figure 3.6: LSS for the intravehicular suit used in Soyuz TM and Mir.](from reference 6, with permission)
a minimum suit pressure of 19.3 kPa [6]. The hydraulic circuit diagram is shown in Figure 3.7.

With an increasing number of EVAs the specific requirements of an EVA LSS and suit accelerated the separate design of the suit and PLSS for EVA. With the Salyut and Soyuz station complexes and US Shuttle and international space station systems, the fully flexible suits of Voskhod, Gemini and Apollo EVA systems gave way to a semi-rigid suit with a hard upper torso, and integrated PLSS for reliability. The LSS has to be portable and independent (PLSS), to provide a pressurised gas supply with an adequate inspired oxygen concentration, adequate ventilation to the face and around the suit, gas recirculation, carbon dioxide absorption, and thermal equilibrium in the face of different work rates. It became clear quite early on in the development of EVA suits that astronaut thermal equilibrium could not be met by gas ventilation of the suits alone; the development of the Liquid Cooling and Ventilation garment (LCVG) allowed the higher thermal conductivity of water to act as a cooler for the body, as well as insuring collection of ventilation gas from the periphery of the suit to be returned to the CO$_2$ absorber.

Within a spacesuit, the breathing gas is 100% oxygen, pressurising the suit to between 25.9 and 40 kPa; these pressures are therefore also the oxygen partial pressures, which are adequate to sustain life with a small margin of safety to minimise the risk of either oxygen toxicity or hypoxia. The suit pressure is kept to these low levels to maintain mobility in all parts of the suit, including the gloved hand, and this necessitates the use of a high inspired oxygen fraction with the attendant risk of flammability. The combination of a low pressure and the absence of nitrogen in the ventilation gas, when the astronaut’s pre-EVA breathing atmosphere in the space station or craft has been air at sea level pressure, means that decompression sickness (DCS) is an issue which must be considered in EVA planning. Safety in this regard is assured by carrying out a prebreathing schedule prior to EVA. Current US suits operate at a pressure of 28.5 kPa, and Russian suits operate at 40 kPa, with an option to drop to 27.6 kPa to improve mobility; the higher the suit pressure, the lower the risk of DCS and the shorter the prebreathe time. The atmosphere of Russian spacecraft has always been air at 101 kPa, while early US spacecraft atmospheres consisted of oxygen only at 34.5 kPa; the lower the cabin nitrogen partial pressure (pN$_2$), the easier it is to avoid DCS; however a pure oxygen environment is always a fire hazard, as was tragically demonstrated in Apollo 1. Current development includes a suit designed to be pressurised to 57 kPa in order to reduce the risk of DCS, and a hard shell suit designated as a ‘zero prebreathe suit’ (ZPS) [5]. The oxygen source in a PLSS has traditionally been oxygen gas stored in a cylinder, but other technologies, which have been considered, include liquid oxygen and water electrolysis, both of which have been used in past or present spacecraft LSSs. Liquid oxygen is being considered for a PLSS for use on Mars, using metabolic heat to vapourise it [7].
Chemical absorption of carbon dioxide from the exhaled breath means that the unused breathing gas can be recirculated around the suit and reused. The technology of CO₂ absorption for space PLSS has hitherto consisted of Lithium Hydroxide crystals, these being lighter and more efficient than Soda lime used in diving and anaesthesia applications. For other waste gases there is an activated charcoal filter. If the absorption system fails, inspired CO₂ starts to rise causing increased ventilation, headache, cognitive impairment and eventually unconsciousness. More recently CO₂ adsorption technology on ISS has consisted of zeolites, but this has not yet been developed for PLSS. In all technologies, the direction of development is towards regenerative techniques, and in habitats (as opposed to suits), towards bioregenerative techniques. Other technologies being investigated for future use in space stations to produce oxygen and absorb carbon dioxide, or sometimes both, include amine desorption, vapour electrolysis, and Sabatier and Bosch reactors.
The PLSS breathing system, which is functionally linked to the ventilation and pressurisation system, allows for recirculation of waste gases through the absorber, with only a relatively small amount of fresh gas being introduced per unit time. A venturi device powers the recirculation of the breathing gas and a fan drives the ventilation gas around the suit. The oxygen is supplied from cylinders and the freshest gas from the system is preferentially delivered near the face; waste gas for carbon dioxide absorption is collected from the extremities of the limbs of the suit. In both Russian and US suits, the PLSS is integrated with the suit. In the US suit it is attached to the outside as a backpack; the astronaut first dons the upper torso and backpack, then dons the lower torso. In the Russian suit, the PLSS is integrated into the back access door to the one-piece suit. It can be seen from the foregone description that while there are many similarities in the requirements for such a PLSS to a medical breathing system, the engineering constraints are essentially much greater for a physically working astronaut, breathing pure oxygen at a low partial pressure, than for a surgical patient breathing an anaesthetic gas mixture; nevertheless, it raises the question about whether a venturi could or should be used in an anaesthetic system.

The modern spacesuit consists of a multilayered structure, the innermost layer of which is a urethane coated nylon bladder, protected mechanically by a Dacron restraint garment and a multilayered micrometeoroid protection garment, consisting of neoprene coated nylon liner and a number of aluminized Mylar insulation layers, covered in an ortho-fabric cover (Figure 3.8). In between is the liquid cooling garment, consisting of a tricot liner, water transport tubing and a nylon/spandex outer layer (Figure 3.9). The multilayered structure also provides radiation protection equivalent to aluminium shielding of 0.5 g.cm\(^{-1}\), adequate for an EVA in an equatorial orbit, but not for EVA over the South Atlantic anomaly or in a polar orbit; nor does it provide protection against solar flares [5].

The average work rates of astronauts are between 400 and 500 W, though this can rise to 600 W. Russian studies have shown an average work rate of 290 W, with short term maxima of 910 W. As well as removing metabolic heat with the liquid cooled garment layer, the PLSS also has a sublimative heat exchanger to offload excess body water and heat [6] (Figure 3.10).

Figure 3.8: The multiple layers of the EVA suit.
[from reference71, with permission]
In the event of any system failure within the suit or PLSS, its design has to allow safe return to the airlock, which should allow entry and exit with minimum use of consumables and minimum effort. If there is a reduction in suit pressure for any reason, a secondary oxygen injector can be activated to allow limited continuation of activities.
If there is a loss of suit pressure, the emergency oxygen supply can be activated to allow return to airlock. If there is loss of glove pressure, an automated cuff is activated to allow immediate return to airlock. If the outer bladder layer fails, then a secondary previously redundant bladder automatically becomes operational. If the primary oxygen supply fails, the emergency supply is activated. If the CO₂ absorption cartridge fails, the secondary oxygen injector is activated to flush out CO₂ to allow return to airlock. If the heat exchanger fails, the injector acts as a backup cooling system until return to airlock (see table 3.2).

A measure of the risk of DCS is the ratio R, of the initial partial pressure of nitrogen being breathed pN₂, to the final suit pressure pₚₚ (whose pN₂ = 0). This risk is reduced by slowly offloading the nitrogen by prebreathing for a sufficient period of time prior to donning the suit. Using a higher suit pressure also reduces the risk. Figure 3.11 shows prebreathing time plotted against suit pressure for different cabin pressures and cabin oxygen concentrations. These curves will have been deduced from human trials using depressurization testing and Doppler detected venous gas bubbles as the surrogate marker for DCS. It demonstrates that both higher suit pressures and lower pre-EVA pN₂’s shorten prebreathe time. Figure 3.12 shows the current US shuttle pre-EVA prebreathe schedule; it shows a rather prolonged period of disruption for the whole crew, in order to minimize all nitrogen previously stored in body tissues, and a lot of effort is being expended in looking ways of shortening the prebreathe time without increasing the risk of DCS. The Russian EVA suit for use in Salyut, Mir, the ISS and Buran, has a rigid upper torso, a non-detachable helmet and a PLSS, with arms and lower torso made of flexible materials, and was first worn in 1977. Access to it is through a rear door, on the inside of which are mounted the main components of the PLSS (Figure 3.13); controls and indicators are mounted on the front of the rigid torso. The advantages of the semi-rigid structure over earlier designs include: smaller dimensions of the pressurised torso; simplicity of operation and enhanced safety and reliability resulting from the elimination of external lines connecting the PLSS to the suit, given that the PLSS is within the pressurised part of the suit, and hence not exposed to vacuum; the use of a single size suit for all cosmonauts; easy access to PLSS components for maintenance and repair; however, this design means electrical systems are exposed to a high oxygen partial pressure with the attendant risk of fire. Other components of the suit include a vacuum shield thermal insulation layer, a liquid cooling garment, special underwear, a biological sensors belt, and a headset and a detachable radio-telemetry unit.
The semi-rigid design of the Orlan-DM suit for EVA use from Mir space station and ISS is designed for the relatively high pressure of 40 kPa, reducing the cosmonaut’s prebreathe time prior to EVA. The shoulder pressure bearings and soft joints elsewhere allow relatively free mobility of the cosmonaut when the suit is pressurised at 40 kPa.
Under normal conditions, oxygen is provided to the cosmonaut, who selects one of two regulators, one of which is set at 39.2 kPa, the other at 26.5 kPa; it provides oxygen as required in response to a pressure drop inside the suit. The oxygen supply can also be initiated manually using an ejector device built into the helmet’s gas inflow duct. As well as providing breathable oxygen, the ejector also provides a gas flow of sufficient magnitude (1 kg.hr\(^{-1}\)) to clear CO\(_2\) or to maintain suit pressure or suit ventilation in the event of failure of any of the primary systems for these functions. In addition, there is a separate backup emergency oxygen supply system in the Orlan-DM and Orlan-DMA systems for use in the contingencies discussed above. If the pressure of the main LSS drops below 22.0 kPa, the emergency system is automatically activated by an appropriate regulator, to supply a maximum oxygen flow-rate of 3 kg.hr\(^{-1}\) [6].

Apart from cosmonaut oxygenation, there is also the matter of suit ventilation to help maintain suit pressure, and to remove CO\(_2\) and other gaseous waste products, as well as metabolic heat; there is a centrifugal fan and its backup driven by a sealed electric motor; under normal conditions the ventilation system provides 150 L.min\(^{-1}\); functioning together, both fans can compensate for a failure in the water cooling system and prevent helmet visor fogging. Figure 3.14 shows the LSS of the EVA suit used out of Mir and ISS. Ducts deliver ventilation to the helmet, and gas re-enters the closed system at the peripheries of the upper and lower limbs. The high concentration of oxygen within the suit is assured by a period of de-nitrogenating pre-breathing of oxygen. CO\(_2\) clearance is mediated primarily by a canister of LiOH crystals, though no detail is given of their function in the Russian literature [70].
Thermal regulation in the Russian suit incorporates a liquid cooling system, which the cosmonaut regulates manually, allowing the circulating water to either bypass or flow through the heat exchanger. Generated moisture is transported by the circulating gas to the heat exchanger, where it is condensed and channelled to the sublimator, shown in Figure 3.10. Both systems together ensure thermal equilibrium to 698 W of metabolic and physical activity [6].

The US Shuttle EVA suit (extravehicular mobility unit [EMU]) was first worn in 1983 and has a hard upper torso (HUT), made of fibreglass, and soft arms, gloves, and a separate lower torso, legs and boots; the helmet is detachable and the PLSS is mounted on the back of the HUT, controls and displays on the front. The astronaut first dons the lower torso, then the HUT through a waist opening. The connections between the PLSS and the HUT are internal, and the PLSS can be easily serviced without detaching it from the HUT. Suit design also includes a thermal and micrometeoroid garment (TMG), consisting of seven layers of fabric. Between the inner layers are five layers of aluminised Mylar film, while the outer layer consists of a woven fabric of Gortex, Teflon, Kevlar, and Nomex fibres. In a thermal environment, which can range from +113 to -118 °C, the maximum heat loss from the suit does not exceed 130 W and the maximum heat gain remains less than 88 W. The energy absorbing characteristics of the TMG layer prevents micrometeoroids from penetrating the suit pressure bladder.

The PLSS is a closed loop system, shown in Figure 3.15, which in conjunction with the suit itself, provides oxygen for breathing, suit ventilation and pressurisation, temperature and humidity control, and CO₂ removal. The suit operating pressure is designed to minimise the chance of decompression sickness, to provide adequate mobility and to provide an adequate minimum and non-toxic maximum level of inspired oxygen. The standard suit operating pressure is 29.7 kPa. There are two oxygen regulators, the primary of which is set to operate within a band of 29.7 ± 0.7 kPa over a flow range of 0.009 – 0.150 kg hr⁻¹ oxygen, and a range of supply pressure of 0.45 – 7.24 MPa. The secondary oxygen regulator operates within a pressure band of 23.0 – 26.9 kPa, and kicks in if suit pressure falls to within this range. There is also an emergency backup purge system to provide minimum suit pressure, ventilation, CO₂ removal and cooling in the event of failure of both oxygen regulators [6].
The primary oxygen supply contains 0.54 kg of oxygen stored at 6.2 ± 0.3 MPa, and is rechargeable from the orbiter’s supply. The emergency supply contains 1.2 kg oxygen and is designed to maintain the suit at the emergency pressure over a flow range of 0 – 2.4 kg. hr\(^{-1}\) of oxygen at a supply pressure range of 2.5 – 51.0 MPa. It provides 30 minutes of oxygen at 2.3 kg.hr\(^{-1}\) or 1 hour at 1.1 kg.hr\(^{-1}\), depending on which of two available purge orifices is manually selected [6].

Control of CO\(_2\) levels within the suit is by means of absorption of CO\(_2\) using a lithium hydroxide canister. LiOH is lighter and has a higher absorption capacity than Soda lime,
being able to absorb 0.917 kg CO₂ / kg LiOH. The canister contains 1.1 kg LiOH, which allows the pCO₂ in the suit to be maintained below 1.01 kPa for at least 7 hours under normal EVA conditions at an average work rate of 293 W at a Respiratory Quotient of 0.90. A warning is given to the crew member when the pCO₂ exceeds 1.07 kPa. CO₂ levels in the oronasal area are controlled by directing ventilation flow of 169.9 L.min⁻¹ over the crew member’s face and head, which has the effect of entraining exhaled air and directing it downwards, where it is returned to the main ventilation gas flow for CO₂, heat and humidity removal. The gas directed over the face is of very low humidity to prevent visor fogging, which is also helped by covering the inner surface of the visor with a detergent before EVA.

To deal with metabolically produced moisture and humidity, the PLSS uses a sublimator to condense water vapour out of the ventilation circuit. Normally the ventilation flow is cooled from 39.4 to 13.9 °C. As the water vapour is condensed it is carried along the walls of the sublimator into a ‘slurper’ array of holes, where a rotary water separator centrifugates the water back to the water storage system for reuse. The sublimator also has the function of using the water as a heat dissipation mechanism; the sublimator uses the phase change characteristics of water under near vacuum conditions. The rest of the device is concerned with storing and routing the heat carrying water from the astronaut to the sublimator and back. The sublimator contains an ice layer encapsulated in a metal porous mesh.
The water circulation loop pumps cooling water around a liquid cooling and ventilation garment (LCVG), which the astronaut wears under the suit, through which 108.9 kg hr\(^{-1}\) water circulates, whose water temperature can be adjusted by the wearer. The water pump is magnetically coupled to the EMU’s ventilation fan and routes the water to a cooling control valve which allows the wearer to divert a controllable and variable proportion of the LCVG cooling water to the sublimator for heat rejection. This gives cooling water in a temperature range of 7.2 - 27.2°C, which provides a greater degree of comfort than changing the water flowrate through the LCVG circuit, which could possibly expose the astronaut to low flow rates of very cold water. Since a sublimator requires a vacuum to function, during prebreathe operations, the EMU cooling water system is connected via an umbilical to a heat exchanger on board the orbiter. The EMU system is designed to deal with metabolic loads of 293 W, with peak loads of 469 W for 1 hour and 586 W for 15 minutes [6].

The LSS is designed to be fail-safe and the PLSS has no built-in redundancy. If the ventilation, cooling or CO\(_2\) removal systems fail, or if the suit develops an oxygen leak up to 2.3 kg.hr\(^{-1}\), then manual operation of a purge valve automatically activates the secondary oxygen pack. This ensures 30 minutes of oxygen in open loop purge mode,
which allows control of CO₂ (up to 2.0 kPa), some cooling, and allows the astronaut to return to the airlock in an orderly fashion. The design philosophy of the EMU is based with a number of considerations in mind. The failure of a single EMU element must not allow a catastrophic event to occur; excessive decompression, fire or explosion must be ruled out in the design. Critical subsystems with a high possibility of failure should have a redundant back up. Particular attention must be paid to fire safety and prevention, in view of the oxygen rich atmosphere. There should be automatic activation of systems to maintain suit pressure to support life. The design should have built in safety margins in terms of suit shell pressure, other mechanical loads and for electrical currents in a pure oxygen atmosphere [6].

Table 3.1 shows a list of design requirements for an EVA suit. Table 3.2 shows the processes within the LSS, which need to be enacted in order to meet a number of emergency situations.

<table>
<thead>
<tr>
<th>Space Suit Requirements</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pressurized volume and</td>
<td>Provision of oxygen, carbon dioxide removal,</td>
</tr>
<tr>
<td>respirable atmosphere</td>
<td>and pressure control</td>
</tr>
<tr>
<td>Thermal control and external</td>
<td>-</td>
</tr>
<tr>
<td>thermal insulation</td>
<td>-</td>
</tr>
<tr>
<td>Physical protection from</td>
<td>Micrometeoroids, sharp edges</td>
</tr>
<tr>
<td>external objects</td>
<td>-</td>
</tr>
<tr>
<td>Communication</td>
<td>With spacecraft and ground control</td>
</tr>
<tr>
<td>Radiation shielding</td>
<td>No shielding from Solar Flares and GCR, possible</td>
</tr>
<tr>
<td>Proper bioengineering</td>
<td>For crew members of different anthropometry (85 percentile)</td>
</tr>
<tr>
<td>characteristics of the space</td>
<td>-</td>
</tr>
<tr>
<td>suit</td>
<td>-</td>
</tr>
<tr>
<td>Proper physiologic and hygiene</td>
<td>Peak metabolic rate of up to 700 W Management of urine and feces</td>
</tr>
<tr>
<td>conditions</td>
<td>-</td>
</tr>
<tr>
<td>High reliability</td>
<td>-</td>
</tr>
<tr>
<td>Fire safety</td>
<td>In 100 % oxygen atmosphere</td>
</tr>
<tr>
<td>Durable lifetime</td>
<td>3-6 years onboard a spacecraft</td>
</tr>
<tr>
<td>Reusability</td>
<td>25-50 sorties with some spares onboard the spacecraft</td>
</tr>
<tr>
<td>Universality</td>
<td>Possibility to provide EVAs with different spacecraft</td>
</tr>
<tr>
<td></td>
<td>Usability in combination with MMU</td>
</tr>
<tr>
<td>Minimum weight of consumables</td>
<td>Feed water for coolant, Absorbent for carbon dioxide removal</td>
</tr>
<tr>
<td>Simplicity in operation and</td>
<td>Repair onboard the spacecraft possible</td>
</tr>
<tr>
<td>maintainability</td>
<td>-</td>
</tr>
<tr>
<td>Minimum weight and overall</td>
<td>-</td>
</tr>
<tr>
<td>dimensions</td>
<td>-</td>
</tr>
<tr>
<td>Minimum costs</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 3.1: Design requirements of an EVA spacesuit. [from reference 5, with permission]
3.3. Closure

A detailed introduction to spacesuits, both intravehicular and extravehicular, has been given. The extravehicular space suit and its associated life support system constitute a low flow closed system, much like the anaesthetic circle system. As such it is considered a subject worthy of study in order to learn what we can from a well engineered system, designed to a high standard.

Table 3.2: Procedures to activate the LSS in an EVA in emergency.[from reference 5, with permission]
Chapter 4

Introduction to Diving Breathing Systems

As with the PLSSs for use in space, life support systems for use underwater have also been much more extensively engineered than medical breathing systems to safely provide breathing gas under the extreme conditions of temperature and pressure encountered underwater. A review of these systems will provide some additional information to inform the subsequent design of medical breathing systems.

4.1. History [from reference 72, Edmonds, Lowry, Pennefeather & Walker, 1992]

Standard rig diving dress was introduced in 1837 by Augustus Siebe, consisting of a rigid helmet attached to a flexible waterproof suit, to which air was pumped down from the surface to the diver, as shown in Figure 4.1. The suit allowed the diver to alter his position without risk of flooding the helmet, and to alter the volume of air in the suit, and thus also the buoyancy and depth. The Siebe suit was found to be the best option by the Royal Navy when exploring the wreck of the ‘Royal George’ at that time, and has evolved and survived into modern times. Modern derivatives of the Siebe apparatus are enshrined in the US Navy Mark 5 and Mark 12 sets, as shown in Figure 4.2. The evolution of the modern diving suit allowed the limits of diving activity to be extended, leading to the occurrence of complications such as decompression sickness. This in turn led to the development of diving medicine, whose originator can be considered to be Paul Bert in the mid nineteenth century. In the early twentieth century, Haldane did the necessary experimental work, using goats to develop safe diving tables to avoid decompression sickness. Diving to depths greater than 60 m led to the recognition of nitrogen narcosis and oxygen toxicity when breathing air.

Figure 4.1. The first diving helmet of Augustus Siebe

Figure 4.2. Diving helmet and dress worn in the nineteenth and early twentieth centuries

[from reference 72, with permission]
The early development of self contained underwater breathing apparatus (SCUBA) dates back to the mid nineteenth century. Condart developed a device in which air was stored in a copper pipe wrapped around the diver’s body, and was released into a hood. Carbon dioxide was washed out through a small hole using the fresh gas flow. The device of Rouquayrol and Denayrouze, developed in 1865, which is shown in Figure 4.3, consisted of an air supply from the surface to the suit (after the manner of Siebe), from which the diver could temporarily disconnect himself for a few minutes at a time to use the air stored in the suit. By 1918, Ohgushi had developed the first system with an independent air supply, the gas flow into the mask being controlled by a valve activated by the teeth. It was not until 1943 that Cousteau and Gagnan had designed the first SCUBA system with a demand valve. It was this valve, which allowed air to be supplied to the mouthpiece, triggered by a fall in airway pressure caused by the diver’s inspiratory effort. Furthermore, it was designed to deliver the air at a pressure appropriate to the depth.

Closed circuit systems were developed during the same period. These are systems in which oxygen is supplied at relatively low flow from a self contained supply, carbon dioxide is chemically absorbed by alkali, and residual gas is recirculated. The first of these, invented by Fleuss in 1878, consisted of an oxygen system, in which the carbon dioxide was absorbed by rope soaked in caustic soda. Because of the increased mobility afforded by this system, it was used in clandestine military operations, fire fighting and
rescue from flooded mines. Considerable risk was entailed in its use, since little was known of the effects of oxygen toxicity at this time.

Mixed gas diving was developed between the wars. Initially, it was a concept designed to reduce the risk from decompression sickness by reducing the nitrogen fraction in the breathing gas, usually in conjunction with a system in which carbon dioxide is absorbed. Similarly, the use of helium with oxygen was developed as a means of reducing the risks of decompression sickness and nitrogen narcosis, and of reducing the work of breathing. Additionally, by using mixtures with low oxygen concentrations, the risk of oxygen toxicity could be minimised. However, helium adversely affects voice communication and has a high thermal conductivity, resulting in significant heat loss. Breathing systems for modern gas mixtures have sophisticated electronic mixing and monitoring systems to ensure the appropriate gas mixtures at different depths. In order to reduce the need for decompression between dives, the concept of saturation diving was developed after WWII, in which divers and tunnel workers could live under increased atmospheric pressure during daily rest periods, and only slowly decompressed at the end of a prolonged duty period.

4.2. Current technology [from reference 73.]

4.2.1. SCUBA systems
The most common underwater breathing system is the SCUBA. Most systems operate using two stage pressure reduction valves. The first stage, shown in figure 4.4, reduces the gas pressure from the cylinder operating pressure of about 15 – 20 MPa (150 – 200 ATA), to a pressure of approximately 10 atmospheres above the diver’s ambient pressure; this ensures optimal performance of the second stage reduction valve. The valve is held open by the force of the spring until the pressure above the piston rises sufficiently to close it. The first stage valve then opens as gas is inspired by the diver and closes on the diver’s expiration. Water entry to the water chamber of the first stage valve helps to keep it open, which ensures that it is depth compensated.
Figure 4.5a and 4.5b shows the second stage regulator, containing a demand valve. On the diver’s inspiratory effort, the pressure in the mouthpiece is reduced and the diaphragm is depressed, thereby moving a lever, which opens the valve to admit gas into the housing from the first stage output. On expiration, the pressure in the housing increases and the diaphragm bends back to close this valve; this increase in pressure also allows the exhaust valve to the water to open to permit exhalation. If the diver increases his depth, the diaphragm is further compressed, which tends to open the inlet valve more readily, to equilibrate the second stage regulator to the increased ambient pressure. Manually pressing the purge button also opens the gas supply to the second stage. Some second stage regulators have a venturi within them, which enhances and prolongs the low pressure generated within the housing with inspiration, thereby assisting inspiration. Some regulators have a pilot valve assembly so that a very small drop in pressure with inspiration results in an amplified flow from the first stage. Both techniques are designed to assist inspiratory flow and reduce respiratory work.
The gas flow through such a regulator is unsteady flow, approximating to a sinusoidal waveform with respiration, with a peak flow rate of about three times the diver’s minute volume flowrate. The maximum flow rates for both first and second stage regulators are determined by orifice theory, and are therefore proportional to their respective upstream pressures, and independent of their downstream pressures. The mass flow rate demanded by the diver’s respiration is directly proportional to ambient pressure, and although the capacity of the regulator also increases with ambient pressure, it does so to a lesser extent. The performance of the regulator can be assessed by the peak pressures achievable during the respiratory cycle if they are linearly related to the flow rate achieved. While this relationship holds for conventional regulators, for those with venturi, vortex or pilot assist devices, such a relationship only exists for a very small fraction of the respiratory cycle, and does not represent a large change to the respiratory work of a breathing cycle. There is a very firm upper limit to the volume flowrate available from the regulator, and that limit decreases with depth and with decreasing cylinder pressure.
4.2.2. Remotely supplied diving helmets

Professional divers do not need to be as mobile as sports divers and are consequently often well served by a gas supply from the surface or a diving bell using an umbilical hose to a professional diving helmet, an example of which is shown in Figure 4.6. The gas supply can be from a high-pressure source, with a first stage regulator close to it; the umbilical hose to the diver then supplies the second stage regulator on the diver’s helmet. Alternatively a low-pressure compressor may be used to feed directly to the helmet. In either case there is normally a backup system from a cylinder on the diver’s back. Traditionally, surface supplied gas was supplied as free flowing gas to the helmet and the suit. The problem with this was that the gas flow also had to be adequate to clear carbon dioxide, which put considerable demand on the gas supply and potentially resulted in gas wastage; some of the dynamics of gas equilibrium in such a helmet are shown in Figure 4.7. Under these conditions it is possible to fit a CO$_2$ absorption system and a rebreathing system in order to recirculate gases. The efficiency of such a rebreather is enhanced by using a venturi device to accelerate the gas flow round the system, as shown in Figure 4.8.

Modern diving helmets of the Kirby Morgan type have a number of advantages over older equipment. A demand valve allows economic use of gas. A sealing mask within the helmet reduces the chance of a diver drowning and allows verbal communication. There is a primary gas supply, along with communications and heating lines from a surface supply or a diving bell. A secondary gas supply is available from a cylinder on the diver’s back. Where appropriate, exhaled gas can be returned to the diving bell for
Figure 4.7. Gas volume and pressure dynamics of a diving helmet.  
$V=\text{volume}, \ P=\text{pressure}$. Suffixes $S=\text{supply}$,  
$E=\text{expired (from lungs)}$,  $H=\text{helmet}$,  
$A=\text{ambient}$,  $X=\text{expired (from helmet)}$,  $T=\text{tidal (volume)}$.  
$C$, $D$, and $G$ are constants.  
[from reference 73, with permission]

recirculation, using two pumps (‘push-pull’ system), the dynamics of which is shown in Figure 4.9. In considering the design of ventilated helmets, the two most important factors are maintenance of a safe CO$_2$ partial pressure (pCO$_2$) and maintenance of safe pressures within the helmet, as shown in Figure 4.7. In general terms the pCO$_2$ within the helmet is directly proportional to metabolic CO$_2$ production and inversely proportional to helmet gas flowrate. The extent to which exhaled gas and fresh gas mix together and the helmet volume also have some bearing on helmet pCO$_2$. An acceptable upper limit to pCO$_2$ is 20 millibar. When a helmet is used with a full volume dry suit, helmet and suit pressure are identical, the suit itself acting as a counterlung. The diver has to manually control both over- and under-pressure of the suit or risk difficulties with buoyancy, but does not usually risk barotrauma. When a helmet is used with a neck dam, any helmet overpressure immediately becomes lung overpressure and a barotrauma risk if this is in excess of 60 – 100 millibar, unless chest expansion is restrained in some way, either actively or passively. The exhaust valve depicted in Figure 4.7 is carefully designed to open at a pressure to prevent barotrauma. Whether this is the actual outcome depends on the diver’s orientation, since the hydrostatic pressure difference between the exhaust valve and the lung centroid is 35 millibar. An additional safety feature to prevent barotrauma is the ability of the neck dam to allow excess helmet gas pressure to be released from the helmet to the suit, assuming it is properly applied during
suit donning. Furthermore, the sinusoidal nature of the diver’s respiratory pattern results in a cyclical change in helmet pressure which is somewhat attenuated by a small amount of cyclical displacement of the neck dam. A breathing pattern which minimises pressure swings within the helmet is likely to be one in which the required minute volume is achieved by lower than normal tidal volumes and higher than normal respiratory rates. This pattern may not be the most efficacious for CO$_2$ clearance. If a venturi device is fitted together with a CO$_2$ absorber in order to recirculate gases as shown in Figure 4.8, the pressure swings within a helmet using a neck dam are likely to be higher.

The description above indicates that, due to the higher values of absolute pressure and gas partial pressures, and the potential for greater pressure swings with breathing, the physiological demand by an exercising diver on such underwater breathing systems is much greater than for medical breathing systems. In addition, the need to change the inspired fractional gas concentrations with depth, and to monitor those concentrations with accuracy, means the engineering quality of such systems is required to be higher than for medical breathing systems.
4.2.3. Closed and semi-closed breathing systems

As with the other applications of closed or semi-closed rebreathing systems under discussion in this thesis, their use in diving facilitates a major reduction in supply gas wastage. In the diving context, this is accentuated by increasing ambient pressure. Their disadvantage is in their additional complexity. The components of a diving rebreathing system are shown in Figure 4.10 and 4.11, and are essentially the same as those found in an anaesthetic or a space suit circle system. Sometimes, for simplicity, the circle is replaced by a single ‘to-and-fro’ tube as shown in Figure 4.10a. Otherwise, as indicated, expired gas from the diver circulates through a CO₂ absorber into a reservoir bag (‘counterlung’) and back to the diver. Fresh gas is supplied from cylinders to the system through a flow regulator into the counterlung or reservoir bag, and one way flow is assured by unidirectional valves. Thus, with each circulation of fresh gas, whether it be an oxygen/nitrogen or an oxygen/helium mixture, the unused portion of oxygen and the inert gas component can be recirculated. There is also a valve to vent excess gas from the counterlung. As with CO₂ absorption systems in other applications, the CO₂ absorber contains hydroxide granules which function chemically according to the following equation [72], where M is a bivalent metal, usually Ca²⁺:

\[
\text{M(OH)₂ + CO₂} \rightarrow \text{MCO₃ + H₂O.}
\]

Where the system is used as a low flow closed system, using only oxygen as the fresh gas, the counterlung is initially filled with oxygen, which poses the potential danger of oxygen toxicity as the oxygen partial pressure (pO₂) rises with increasing depth. For this
reason the operational depth of this system is limited. However, unless the diver prebreathes oxygen, the gas to the counterlung gradually becomes diluted with the diver’s exhaled nitrogen, previously stored in the lungs and dissolved in other body tissues. Another possible danger therefore is that the gas mixture in the counterlung may become an hypoxic one. However the lack of exhaled gas bubbles from the system means that it is advantageous for shallow, clandestine military operations.

The greatest economy of gas usage is achieved with the low flow closed system, using a pair of gases, say oxygen and helium, supplied to the counterlung from separate cylinders on demand, as shown in Figure 4.12. Against gas economy, greater complexity is introduced: there is a fuel cell oxygen sensor at the outlet from the counterlung to the diver’s inspiratory limb, which triggers further oxygen supply to the counterlung to maintain an appropriate pO$_2$. It does this by comparing the measured pO$_2$ against a desired pO$_2$ and operates a solenoid valve to add more oxygen if necessary, or to operate an alarm. To add to safety there are three oxygen sensors. There is also a counterlung volume detector, which triggers further helium supply to the counterlung as necessary to maintain its volume and pressure, or triggers a valve to release inert gas from the counterlung in the event of excess pressure in the system. Should these devices fail, it is possible to manually override them.

Such mixed gas systems can be used at somewhat higher FGF in a semi-closed way with some gas loss through a relief valve; this allows some of the gas savings of a closed system to be realised without the depth limitation of the oxygen closed system, and without the relative complexity of the mixed gas closed system. The oxygen concentration in the diver’s inspired gas is a balance between the gas flow into the counterlung, the diver’s gas consumption, and gas loss through the relief valve. The gas flow and composition are set to accommodate the proposed dive. The composition is set first to allow as high a pO$_2$ as possible to avoid oxygen toxicity at depth. The flow is then chosen to ensure adequate oxygen supply at the surface. There are several methods for metering fresh gas supply in semi-closed systems. One is to use a constant mass flow rate of a fixed gas mix through a sonic orifice, whose flow rate therefore depends only on upstream pressure. Fig 4.13 shows the representation of such a system and the calculations used to set flow and oxygen concentration.
Figure 4.10. Semi-closed rebreathing systems. a) to and fro; b) circle.[from reference 72, with permission]

Figure 4.11. Closed circuit, mixed gas rebreathing system.[from reference 72, with permission]
Another method of setting flowrate and oxygen concentration in semiclosed systems is the constant volume ratio device. This is a design of the French Navy and follows oxygen requirement in response to workload. The exhaust bellows and the counterlung are mechanically linked, creating a constant ratio between the exhaled volume and the counterlung volume. The design works most economically between specified depths.
A third method of gas metering for these systems is the constant mass ratio device. The gas mixture is supplied to the counterlung via a dosage cylinder whose operation is controlled by the movement of the counterlung bellows. This gives a more economical use of fresh gas than the constant volume ratio device, since gas flow does not vary with depth. It is also more economical than the constant mass flow device. All of the systems described give a gas supply of fixed oxygen concentration, which means an increasing pO$_2$ with depth. Therefore the diving system has to include two or three different gas mixtures for use at different depth ranges.

It is possible to achieve a constant pO$_2$ irrespective of depth by metering the oxygen to the counterlung at a constant mass flowrate, while the inert gas supply varies with absolute pressure to give a constant volume flowrate. In a device for ensuring these conditions, the gas mixture is fed to the counterlung via a laminar flow element, across which there is a constant pressure drop. This is maintained by feeding back the output pressure as a reference ambient pressure to the regulator controlling the inert gas flow. Through the laminar flow element, the gas flowrate is a function of the pressure drop across it, its geometry and gas viscosity, all of which are independent of ambient pressure, resulting in a constant volume flowrate. Another variant on this device achieves
a constant volume flow rate of inert gas from a mass flow jet supplied by a pneumatic amplifier, whose output pressure has a constant ratio to ambient pressure. This allows the inert gas mass flow rate, which is proportional to the upstream pressure, to rise in proportion to ambient pressure.

A variant on the closed system design is the ‘push-pull’ apparatus. The supply gas is ‘pushed’ from a central processing station to the helmet, as in a surface supplied helmet. However the exhaust gas is ‘pulled’ from the helmet to the central processing station, where the CO₂ is absorbed and fresh gas recirculated. The central processing station in this case is often the diving bell itself, and can be used to supply several divers, providing their operating depths are not too dissimilar. The exhaust side of the system must be carefully designed to avoid over- or under-pressure of the helmet, and this is governed by the presence of an overpressure exhaust valve and an exhaust back pressure regulator to prevent under-pressure, as shown in Figure 4.9.

4.2.4. Carbon dioxide absorption systems

In a similar way to breathing systems for other uses such as anaesthesia and space portable life support systems, the performance of CO₂ absorber system is crucial to the efficacy of low flow closed circle systems. The materials frequently used are ‘Baralyme’ and ‘Sodasorb’, the latter being available as type A, medical grade and high performance. The latter two variants of ‘Sodasorb’ have a higher moisture content (14-19%) than type A or ‘Baralyme’ (12-14%), and should therefore have a better performance at low temperatures or relative humidities. High performance ‘Sodasorb’ has a greater porosity than medical grade and should therefore have a higher absorption rate at high gas flow rates. The endurance of canisters of type A or medical grade Sodasorb have been found to be between 30 and 110% greater than that of a canister of Baralyme, particularly at low water temperature or low gas humidity. High performance Sodasorb gave 100% greater duration than Baralyme at similar water temperature and gas inlet humidity. The presence of helium rather than nitrogen reduces canister duration by 15%. High performance Sodasorb has been found to be 70 – 300% more efficient than medical grade Sodasorb, the difference being greatest at low water temperature and relative humidity. High performance Sodasorb therefore appears to have the longest duration.

The efficiency of absorbent is directly proportional to inlet gas temperature and relative humidity; it is also proportional to external water temperature, which can increase heat loss and thus absorbent bed temperature. A gas inlet temperature and relative humidity of at least 20°C and 50% respectively can maintain maximum absorbent efficiency of 100 – 240 litre CO₂ per kg of absorbent, providing poor design and cold water are not also present.
There are three possible gas flow configurations through the absorber, horizontal, vertical and annular. It has been shown [73] that the performance of all three is similar unless low gas inlet temperature is present (<10°C), in which case the vertical flow design experiences a 20% reduction in efficiency, whereas the other flow designs experience a 80% efficiency reduction. This is presumably because the vertical flow design allows moisture condensate to collect across the gas flow path to contribute to an overall increase in relative humidity of the gas, even in the face of an increasing gas temperature through the absorber, thus increasing absorber efficiency. In order to minimise heat loss from the breathing system, either the umbilical supplied gas must be heated, or careful consideration must be given to breathing system configuration of the self contained diver.

4.3. Closure
A summary of diving life support systems has been given, including the closed and semi-closed low flow systems in which carbon dioxide is absorbed. These circle systems therefore lend themselves to further study in order that functional improvements may be suggested for other circle systems, such as the ones used in anaesthesia.
Chapter 5.

Modelling breathing systems: Bath fp

5.1 Introduction

A breathing system is a reasonably complex collection of straight and curved tubes, of relatively wide diameter, corrugated to encourage gas mixing, unidirectional valves, right angle turns, T pieces, a fresh gas entry point, an overflow valve, and a reservoir bag to act as a buffer against gas concentration changes. The mechanical characteristics of these individual components were fully investigated by M. Jones in this department, and are reported in reference 77. Figure 5.1 shows the Bath fp icon for such a system, with components labelled to correspond to the Bath fp model; this is a reasonably accurate diagram of such a system as used clinically. Some systems might also contain a venturi to accelerate flow round the system (see chapters 8 and 9), and low flow systems generally also have a CO₂ absorption canister, packed with hydroxide crystals as indicated in figure 5.1.

Looking at figure 5.1, the iconic Bath fp model of the standard configuration of the anaesthetic circle system, FGF enters the system where indicated in the top right hand corner. At the far left, the patient takes an inspiratory breath, which opens up the
unidirectional inspiratory valve AOV2(1), and encourages the FGF towards the patient along the (upper) inspiratory limb of the system, tubing APO7(1); however, depending on the relative magnitudes of the FGF and the patient’s inspiratory flow rate, a proportion of the FGF is diverted retrogradely (downwards) through the carbon dioxide absorber ACS1(1), against a closed expiratory unidirectional valve AOV1(1), some for storage in the reservoir bag ARB1(1) and some for excretion through the overflow valve AEV1A(1). On the patient’s expiration, the inspiratory unidirectional valve AOV2(1) closes, the expiratory unidirectional valve AOV1(1) opens and expiratory gas flows down tubing APO7(2); depending on the relative magnitudes of the FGF and the patient’s expiratory flow rate, some of the expiratory gas is excreted through the overflow valve AEV1A(1), some is stored in the reservoir bag ARB1(1), and a proportion is recirculated through the carbon dioxide absorber ACS1(1), where the CO$_2$ is absorbed, the remainder of the expiratory gas being recirculated for the next inspiration.

Each of these components has its own thermodynamic and hydraulic characteristics when tested in isolation, and equations are implemented in Bath fp, which can model the component in general terms as either a capacitive or a restrictive element. Capacitive elements include tubes and reservoir bag, and require analysis in terms of pressure, volume and gas concentration changes. Restrictive elements include the venturi, tubes, valves and orifices and require analysis in relation to pressure drop and associated flow rate; orifices can also occur at entry to and exit from capacitive elements; some components have behavioural characteristics of both elements. The model then has to define hydraulic characteristics of the linkages between components, and finally to characterise the behaviour of the whole system. This means doing calculations, which produce pressures and pressure drops, flow rates and volumes, and gas concentrations, for a large number of points in the system. This is the essence of Bath fp, which has been used to model diving breathing systems [75], submarine escape systems [76] and high flow anaesthetic breathing systems [77], from which more details of the derivation of Bath fp equations may be found. The rest of this chapter contains a relevant summary of these works.

Certain assumptions are made in relation to modelling of the system. Some energy sources are ignored, including heat generation due to vibration and radiative heating, and kinetic and potential energy, because these changes are negligible compared to the other energy terms of internal energy and that due to volume and pressure changes. A perfect gas obeying the gas laws is assumed, undergoing processes, which are isothermal and including a mix of isothermal and adiabatic changes. Uniform gas velocity across the cross section is assumed, since this simplifies all subsequent calculations. A lumped parameter model can be used with the assumption of perfect gas mixing within the capacitive elements, resulting in uniform pressure within the element. Incompressible gas
flow is assumed for small pressure drops, although high flow rates may result in some compressibility, which is accounted for.

5.2 Capacitive Elements
A capacitive elemental volume undergoing change in a small period of time is shown in figure 5.2. The energy change across its boundaries is given by (see list of symbols)

\[ \Delta E_v = E_{in} - E_{out} - \Delta W_{out} + \Delta H_{in} \]  \hspace{1cm} (1)

For the application under investigation, it is possible to neglect internal heat generation due to vibrational or radiative heating, or due to kinetic and potential energy (except in the case of a nozzle or a diffuser). These are small in relation to internal energy and specific enthalpy terms. With these limits in mind it is possible to deduce the rate of change of energy terms as

\[ \dot{m}u_v + \dot{m}_v = \Sigma \dot{m}_{in}h_{in} - \Sigma \dot{m}_{out}h_{out} - \dot{W}_{out} + \dot{H}_{in} \]  \hspace{1cm} (2)

Using the general thermodynamic relationships in Table 5.1, it is possible to deduce from this expression, the rate of change of temperature in the gas volume.

\[ \dot{T} = \frac{1}{mC_v} \left\{ \sum \dot{m}_{out}C_pT_{out} - \dot{m}C_vT - \dot{W}_{out} + \dot{H}_{in} \right\} \]  \hspace{1cm} (3)
and it is noted that $T_{\text{out}} = T$. The rate of heat transfer, $\dot{H}_{\text{in}}$, can be represented by consideration of heat conduction, convection and radiation. Given the complexity of determining heat transfer coefficients, and because this energy term is small, in this application it is possible to neglect it. For breathing systems, the work done on or by the gas is usually attributable to the change of gas volume, so that $\dot{W}_{\text{out}} = PV$, remembering that negative values represent work done on the gas.

By differentiating the gas law $P.V = m.R.T$ and rearranging, it is possible to get an expression for the rate of change of pressure in the volume.

$$
\dot{P} = \dot{m} \frac{RT}{V} - \dot{V}P + \dot{T} \frac{mR}{V}
$$

(4)

If a polytropic process is assumed, and if the substitutions in equation (3) and the assumptions described above, are put into equation (4), then rate of change of pressure becomes:

$$
\dot{P} = \frac{n}{V} \left\{ R \left[ \sum \dot{m}_{\text{in}} T_{\text{in}} - \sum \dot{m}_{\text{out}} T_{\text{out}} \right] - \dot{V}P \right\}
$$

(5)

<table>
<thead>
<tr>
<th>Definition</th>
<th>Specific internal energy, $u$</th>
<th>Specific enthalpy, $h$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$C_p - C_v = R$</td>
<td>$C_v = (du/dT)_v$</td>
<td>$C_p - (dh/dT)_p$</td>
</tr>
<tr>
<td>$\gamma = C_p / C_v$</td>
<td>$u = C_v T$</td>
<td>$h = C_p T$</td>
</tr>
<tr>
<td>$h_o = u_o + p_o V_o$</td>
<td>$u = u_o + C_p (T - T_o)$</td>
<td>$h = h_o + C_p (T - T_o)$</td>
</tr>
<tr>
<td>$p_o V_o = RT_o$</td>
<td>$u = h_o - C_p T_o + C_v T$</td>
<td>$h = h_o - C_p T_o + C_v T$</td>
</tr>
</tbody>
</table>

*Table 5.1. Some general thermodynamic relations (from reference [76])*
Where $\dot{P}$ is small, with slow gas movement, heat transfer results in constant temperature, an isothermal process, and $n = 1$. For rapid gas movement $\dot{P}$ is large, heat transfer is small in the time element, the process is adiabatic, and $n = \gamma$. Once $\dot{P}$ is determined, $\dot{T}$ can be determined by rearranging equation (4). In this ‘lumped parameter theory’, it is assumed that pressure and temperature throughout the gas volume be uniform.

5.3 Gas Mixtures

Bath fp includes consideration of the effect of the changing composition of four different gases in the mixture, $O_2$, $N_2$, $CO_2$, and $N_2O$, and their fluid and thermal properties. Based on Dalton’s law, it is possible to derive an expression for the molecular weight of a gas mixture, and its specific heat at constant pressure:

$$C_p = \frac{1}{V.Mo} \sum \left[ V_i.Mo_i.C_{p_i} \right]$$

(6)

whence the other thermodynamic constants, $C_v$ and $\gamma$, for the mixture can be obtained. In addition Bath fp uses reference [78] to determine viscosity of the gas mixture from the volume fractions and viscosities of the individual gases in the mixture. Reid et al [78] also suggest that the viscosity of gas constituents is independent of pressure, but gives a relationship for viscosity variation with temperature. Table 5.2 (in appendix) gives the information needed to determine all the properties of a gas mixture.

5.4. Restrictive Elements

A restrictive element is defined as one, which requires a significant local pressure drop to produce gas flow through that element. An example of a fixed restrictive element is an orifice; an example of a variable restrictive element is a valve. For breathing systems, compressibility is a relevant consideration, and the theory of compressible fluid dynamics is important. A standard expression for mass flow rate through an orifice is:

$$\dot{m} = C_d C_m A \frac{P_u}{\sqrt{T_u}}$$

(7)

and the value of $C_m$ depends on whether the flow is subsonic or sonic. If $(P_d/P_{up} \geq [2/((\gamma+1))]^{\gamma/(\gamma-1)}$ then the flow is subsonic and
Otherwise the flow is considered sonic and therefore choked, in which case:

\[ C_m = \frac{2\gamma}{R(\gamma-1)} \left[ \left( \frac{P_d}{P_u} \right)^{2/\gamma} + \left( \frac{P_d}{P_u} \right)^{(\gamma-1)/\gamma} \right] \]  

(8)

although, of course, sonic flow is unlikely in this particular setting. At a maximum volumetric flow rate of 25 L.min\(^{-1}\), the mass flowrates being considered here are unlikely to exceed between 0.5 \(\times\) 10\(^{-3}\) and 0.75 \(\times\) 10\(^{-3}\) kg.s\(^{-1}\), depending on the gas mixture being used.

By considering enthalpy and kinetic energy changes through the orifice, a relationship between velocity and temperature can be deduced, assuming the upstream and downstream velocities are negligible compared to the throat velocity:

\[ C_p \cdot T_u = C_p \cdot T_f + \frac{v_f^2}{2} = C_p \cdot T_d \]  

(10)

For a valve it is possible to derive its equation of motion taking account of the mass of the plate, the force of the spring if one exists, friction force, and viscous and momentum forces. For an APL (airway pressure limiting) valve:

\[ m.\ddot{x} = F_{press} - F_{spring} - m.g - F_{flow} - F_{damp} \]  

(11)

For a unidirectional plate valve, a similar equation holds, without the presence of \(F_{spring}\).
5.5. Tubes

In modelling tubes, as their volumes can be considered fixed, it is possible in the first instance to treat them as capacitive elements [76]. The same expressions for rates of change of pressure and temperature as discussed above can therefore be used. However, flow down any pipe, especially a corrugated one, is not frictionless and this effect must also be included. It can be assumed that the time delay for a pressure wave along a length of anaesthetic tubing is negligible and therefore a ‘lumped parameter’ model can be used, in which the pressure variation along the tube may be assumed to be uniform, and the flow may be considered to be one-dimensional. Under these conditions the momentum equation may be written as:

\[
\frac{dP}{\rho} + v \cdot \frac{dv}{dl} + \frac{4f_c \cdot D_p}{2} \cdot \frac{v^2}{l} = 0
\]

where the final term represents frictional loss. The solution of the equation requires the relationship between pressure and density, and this depends on the amount of heat transferred in the process. Solutions can be derived for both adiabatic and isothermal conditions, and a combined equation for the mass flow can be obtained:

\[
q_f = A_p \sqrt{\frac{\left( \frac{P_u^2}{R.T_u} - \frac{P_d^2}{R.T_d} \right)}{4f_c \cdot L + \frac{n+1}{n} \ln \left( \frac{P_u}{P_d} \cdot \frac{T_u}{T_d} \right)}}
\]

where \( n = \gamma \) for adiabatic condition and \( n = 1 \) for isothermal. There are additional pipe losses due to bends and changes in diameter, which are accounted for by calculation of the effective length \( L_{\text{eff}} \):

\[
L_{\text{eff}} = L + \frac{K}{4f_c \cdot D_p}
\]

where the frictional loss can be considered to be concentrated in one part of the tube and the value of K depends on the change of flow angle. Even in a straight tube, the lumped parameter analysis allows for the modelling of a friction orifice as a restrictive element, either at one end or other of the tube, or at both ends or in the middle of the tube. Such K factor losses also occur at right angled connection pieces, T-pieces and Y-pieces.

The friction factor \( f_c \) varies depending on whether flow is laminar or turbulent. If laminar, in which case Reynolds number \( Re < 2000 \), then
\[ f_c = \frac{16}{\text{Re}} \]  

(15)

If turbulent flow exists, then the Colebrook equation is used in a modified form as a function of \( f_c, F(f_c) \), in order to use Muller’s iterative method to deduce \( f_c \):

\[
\frac{1}{\sqrt[4]{4f_c}} = -1.74 + 2\log_{10}\left(2r + \frac{18.6}{\text{Re}\sqrt[4]{4f_c}}\right) = F(f_c)
\]  

(16)

At the beginning of the iteration the value of \( f_c \) is set to be the mean of the lowest and highest values, which can be estimated from the Moody diagram. As the iteration proceeds and \( F(f_c) \) approaches zero within an acceptable tolerance, the iterative process is terminated, and the last estimated value of \( f_c \) is used. It is also necessary to consider the situation where the flow is choked, since this will occur when the ratio of downstream to upstream pressure is less than a certain value. The choked mass flow rate is given by:

\[ q_c = \rho_c v_c A_p = P_c \sqrt{\frac{n}{R.T_d}} A_p \]  

(17)

where \( n = \gamma \) for adiabatic and \( n = 1 \) for isothermal conditions. If this is put back into equation (13), and calling the downstream pressure the choked pressure \((P_d = P_c)\), then an implicit relationship for the minimum pressure ratio at which the flow starts to choke \((P_c/P_u)\) is obtained:

\[
\frac{1}{n(\frac{P_c}{P_u})^{\frac{n}{D_p}}} - \frac{4f_{ch}}{L} - k + \frac{n+1}{n} \ln\left(\frac{P_c}{P_u}\right) = 0
\]  

(18)

In this case the Colebrook equation \{(16)\} is also inappropriate to use to work out \( f_c \) for choked flow in equation (18), because this has an implicit relationship also. A convenient alternative is the explicit formula of Haaland:

\[
\frac{1}{\sqrt[4]{4f_{ch}}} = -1.8\log_{10}\left[\frac{6.9}{\text{Re}_{ch}} + \left(\frac{r}{3.7}\right)^{1.11}\right]
\]  

(19)

where the Reynolds number for choked flow \( \text{Re}_{ch} \) is given by:
In this way the minimum pressure ratio for choked flow can be determined from equation (18), using Muller’s iterative method. If the actual pressure ratio \( \left( \frac{P_d}{P_u} \right) < \left( \frac{P_c}{P_u} \right) \) then the flow is said to be choked and the mass flow rate is calculated using the choked pressure ratio.

The above process for determining the mass flow down a tube is reasonably complex, even within a simulation package like Bath fp. There is an alternative empirical equation for convoluted metal tubes after Yeaple [79]:

\[
\Delta P = N \left[ 1 - \left( \frac{D_H}{D_H + 0.438s} \right)^2 \right] \rho U \frac{2}{2}
\]  

from which it is possible to derive mass flow rate. The advantage of equation (21) is that an empirical relationship exists between pressure and flow, which leads to shorter simulation run times than the theoretical analysis above, which requires an iterative procedure. In an earlier study by Tilley [80], a comparison was made between use of the empirical equation and the theoretical analysis on corrugated tubes used in diving equipment. Initially there was poor agreement between them, which led to a modification by Tilley of equation (21), which was further modified to include K factor pressure losses:

\[
\Delta P = K \frac{\rho U^2}{2} + N \left[ 1 - \left( \frac{D_H}{D_H + 0.35s} \right)^2 \right] \rho U \frac{2}{2}
\]

This can be rearranged to give mass flow rate:

\[
q_f = \frac{\sqrt{b^2 + 4a|\Delta P|} - b}{2a}
\]

where \( a = \frac{KRT}{2PM^2} \)

and \( b = \frac{N}{A} \left[ 1 - \left( \frac{D}{D + 0.35s} \right)^2 \right]^2 \)
In modelling coaxial tubes, which play an important part in some anaesthetic breathing systems, the same individual laminar-turbulent tube models just described are used, bearing in mind that the inner tube is usually smooth and the outer is usually corrugated. The modelling occurs as two distinct volumes, each with a pressure drop modelled as a restriction in the middle of the tube. The model must have appropriate coaxial connection pieces connected to each port. The models of some of these describe the mixing of gas from outer and inner tubes [77], while pressure losses are determined using the K factor method. Pressures and compositions of the gaseous mixture are calculated by integrating equation (4).

5.6. Reservoir Bag

The reservoir bag can be modelled as a capacitative element with a variable volume, and from this it is possible to derive equations relating bag pressure and volume and their rates of change. The net opening force acting on the reservoir bag is the distending pressure x its surface area:

\[ F_{\text{press}} = A_{\text{surf}} \left( P - P_{\text{atm}} \right) \]

Forces due to elasticity of the bag tend to contract it. The elastance, \( \varepsilon \), is assumed to have a nominal value until the bag is stretched, after which point it varies as the bag is stretched until it reaches a maximum value. The elastic force \( F_{\text{elas}} \) varies as indicated in equation (24) accordingly as the volume of the bag varies between two defined limits, \( V_A \) and \( V_B \), where \( V_A \) is the maximum volume reached when the bag is slack, and \( V_B \) is the maximum volume reached when the bag is stretched:

\[ F_{\text{elas}} = A_{\text{surf}} \varepsilon_{\text{nom}} V \]

for \( V < V_A \)

\[ F_{\text{elas}} = A_{\text{surf}} \left[ \varepsilon_{\text{nom}} V_A + \left( \varepsilon_{\text{max}} V_B + \left( \varepsilon_{\text{nom}} - \varepsilon_{\text{max}} \right) V_{\text{str}} - \varepsilon_{\text{nom}} V_A \right) \left( \frac{V - V_A}{V_B - V_A} \right)^2 \right] \]

for \( V_A < V < V_B \)

\[ F_{\text{elas}} = A_{\text{surf}} \left[ \varepsilon_{\text{max}} V + V_{\text{str}} \left( \varepsilon_{\text{nom}} - \varepsilon_{\text{max}} \right) \right] \]

for \( V > V_B \)
Figure 5.3 shows the results of bag pressure (a surrogate for $F_{\text{elas}}$ in the equations above), plotted against bag volume, showing two distinct regions of the bag’s mechanical behaviour, where $V_A$ is about 1.6 L, for $V<V_A$ and for $V_A<V<V_B$ as described above. The plotted line is the result of mechanical testing carried out by M.Jones in reference 77, fitted by D.G. Tilley to the triangular points derived from the equations above.

Another retarding force in the bag’s expansion is frictional force, which is proportional to the velocity of movement of the bag material, and the sum of all these forces acts to describe the equation of motion of the bag:

$$m\ddot{r} = F_{\text{pres}} - F_{\text{elas}} - F_{\text{fric}}$$

This can be integrated once to provide velocity, and again to calculate radius of the bag, whence volume and surface area can be deduced. This allows rate of change of volume to be calculated, which can be put back into equation (4). The change in gas composition in the bag can be determined using the same equations as for other capacitative elements.

The reservoir bag also exhibits hysteresis, which means that the pressure-volume relationship is different when the bag volume is increasing compared to when it is decreasing, and depending also on the rate of change of the bag’s volume. The effect is
best included in this analysis by the inclusion of a pressure term $P_{hyst}$ in a modification of equation (24). However, in the event, it was found that omitting hysteresis from the bag avoided spiky anomalies in the results.

5.7. Carbon dioxide absorber

The model for the carbon dioxide absorber has to characterise both the hydraulic pressure drop down its length and the absorption of carbon dioxide (CO$_2$) by the soda lime crystals. The modelling of the hydraulic pressure drop is complicated by the fact that the absorber consists of many spherical crystals filling the void of the container, kept in place by baffles at each end; the pressure drop is therefore influenced by the ratio of surface area of crystals to volume of the intervening void. To model the CO$_2$ absorption within the absorber, the canister is divided into a number of elemental sections each of length $dl$, and a dynamic mass balance for each element can be determined, which leads to expressions of CO$_2$ concentration changes throughout the absorber. The frictional pressure drop through the porous material is defined using the D’Arcy equation [81]:

$$\Delta P = f_{c} \cdot \rho \cdot U^2 \cdot L \left( \frac{A_{surf}}{V_{void}} \right)$$

where $A_{surf}$ is the total surface area of the absorbent particles and $V_{void}$ is the volume of the voids. By noting that $V_{void}/A_{surf}$, the hydraulic mean depth of the canister, can be related to the diameter of crystal spheres $D_{abst}$ and the ratio $\xi$ of $V_{void}$ to total canister volume, and defining friction factor $f$ in terms of Reynolds number, it is possible to deduce an equation for the frictional pressure drop for both laminar and turbulent flow, $\Delta P$:

$$\Delta P = 2L \left[ \frac{75 \mu U (1-\xi)}{D_{abst}^2} + \frac{0.875 \rho U^2}{D_{abst}} \right] \frac{1-\xi}{\xi^3}$$

This can be added to the K factor loss produced by the baffle plates at each end of the canister to give total pressure loss in the canister. By rearranging the resulting equation, the flow velocity $U$ may be calculated, from which mass flowrate $m_{abst}$ may be deduced as follows:

$$m_{abst} = \left\{ \frac{\sqrt{b^2 + 4a|\Delta P|}}{2} \right\}$$

where $a$ and $b$ are factors of the squared and linear terms respectively, in the quadratic equation concerned:
such that
\[ a = \frac{1}{A^2} \left[ \frac{K\rho}{2} + 1.75\rho_M L \frac{(1 - \zeta)}{\zeta^2} \right] \]

and
\[ b = \frac{150\mu L (1 - \zeta)^2}{A.D_{abst}^2 \zeta^3} \]  
(28)

In order to model the absorption of carbon dioxide, the canister is divided into a number of elemental sections as shown by Lo [75]. The mass balance of the gas and absorbent media within the \( i \)th element is:

\[ \dot{c}_i = -\frac{U_E}{d} (c_{i-1} - c_i) - \frac{\rho_B}{\zeta} \dot{w}_i \]  
(29)

where \( U_E \) is the average gas velocity through the voids, so that \( \xi U_E = U \). According to Lo, the absorption rate in the \( i \)th element can be described by:

\[ \dot{w}_i = \frac{k_M A_{abst}}{\rho_B} c_i \left\{ 1 - \frac{w_i}{w_{max}} \right\} \]  
(30)

where \( k_M \) is the mass transfer coefficient. The gas transfer consists of two phases: the transfer of the gas to the absorbent surface and the chemical absorption process. The latter process occurs almost instantaneously, so that the rate limiting step in the combined process is the former; the gas mass transfer coefficient \( k_{M,G} \) can be determined by an empirical relationship between the Stanton number for mass transfer, \( k_{M,G}/U \), Reynolds number and Schmidt number:

\[ St.(Sch)^{0.6} = \frac{0.438}{\zeta} (Re)^{-0.4} \]  
(31)

Hence the response of the canister to absorption of carbon dioxide can be determined from equations (29) and (30), subject to initial and boundary conditions. To determine the concentration of \( CO_2 \) in the first element, it is necessary to determine its concentration upstream of the canister, which can be deduced from the mass fraction in the component preceding the canister. In this way the concentration and absorption in each element can be determined by integrating equations (29) and (30). The canister model also provides the \( CO_2 \) mass fraction to the model downstream from the canister. The total absorption rate of \( CO_2 \) is the sum of all absorptions in each of the \( i \) elements, \( i=1, n \):

\[ m_{CO_2} = \sum_{i=1}^{n} \frac{d\dot{w}_i}{dt} \rho_M A_{abst} dl \cdot M_{CO_2} \]  
(32)
from which the mass flowrate of gas through the canister may be determined.

5.8. Sources
Bath fp also models pressure and flow breathing cycles to add some input and boundary conditions to the structural model elements described above. The waveform patterns of these variables can be changed, but in general a sinusoidal waveform is the most useful. However, given that the human airway pressure waveform in normal quiet breathing has unequal negative and positive pressure magnitudes in inspiration and expiration respectively, this was built in to the patient breathing model; an inspiratory airway pressure of \(-5\) cm H\(_2\)O, and an expiratory pressure of \(+2\) cm H\(_2\)O were selected [96]. A tidal volume can be selected, and account is taken of the respiratory quotient (CO\(_2\) produced/O\(_2\) consumed), and of lung airway resistance and alveolar compliance. Another subsequent amendment made to the Bath fp model was to include the presence of tracheal deadspace (absent in earlier versions of Bath fp), which gave more realistic carbon dioxide fractions within the system.

5.9. Nitrous oxide absorption
Anaesthetic textbooks tell us that absorption of a breathable anaesthetic agent depends on its blood/gas solubility, cardiac output and alveolar to venous partial pressure difference [82]. From some breath by breath measurements on six patients, of inspired and expired N\(_2\)O carried out by Severinghaus in 1954 [83], the following formula was deduced to describe N\(_2\)O absorption by body tissues, which has been quoted more frequently than any others to describe nitrous oxide uptake:

\[
\text{Absorption rate } N_2O = \frac{1000}{\sqrt{\text{time}}} \text{ ml.min}^{-1}
\]

Severinghaus noted that the rate of uptake of N2O was thirty times the excretion rate of N\(_2\), which coincides with their relative solubilities in blood. Since then, several studies have both supported and challenged this model of N\(_2\)O uptake. Virtue in 1982 [84], did measurements on six subjects breathing 25% N\(_2\)O in oxygen, and projected his results to estimate absorption for patients breathing 75% N\(_2\)O in O\(_2\); he took no measurements in the first four minutes, treating this as a “wash in” period, but found that the results for the recording period were reasonably close to those predicted by the Severinghaus equation, while recognising that there was considerable individual variation; he noted that from four minutes the alveolar uptake corresponded to the body uptake, whereas before this, uptake was governed by alveolar uptake, saturating the lungs (particularly the FRC) with N\(_2\)O; he also thought that beyond two hours, the Severinghaus equation may not apply as saturation is approached beyond two hours. Bengtson et al [85] did measurements on 40
orthopaedic patients and found a reasonable correlation with Severinghaus’ original equation.

In 1984 Beatty et al [86] did some measurements on twenty patients undergoing abdominal surgery; their criticism of previous studies concerned the wide range of techniques and results, suggesting some lack of attention to detail in terms of leaks in circle systems and assumptions. By particular attention to such detail, they found that N₂O uptake was less than previous studies had suggested, and that their clinical results corresponded to those predictable by a multi-compartmental mathematical model as described by Mapleson in 1963 [87]. They were able to fit their measurements to the equation:

\[
\text{N}_2\text{O uptake} = 412 \cdot \text{time}^{-0.37} \text{ mL.min}^{-1}.
\]

This equation was used in Bath fp for this work, because it seemed to offer a better fit to some carefully planned clinical measurements, and because it fitted more closely with Mapleson’s mathematical analysis of uptake and excretion of inert gases using a four compartment model [87]. However, this equation will not apply instantaneously, at least until there is some nitrous oxide in the body, as Beatty took no measurements for the first two minutes. For the purpose of applying the this equation in Bath fp modelling, a linear increase in nitrous oxide absorption is assumed from time zero through the first thirty seconds of the run (five breaths). The justification for this is the estimate of the time taken to saturate a 1500ml FRC with the 50% N₂O mixture, using five breaths (at 10 breaths/min), each containing 250ml of N₂O (35 sec). A graphical representation of this process is shown in figure 5.4 below.

![Figure 5.4. Graphical representation of Bath fp model of N₂O absorption.](image-url)
5.10. Venturi.
See chapter 8 for an introduction to the venturi, and some aspects of its mathematical modelling in Bath fp.

5.11. Breathing trial.
See chapter 9 for assumptions made about residual nitrous oxide levels after each breathing run in relation to what is known about the mathematical modelling of the pharmacokinetics of such gases.

An outline has been given of the Bath fp modelling process for breathing systems. This is mainly the work of others [75, 76, 77, 80], but an understanding of the principles of the modelling process is important to the current work. Other aspects of the mathematical pharmacokinetics of nitrous oxide is also introduced, as this information was relevant to further Bath fp development by Dr. D.G. Tilley for this project.
Chapter 6

Bathfp Simulation: Circle System, standard configuration

6.1. Introduction
Based on the author’s wish to use the Bath fp model described in chapter 5 to simulate the behaviour of a standard anaesthetic circle breathing system under different sets of conditions, in order to improve its design as indicated in the earlier chapters, simulations were carried out according to the plan outlined in this chapter. Except where indicated, the experimental runs were all carried out with a number of variables standardised as indicated in the previous chapter. This was to allow comparisons to be made between subsequent changes in geometrical configurations of the breathing system.

6.2 Methods
Results are presented from Bath fp runs for the standard configuration of the circle system as shown in figure 5.1. Within the patient model, ABW99, the tidal volume is set at 500 ml, functional residual capacity at 1500 ml, respiratory quotient at 0.84, respiratory rate 10 breaths/minute. The standard gas mixture from the FGF source was set at 50% oxygen, 50% nitrous oxide unless otherwise indicated. Fresh gas flows used were 0.5, 1.0, 3.0, and 10.0 L.min\(^{-1}\). At the start of every run the gas content of the system was set to be air, and it is the change in concentrations of gas content as each run progresses which is of interest.

The initial geometrical settings of the circle system, which were subsequently changed to compare functional efficiency, include:
- APO7(1) and APO7(2) length 1 m, internal diameter 22 mm (corresponding to that of current standard tubing used clinically);
- ACS1(1) diameter 140 mm, length 70 mm.

Initial runs were short 12 second runs to explore the morphology of the uptake curves for the different gases into the breathing system, and at first did not include a Bath fp model for N\(_2\)O uptake. This was subsequently added and it was possible to comment on the difference this made to the results. Longer, 120 second runs were then carried out to explore in more detail the likely equilibrium behaviour of the breathing system. Then system geometry was changed by halving and halving again AP07 tubing diameter, halving and doubling AP07 tubing length, and by changing the volume and shape of CO\(_2\) absorber ACS1(1).
6.3 Results and Discussion

6.3.1 Short runs (12 seconds)

Some examples of results obtained from Bath fp modelling of an early model of a circle system are presented and discussed piecemeal. This first model was developed without taking account of the uptake of nitrous oxide by the body, but the results are nevertheless useful in showing differences in variables with regard to changes in FGF, and will be compared to later results with the N₂O absorption accounted for.

The first results to be considered concern the early changes in gas concentrations, those within the patient’s first two breaths of fresh gas containing 50% O₂ and 50% N₂O, in a system initially filled with air. Figures 6.1a and 6.1b show the gas concentrations just outside the entrance (port2) to the tube AP07(1) at the proximal end of the inspiratory limb of the system. At a very high FGF of 10 L min⁻¹ (figure 6.1a), early inspiration is characterised by a sharp rise in O₂ and N₂O mass fractions to levels corresponding to 50% each by volume, and a correspondingly rapid fall in the N₂ mass fraction; in general Bath fp calculates mass fractions of gases as outlined in chapter 5, but these are acceptable surrogates for partial pressure or concentration. The FGF is sufficiently high that the patient’s own respiratory pattern does not affect the gas mass fraction waveforms at this point of the system; this pattern occurs because the high FGF exceeds the patient’s own inspiratory flow demand. As expiration starts, the O₂ mass fraction remains constant and high, since the inspiratory valve AOV2(1) is closed. If this is compared with the early fate of oxygen mass fraction at the same location using a much lower FGF of 1.0 L min⁻¹ (figure 6.1b), it can be seen that the respiratory pattern imposes itself on the FGF waveforms. The slight initial rise in the O₂ mass fraction followed by a fall as the patient’s inspiratory flowrate increases towards the middle of inspiration; the pattern is dominated by the initial air contained within the system rather than the FGF. Even a sharper rise in O₂ mass fraction at the beginning of the second breath is short-lived in the next inspiratory cycle. The clinical message is that high FGF at the beginning of the anaesthetic procedure is important to deliver adequate oxygen to the patient from the outset; as will be noted, there are other reasons too for a high FGF at the start.

If we look at other gas mass fraction changes, it can be seen that the nitrogen and nitrous oxide curves are near reflections of each other, the one being offloaded from a concentration of 78% by volume from the patient’s lungs, the other being loaded from zero towards the fresh gas concentration of 50%. At a high FGF of 10 L.min⁻¹ (figure 6.1a), the [N₂] falls rapidly as the fresh gas dominates the flow at this point; in a way the N₂ curve reflects the O₂ curve. In expiration, the upstream valve (AOV2(1)) is closed, so further changes in [N₂] from a residual value is not possible. At this high FGF denitrogenation of the patient occurs quickly. Looking at figure 6.1b, at low FGF, there is an initial fall in the [N₂], followed by a rise as inspiration proceeds, air in the system
dominating the inspiratory contents rather than the meagre FGF contribution; as the end of inspiration and expiration proceeds, the [N₂] falls again and remains level as the FGF predominates over the respiratory flow; even in the pause at the start of the second breath, the reduction in [N₂] is short and temporary. This shows that effective excretion of dissolved nitrogen from body stores as well as nitrogen from the lungs depends on a high FGF, a fact which is often forgotten by anaesthetic practitioners who may start using low FGF prematurely. In considering loading anaesthetic agent, the same, but converse argument applies, that low FGF will fail to load the patient quickly enough, and a high FGF is required for this purpose too. Carbon dioxide levels remain very low throughout the cycle, as they do in the expiratory pipe AP07(2), the pipe which lies immediately downstream of the patient. Maximum volumetric flowrate through inspiratory pipe AP07(1) is constant at 23 L.min⁻¹ in the presence of FGF of 0.5, 1.0, 3.0 L.min⁻¹, dropping to 20 L.min⁻¹ at a FGF of 10 L.min⁻¹. This demonstrates that the patient’s respiratory waveform dominates the gas flow patterns at all but the highest FGF. 

If the inspiratory unidirectional valve, AOV2(1), is considered, the maximum inspiratory pressure drop across the valve seems constant at 0.5 – 0.55 mbar, irrespective of FGF. As this valve is closed by the pressure difference across it during expiration, the calculations made by the model for expiratory pressure drops should perhaps be ignored. The same is true if the expiratory unidirectional valve, AOV1 (1) is considered in expiration; namely that the maximum expiratory pressure drop across this valve is fairly constant at 0.5 – 0.6 mbar, irrespective of FGF. As expected by conservation of mass flow, the maximum flowrates through these valves follow the same pattern as through the both pipes AP07(1,2).

Consider next the T-connector, ATP2(1), which connects the fresh gas inflow and the expiratory gas as it emerges from the CO₂ absorber, and delivers this mixture towards the patient via AOV2(1) and AP07(1). Figure 6.2 shows gas mass fractions outside port 1, which is the inflow to the T-piece from the CO₂ absorber, figures 6.2 a) and b) referring to FGF of 10 L.min⁻¹ and 1.0 L.min⁻¹ respectively. On looking at Figure 6.2a and the O₂ mass fraction at high FGF, it is observed that early on in inspiration there is a rise in the [O₂], which dips again as maximum inspiratory flow rate occurs; this must be because early on, when the patient’s inspiratory flow is low, some fresh gas flows retrogradely towards the CO₂ absorber and beyond, towards the reservoir bag. As the inspiratory flowrate increases, [O₂] at port 1 dips as some stored air and mixed with this additional store of fresh gas is drawn back towards the patient. On expiration, with valve AOV2(1) closed, fresh gas is once again directed retrogradely, and [O₂] at port1 rises towards the level found in the FGF, though diluted by expiratory gas. If this is compared with figure 6.2b) at a low FGF, [O₂] at port 1 during inspiration does not change from that of air; all the low FGF goes towards the patient, none goes retrogradely towards the CO₂ absorber. As with the higher FGF, on expiration when the valve AOV2(1) is closed, FGF flows
retrogradely so the $[O_2]$ at port 1 rises towards that of the FGF. Note that on the next inspiration, the $[O_2]$ at port 1 again remains low, since all of the low FGF is directed towards the patient and none retrogradely.

If the other gas mass fractions are observed in figure 6.2a) and 6.2b), it can be seen that the nitrous oxide levels strongly support the explanation on gas pathways given in the preceding paragraph in relation to oxygen. The nitrogen levels, as indicated before, are a mirror image of the $N_2O$ levels, and $CO_2$ levels are acceptably low throughout.

If we consider port 3 of the same T-piece, ATP2(1), the point which describes the effect of mixing within the T-piece of the FGF from port 2 and the expiratory gas from port 1, we find that the gas mass fractions are almost identical to those at port 2 of the pipe AP07(2). The pressure drop between ports 1 and 3 of ATP2(1) is constant at 0.07 – 0.08 mbar for FGF’s of 0.5, 1.0, 3.0 Lmin$^{-1}$, which encourages the gas from the expiratory limb of the system to be taken up by the inspiratory limb; when the FGF is 10 Lmin$^{-1}$, the pressure drop is less favourable in this respect, and much of the fresh gas flows towards the expiratory limb, and the remainder flows towards the patient; such a high FGF is not economical.
If attention is now directed to the other T-piece connection in the system, ATP2(2), which is situated downstream of the expiratory unidirectional valve AOV1(1), and upstream of the carbon dioxide absorber ACS1(1), then it can be seen that the side arm of
the T-piece allows some of the gas on the expiratory side of the system to be directed towards the reservoir bag ARB1(1), via the overflow valve AEV1A(1) (see figure 5.1). On the one hand, a low FGF allows most of the expiratory gas to be recirculated, thereby enhancing the economy and efficiency of the system, but thereby also loading most of the \(\text{CO}_2\) through the \(\text{CO}_2\) absorber, and providing a relatively small volume of stored gas in the reservoir bag for the system to draw on. On the other hand, a high FGF means that there will be a large wastage of gas excreted out of the system through the overflow valve, taking with it some of the \(\text{CO}_2\) load, but also providing a large volume of stored gas in the reservoir bag. The gas mass fractions at the entrance to the T-piece (port 3), are fairly constant at all flows, at values very close to the starting values in the system, and in common with all other points within the systems downstream of the patient up to this point, and will therefore not be commented on further.

If the gas mass fractions at port 1 of ATP2(2) are examined, port 1 being the side arm of the T-piece, which allows gas to leave the system and go towards the reservoir bag and the overflow valve, then the ability of the system to preserve or excrete gas may be commented on. Figure 6.3 shows gas mass fractions at port 1 of ATP(2)2 at a FGF of 10 L.min\(^{-1}\) and at the very low flow of 0.5 L.min\(^{-1}\). The low FGF shows very little change (figure 6.3b) in the gas concentrations reaching this point before the end of expiration of the first breath; this point of the system is a very long way from the fresh gas supply, and the FGF is very low, so this result is not surprising. At a high FGF on the other hand, the changes in gas mass fractions or concentrations are more dramatic (figure 6.3a) than points upstream, shown in figures 6.1a and 6.2a. The rise in [\(\text{O}_2\)] early in inspiration of the first breath must be due to retrograde flow of fresh gas through the \(\text{CO}_2\) absorber, for storage in the reservoir bag. As inspiration proceeds, the [\(\text{O}_2\)] falls as the majority of the fresh gas flows directly towards the patient on the inspiratory side of the system, with the expiratory unidirectional valve AOV1(1) remaining closed. Towards the end of inspiration, as the patient’s inspiratory demand falls, the [\(\text{O}_2\)] at port 1 of ATP2(2) rises again as an increasing proportion of the fresh gas flows retrogradely towards ATP2(2). During expiration, the [\(\text{O}_2\)] falls as the expired gas finds its way towards port1, and rises again towards the end of expiration as the patient’s respiratory flow diminishes and an increasing proportion of the gas at port 1 is once again retrograde fresh gas. As the emerging pattern of gas mass fractions has revealed, the [\(\text{N}_2\)]s are a reflection of the [\(\text{O}_2\)]s, and the [\(\text{N}_2\)]s are a mirror image of both. The [\(\text{CO}_2\)] levels remain low throughout, at both high and low FGF.

The behaviour of the carbon dioxide absorber, ACS1(1), will now be commented on. Clearly its most important function is to absorb \(\text{CO}_2\) efficiently, without providing excessive resistance to gas flow. It is also interesting to observe how the \(\text{CO}_2\) mass fractions vary as the gas travels through the absorber. Figure 6.4 shows the \(\text{CO}_2\) mass fractions at the inlet to (port 1) and the outlet from (port 2) ACS1(1). Common to both
high (6.4a) and low (6.4b) flowrates, CO$_2$ levels are higher at port 1 than at port 2, and there are maximum port 1 levels on expiration, both of which are expected. Note the factor of $10^{-6}$ on the values, indicating low values in absolute terms. The other feature is that the inspiratory value of CO$_2$ at port 1 at the start of the second breath is higher than in the first, and this pattern will be examined further in the 30 second runs to determine if an acceptable equilibrium is reached. To explain the CO$_2$ concentrations on either side of ACS1(1), it is worth considering the gas pathways in the neighbourhood of ACS1(1) on inspiration and on expiration. On inspiration, gas is drawn into the inspiratory limb, not only from the FGF, but at peak inspiratory flow it is also drawn from gas stored in the expiratory limb from the previous breath, including expired gas from the reservoir bag. At low levels of inspiratory flow, some FGF may be diverted towards the expiratory limb, the reservoir bag and the expiratory valve, especially when the FGF is high in relation to inspiratory flow; under these conditions a significant amount of FGF is diverted retrogradely through ACS1(1) towards the expiratory valve and reservoir bag, thus diluting the reservoir bag’s expiratory gas content, while with low FGF, a smaller proportion of FGF finds its way to the reservoir bag. On expiration, some expiratory gas goes anterogradely through ACS1(1), while some goes to the reservoir bag and the expiratory valve. If figure 6.4b (low FGF) is observed, on first inspiration the small but non-zero [CO$_2$] at port 1 is due to air (which contains a small amount of CO$_2$), from the reservoir bag being drawn towards ACS1(1). On expiration the [CO$_2$] at port 1 rises, because of expiratory gas being delivered to ACS1(1), although a proportion is also delivered to the expiratory valve and the reservoir bag. The [CO$_2$] at port 2 is close to zero because although an anterogradely advancing CO$_2$ containing front appears at port 2 there is also a retrogradely flowing fresh gas front, which results in a [CO$_2$] at port 2 of close zero. In the presence of high FGF (figure 6.4a), this profile is slightly modified; early and late in the inspiratory phase, when the inspiratory flowrate is low, [CO$_2$] at port 1 drops to zero as FGF flows retrogradely through ACS1(1).

If the [CO$_2$] within the body of ACS1(1) in the absorbent phase is considered, there is the expected result of a breath by breath increase, irrespective of the FGF, and to a greater extent towards port 2 than towards port 1. If the [CO$_2$] in the gas phase within ACS1(1) is considered (figure 6.5), the CO$_2$ concentrations at point (1) (near port1) are somewhat different to those at port1 (but note different units); figure 6.5b at low FGF, shows that throughout inspiration [CO$_2$] remains low at point(5) as it is absorbed, although by point (10) a small proportion of the expired CO$_2$ remains unabsorbed. It seems that only by the start of the second breath do noticeable amounts of CO$_2$ appear...
a) $FGF = 10 \text{ L.min}^{-1}$  


b) $FGF = 0.5 \text{ L.min}^{-1}$

Figure 6.4. $CO_2$ mass fractions at inlet (port1), and outlet (port 2), of $CO_2$ absorber, ACS1(1). Curves [1] Mass fraction entering port 1; [2] Mass fraction leaving port 2
At point 10 for an initial period, presumably as the remainder of the preceding expiratory gas stored in the reservoir bag is drawn through the absorber. Figure 6.5a shows that the CO₂ mass fraction at port 1 is modified by the high FGF; at the beginning of the first inspiration, the high FGF encourages retrograde flow through ACS1(1), and the small amount of CO₂ present in the residual air reappears at point 1, and only diminishes as the retrograde FGF goes through the absorber; the mid-inspiratory blip in CO₂ at point 10 is presumably due to a reduction of that retrograde flow; the difference between proximal and distal parts of the absorber is noted. It is also noted that the curves for point 5 of the absorber (blue curves) show unrecordably low magnitudes; presumably at point 1 the absorption process has only just started, by point 5 it is complete, and by point 10 the retrograde flow referred to is apparent. The results suggest a complex relationship between retrograde FGF on inspiration, anterograde expiratory flow, and the process of CO₂ absorption.

The majority of the pressure drops across ACS1(1) are across the absorbent crystals rather than across the canister; at a maximum value of 0.25 mbar at low FGF, it is an order of magnitude greater than that across the canister itself; due to the significant retrograde flow that occurs, the pressure drop is modified to lower values at high FGF. A similar difference in volumetric flowrate through ACS1(1) occurs, with peak values of –6 to +23 L.min⁻¹ occurring at low FGF, and –15 to +19 L.min⁻¹ at high FGF. This difference is because of the addition and subtraction of the respiratory flows from the constant value of FGFs in one direction.
6.3.2. Long runs (120 seconds).

The one hundred and twenty second runs were carried out to get an idea of the trends of gas mass fractions towards equilibrium within the different components of the system at different FGF’s. Starting with the inspiratory tube APO7(1), the first thing that stands out is the difference in waveforms of the mass gas fractions at the inlet end (port2) and the outlet end (port1) of APO7(1); the second thing is that, as expected, there is a difference in waveforms between high and low FGF’s; both aspects will be commented on.

Figure 6.6a) and b) show the differences in waveform, respectively, between entrance to (port2) and exit from (port1) the inspiratory tubing APO7(1). The reasons for the shape of the waveforms at port 2 have already been discussed in the previous section, but it is notable that they are very different from those at port 1. By the time the inspiratory gas has travelled the length of the tubing APO7(1) a certain amount of gas mixing has taken place and the waveform picture at port1 is dominated by the patient’s (asymmetric) sinusoidal respiratory flow pattern, rather than by the presence of the upstream valve AOV2(1), or the relationship between FGF and the inspiratory flowrate; the waveforms become an almost square wave pattern, with inspiratory peaks in O₂ and N₂O, inspiratory troughs in N₂ and CO₂, and converse troughs and peaks in expiration; figure 6.6 shows results at a median FGF of 3 L.min⁻¹ and demonstrates a reasonably rapid rise in O₂ to its equilibrium level, a gradual rise in N₂O mass fractions, mirrored by a gradual decline in N₂ mass fraction. There is incidentally a significant expiratory value of CO₂ mass fraction at port1, which is not apparent at port2, which represents the proximity of the patient’s source of CO₂ production; note that at this FGF, the inspiratory value of CO₂ remains close to zero.

In order to compare rates of change of gas mass fractions in relation to FGF, figure 6.7a) and b) should be observed; it can be seen in figure 6.7a, that at high FGF, there is an increase from the first inspiration in peak O₂ mass fraction from 0.24 to 0.41, and this occurs within the first three breaths. In contrast, at low FGF, the same variable essentially declines gradually from 0.24, reaching a value of about 0.20. The rise in peak inspiratory values of N₂O mass fraction at high FGF is 0.38 at the first breath to 0.56 by the third, whereas at low FGF, this variable goes from 0.05 at the first breath, to 0.20 by the twentieth. Similarly, at high FGF the expiratory maxima of N₂ mass fractions decreases from 0.72 at the end of the first breath to less than 0.10 on the ninth, reaching a value close to this by the third, while at low FGF this variable decreases from 0.75 at the end of the first breath to about 0.6 by the twentieth.

These results emphasise that at the start of the anaesthetic process, when it is necessary to change gas concentrations reaching the patient to the greatest extent and with the greatest rapidity, high FGF is very much more efficacious than low FGF for quickly increasing
inspired oxygen concentration, denitrogenating the patient, and delivering anaesthetic agent to the patient. There is a tendency amongst some anaesthetists using the circle system either to use low FGF from the start or to reduce FGF rates prematurely. Taking the rate of fall of \([N_2]\) as the indicator for a kind of ‘half life’ of the system at 10 L.min\(^{-1}\) FGF, it can be seen that \([N_2]\) halves in four breaths, suggesting that the process of denitrogenation is 94% complete in about sixteen breaths, about 105 seconds; in actual fact nitrogen, and indeed nitrous oxide, are rather soluble gases in body tissues (\(N_2O\) is thirty times more soluble than \(N_2\)) and the response of the breathing system is modified by this; furthermore all the anaesthetic drugs given are respiratory depressants, so this lengthens the total time to an even greater extent. Note also that the peak expired CO\(_2\) mass fraction rises to 0.05 at high FGF and to about 0.1 at low FGF.

The remainder of the results for this run relate to how the system handles CO\(_2\). Figure 6.8 a and b show the mass fractions of CO\(_2\) at entrance to (port1) and exit from (port2) the carbon dioxide absorber ACS1(1); at high FGF the inspiratory value after is unrecordably low at both ports, while at low FGF it reaches 0.07 at port1 and remains negligible at port2; this indicates that the CO\(_2\) absorber does not absorb all of the CO\(_2\) at low FGF. If figure 6.1 is referred to, it is seen that the CO\(_2\) mass fractions further away from ACS1(1), namely at the entrance to APO7(1), are indeed low in the context of the other gases.

![Figure 6.6. Gas mass fractions into and out of tubing APO7(1). FGF = 3 L.min\(^{-1}\).](image)

\(a)\) At entrance to (port2) APO7(1)  \(b)\) At exit from (port1) APO7(1)

a) $FGF = 10 \text{ L.min}^{-1}$

b) $FGF = 0.5 \text{ L.min}^{-1}$

Figure 6.7. Gas mass fractions at exit (port1) from inspiratory tubing APO7(1).


a) $FGF = 10 \text{ L.min}^{-1}$

b) $FGF = 0.5 \text{ L.min}^{-1}$

Figure 6.8. Carbon dioxide mass fractions at entrance to (port1)(curve [1]) and exit from (port2)(curve [2]) carbon dioxide absorber ACS l(1).

6.3.3 The nitrous oxide absorption model
The results described above were based on a model, which did not account for the uptake
of nitrous oxide by the body. Section 5.8 of chapter 5 describes how an equation to
describe the uptake of nitrous oxide was arrived at, based on the work of Beatty [86],
who derived the following equation to describe this process, which was assumed to start
occurring after thirty seconds of breathing, to allow the FRC to become saturated with
\( \text{N}_2\text{O} \):

\[
\text{N}_2\text{O \ uptake} = 412. \text{time}^{-0.37} \text{mL.min}^{-1}.
\]

This equation was added to the Bath fp software, to the patient component ABW99, and
all further results were obtained including this amendment. Figure 6.9 shows that the in
the first thirty seconds a linear increase in \( \text{N}_2\text{O} \) uptake is assumed, followed by
application of the Beatty equation.

Figure 6.10 a) and b) below show the difference at the T-piece closest to the patient,
ATP1(1), between assuming no \( \text{N}_2\text{O} \) absorption (6.10a), and \( \text{N}_2\text{O} \) absorption according to
the model shown in figure 6.9 below. It shows a slight lowering of the \( \text{N}_2\text{O} \) gas fractions
near the patient after twenty breaths if absorption is assumed; additionally it can be seen
that the lower \( \text{N}_2\text{O} \) fractions lead to higher \( \text{O}_2 \) and \( \text{N}_2 \) fractions (beneficial and non-
beneficial respectively).

Another run that was done with this model at this stage, prior to changing system
geometry, was a 30 second run, using 100% oxygen at a FGF of 10 L.min\(^{-1}\). This was worth
doing because clinically at the beginning of an anaesthetic a patient is often pre-oxygenated
before any anaesthetic is given. The result was an increase in inspired \( \text{O}_2 \) mass fraction to more
than 0.8 by the third breath, and 0.85 by the eighth, with the expired \( \text{O}_2 \) concentration
exceeding 70% by the eighth breath. This supports clinical practice.

*Figure 6.9. Curve showing the course of \( \text{N}2\text{O} \) uptake at patient model ABW99 assuming a linear rise in the first thirty seconds, followed by a\ncourse predicted by the Beatty equation (see text).*
6.3.4. Change of length of tubes APO7(1,2)

For these runs, the length of APO7 (1) and (2) is doubled from 1 m to 2 m, and halved to 0.5 m. Do tubes of different lengths alter the response of the system? It would be expected that increased volume of the systems due to longer tubes would increase the initial response time of the system, and shorter tubes may shorten the early response, but in both cases perhaps not change the response of the system in equilibrium.

If port 1 (outlet) of APO7(1) is examined, figure 6.11 shows that, at a FGF of 10 L.min$^{-1}$, the peak inspired oxygen mass fraction rises from 0.24 to its peak of 0.40 on the first breath when length of APO7(1 and 2) is 0.5 m, and by the fourth breath when the length is 2.0 m. Figure 6.12 shows that at a FGF of 0.5 L.min$^{-1}$ mass fraction of O$_2$ remains constant and low at 0.24. In other words, doubling and halving the length of the tubing at low FGF does not significantly alter the dynamics of the oxygen mass fraction.

If figures 6.11 and 6.12 are re-examined in relation to nitrous oxide mass fraction increase up to the third breath, it is found that at a FGF of 10 L.min$^{-1}$ (figure 6.11) there is an increase to its maximum value of 0.58 by the second breath at a tube length of 0.5 m, while this process takes five breaths at a tube length of 2 m. At a FGF of 0.5 L.min$^{-1}$ (figure 6.12) an inspired value of N$_2$O of 0.1 is reached by four breaths for a tube length of 0.5 m, whereas this value is not reached in twenty breaths for a tube length of 2 m. This demonstrates that at a high FGF, there is a delay in raising N$_2$O mass fraction as L is
increased, and for this reason it would be worth keeping the tubing as short as possible, assuming high FGF is used at the beginning of an anaesthetic.

Figure 6.11. Gas mass fractions at port 1 (exit from), inspiratory tubing APO7(1) for three different lengths L of APO7(1) and APO7(2), diameter 22 mm, FGF = 10 L.min⁻¹.

Figure 6.12. Gas mass fractions at port 1 (exit from), inspiratory tubing APO7(1) for three different lengths L of APO7(1) and APO7(1), diameter 22 mm, FGF = 0.5 L.min\(^{-1}\). Curves: [1] Oxygen; [2] Nitrogen; [3] Carbon dioxide; [4] Nitrous oxide.
6.3.5. Change of diameter of tubes APO7(1, 2)

If now the diameter of the tubes is halved to 11 mm (the length of the tubes being left at the original standard 1 metre), figure 6.13 shows the same mass gas fractions at the same location, port1 (exit from) APO7(1). If figures 6.11a) (diameter 22 mm) and 6.13a) (diameter 11 mm) are compared (both refer to high FGF), it is seen that the peak O$_2$ mass fractions of 0.42 are achieved by the third breath at a tube diameter of 22 mm, and achieved immediately on the first breath at a tube diameter of 11 mm. If figure 6.12a) (diameter 22 mm) and figure 6.13b) (diameter 11 mm) are compared (both refer to low FGF), there is a failure for the inspiratory O$_2$ mass fraction to increase over twenty breaths in both cases; the 22 mm tube is associated with a peak inspiratory value of no more than 0.24, and the 11 mm tube is associated with a peak of very short duration with each inspiration of 0.27.

\[ a) \text{FGF} = 10 \text{ L.min}^{-1} \quad b) \text{FGF} = 0.5 \text{ L.min}^{-1} \]


If the same figures are compared in relation to nitrous oxide mass fractions, it is seen that at high FGF, the peak inspiratory N$_2$O mass fraction of 0.58 is reached after three breaths for the tube of diameter 22 mm and reached immediately on the first breath for the 11 mm diameter tube. At low FGF the same comparisons reveal a value of only 0.09 being reached after twenty breaths for the tube of diameter 22 mm, and a short duration spike of no more than 0.26 being reached immediately in the 11 mm tube. The pressure drop along the tube is not increased significantly by halving the diameter, either at low or high FGF.
6.3.6. Change of length of carbon dioxide absorber ACS1(1)

The next change in system geometry to be made is to the carbon dioxide absorber, ACS1(1). The standard shape of the absorber is cylindrical, with a length of 70 mm and an internal diameter of 140 mm, the gas flowing predominantly longitudinally through it. All the runs to this point have been carried out with these dimensions. This section looks at the effect of doubling the length to 140 mm, keeping the diameter at 140 mm. The main interest here is the way CO$_2$ is handled with a doubling of the CO$_2$ absorber’s length. A set of runs was done at a single FGF of 3 L.min$^{-1}$. Figure 6.14 shows the carbon dioxide at input to (port1) and exit from (port2) the carbon dioxide absorber ACS1(1); Figure 6.14 shows two such runs, with the FGF standardised at 3 L.min$^{-1}$, one with the standard length of absorber (70 mm, figure 6.14a), the other with the absorber length doubled (140 mm, figure 6.14b), and shows an equilibrium value of CO$_2$ mass fraction of 1.8x10$^{-2}$ for the original, shorter absorber, and a value of about 1.0x10$^{-2}$ for the longer absorber, after 120 seconds in both cases. This means that doubling the absorber path length does indeed confer significant additional benefit as far as carbon dioxide management is concerned.

![Figure 6.14. Effect of doubling the length of absorber ACS1(1) on its handling of CO$_2$. FGF = 3L.min$^{-1}$. Mass fraction entering port 1 (curve [1]); mass fraction leaving port 2 (curve[2]).](image)
It was thought worth doubling length to 140 mm (from standard length of 70 mm), reducing diameter to $1/\sqrt{2}$ of original diameter of 140 mm, thereby keeping the absorber’s cylindrical volume the same, to see if this changed CO$_2$ handling. Figure 6.16 shows these results. If figure 6.15a is compared with figure 6.8a (both refer to high FGF), no difference is seen in either the outlet CO$_2$ fractions. However, if figure 6.8b is compared to figure 6.15b, it can be seen that increasing absorber length, while keeping absorber volume unchanged, results in a maximum expiratory CO$_2$ fraction of 0.02 (longer absorber), rather than 0.03 (shorter absorber). While this series of 180 second runs was being carried out, it was thought worth collecting data on how CO$_2$ equilibrium was achieved at other system components, using a FGF = 3 L.min$^{-1}$, and a ACS1(1) length of 140 mm (although no comparison has been made with the 70 mm ACS1(1) in this respect). Figure 6.16 shows how the carbon dioxide mass fractional equilibrium over a 180 second period at port 1 (exit from) APO7(1) (figure 6.16a), and at the two T pieces ATP2(1) and ATP2(2); at ATP2(1), CO$_2$ mass fractions at port 1 (outlet from CO$_2$ absorber, inlet to inspiratory side of system) and port 3 (exit to inspiratory unidirectional

Figure 6.15. CO$_2$ mass fractions at entrance to (port 1)(curve [1]), and exit from (port2) (curve[2]) of CO$_2$ absorber ACS1(1). Change of ACS size to maintain same volume, diameter=100 mm, length = 140 mm. APO7 L=2 m.
valve and inspiratory side of system) are shown (figure 6.16b); at ATP2(2), CO₂ mass fractions at port 1 (exit to overflow valve and reservoir bag) and port 2 (exit to inlet

Figure 6.16. Mass fractions of CO₂ at various components of the system; FGF = 3 L.min⁻¹:

a) At port 1 (exit from) APO7(1). (curve [1])

b) At port 1 (inlet from absorber to inspiratory limb of system) (curve [1]) and at port 3 (exit to inspiratory limb) at T piece ATP2(1) (curve [2]).

c) At port 1 (exit for waste gases to reservoir bag and overflow valve) (curve [1]) and at port 2 (exit to CO₂ absorber) at T piece ATP2(2) (curve[2]).
side of CO₂ absorber) (figure 6.16c). In all cases a CO₂ mass fraction equilibrium is obtained, and in fact there appears to be slight reduction in CO₂ at some points with the passage of time. For the exit from the inspiratory tube APO7(1), this shows a CO₂ mass fraction of 70 x 10⁻³ on expiration and a value of 5 x 10⁻³ on inspiration is achieved. Thus the CO₂ delivered to the patient on inspiration is kept low, even if the expired values are high, since this port is in the immediate vicinity of the patient’s own CO₂ production. For ATP2(1), it shows a CO₂ mass fraction on inspiration of less than 5 x 10⁻³ at port 3 (exit from the T piece towards the inspiratory limb) is achieved. For ATP2(2), it shows that at this high FGF, by far the larger portion of CO₂ is excreted towards the reservoir bag and overflow valve, rather than to the absorber.

In addition a series of curves was drawn for APO7(2), the expiratory tube, plotting pressure drop down the tube against volume flowrate, for low and high flowrates (0.5 and 10 L.min⁻¹), and for the range of diameters for the tube of 5.5, 11 and 22 mm. The area under these curves would therefore give a measure of power expended in the expiratory cycle of the breath, and the slope of the curves gives a measure of the resistance to expiration. The results, shown in figure 6.17, show no real difference between these parameters for the range of variables mentioned at high and low FGFs.

![Graphs showing pressure drop vs FGF down pipe AP07(2) at high and low FGFs for 5.5, 11 and 22mm diameter tubes](image)

**Figure 6.17.** Power curves, showing pressure drop vs FGF down pipe AP07(2) at high and low FGFs for 5.5, 11 and 22mm diameter tubes

a) FGF=10 L. min⁻¹  

b) FGF=0.5 L. min⁻¹
6.4 Closure

This section has described results obtained from Bath fp software on a standard configuration of the anaesthetic circle breathing system. From these results and their discussion it is possible to draw the following conclusions:

• The importance is reinforced of starting the anaesthetic with high FGF in order to enhance the patient’s oxygenation, to denitrogenate the patient quickly, and to efficiently and quickly deliver the inhaled anaesthetic agent to the patient.
• The software reveals that the exact fate of the FGF depends on its magnitude in relation to the patient’s inspiratory flowrate, which varies throughout the inspiratory cycle; at high FGF more of the FGF flows retrogradely in the inspiratory part of the cycle through the CO₂ absorber for storage, which not only has an effect on the manner in which CO₂ is handled by the absorber, but also alters the subsequent inspired O₂ concentration.
• The doubling and halving of the length of the breathing tubes shows that, while it does not markedly alter the rate of patient oxygenation (compared to how a high FGF achieves this), it could be said that halving the length improves the rate of anaesthetic agent delivery. The same could be said of halving the diameter of the tubes.
• Long run durations of Bath fp show that an acceptable level of CO₂ is achieved in the system, and that not only does doubling the volume of the absorber improves CO₂ handling as expected, but also doubling the path length of the CO₂ absorber significantly improves this, even if the absorber volume is kept constant.
• The addition of the nitrous oxide absorption model shows that this significantly alters N₂O concentrations in the system at low FGF’s.
• There are no significant variations in volume flowrate or pressure drops at key points in the breathing system with changes in FGF, nor are there any changes to these variables with changes in system geometry. Therefore it can be recommended that a more compact standard circle system could be achieved by using shorter, narrower tubing without any increased hazard to the patient.
• It is clear that doubling the volume of the CO₂ absorber gives more capacity for absorbing CO₂, but it was also shown that doubling the absorber gas path length isovolumetrically, also improves CO₂ absorbing capacity.
Chapter 7.
Bathfp Simulation: Coaxial system

7.1. Introduction
The concept of coaxial breathing systems was introduced in chapter two, and when used as a circle system it forms a more compact version of the circle. There is no doubt that the coaxial circle system has been under-used in clinical anaesthetic practice, probably because of a lack of understanding of its function and its resemblance to the high flow coaxial Mapleson D system (Bain) with which it might therefore be confused by an indiscriminate user. The availability of Bathfp modelling for coaxial tubes provides an opportunity to assess the function of such a coaxial circle system, as with the standard system to determine the function in detail, and to understand ways, if any, in which it might function differently to the standard system. The iconic model of the Bath fp representation of the coaxial breathing system is shown in figure 7.1.

![Figure 7.1](image)

The iconic model of the Bath fp configuration of the coaxial breathing system.

As with the standard circle system, fresh gas enters the system from the top right hand corner. As the patient at the left hand end of the diagram takes an inspiration, the inspiratory valve AOV2(1) opens, allowing fresh gas to enter the inspiratory limb of the system. As before, if the FGF is high enough, in the early and late parts of the patient’s inspiration, when the patient’s inspiratory flow rate is relatively low, some of the fresh gas will be diverted downwards, retrogradely through the CO$_2$ absorber towards the reservoir bag and the overflow valve AEV1A(1) against a closed expiratory
unidirectional valve AOV1(1). This aspect of the system’s function is supported by the results and will be discussed. The inspiratory limb of the system is the inner concentric tube, delivering inspiratory gas close to the patients at ACA1(1). The expiratory limb is the external concentric tube, directing expiratory gas towards the expiratory unidirectional valve AOV1(1), the reservoir bag, the overflow valve AEV1(1), and towards the CO$_2$ absorber ACS1(1). This direction of expired gas flow is ensured by the closure of the inspiratory unidirectional valve AOV2(1).

7.2 Methods
In this series of runs, the default length of the tubing is 1 m, and this was subsequently altered to 2m and 0.5 m. The default settings for the internal diameter of the external (expiratory) tube, and of the internal (inspiratory) tube were 21 mm and 10 mm respectively. The diameters were subsequently changed to the following outer/inner diameter pairs (labelled as D/d mm in figures): 21 and 5 (or 7, see below) mm; 21 and 15mm; 21 and 10 mm; 15 and 10 mm; and 25 and 10 mm. Both sets of geometrical changes were put in place to see if geometrical alteration changed function. Simulations were done at the same FGFs, namely 0.5, 1.0, 3.0 and 10 L.min$^{-1}$. Most simulations were done for between 100 and 120 seconds.

7.3. Results and Discussion

7.3.1 Gas mass fractions at patient end, attachment ACA1(1)
Figure 7.2 shows the breath-by-breath gas mass fractions at the patient end of the end attachment ACA 1(1) of the coaxial tubing ACP 2(1), over 120 sec, at a high FGF of 10 Lmin$^{-1}$, for baseline values of expiratory outer tube and inspiratory inner tube diameters of 21 mm and 10 mm respectively. The upper panel [a) and b)] of figure 7.2 corresponds to a standard tube length of 1.0 m; a) shows the gas mass fractions at the outlet of the (inspiratory) inner tube, b) shows the same results for the (expiratory) outer tube. The lower panel [c) and d)] shows gas mass fractions for the inner tube only; c) shows gas mass fractions for a tube length of 0.5 m, d) shows the same results for a tube length of 2.0 m. In comparing only inner (inspiratory) tube gas mass fractions to start with [a), c) and d)], looking at the oxygen mass fractions, there is very little difference in their profiles; the high FGF dominates the picture, rather than the tubing geometry. The same characteristics are reflected in the nitrous oxide curves, and the nitrogen curves, as always, somewhat mirror the nitrous oxide curves. For O$_2$ and N$_2$O, the high FGF ensures a rapid rise towards FGF fractions; for N$_2$, there is an initial rapid fall, followed by a rise as air in the system is circulated around it with the FGF. Figure 7.2 b) is included to compare gas mass fractions in the outer tube with those in the inner at standard length and diameters of tubes, using a high FGF. The first thing that is noticeable is the loss of the respiratory swing on gas mass fractions as time progresses, due to gas mixing of a steady stream of high FGF with expiratory gas fractions superimposed; the initial respiratory variation that exists occurs before the high FGF starts to dominate. The next
set of features to observe is the gradual change of all the gas mass fractions towards an equilibrium in the first 70 seconds or so, compared to the immediate changes, which occur at the inner tube due to the presence of the high FGF. Finally, the carbon dioxide mass fractions are unrecordably low at the inner tube, but reach an equilibrium of about 0.06 with the first expiration at the outer tube.

Figure 7.3 shows the same sets of results of gas mass fractions at the patient end of the inner inspiratory tube of ACA1(1) under standard conditions of a 1m length of tube, a FGF of 10 Lmin\(^{-1}\) for 120 sec, but with different diameters of internal and external tubes. While the outer and inner diameters in figure 7.2 were held at 21 and 10 mm, figure 7.3 a), b) and c) shows inner tube results for outer and inner tube diameters of 21mm & 5 mm, 15mm & 10 mm, and 25 mm & 10 mm respectively.

Looking at the oxygen mass fractions of figure 7.3, there is very little difference between the curves in a), b) or c), so the different ratios of external to internal diameter do not alter oxygen handling by the system at this point. With regard to the expired levels of nitrous oxide mass fractions, it is noted that the 21/5 curve in a) has a more prominent trough at the beginning of expiration; this may be because the initial gas composition at the exit of the narrow inspiratory tube is dominated by patient’s expired gas to a greater extent than at the wider inspiratory tube. The rate of rise of expired nitrous oxide levels does not seem to vary much between the curves in figure 7.3 a), b) and c) and figure 7.2a). Looking at the profile of inspired nitrogen mass fractions, the only notable difference between all of them is the brief inspiratory spike with each breath in the 21/7 mm configuration; this may be because the narrower inner tube delivers residual air in the system to the patient at higher velocity. The overall rates of rise and fall of the gases is no different between configurations, suggesting that this has no bearing on the system’s behaviour, which, as observed, is dominated by the high FGF. CO\(_2\) fractions remain unrecordably low for all configurations.
a) Inner gas mass fraction

\[ L=1\text{m} \]

c) \( L=0.5\text{m} \);

Inner gas mass fractions

d) \( L=2\text{m} \);

Figure 7.2. \( O_2, N_2, CO_2 \) and \( N_2O \) gas mass fractions at the patient end of the delivery tube ACA1(1). Upper panel: \( L=1\text{m} \): a) inner inspiratory tube, b) outer expiratory tube. Lower panel: inner inspiratory tube c) \( L=0.5\text{ m} \); d) \( L=2.0\text{ m} \).

Figure 7.3. $O_2$, $N_2$, $CO_2$ and $N_2O$ gas mass fractions at the patient end of the inspiratory inner delivery tube ACA1(1). FGF = 10 $L/min^{-1}$, 120 sec run, $L = 1m$, external & internal diameters a) 21 & 7 mm, b) 15 & 10 mm, c) 25 & 10 mm. Curves: [1] = $O_2$, [2] = $N_2$, [3] = $CO_2$, [4] = $N_2O$
Turning now to results using a low FGF of 0.5 L/min for 120 sec, figure 7.4 shows gas mass fraction results at the patient end of the inner tube of ACA 1(1) for external and internal tube diameter pair of 21 and 10 mm, using the default length of 1 m. While the graphical simulation data were collected for the other tubing D/d diameter pairs, since no difference in behaviour was found between the 21/10 mm pair and the 21/7, 15/10 and 25/10 mm pairs, the latter three are not included here. The choice of 7 mm internal tubing diameter for the second tubing diameter combination occurred because it was found that Bath fp encountered some numerical stability problems when 5 mm diameter was chosen using this low FGF, which did not occur for the 7 mm diameter. When such problems occurred with Bath fp simulation, it was generally to do with boundary conditions between components; this was dealt with when Dr DG Tilley subsequently adjusted those conditions or altered the mathematical iteration interval. As with the standard version of the circle system it can be seen that the low FGF does nothing to change the gas mass fractions quickly to desired levels. It is noticeable that carbon dioxide mass fractions are recordable compared to those at higher FGFs.

Figure 7.4. O₂, N₂, CO₂ and N₂O gas mass fractions close to the patient, of the inspiratory inner delivery tube ACA1(1). FGF = 0.5 L/min⁻¹, 120 sec run, L = 1m, external & internal diameters 21 & 10 mm. Curves: [1] = O₂, [2] = N₂, [3] = CO₂, [4] = N₂O.
Under the same conditions of FGF and tube length, figure 7.5 looks more closely at oxygen mass fractions at the patient end of attachment ACA1(1), looking at those fractions at both inner and outer tubes; once again, only the 21/10 mm D/d tubing diameter ratios is shown here, as the collected simulation data showed no difference in form between this and the other three different diameter ratios.

As expected, it is clear that the oxygen mass fractions at the inner tube are significantly greater than at the outer. Additionally, at this low FGF rate the inspiratory to expiratory difference is much greater at the inner than the outer tube, and there is in all cases an initial fall of about 5% of the starting value in oxygen mass fractions at the inner tube in the first five breaths, and an initial step fall in the outer tube to reflect the addition of expired air to the outer tube mixture.

Figure 7.5. $O_2$ gas mass fractions at the patient end of the inspiratory inner and expiratory outer tubes of the tubing ACA1(1). FGF = 0.5 $\text{L min}^{-1}$, 120 sec run, $L = 1m$, external & internal diameters 21 & 10 mm. Curves: [1]=outer tube, [2]=inner tube.

Figure 7.6 shows results for nitrous oxide under the same conditions as for oxygen in figure 7.5. These show a mirror image of the oxygen mass fractions of figure 7.5. It was noted that the maximum inspired $N_2O$ mass fractions in the early period for the 21/7 mm tube diameter ratio is higher than for the other tube diameter ratios. This is because the small inspiratory tube delivers nitrous oxide to the patient at the highest velocity. It
remains to be seen whether there are any disadvantages to this diameter of inspiratory tube, such as increased resistance to inspiration, although figure 6.18 would suggest that there is no significant increase in resistance to gas flow down a 7mm tube.

7.3.2 Oxygen mass fractions at T-piece ATP2(1) on inspiratory loop

It was thought to be worthwhile looking at the fate of oxygen mass fractions on the return inspiratory loop, from the outlet of the CO\(_2\) absorber ACS1(1), through the Tpiece ATP2(1) where the FGF enters the system, back along the inspiratory tubing AP07(1) to the patient model ABW99. Figure 7.7a) shows the oxygen mass fractions around the T-piece on the inspiratory side of the system ATP2(1), for standard conditions of tubing length 1m, diameter ratio of 21/10 mm and FGF of 0.5 L\(\text{min}^{-1}\). Once again it was found that there is no difference in oxygen mass fractions over this time period between the differing geometries, and the results of the other coaxial diameter ratio results are therefore not presented here. Notice that the oxygen mass fraction of 0.42 at port 2, the fresh gas inlet, corresponds to a volumetric concentration of 50% oxygen at this point. ATP2(1) port 1 is where the residue of expiratory gas with CO\(_2\) already absorbed is joined by the FGF at port 2, and the mixture is delivered to the inspiratory side of the system via port 3, where the outlet flow from the CO\(_2\) absorber ACS1(1) has been joined.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure7.6.png}
\caption{\(\text{N}_2\text{O}\) gas mass fractions at the patient end of the inspiratory inner and expiratory outer tubes of the tubing ACA1(1). FGF =0.5 L\(\text{min}^{-1}\), 100-120 sec run, \(L=1\text{m}\), external & internal diameters 21 & 10 mm. Curves: [1]=outer tube, [2]=inner tube.}
\end{figure}
by the FGF, whence it goes (on inspiration) towards the inspiratory unidirectional valve AOV2(1). It is seen more clearly from figure 7.7b, showing 10 seconds of run, that on inspiration there is a peak rise in port 1 oxygen mass fraction, which is greater than, and earlier than that at port 3. This is because of the partial retrograde flow of FGF that occurs early in inspiration, which enriches the oxygen mass fractions at port 1; a bit later in inspiration, when the inspiratory demand is greater, a larger proportion of the FGF goes to port 3, making the oxygen mass fraction at port 3 high, but with a supplemental supply from that part of the system dominated by air in the early breaths and by expired gas in later breaths, at which time the oxygen mass fraction at port 1 is also dominated by expiratory gas. In the peak expiratory phase, fresh gas is prevented from flowing retrogradely, hence the oxygen mass fraction at port 1 falls lower than that at port 3 because expired gas dominates the fractions at port 1, while port 3 also consists of some FGF. As figure 7.7 a) shows, the oxygen mass fractions at ports 1 and 3 do not rise at all to reach the FGF values at port 2.

By way of comparison, figure 7.8 shows the same results as figure 7.7, the oxygen mass fractions at ATP2(1), but with the FGF at 10 L.min⁻¹, with the original standard outer and inner tube diameters of 21 and 10 mm, and a tubing length of 1m; a) shows a 120 second run and b) shows a 10 second run. Firstly it was found that there were no differences in the data for 1m, 0.5m and 2m length tubing, and only data for the 1 m tubing length is presented here.; secondly, in contrast to the curves in figure 7.7, there is an immediate rise in the inspired oxygen mass fractions at ports 1 and 3 to the FGF value of 0.42 at port 2. At port1 this is due to the overwhelming effect of retrograde flow, and at port3 it is due to the overwhelming predominance of FGF. The expiratory oxygen mass fraction at port1 is lower, and takes longer to rise, than that at port3, because of the effect of the initially air filled system, and subsequently the effect of expiratory gases in this part of the system. Figure 7.8 b) shows the waveforms more clearly using a 10 second run; this shows the initial rise of O₂ mass fractions from the starting value of 0.24 attributable to air, rising towards the FGF value of 0.42; as inspiration proceeds the O₂ mass fraction falls again at port 1 as inspiratory demand draws air or expired gas towards the T piece from the CO₂ absorber; the fall does not happen at port3 because of the dominance of the high FGF there
Figure 7.7. Oxygen mass fractions at the inspiratory T-piece, ATP 2(1); port1=exit from CO₂ absorber, port2=fresh gas inlet, port(3) exit to inspiratory limb of system, L=1m, FGF=0.5 L.min⁻¹, D,d=21,10mm, a) 120 second run, b) 10 second run. Curves: [1]=port1, [2]=port2, [3]=port3.
7.3.3 Nitrous oxide mass fractions at T-piece ATP2(2) on expiratory loop

It was thought worth looking at the state of nitrous oxide fractions on the expiratory side of the system, and figure 7.9 shows the nitrous oxide mass fractions at a FGF of 0.5 L.min⁻¹, at the T piece on the expiratory side of the system, at ATP2(2), where the gas in the expiratory limb comes from the expiratory unidirectional valve AOV1(1) to enter the T piece at port 3, to be divided between: port 1 which delivers part of the expiratory gas to the reservoir bag and the overflow valve AEV1(1); and port 2 to the carbon dioxide absorber ACS1(1) and return to the inspiratory side of the system.

Figure 7.9. Nitrous oxide mass fractions at the expiratory T-piece, ATP2(2) on expiratory loop, at FGF of 0.5 L.min⁻¹, D=21,10mm, L=1m. a) 120 second run b) 10 second run Port1=exit from CO₂ absorber, port2=fresh gas inlet, port3=exit to inspiratory limb of system. Curves: [1]=port1, [2]=port2, [3]=port3.
Figure 7.9. Nitrous oxide mass fractions at the expiratory T-piece, ATP 2(2), L=1m, D,d = 21,10mm, FGF = 0.5 Lmin⁻¹; port1=exit to overflow valve AEV1(1), port2=exit to CO₂ absorber ACS1(1), port(3)= from expiratory unidirectional valve AOV1(1). Curves: [1]=port1, [2]=port2, [3]=port3.
Figure 7.10. Nitrous oxide mass fractions at the expiratory T-piece, ATP2(2), FGF=10 Lmin\(^{-1}\), 120 sec run, L=1m, D,d = 21,10mm; port1=exit to overflow valve AEV1(1), port2=exit to CO\(_2\) absorber ACS1(1), port(3)= from expiratory unidirectional valve AOV1(1). a) 120 second run, b) 10 second run Curves:[1]=port1, [2]=port2, [3]=port3.
As with all other sets of data, there were no significant differences here between different inner and outer tube pairs, and between different tubing lengths, and only the 21/10 mm diameter pair data and 1 m tubing length are shown. It is perhaps less surprising that such differences do not exist from this distance of the concentric tube system itself, compared to differences which might be expected (but shown not to exist) near the concentric tube exits, such as at ACA 1(1). The first thing that is noted is the loss of respiratory variation in the gas mass fractions at port 3; having travelled through a wide bore expiratory tube ACP2(1) (even with a narrowed annular lumen), and through the unmarked capacitive volume, this is probably because of gas mixing, and the distance from the sinusoidal source (the patient). If this is the case, then how is the respiratory variation between inspired and expired mass fractions restored at port 1 and port 2? It must be due to the retrograde flow, which occurs according to the same pattern, from the inspiratory side. The inspiratory to expiratory variation is much greater at port 2 than at port 1, because the former is more likely to be dominated by retrograde FGF, noting the maximum 0.35 N₂O inspiratory mass fraction at port 2. There is an initial slight rise in nitrous oxide mass fractions at all three ports, due to a combination of retrograde flow of fresh gas before much is absorbed by the patient, and some return of nitrous oxide on expiration. This settles as the effect of moving the system’s residual air around the system dominates subsequent breaths. The nitrous oxide mass fractions, both inspired and expired, at all three ports never really rise significantly.

By way of comparison, figure 7.10 shows the same set of results as figure 7.9, but for a FGF of 10 L.min⁻¹. There is again no significant difference between the different geometries. As at the lower FGF, there is also still a relative loss of respiratory variation in nitrous oxide mass fractions at port 3, with strong respiratory variation at ports 1 and 2. The difference here is that the FGF is high enough that the nitrous oxide mass fractions rise immediately at port 2 and fairly rapidly at port 1 to the level in the FGF of 0.58, corresponding to 50% by volume. This is because the effect of retrograde flow of the high FGF predominates. Once again there is no real difference between waveforms representing the different configurations. Figure 7.10 b) shows the waveform more clearly as it is a sample of the first 10 seconds only (1½ breaths). On the first inspiration the initial rise in nitrous oxide mass fractions at ports 1 and 2 due to retrograde FGF is reversed as the patient’s own inspiratory flowrate increases, and decreases again in late inspiration. On expiration the mass fractions increase again; that at port 2 rises directly to the FGF mass fraction of 0.58 due to high retrograde FGF; that at port 1 rises initially towards this but is modified and reduced by the presence of the initial air in the system, and on subsequent breaths by the mass fractions in the patient’s expired breath; towards the end of expiration retrograde FGF once again dominates. At port 3, in the inspiratory
phase the nearby expiratory unidirectional valve AOV1(1) is closed, and this is what keeps the N2O fraction at a single low level (equal to that of the previous expired breath); on expiration it rises to a relatively constant expiratory level, relatively unaffected by the patient’s respiratory pattern.

7.3.4 Carbon dioxide management by CO2 absorber ACS1(1)

Turning now to the management of carbon dioxide by the CO2 absorber ACS1(1), figure 7.11 a) shows the CO2 mass fractions at the entrance to (port1), and the exit from (port2) of the CO2 absorber ACS1(1) at a FGF of 0.5 L.min⁻¹ for standard tubing length of 1 m and for the standard tubing diameter pair of D/d=21/10mm and standard length of 1m. As with all other elements on the expiratory side of the system, there was no difference in CO2 mass fractions between data representing the different inner and outer tube diameter pairs at a low FGF, so these graphical data were omitted. In these curves the peaks occur on expirations, the troughs on inspirations; note the peaks at port 1 occur fractionally before those at port 2, confirming that the direction of gas flow during the expiratory phase is from port 1 to port 2, upwards in figure 7.1. Note the overall low values of mass fraction, with a maximum expiratory value at port 2 of 0.03 (2½%), and the marked differences between ports 1 and 2. Clearly the CO2 absorber does the job it is designed to do, in that maximum expiratory CO2 mass fraction at port 1 after 120 sec is about 0.1, while at port 2 it is 0.03 (not zero as a perfect system would achieve). It is notable that inspiratory CO2 mass fractions at port 2 do fall to zero on each inspiration, presumably aided by stored FGF. Referring back to figure 7.4, this remains as an average value of about 0.03 at the patient end of the system at ACA1(1). By comparison, figure 7.11 b) shows the CO2 mass fractions at ports 1 and 2 for a FGF of 10 L.min⁻¹; the expired CO2 mass fractions at ports 1 and 2 are 0.035 and 0.002 respectively. Clearly, the high FGF itself plays an important role in contributing to lowering the CO2 by a factor of ten compared to that at low FGF; this is not least because at high FGF, more gas is excreted out of the overflow valve AEV1A(1).

7.3.5 Gas mass fractions at overflow valve AEV1A(1)

Figure 7.12a) shows the gas mass fractions at the overflow valve AEV1A(1), for a standard length of tube of 1 m, for the D/d=21/10 mm outer and inner tube diameter ratios, and at a FGF of 10 L.min⁻¹, since much more gas will be lost through the overflow valve at high FGF than at a low FGF. As elsewhere, there is no significant difference between the curves of different diameter ratios, and hence the data for the other diameter ratios are not shown. The starting values of the N2 is lower than that in air; this must be because at this high FGF, gas overflow from AEV1A(1) consists in part of retrograde FGF. The early peaks and troughs of O2, N2 and N2O are higher and deeper than one might expect, as retrograde FGF dominates the inspiratory phase here, and dilution with the initial air filled system and the later expiratory gases occurs in the expiratory phase. As time progresses the inspiratory to expiratory differences in these gases reduces
towards mean values of mass fractions of $0.42 - 0.35 = 0.07$ for oxygen and $0.58 - 0.35 = 0.23$ for nitrous oxide, which correspond to volumetric

\[ FGF = 0.5 \text{ L.min}^{-1} \]

\[ FGF = 10 \text{ L.min}^{-1} \]

Figure 7.11. Carbon dioxide mass fractions at the entrance to (port 1) and exit from (port 2) the CO$_2$ absorber ACS1(1). $L=1m$, $D/d=21/10$ mm. a) $FGF = 0.5 \text{ L.min}^{-1}$, b) $FGF = 10 \text{ L.min}^{-1}$. Curves: [1]=port1, [2]=port2
Figure 7.12. Gas mass fractions at the overflow valve AEV1(1). L=1m; a) FGF = 10 L.min⁻¹. b) FGF = 0.5 L.min⁻¹ (shown at ATP3(1)). Curves: [1] = O₂; [2] = N₂; [3] = CO₂; [4] = N₂O.
concentration differences of 8% and 19% respectively; these are figures which represent continuing oxygen consumption and nitrous oxide absorption. Although nitrogen levels reduce with time, the mass fraction in the valve expirate has only halved by about 70 seconds and still has some way to go before reaching zero at this high FGF; this demonstrates just how much N₂ is stored in the body, both in the lungs and dissolved in body tissues, and therefore how much time is required early in the anaesthetic to denitrogenate completely if this is required. It is clear that carbon dioxide is also excreted from the valve, reaching an equilibrium mass fraction of 0.02. Figure 7.12 b) shows the same results (actually at T-piece ATP3(1) adjacent to the valve, but is equivalent) for a FGF of 0.5 L.min⁻¹; by contrast this shows very little change in the overflow gas composition from air which is what initially fills the system; a significant amount of overflow carbon dioxide is noted at a mass fraction of about 0.04

7.3.6 Pressure drops across different system components
Turning now to a discussion of pressure drops through different components in the system, figure 7.13 shows pressure drops attributable to the different components of the carbon dioxide absorber at low FGF a) and at high FGF b), using standard length and diameters of tubing elsewhere in the system. The total pressure drop between ports 1 and 2 on inspiration at a FGF of 0.5 L.min⁻¹ ranges between 0 and 0.16 mbar, while at a FGF of 10 L.min⁻¹, it ranges between −0.1 and +0.01 mbar, a range of 0.11 mbar. At the higher FGF it can be seen that at the very beginning of inspiration the pressure drop falls rapidly from 0 to −0.1 mbar with retrograde flow of the high FGF. In both graphs, this pressure drop between ports 1 and 2 (black curve) is made up of components from the absorbent media (blue curve) and the canister (red curve). At both FGF’s, the majority of the pressure drop is through the absorbent, with a small contribution through the canister.

Clearly there are direction changes in the flow at high FGF, which are not present at low FGF, although the total range of pressure drop is similar at both FGF’s. In the expiratory phase at high FGF, there is a significant reverse component of pressure drop due to retrograde flow, despite the presence of anterograde expiratory flow. The pressures across ACS1(1) for all other selected geometries, lengths of ACP2(1) of 0.5 and 2.0 m, and of different diameter combinations of outer and inner tubes 21 and 7 mm, 15 and 10 mm, and of 25 and 10 mm, at both low and high FGF, are almost identical to those shown in figure 7.13.

With regard to pressure drops across the unidirectional valves, Figure 7.14 shows the pressure drops across the expiratory unidirectional valve AOV1(1) in the first 10 sec. Figure 7.14a) shows the results at low FGF, and on inspiration, when this valve remains closed, the pressure on the upstream side of the valve (port2) falls to −1.2 mbar as a
result of the inspiratory effort from the patient against the closed valve; the pressure on the downstream side (port1) hardly changes because of the distance of this point from the source of the negative pressure (the patient’s inspiratory effort) and the presence of other relatively high resistance components such as the carbon dioxide absorber. There is initially a very slight rise in pressure at the beginning of the inspiratory effort because of the small retrograde flow of fresh gas

\[ FG F = 0.5 \text{ L.min}^{-1} \]

\[ FG F = 10 \text{ L.min}^{-1} \]

\emph{Figure 7.13.} Pressure drop across carbon dioxide absorber ACS1(1). \( L = 1 \text{m}; \) \( D,d=21,10 \text{ mm}; \) a) \( FG F = 0.5 \text{ L.min}^{-1}; \) b) \( FG F = 10 \text{ L.min}^{-1}. \) Curves: [1]=port 1 to 2, [2] absorbent media, [3] canister.

before the inspiratory effort becomes significant. On expiration there is a rapid rise in the upstream (port 2) pressure to an immediate maximum value of about + 1.2 mbar although it does not seem that this valve is a site of flow limitation under these circumstances; the downstream (port1) pressure rises to a small positive value in the same sinusoidal fashion as the expiratory waveform itself. Looking at figure 7.14 b), showing the pressures at this valve at a high FGF, in the inspiratory phase the pressure on the upstream side of the valve (port2) falls to about –0.9 mbar on the first breath, somewhat less than at lower FGF, and the pressure on the downstream side (port1) does not change from zero. The presence of a high FGF continues to dominate the absolute pressures upstream and downstream of the valve which both rise with subsequent breaths to between 6 and 7 mbar by the second breath, although it is notable that the profile of the pressure drop
across the valve is almost identical at both high and low FGFs, despite the absolute pressures either side of the valve being very different at both FGFs. With regard to the inspiratory unidirectional valve AOV2(1), the pressure profiles are very similar; at low FGF, the pressures on the upstream side (port1) and the downstream side (port2) on inspiration fall slightly and to −1.1 mbar respectively, giving a maximum inspiratory pressure drop of +0.9 mbar; the pressure waveforms across this valve on inspiration follow the same pattern as the patient’s respiratory waveform, so flow limitation is unlikely; on expiration this valve closes, due to a pressure of about +1 mbar on the downstream side (port2) and of +0.3 mbar on the upstream side (port1). At high FGF

\[ FGF = 0.5 \text{ L.min}^{-1} \]
\[ FGF = 10 \text{ L.min}^{-1} \]

Figure 7.14. Pressure drop across expiratory unidirectional valve AOV1(1). L =1m; \( D,d=21,10 \text{ mm} \); 10 sec; a) FGF=0.5 L.min\(^{-1}\), b) FGF=10 L.min\(^{-1}\).

the upstream and downstream pressures are similar to those at low FGF on the first breath, as is the pressure drop across the valve; as with AOV1(1), the pressure waveforms across the inspiratory valve AOV2(1) rapidly rise with subsequent breaths and are dominated by the FGF itself, with peak values between 6 and 7 mbar. Once again the pressure drop profiles are identical at high and low FGFs, despite different values of absolute pressure.
7.4. Closure

- The concept and function of a coaxial version of the circle breathing system was introduced, noting that its underuse by the anaesthetic community may be because of a lack of understanding of its function, and due to the possible risk of confusion with other coaxial systems. The standard coaxial configuration was considered to be with a coaxial tube of 1 m in length, with coaxial diameter pairing of outer tube diameter of 21mm and an inner of 10mm (21/10mm).
- The tubing length was changed to 0.5 and 2 m, and the coaxial pairings were changed to 21/7mm, 15/10mm and 25/10mm. Two FGFs were used, 0.5 L.min\(^{-1}\) and 10 L.min\(^{-1}\).
- Results were obtained for gas fractions at the coaxial connector nearest the patient ACA1(1), at both inner and outer tubes, when it was found that changing coaxial pairing geometry or tubing length did not make any significant difference to system performance, though as with the standard circle system, big differences in function were demonstrated between high and low FGFs.
- Results obtained at the T-piece on the inspiratory side ATP2(1), and on the expiratory side ATP2(2) drew similar conclusions, namely that there were no differences in gas mass fractions with different tubing geometries, but that FGF played a significant role in altering system function.
- However, all results allowed a detailed insight into the function of the system. In particular, the direction of FGF at high and low FGFs, indicating the storage capacity of the system, and the detailed distribution of different gas mass fractions at different FGFs.
- Results obtained for the CO2 absorber ACS1(1) showed the difference in carbon dioxide handling at high and low FGFs, and that it was adequate in both cases; that the bulk of the pressure drop down the absorber is across the absorbent crystals, and differs significantly depending on FGF.
- Results obtained for the pressures at the unidirectional valves also gave further insight into the function of the system; in particular, there was no difference in pressure distributions between the inspiratory and expiratory unidirectional valves; in addition, while the absolute pressures upstream and downstream of the valves were very different at high and low FGFs, the overall pressure difference across the valves were identical at both FGFs.
- Overall, it can be stated from the simulations that the results are very similar to those of the standard circle system described in chapter 6. Since the coaxial system is more compact than the standard system, its use for anaesthesia should be encouraged.
Chapter 8.
The venturi device in breathing systems

8.1 Venturi theory and modelling
When delivering oxygen-enriched air to a patient by mask, the actual concentration of oxygen received depends to some extent on the patient’s inspiratory flow rate. In 1960 it was thought desirable by Nunn to provide a means of improving the stability of the oxygen concentration in such a mask by the use of the ‘Venturi’ principle [15] (after Giovanni B. Venturi 1746-1822). A venturi is a device in which flow through a throat (a narrowing), accelerates and results in a fall in pressure in that area, as predicted by Bernoulli’s equation; this mechanism is often used as a flow measurement device [88], and fluid flow through a pipe of varying diameter forms the basis of jet propulsion [89]; in addition jet pumps of similar design are used to transport solids [90].

![Venturi Diagram]

Figure 8.1
Picture representing venturi function:
1) normal function,
2) stall,
3) start of reverse flow,
4) full flow reversal.
Reproduced with permission of Dr D.G. Tilley
If a jet flow of gas is inserted into the throat, entrainment of gas from the immediate vicinity takes place and the low pressure in the throat amplifies the entrainment process, as shown in figure 8.1(1). The simplicity of the venturi to aid gas entrainment means that it has found its use in aiding gas flow around human breathing systems, whether used in underwater or space life support systems (see chapters 3 and 4); as alluded to earlier, relatively little use has been made of them in anaesthetic breathing systems [43, 44, 45]; the venturi is more frequently found in simple air entrainment systems used on clinical wards to provide oxygen-enriched air for patients. However, the simplicity of design of such an entrainment system for this use can lead to failure of function of the entrainment duct when a raised downstream pressure, or a fall in the upstream pressure leads to stalling of the duct [figure 8.1(2), and indeed reversal of the direction of gas flow [figure 8.1(3,4)]. A well-designed entrainment duct with a diffuser (a gradual enlargement from throat to outlet) is less likely to stall and reverse [15]. Such a venturi accelerates flow of gas around a breathing system in the direction of the FGF delivery, and may very well improve delivery of anaesthetic gases to and from a patient with fewer moving mechanical parts which might be prone to failure. The question is how low a FGF it is possible to achieve with such a system, given that a venturi usually requires significant FGF to function as such, even if in doing so it results in some gas economy [43]. A simplified diagram of a venturi is shown in figure 8.2 below, in which equations at the following points in the venturi are initially identified:
Inlet: \[ P_1 = P_{in} - \frac{\rho_1 u_1^2}{2} \]

Throat: \[ (P_1 - P_2)A = \rho_2 u_2^2 A - \rho_0 u_0^2 a_0 - \rho_1 u_1^2 a_1 \]

Outlet: \[ P_2 + \rho_2 u_2^2 = P_3 + \rho_3 u_3^2 + k_L \rho_2 u_2^2 \]

where \( k_L \) is the loss factor to account for the enlargement from the throat.

Continuity: \[ \rho_2 u_2 A = \rho_0 u_0 a_0 + \rho_1 u_1 a_1 = \rho_3 K A u_3 \]

Early efforts to work these equations into forms, which may be used by Bath fp, were based on Dorrington [15] in his chapter on venturi breathing systems. This was developed by Tilley (see appendix 8.1 for full derivations); in the first instance a solution is found for \( u_2 \) from a quadratic equation derived from those above:

\[ a' u_2^2 + b' u_2 + c' = 0 \]

where \[ a' = \rho_2 + M \left[ \left( \frac{\rho_2}{\rho_1} \right) \left( \frac{A}{a_1} \right) \right]^2 + \frac{N \rho_2}{2} \]

\[ b' = -2M \left[ \left( \frac{\rho_2}{\rho_1} \right) \left( \frac{a_0}{a_1} \right) \right] u_0 \]

\[ c' = M \left[ \left( \frac{\rho_0}{\rho_1} \right) \left( \frac{a_0}{a_1} \right) \right]^2 - \rho_0 \frac{a_0}{A} u_0^2 - (P_{in} - P_3) \]

and where \[ M = \frac{\rho_1}{2} - \frac{\rho_1 a_1}{A} \]
and \[ N = \frac{\rho_2}{\rho_3 K^2} + k - 1 \]

The mass flowrate \( q_2 \) at a point such as the throat is calculated as:

\[ q_2 = C_d a_2 \sqrt{\frac{2(P_2 - P_1)}{\rho_2}} \]

Note the inclusion of a discharge coefficient \( C_d \), usually less than 1, depending on Reynolds number \([91]\), to allow for velocity change and vena contracta.

It can be seen from figure 8.2 that if the outlet pressure \( P_3 \) increases, or the inlet pressure \( P_{in} \) decreases, then a point will be reached where \( u_1 = 0 \), at which point the venturi is said to have stalled, and ceases to function reliably and appropriately. Tilley derives the condition of stalling:

\[ P_3 - P_{in} = -\rho_0 u_0^2 \left[ \left( \frac{\rho_0}{\rho_2} \right) \left( \frac{a_0}{A} \right)^2 - \left( \frac{a_0}{A} \right) + \left( \frac{N}{2} \right) \left( \frac{\rho_0}{\rho_2} \right) \left( \frac{a_0}{A} \right)^2 \right] \]

A further increase in outlet pressure, or a further fall in inlet pressure, will result in flow reversal. At this time the FGF and the flow from the outlet port will pass directly back to the inlet. This situation is shown in figure 8.3.

![Figure 8.3](image)

**Figure 8.3**

Variables in equations for venturi function at reversal (see text)

In analysing this scenario, Tilley assumes that \( u_1 \) and \( u_2 \) have positive values initially, and the change of direction is accounted for at a later stage of the analysis. It is also assumed that the gases are completely mixed in the throat, and that \( P_1 = P_2 \) so that \( \rho_1 = \rho_2 \).

By considering the point at which \( u_2 = 0 \), when \( P_1 = P_2 = P_3 \), and when \( P_1 > P_{in} \), when:

\[ P_3 - P_{in} = \rho_2 \frac{u_1^2}{2} = \frac{\rho_2}{2} \left[ \left( \frac{\rho_0}{\rho_2} \right) \left( \frac{a_0}{A} \right) \right]^2 u_0^2 \]

at flow reversal.
Should \((P_3 - P_{in})\) increase further, then flow will be drawn from the outlet, \(u_2\) changes direction, and the continuity equation for this situation becomes:

\[
\rho_2 a_1 u_1 = \rho_0 a_0 u_0 + \rho_2 A u_2
\]

Using a similar equation of energy between outlet and inlet as before:

\[
P_1 + \frac{\rho_3 u_3^2}{2} = P_3 + \frac{\rho_3 u_3^2}{2} - \frac{\rho_2 \cdot K_{LR} u_3^2}{2}
\]

where \(K_{LR}\) is the loss coefficient for the gradual contraction into the throat. Once again a quadratic equation in \(u_2\) evolves:

\[
a' \cdot u_2^2 + b' \cdot u_2 + c' = 0
\]

where

\[
a'' = \frac{S \cdot \rho_2}{2} - \frac{\rho_0}{2} \left(\frac{A}{a_1}\right)^2
\]

\[
b' = -\rho_0 \left(\frac{a_0 \cdot A}{a_1^2}\right) u_0
\]

\[
c' = -\frac{\rho_2}{2} \left(\frac{\rho_0 a_0}{\rho_2 a_1}\right)^2 u_0^2
\]

and

\[
S = \left(\frac{\rho_2}{\rho_3}\right) \left(\frac{1}{K_{LR}^2}\right) - 1
\]

\(u_1\) can be calculated from continuity and \(P_1\) can be found from:

\[
P_1 = P_3 + \frac{S \cdot \rho_2 u_2^2}{2}
\]

When flow reversal occurs \(u_1\) and \(u_2\) become negative compared to normal operation, and this is accounted for in the Bath fp model by simply changing their signs from + to − since the velocity vector changes direction. This change of sign also ensures that mass flows in the model change direction as well.
In the transitional region, between normal function and reverse flow operation, the velocity $u_2$ will reduce from the stall condition, when $u_1 = 0$, to the point where reverse flow occurs. As this is likely to be a small flow region, Tilley proposed the assumption that the velocity $u_1$ varies linearly with the pressure difference acting across the venturi $(P_{in} - P_3)$ from $u_1 = 0$ at stall to $u_1 = (\rho_0.a_0.u_0/\rho_1.a_1)$.

When testing the equations associated with flow reversal, it was found that the pressure difference $(P_3 - P_{in})$ at stall was greater than at onset of reverse flow. This presented difficulties within the venturi model, and the pressure drop was converted into a rising characteristic when changing from normal to reverse flow. This was achieved in the transition region as follows:

At stall ($u_1 = 0$) \hspace{1cm} pressure difference (p.d.) = stall p.d.
At reverse flow ($u_2 = 0$) \hspace{1cm} p.d. = stall p.d. + reverse flow p.d.

The velocities $u_1$ and $u_2$ were assumed to vary linearly across the region. The modified pressure drop was then included in the reverse flow equations to ensure that pressures and flows agreed at the junction between the transitional and reverse flow regions.
8.2 Bath fp Modelling of circle system with venturi

8.2.1 Introduction
Part of this thesis is to examine whether a venturi, as described above, can be used to accelerate gas flows around a circle anaesthetic breathing system, the venturi acting as a flow director, and therefore replacing the unidirectional valves. While it has been shown in the past that a venturi can act as a FGF economising device, it remains to be shown how low the FGF can go for the venturi to function effectively. Furthermore having shown that a coaxial circle system functions as a circle system should, it was decided to use the coaxial model of the circle system for the venturi simulations. Bath fp simulation provides an opportunity to show this.

8.2.2 Methods
In this section results are presented from Bath fp runs for the circle system with a venturi to accelerate flow around the system; therefore no unidirectional valves were included in the model, and a coaxial version of inspiratory and expiratory limbs, was used, as shown in figure 8.4 with labelled components. Initially a number of variables were standardised as indicated, to allow comparisons to be made between subsequent changes in geometrical configurations, as before. The standardised configuration included: the coaxial tube ACP2 with length 1m, internal diameter of the outer (expiratory) tube of 21 mm, internal diameter of the inner (inspiratory) tube of 10 mm referred to as the 21/10mm coaxial model); the venturi with a throat diameter of 6mm and a FGF delivery nozzle of 2 mm diameter (referred to as the 6/2mm venturi model); FGF of 3 L/min; a tracheal model was also included, as it was found that this gave more appropriate values of expired CO₂. Subsequent changes in geometry included coaxial models of 15/10mm, 25/10mm and 25/12mm; although such changes in coaxial geometry were not shown to change function in the standard coaxial system, it was thought worth investigating the effect of such change on the function of the venturi. Changes in venturi geometry included the 6/1mm model, in other words the throat diameter was kept the same, and the FGF nozzle diameter was halved; this was to see if the possible increase in gas velocity would alter entrainment around the breathing system. Subsequent changes in FGF included FGFs of 0.5, 6 and 10 L.min⁻¹.
8.2.3 Results and Discussion

8.2.3.1 Gas fractions at patient attachment ACA1

Figure 8.5 shows gas mass fractions outside port 1 (nearest the patient) of ACA1, the last coaxial connector before the patient’s mouth. All results shown here refer to a geometrical configuration, which includes ACP2 coaxial tubing of standard external and internal diameters of 21 mm and 10 mm respectively (21/10mm). Figures 8.5 a) to c) refer to FGFs of 3, 0.5 and 6 L.min$^{-1}$ respectively (standard, low and high FGF respectively, to compare to standard flow), and a venturi VEN01R with throat diameter 6mm and FGF nozzle diameter of 2mm (6/2mm); in figures 8.5 d) to f) the same three FGFs are depicted, this time with VEN01R dimensions of 6/1mm (same throat diameter, smaller diameter FGF nozzle). At this point in the system there is a mixture of inspired and expired gases, so these figures give a superficial view of changes in gas mass fractions. While a FGF of 6 L.min$^{-1}$ leads to a more rapid rise in nitrous oxide, a more rapid fall in nitrogen, and lower carbon dioxide fractions than at a FGF of 3 L/min (as with other systems, and as expected), the rate of rise of oxygen is about the same at both FGFs; this applies to both the 6/2mm and the 6/1mm configuration. At a FGF of 3 L.min$^{-1}$, but not at 6 L.min$^{-1}$, the 6/2mm model results in a more rapid rise in nitrous oxide, and fall in nitrogen, than the 6/1mm model. At a FGF of 0.5, L.min$^{-1}$ the oxygen fractions actually fall, the nitrous oxide fractions rise very slowly (as with other systems), and the carbon dioxide fractions rise, these changes happening to the same extent in both 6/2mm and 6/1mm configurations.
Figure 8.5 a), b), c)
Gas mass fractions at port 1 of ACA1 for coaxial tube of 21/10 mm configuration and venturi of 6/2mm configuration. a) FGF = 3 L.min$^{-1}$, b) FGF = 0.5 L.min$^{-1}$, c) FGF = 6 L.min$^{-1}$. Curves: [1] = O$_2$, [2] = N$_2$, [3] = CO$_2$, [4] = N$_2$O.
Figure 8.5 (cont’d) d), e), f)
Gas mass fractions at port 1 of ACA1 for coaxial tube of 21/10 mm configuration and venturi of 6/1 mm configuration. d) FGF = 3 L/min, e) FGF = 0.5 L.min⁻¹, f) FGF = 6 L.min⁻¹. Curves [1] = O₂, [2] = N₂, [3] = CO₂, [4] = N₂O
Figure 8.6 also shows gas mass fractions outside port 1 of ACA1, this time using the model in which ACP2 has diameters 25/10 mm configuration, FGF=3 L.min$^{-1}$. a) 6/2mm venturi mode, b) 6/1mm venturi model. Curves [1] = O$_2$, [2] = N$_2$, [3] = CO$_2$, [4] = N$_2$O.

Figure 8.6 also shows gas mass fractions outside port 1 of ACA1, this time using the model in which ACP2 has diameters 25/10mm (wider diameter outer tube), at a FGF of 3 l.min$^{-1}$. Results for the 6/2mm venturi model are shown in a) and for the 6/1mm venturi model in b). It seems that this change in geometry does not change the mass fraction profiles described for the 21/10mm model.

Figure 8.7 shows the same results, this time for ACP2 model 25/12mm, once again with no significant difference between this model and the others in terms of the gas fraction profiles described. For all models of ACP2 at a FGF of 6 L.min$^{-1}$, although not shown here, it was found that there is smaller inspiratory to expiratory variation in oxygen profiles at ACA1 in all cases, as equilibrium is approached, due to the FGF dominating the function of the system. A typical range of pressures inside ACA1 at a FGF of 3 L/min was found to be about 0.3 to 3.0 mbar for the 25/12mm tube model at a FGF of 3 L.min$^{-1}$.\[138\]
Figure 8.8 shows a comparison of gas mass fractions at a FGF of 10 L.min$^{-1}$ for the 6/2mm venturi configurations of the 21/10mm, 25/10mm and 25/12mm ACP2 coaxial tube models. Although the intention would be to use such a breathing system at low FGF, this run was carried out to see if the venturi function at high FGF could lead to differences depending on coaxial tube configuration. However, figure 8.8 reveals no such differences. If figure 8.8a) is compared to figures 8.5, 8.6 and 8.7, there is merely a more rapid change in gas fractions, as expected.

Figure 8.7. Gas mass fractions at port 1 of ACA1 for coaxial tube of 25/12 mm configuration, FGF=3 L.min$^{-1}$. a) 6/2mm venturi mode, b) 6/1mm venturi model. Curves [1] = O$_2$, [2] = N$_2$, [3] = CO$_2$, [4] = N$_2$O.
Figure 8.8
Gas mass fractions at port 1 of ACA1, using 6/2mm venturi model, $FGF = 10 \text{ L.min}^{-1}$.

a) 2/10mm model,

b) 25/10mm model,

c) 25/12mm model.

8.2.3.2 Gas fractions in coaxial pipe volume ACP2

Figure 8.9 shows gas mass fractions in the main coaxial tube ACP2, in the outer and inner pipes (outer expiratory and inner inspiratory tubing, respectively), for ACP2 coaxial model 21/10mm, at two FGFs of 3 and 6 L.min\(^{-1}\) for the venturi 6/2mm model [a) to d]), and for the venturi 6/1mm model at a FGF of 3 L.min\(^{-1}\) [e) and f]). ‘In the pipe’ in each case means at a point closer to the valves than the patient, as ACP2 was set up with a restrictor at its mid-point. Comparing first of all figures 8.9 a) and b), at 3 L.min\(^{-1}\) FGF, there is a more rapid initial rise in oxygen and nitrous oxide, and a more rapid initial fall in the nitrogen gas mass fractions in the inner pipe than the outer, and a much lower carbon dioxide fraction in the inner pipe than the outer, as would be expected. Comparing curves at FGF of 3 L.min\(^{-1}\) [a) and b)] with curves at FGF of 6 L.min\(^{-1}\) [c) and d)], there is once again a more rapid rise and fall in gas fractions at the higher FGF, as expected. Comparing curves of the 6/1mm model [e) and f]), with a) and b) (6/2mm model, both at FGF 3 L.min\(^{-1}\)), there is no difference in the rates of rise and fall of gas mass fractions, with one exception; the carbon dioxide fractions in the outer pipe are lower for the 6/1mm model than the 6/2 model at a FGF of 3 L.min\(^{-1}\); this might suggest that, with the narrower 1mm venturi FGF delivery tube, there is an advantage in carbon dioxide handling, but not over the handling of the other gases. The other notable difference between outer and inner tube mass fraction profiles is that the inspiratory to expiratory waveform differences are more marked in the inner tube than the outer tube for the 6/2mm model, presumably due to more gas mixing; this difference is less noticeable for the 6/1mm model. In considering the same sets of results for gas mass fractions in the outer and inner pipe volumes of ACP1, this time using the 25/10mm and 25/12 mm models of the coaxial tube, standardised with venturi model of 6/2mm and a FGF of 3 L.min\(^{-1}\), no noticeable differences in gas mass fractions in the pipe volumes were found.

Figure 8.10 shows the fall in nitrogen mass fractions in the outer and inner pipes of coaxial pipe ACP2, for the 21/10mm coaxial tube model; a) and d) are for both at a FGF of 3 L.min\(^{-1}\), b) is at a FGF of 0.5 L.min\(^{-1}\), and c) is at a FGF of 10 L.min\(^{-1}\) (standard FGF followed by low and high FGFs for comparison); all curves show a more rapid fall in the inner pipe, compared to the outer pipe, which is as expected, and this difference is more marked at low FGFs than higher, while the overall fall is more marked at the higher FGF, as expected. When comparing a) the 6/2mm venturi model with d) for the 6/1mm venturi model both at FGF of 3 L.min\(^{-1}\), the difference in fall between outer and inner pipe is greater with the 6/2mm model than with the 6/1mm model, although the overall fall in both models is the same. Looking at curve b) where the FGF is 0.5 L.min\(^{-1}\), the outer volume N\(_2\) mass fraction does not fall quickly, while the inner pipe N\(_2\) initially falls much more quickly; this is presumably because of the addition of FGF to the gas mixture,
and because of the venturi effect, even if this is small. Curve c) shows how rapidly both outer and inner

![Graph a) outer pipe volume](image1.png)

![Graph b) inner pipe volume](image2.png)

![Graph c) outer pipe volume](image3.png)

![Graph d) inner pipe volume](image4.png)

Figure 8.9. Gas mass fractions at outer and inner pipe volumes of coaxial tube ACP2, 21/10mm model, with 6/2mm venturi model. Upper pair a) and b) FGF = 3 L.min⁻¹
N₂ mass fractions fall at high FGF of 10 L.min⁻¹, because the whole system is dominated by the high FGF. No differences were found for the 25/10mm and 25/12mm models of ACP2, from the 21/10mm model, and the results are therefore not shown here.

Figure 8.11 shows the change in oxygen mass fractions in coaxial pipe ACP2. To some extent these are mirror images of the nitrogen curves shown in the previous figures, differences of course lying in the equations of their individual kinetics. a) shows a faster rise of mass fraction in the inner pipe than the outer at a FGF of 3 L.min⁻¹, which is to be expected; b) shows that at a FGF of 0.5 L.min⁻¹, the oxygen mass fraction in the outer pipe volume actually falls, while that in the inner pipe rises rapidly to a plateau, presumably due to the effect of the accelerated FGF by the venturi, then starts a slow decline; the inspiratory to expiratory variation is greater at this low FGF than at higher FGFs; c) shows a rapid rise in both outer and inner oxygen mass fractions, as the system is swamped by the high FGF; d) shows that the 6/1mm model of the venturi results in a rise of both outer and inner oxygen mass fractions together compared to the 6/2mm
venturi model. Once again it was found there is no difference in oxygen mass fractions at corresponding FGFs and venturi model.

Figure 8.10. Nitrogen mass fractions in outer and inner volumes of coaxial pipe ACP2, 21/10mm coaxial model. a), b) and c) 6/2mm venturi model; d) 6/1mm venturi model.

a) FGF = 3 L.min⁻¹  6/2mm venturi  b) FGF = 0.5 L.min⁻¹  

c) Venturi 6/2mm, FGF=10 L.min⁻¹  d) Venturi 6/1mm, FGF=3 L.min⁻¹

Figure 8.11. Oxygen mass fractions in outer and inner volumes of coaxial pipe ACP2, 21/10mm model. a), b) and c): 6/2mm venturi model: a) FGF = 3 L.min$^{-1}$, b) FGF = 0.5 L.min$^{-1}$, c) FGF = 10 L.min$^{-1}$; d) 6/1mm venturi model, FGF = 3 L.min$^{-1}$
Figure 8.12. Nitrous oxide mass fractions in outer and inner volumes of coaxial pipe ACP2 21/10mm model, a), b) and c) 6/2mm venturi model. a) FGF 3 L.min\(^{-1}\). b) FGF 0.5 L.min\(^{-1}\) c) FGF 10 L.min\(^{-1}\). d) 6/1mm venturi, FGF = 3 L.min\(^{-1}\)
Figure 8.13. Carbon dioxide mass fractions in outer and inner volumes of coaxial pipe ACP2 21/10mm model, 6/2mm venturi model. a) FGF 3 L.min$^{-1}$, b) FGF 0.5 L.min$^{-1}$, c) FGF 10 L.min$^{-1}$, d) FGF 3 L.min$^{-1}$, 6/1mm venturi model. Curves: [1] outer pipe volume, [2] inner pipe volume.
Figure 8.13 (cont’d). Carbon dioxide mass fractions in outer and inner volumes of coaxial pipe ACP2, FGF 3 L min\(^{-1}\). ACP(1)25/10mm model; e) 6/2mm venturi model; f) 6/1mm venturi model. Curves: [1] outer pipe volume, [2] inner pipe volume.

Figure 8.13 (cont’d). Carbon dioxide mass fractions in outer and inner volumes of coaxial pipe ACP2 25/12mm model, FGF 3 L/min. g) 6/2mm venturi model; h) 6/1mm venturi model. Curves: [1] outer pipe volume, [2] inner pipe volume.
dimensions between the three coaxial tube model configurations of 21/10mm, 25/10mm and 25/12mm, which corresponds to the other findings thus far.

Figure 8.12 shows the nitrous oxide mass fractions in coaxial pipe ACP2, which are broadly similar to those of the oxygen curves. The exception is in b) at a FGF of 0.5 L.min$^{-1}$ where the inner volume $N_2O$ rises and plateaus, and the outer pipe does not rise; the nitrous oxide curves slowly rise unlike the oxygen curves which decline slowly after an initial rise. This difference will lie in the different equations in Bath fp governing $O_2$ consumption and $N_2O$ uptake; the exact profile of both sets of curves will depend on precise values of $O_2$ consumption and $N_2O$ uptake in an individual patient.

Figure 8.13 shows carbon dioxide mass fractions at outer and inner pipes of the coaxial pipe ACP1(1). a) to c) show, as expected, that the lower the FGF, the higher the carbon dioxide mass fractions, in both outer and inner pipe volumes. It is worth noting that the peak inspiratory values in the inner pipe at the end of the 120 second run, respectively at 10, 3 and 0.5 L/min are approximately 0, 0.5 x$10^{-2}$ and 1.5 x$10^{-2}$ (corresponding to 0%, 0.7%, and 2.1%) respectively, with the 0.5 L.min$^{-1}$ FGF value still rising.

Interestingly, the curves, which are associated with a 6/1mm venturi model (for the three coaxial tube models), show significantly lower inspiratory and expiratory CO2 mass fractions in the outer pipe, at least at the standard FGF of 3 L.min$^{-1}$. In the inner pipe, the inspiratory values of carbon dioxide are lower with the 6/1mm model than with the 6/2mm model, even if the expiratory value here is unchanged. This must be due to the accelerating effect of the narrower venturi in each case.

8.2.3.3 Gas behaviour at the venturi VEN01R

Figure 8.14 shows gas mass fractions in the connectors, CONN1(2) and CONN1(1), upstream and downstream respectively of the venturi VEN01R. As expected, comparing a) and b) shows more rapid initial changes in gas fractions in the downstream connector CONN1(1) than in the upstream connector CONN1(2). Comparing c) and d), as usual the high FGF of 10 L.min$^{-1}$ results in more rapid gas mass fraction changes than low FGF of 0.5 L.min$^{-1}$, in which in fact there is almost no change in mean gas mass fractions despite the respiratory cyclical variation; at such a low FGF the respiratory pattern is bound to dominate the venturi itself, even this far from the patient. e) shows that the initial rate of change of gas mass fractions is lower with the 6/1mm venturi model compared to the 6/2mm venturi model, even though the overall changes are no different; presumably this is because of a lower FGF emerging from the venturi itself; there is also a greater loss of inspiratory to respiratory swing with the 6/1mm model, presumably because the narrower venturi makes its function more
Figure 8.14. Gas mass fractions at the connectors upstream [CONN1(2)] and downstream [CONN1(1)] of the venturi VEN01R model 6/2mm, using the coaxial pipe model 21/10mm. a) CONN1(2) [upstream] b), c), d) CONN1(1) [downstream]. a) and b) FGF 3 L.min⁻¹. c) FGF 0.5 L.min⁻¹ d) FGF 10 L.min⁻¹. Curves [1] = O₂, [2] = N₂, [3] = CO₂, [4] = N₂O.
e) 21/10mm coaxial, 6/1mm venturi model

f) Partial pressures in CONN1(1), 25/12mm, 6/2mm venturi model, FGF=3L.min⁻¹

Figure 8.14 (cont'd). e) Gas mass fractions at the connector downstream [CONN1(1)] of the venturi VEN01(1), FGF 3 L.min⁻¹. Using coaxial tube model 21/10mm, VEN01(1) model 6/1mm. f) Partial pressures of gases in mbar, in connector CONN1(1) for 25/12mm coaxial model, 6/2mm venturi model, FGF=3 L.min⁻¹ (compare with 8.14b) above.

effective. No differences were found in the rate of change of gas mass fractions in the downstream connector CONN1(1) where coaxial pipe ACP1(1) models 25/10mm and 25/12mm models are used. f) is the same as b) except given as partial pressures in mbar.

In describing and discussing results for the venturi itself, new terminology must be explained. It has been discussed that the purpose of a venturi is to accelerate flow around a system by entraining upstream gas into the venturi flow (in this case the venturi flow is the FGF); the volumetric flow entrainment ratio is defined as the ratio of entrained volume flow to venturi volume flow. The pressure drop across the venturi is the actual pressure difference between the output and the input of the whole venturi component (between port 3 and port 2 in figure 8.4); the stall pressure drop indicates how big that pressure difference would have to be, all other conditions of geometry and gas flowrate being unchanged, for the venturi to stall, i.e. not function as a venturi. Figure 8.15 shows the volumetric flow entrainment ratios, pressure drops and stall pressure drops in the venturi VEN01R. a) to d) are for the 21/10 mm tube model, a) to c) for the 6/2mm venturi model at different FGF’s, d) for the 6/1mm model at a FGF of 3 L.min⁻¹. Looking at a) to c) it can be seen that the higher the FGF, the more closely the volumetric flow entrainment ratio adheres reliably to a value of about 1.9; in fact b) shows that at the low FGF of 0.5 L.min⁻¹ the flow entrainment ratio actually varies between large positive and negative values as the breathing cycle goes between inspiration and expiration; in other words the venturi effect is undermined by the low FGF. Comparing a) and d), it can be seen that the 6/1mm venturi model with a FGF of 3 L.min⁻¹ results in a volumetric flow entrainment ratio of between 4.8 and 6.0, whereas the 6/2mm venturi model under the same FGF results in a volumetric flow ratio of 1.3 and 2.1. Comparing a) for the 21/10 tube model, e) for the 25/10mm tube model, and g) for the 25/12mm model, both the latter two representing the 6/2mm venturi model at a FGF of 3 L.min⁻¹, it can be seen that there is a slight rise in the volumetric entrainment ratio as the inner and outer coaxial tubing increases in diameter; increasing the outer tube diameter to 25mm increases the ratio to 1.7 on expiration, and increasing the inner tube to 12mm increases both expiratory and inspiratory ratios, to 1.9 and 2.2 respectively. When comparing d), f) and h) for the 6/1mm venturi, this effect is even more marked, with mean entrainment ratios rising from 5.5 to 6.0 to 7.0 respectively for the 21/10mm, 25/10mm and 25/12 mm tube models. Turning to the matter of the pressure drop across the venturi and the stall pressure drop, clearly the higher the stall pressure drop in relation to the actual pressure drop, the less likely the venturi is to fail in its function of entrainment. Comparing a) to c) it can be seen that the stall pressure drop rises from 0 to 0.3 to 4.5 mbar as FGF increases from 0.5 to 3.0 to 10 L.min⁻¹, with the gap between actual pressure drop and stall pressure drop also increasing. If a) and d) at the same FGF are compared, the stall pressure drop increases from 0.3 to 1.8 mbar between the 6/2mm and 6/1mm venturi models respectively, with the gap between actual pressure drop and stall pressure drop.
also increasing. Comparing a), c) and g) with each other (6/2mm venturi), or d), f) and h) with each other (6/1mm venturi model), it can be seen that changing coaxial tubing

Figure 8.15. Flow entrainment ratio pressure drop and stall pressure drop at venturi VEN01R, using coaxial pipe model 21/10mm. a) to c) 6/2mm venturi model. a) FGF 3 L.min⁻¹, b) FGF 0.5 L.min⁻¹, c) FGF 10 L.min⁻¹. d) 6/1mm venturi model, FGF 3 L.min⁻¹. Curves: [1] = volumetric flow ratio; [2] = pressure drop, [3] = stall pressure drop.
Figure 8.15 (cont’d). Flow entrainment ratio, pressure drop and stall pressure drop at venturi VEN01R, using coaxial pipe model 21/10mm, FGF=3 L.min⁻¹. e), f) 25/10mm coaxial model; e) 6/2mm venturi, f) 6/1mm venturi g), h) 25/12mm coaxial model; g) 6/2mm venturi, h) 6/1mm venturi . Curves: [1] volumetric flow ratio; [2] pressure drop; [3]=stall pressure drop.
diameters from 21/10mm to 25/10mm to 25/12mm does not change the stall pressure drop, even if the gap between stall pressure drop and actual pressure drop increases slightly. Nevertheless, these results all suggest that the highest volumetric entrainment ratio with the greatest difference between stall pressure drop and actual pressure drop is achieved by having the smaller venturi FGF nozzle with the largest external and internal diameter of coaxial tube. Table 8.1 below shows the minimum FGFs required to prevent stalling of the venturi for the 21/10mm and 25/10mm tube models. In each case the 1mm venturi FGF nozzle requires a lower FGF than the 2mm nozzle, and the 25/10mm tube model requires a lower FGF than the 21/10mm tube model. It is interesting to note also that a lower respiratory rate by the patient is less likely to destabilise the venturi. The table also suggests that a FGF of slightly more than 0.5 L.min⁻¹ is required to prevent stalling. The table also supports the result displayed in figure 8.1b) and which suggests that very low FGF cannot be used with any of the venturi models used in this work, and it would certainly indicate why engineers have not thus far encouraged anaesthetists to use a venturi in a clinical setting.

In examining the behaviour of the venturi, it is also interesting to note what happens if the venturi flow coefficients are varied. All the tests done in this work have been with a $C_q$ of 0.9. Figure 8.16 shows what happens if $C_q$ varies between 0.8 and 1.0. It can be seen from a) that, while an increase in $C_q$ will increase volumetric entrainment ratio, b) shows that this has a negligible effect on the mass fraction of oxygen in the connector immediately downstream of the venturi. It will be recalled from chapter 5 that although hysteresis in the behaviour of the reservoir bag was considered, it was not included in the results, as there was a problem avoiding ‘spiky’ traces when it was included. It was found that, while ignoring reservoir bag hysteresis caused a 5% difference in calculating reservoir bag

<table>
<thead>
<tr>
<th>Venturi throat dia (mm)</th>
<th>Venturi FGF dia (mm)</th>
<th>10 Breaths/min Fresh Gas Supply (L.min⁻¹)</th>
<th>15 Breaths/min Fresh Gas Supply (L.min⁻¹)</th>
<th>Co-axial tube dia (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>2</td>
<td>1.7</td>
<td>2.5</td>
<td>21/10</td>
</tr>
<tr>
<td>6</td>
<td>1</td>
<td>0.8</td>
<td>1.2</td>
<td>21/10</td>
</tr>
<tr>
<td>6</td>
<td>2</td>
<td>1.1</td>
<td>1.7</td>
<td>25/10</td>
</tr>
<tr>
<td>6</td>
<td>1</td>
<td>0.55</td>
<td>0.8</td>
<td>25/10</td>
</tr>
</tbody>
</table>

Table 8.1 – Minimum fresh gas supply to prevent stall using 21/10mm and 25/10mm coaxial tubing. Tidal volume = 0.5L (without bag hysteresis) (initial bag vol = 1L, venturi $C_{q1} = 0.9$, $C_{q2} = 0.5$, $k_c = 0.3$, $APL = 2mbar$)
Figure 8.16. Varying venturi flow coefficient 6mm/2mm venturi, FGF=3 L.min$^{-1}$, 21/10mm coaxial tubes. Curves: [1] Cq=0.9, [2] Cq=1.0, [3] Cq=0.8.

Figure 8.17. Oxygen mass fractions downstream of the venturi [at CONN1(1)] for different APL Valve Settings, 6mm/2mm venturi model, FGF= 3 L.min$^{-1}$ 21/10mm coaxial model. Curves: [1] 1mbar, [2] 2mbar, [3] 3mbar, [4] 4mbar
volume, this had a negligible effect on the oxygen fractions in the connector downstream of the venturi. A typical range of values for the reservoir bag’s volume was 1.5 to 1.9 L with respiration.

Another consideration which was modelled is the setting of the pressure limiting (overflow) valve AEVIa(1), which was set at 2mbar throughout. Figure 8.17 shows the variation in oxygen mass fraction downstream of the venturi at connector CONNi(1) if this valve pressure was set in the range 1mbar to 4mbar. These curves show that on each expiration, and to a lesser extent on each inspiration, higher APL valve pressure settings result in lower values of expired oxygen. This is because more of the expired gas is retained in the system for recirculation around it.

**8.3. Closure**
The Bath fp modelling results for the coaxial circle system with a venturi placed adjacent to the FGF input were presented. Runs were done at FGFs of 0.5, 1, 3, 6, and 10 L.min⁻¹, using venturi throat/FGF nozzle ratios of 6/2 and 6/1mm, and a coaxial tubing outer/inner diameter ratios of 21/10mm, 25/10mm and 25/12mm.

- As with the other systems examined, more rapid changes in gas mass fractions occurred at higher FGFs, and in the inner coaxial tube than the outer.
- It was found that changing the coaxial tubing geometry did not change the gas mass fractions at four main sites in the system, consistent with other systems modelled.
- Changing the venturi geometrical configuration from 6/2mm to 6/1mm resulted in a reduction in gas mass fraction only for carbon dioxide, but not the other gases, all other variables being kept constant.
- With regard to volumetric entrainment ratios, it was found that this was more reliably constant through the respiratory cycle at higher FGFs, and was increased with the larger coaxial tubing configurations (25/12mm > 25/10mm > 21/10mm).
- Although the actual pressure drop across the venturi increased somewhat with increased FGF, it was shown that the venturi is less likely to stall than at lower FGFs.
- Although increasing venturi flow coefficient increased volumetric entrainment ratio, this did not alter the downstream oxygen gas fraction significantly.
- Minimum FGFs to avoid stalling of the venturi are presented, and were found to be above 0.5 L.min⁻¹; stalling was also found to be more likely to occur at higher ventilation frequencies.
- Patient variation in respiratory patterns are not modelled by Bath fp and this may mean that further limitations of the valveless circle system are not elucidated.
• In summary, the function of the valveless, coaxial venturi circle system was not shown to work well at low FGFs, but worked as well as a valved coaxial system at moderate FGFs.
Chapter 9.
Clinical Validation

9.1 Introduction
At an early stage of the mathematical modelling of breathing systems using Bath fp, it was considered desirable to provide some clinical validation for its use in this context, since this had not occurred in the history its development. Such validation studies provide the variance in human behaviour when humans are attached to breathing systems like the ones modelled by Bath fp, to validate the model, and to see if the results obtained by Bath fp describing the system’s behaviour, also pertain in human use of such breathing systems. In particular, it was thought that gas mass fractions (usually measured clinically as percentages), and system pressures would be most easily accessible. To this end it was planned to carry out validation studies using volunteers breathing the same mixture as was used in the models, namely 50% nitrous oxide in oxygen, known clinically as “entonox”. This is a gas mixture, which is used in emergency departments and obstetric labour wards to provide pain relief without the need to fully anaesthetise the patient, as well as being the gas medium in which more potent volatile anaesthetic agents may be added. It is desirable to use the same gas mixture as in the Bath fp model in order to make appropriate comparisons, and is an acceptable surrogate for modelling other volatile agents. This is an easy and relatively safe gas mixture to have volunteers breathe through a standard anaesthetic circle breathing system, delivered from a standard anaesthetic machine. In normal clinical practice the gases are sampled for gas monitoring purposes from a single point in the breathing system close to the patient. It was planned to recruit the author’s own anaesthetic departmental colleagues as participants, on the basis that they were non-naïve to what was required, and that they were locally available. The author’s own hospital anaesthetic rooms would be used for familiarity and safety. The proposal was to recruit ten to fifteen volunteers and to use three different types of circle system, each at three different FGFs, and to sample gases from three easily accessible points in the system.

Any research involving human volunteers is subject to approval by a designated ethics committee, to which the researcher must apply and seek approval before being allowed to proceed. There is a sixty-three question, on-line application form, the IRAS form (Integrated Research Application System), which must be submitted to the applicant’s regional ethics committee of choice, with a subsidiary copy to the applicant’s local hospital research and development (R&D) office for local approval. The form asks detailed questions concerning the nature and process of the proposed study, and it took the author the first half of 2012 to complete it successfully. In addition the applicant has to submit for review by the committee, a protocol for the study, a participant information
sheet, and a consent form. The applicant then has to telephone the administration office of the ethics committee and book a date for a hearing.

At the first hearing for ethical approval, which took place at Frenchay hospital Bristol in September 2012, which the author attended, the application was turned down by the committee for a number of reasons, including the following:

- it was thought that the author would coerce colleagues into participating, and the committee asked that recruitment be extended more widely.
- the committee was concerned that the author would ask colleagues to participate on a day when they were required to work, with obvious hazard to patients;
- the committee felt that there was not enough back up for the safety of the participants if anything went wrong.
- the committee doubted the necessity for human validation trials, suggesting that the author should instead use an ‘artificial lung’ to ‘test the breathing system’.

The author undertook a number of changes, and gathered supporting letters from engineering and clinical colleagues to reassure the committee. The whole application package was resubmitted and at the second hearing in January 2013, approval was granted with some further provisos.

A first batch of seventy invitations was sent to potential participants in April 2013, and only five expressions of interest were received, one of which was withdrawn. The study protocol, the participant information sheet, the opt-in slip, and the consent form are included as appendix 9.1. The responding participants indicated their willingness by signing the opt-in slip. They were then contacted by the author, when a conversation took place, in which the participant information sheet was gone through in some detail, in order to appraise the participant of the details of the process of the trial, the potential hazards and the safety measures in place. Emphasis was placed on the author’s own extensive experience as an anaesthetist himself, and the adherence to internationally approved standards of anaesthetic care and monitoring [92], as the means of providing due care and safety for the participants, as if they were the author’s own patients. A consent form was then signed and a date for each trial was arranged. A further batch of twenty-five invitations to participate was sent out in July 2013, and this resulted in two more expressions of interest, one of which was withdrawn. Although the initial recruitment response was poor, as time went by more colleagues, friends and family volunteered to take part, and ultimately there were nineteen volunteers recruited to the study between April and December 2013, with the first trial being carried out in July, and the last in December.
In carrying out such a clinical trial, it was necessary to find a means of capturing the data recorded from a participant attached to the breathing system, which in normal clinical practice is recorded on the monitors of the anaesthetic machine itself. To this end the

Figure 9.1. Examples of live (online) data recorded from the AS5 anaesthetic machine by S5 Collect and downloaded to Windows XP platform.
local sales representative, Mr Stephen Hancock, of Datex-Ohmeda, sourced the software, to capture such data, both from the participant’s physiology and from the anaesthetic machine and breathing system. The software, ‘S5 Collect’ (93, 94), allows the data to be transferred from the anaesthetic machine monitors to a laptop computer where it can be stored for further analysis; the format of the data collection is shown in figure 9.1. This software was bought and tested by the author, and after some delay in sourcing the correct series of cables and RS232-USB connectors, and in recognising the correct port on the appropriate anaesthetic machine from which to download the data (not all AS5 machines were so equipped), the data collection system worked appropriately. ‘S5 Collect’ software only works with a Windows XP platform.

Due to the initial poor response to the invitation to participate, the author thought a back-up plan for the clinical validation study would be desirable. The Bristol Medical Simulation Centre (BMSC) was contacted, who offered to help with the trials using their physiologically sophisticated mannequin (95). BMSC is one of only a handful of medical simulation centres in the UK, whose purpose is to help train junior medical and other clinical staff. The mannequin was developed by Medical Education Technologies Inc. (METI), to allow medical staff to learn from potentially hazardous clinical scenarios without endangering real patients. The mannequin, named ‘Stan D. Ardman’ (stan-d-ard man), is designed to be a 70 kg, physiologically fit 33 yr old adult male. One of the features of sophistication of the mannequin is that it possesses its own realistic physiology, and does not allow unrealistic physiological models to be pre-set. At the first visit to BMSC in June 2013, when the author tried to set up a tidal volume of 500 ml fixed to a respiratory rate of 10 breaths/minute in order to match those variables used in Bath fp, Stan D. Ardman went into an unrecoverable physiological decline. It emerged that the feedback loop on the gas sampling of Stan’s function was quite crude in the nature and timing of its response; the expired CO\textsubscript{2} was fed back along a tube about one metre long and fed into the simulator to invoke changes in ventilation, resulting in a seven minute cycle time between cyclical maximum values of CO\textsubscript{2}; it was therefore not possible to make meaningful comments on the handling of CO\textsubscript{2}, or on O\textsubscript{2} and N\textsubscript{2}O changes, or on information about pressure, volume and flow rate in the system. It subsequently transpired that there was a fault with the pump, which transmitted sampled gas back to the analysers, which required a major overhaul of the mannequin. The results for this visit to BMSC were abandoned, the overhaul of the mannequin took two months, and a further visit by the author to BMSC took place in October 2013, when results were successfully obtained.
9.2. Human volunteer breathing trials

9.2.1 Recruitment
As indicated above, although the first two batches of invitations to recruitment had a poor uptake (five recruited out of ninety-five invitations), by further informal communication, a total of nineteen volunteers was recruited. Of these, nine were female, and ten were male. Six were born prior to 1980 (>32 years old, mean age 48.1 yrs, age range 38-54yrs), and thirteen were born in or after 1980 (≤32 years old, mean age 26.6 yrs, age range 20-32yrs). All were physically fit, with no intercurrent illnesses of note. None was due to work until the following day at the earliest. There were three postponements, on account of temporary upper respiratory tract infections or lack of anaesthetic machine availability, but there were no further withdrawals or cancellations.

9.2.2 Methods
The volunteers were encouraged to read the patient information sheet, giving a detailed explanation of the procedure, emphasising that they could withdraw their consent at any time; the information sheet was gone through, and finally a formal confidential medical history, relevant to anaesthesia, was obtained prior to obtaining written consent. Pre-trial starvation was not required, since this was not anticipated to have the same ‘pre-operative’ requirements as a general anaesthetic. It was also not deemed necessary to obtain intra-venous access unless it became necessary.

The volunteer was made comfortable on an operating table, usually in a semi-recumbent or near-sitting position. An automatic (non-invasive) blood pressure cuff was wrapped around an upper arm, and set to measure every fifteen minutes. A pulse oximeter probe was attached, either to a finger on the opposite hand or to an ear lobe to monitor oxygenation and pulse rate.

A standard anaesthetic breathing mask was placed on the volunteer’s face and the system was made gas tight by an elastic harness around the head, as shown in figure 9.2.

Figure 9.2. Participant with breathing mask attached. With permission from Ms. R.I. Magee
The volunteer was attached to three different breathing systems in turn:
- the 2m standard adult circle system of tubing diameter 22mm, shown in figure 9.3a;
- the 2m paediatric circle system of tubing diameter 15mm shown in figure 9.3b;
- the 1.6m coaxial circle system of outer diameter 25mm and inner diameter 15mm, shown in figure 9.3c.

After ensuring comfort of the volunteer, (s)he was then asked to breath air at high FGF (>10 L/min) for several minutes, in order to get used to breathing through the system, and to ensure that the system itself was primed with air. Baseline readings of gas concentrations were taken. The gas mixture was then changed to 50% oxygen in nitrous oxide, at a FGF of 10 L/min for between 70 seconds and 180 seconds; the entonox breathing period was then terminated, by changing the gas mixture to high flow oxygen in air, at least 12 L/min. The oxygen enrichment in this case was to counter any ‘diffusion hypoxia’, which might otherwise happen as high concentrations of nitrous oxide are excreted by the lungs, diluting any air being inspired at that time; this occurs because nitrous oxide is thirty times more soluble in body tissues than nitrogen, the biggest
component of air. After several minutes (usually between one and three minutes), when the inspired nitrous oxide concentrations had fallen to less than 1%, the gas mixture was changed to high flow air once again, until expired nitrous oxide gas concentrations were reduced to between 1 and 2%, and the inspired and expired oxygen gas concentrations reached 20 - 21% and 15 – 17% respectively, this usually occurring after a further three to five minutes. As time progressed throughout the trial, the volunteer’s body tissues absorbed increasing amounts of nitrous oxide, and the time required to excrete the nitrous oxide during the air breathing periods became prolonged; in fact a pragmatic compromise was required to avoid prolonging the trial to an unacceptable length of time, and later entonox breathing periods started with an expired N₂O concentration of between 2 and 4%; this was accounted for in the calculations associated with the results, after discussion with a colleague, a national expert in respiratory physiology applied to anaesthesia (Dr A. Lumb [96]).

For each breathing system, intermittent entonox breathing periods were carried out, at three FGFs, 10 L/min, 3 L/min, and either 1 /min or 0.5 L/min, with appropriate high flow air/oxygen and air between these FGFs, as described above. In addition, for each breathing system at all three entonox FGFs, gas sampling took place at three different locations: near the volunteer (as is the practice clinically, referred to as ‘patient sampling’), at the end of the inspiratory breathing tube furthest from the volunteer (‘inspiratory sampling’), and at the end of the expiratory breathing tube furthest from the volunteer (‘expiratory sampling’). These points were chosen for ease of access, as the latter two are adjacent to the tubing attachments to the carbon dioxide absorber. The trial for each breathing system took about one hour, so a complete trial for three systems took three hours. The data was recorded by the standard monitoring systems available as modules integrated with the anaesthetic machine, which in all cases was the Ohmeda – Excel AS5 machine [97], shown in figure 9.4 below, and downloaded to a laptop PC using the S5-Collect software. At the end of each session a set of calibration results was carried out to obtain 25%, 50%, 75% and 100% oxygen with converse values of nitrous oxide; this was done by setting constant, high FGFs on the anaesthetic machine rotameters of the appropriate gas concentrations.

Every volunteer was cared for by the author as if (s)he were a surgical patient undergoing anaesthesia. This involved frequent and close visual observation, intermittent verbal communication, and the recording of non-invasive blood pressure, pulse rate and oxygen saturation, as well as the information on inspiratory and expiratory oxygen, nitrous oxide and carbon dioxide, airway pressure, tidal volume and respiratory rate. A formal written anaesthetic chart of the first three data variables, and the timing of changes in breathing gas mixtures, was kept, which also helped in tracking the FGFs and breathing systems.
9.2.3 Results
Two volunteers on the day had less time available than they had anticipated. In this case the amount of data collection was reduced, usually to two rather than three different FGFs for each breathing system (10 L/min and 1 or 0.5 L/min); this reduced the trial time to two hours.

The first set of data collected was qualitative information about the comfort and effort of breathing through different systems at different flow rates. In general, everybody found the systems equally (un)comfortable to breathe through; no-one found the paediatric system or the coaxial system more difficult to breathe through than the standard system; no-one found more difficulty breathing at low FGFs than at high FGFs; three volunteers found slightly higher (though acceptable) resistance to expiration at high FGF, presumably due to the intrinsic positive end expiratory pressure so caused.

Apart from matching the gas mixture to that modelled in Bath fp, the advantage of using entonox in this trial was that it was possible to maintain constant verbal communication with the volunteer. This fact helped early identification of problems, which occurred with two volunteers. Volunteer F4 momentarily lost consciousness at one point during high FGF using the standard circle system; the author applied jaw lift to ensure the airway was not lost, and changed the entonox mixture to oxygen with air; recovery was rapid and the trial continued with somewhat reduced data collection. Volunteer F9 lost consciousness after just ten breaths of high FGF entonox breathing using the standard circle system; the author changed the gas mixture to high flow air and oxygen, and applied chin lift until consciousness was regained three minutes later; the airway remained patent throughout,
but it was judged prudent to abandon the trial. Most of the younger volunteers appeared to enjoy entonox breathing, describing it as “like a Saturday night”, some laughing at points throughout the trial (nitrous oxide’s historical name is ‘laughing gas’, describing its social use after discovery by Lavoisier in the 1840’s, prior to its use as an anaesthetic in 1846; it is still referred to in German anaesthetic textbooks as ‘Lachgas’). Some volunteers chatted throughout, despite requests not to, as it altered the gas breathing patterns; some dozed and were rousable, but this was judged to be light sleep brought on by fatigue rather than anaesthesia, and the airway in all cases remained uncompromised.

In most cases, the author did take the volunteer home to ensure their post-trial security, unless they had made their own arrangements to be taken home. In all cases the author sent a follow-up text or phone call the next day to ensure the volunteer was fit and well. Only one volunteer, (F5) had nausea and vomiting, which lasted three days.

After the very first volunteer trial in July 2013 (M0), the author was attempting to copy the data from the S5-Collect files to put into other folders, when he lost all of that volunteer’s data. Despite a thorough search, and a visit from the GE representative to try to retrieve this data, sadly it all remained unrecoverable. This prompted a short delay in the trial programme while the author familiarised himself with more reliable methods of data transfer and copying.

The S5-Collect software is capable of collecting data on airway pressure, flow-rate, tidal volume, respiratory rate, gas concentrations of O₂, CO₂, and N₂O; it is also capable of collecting more physiological data variables than this, but only these were relevant to this study. Figure 9.1 above shows how the live data is recorded; figure 9.5 shows an example of how that data is reproduced off-line in order to make it useful to compare with Bath fp results.

Figure 9.5 a) shows inspired and expired oxygen curves (green and black curves respectively), and inspired and expired nitrous oxide curves (blue and red curves respectively) at a FGF of 10 L/min. Looking in more detail at a), it can be seen that air was being breathed prior to the start of this entonox breathing period, which started at about 17.34:30 and finished just after 17.35:50; the curves clearly show the difference between inspired and expired values; inspired values come very close to the delivered values of 50% for each gas at this high FGF breathing period; expired gas
a) Inspired and expired oxygen and nitrous oxide gas concentrations at high FGF.

b) End tidal CO₂, respiratory rate, peak airway pressure, tidal volume.

c) Inspired and expired oxygen and nitrous oxide gas concentrations at medium FGF.

d) Inspired and expired oxygen and nitrous oxide gas concentrations at low FGF.

Figure 9.5. Off line summary and presentation of data collected for one volunteer by S5- Collect in figure 9.1
concentrations do not reach an equilibrium in this short entonox breathing period, even if inspired values do; note that the oxygen concentrations increase after the entonox period as high flow oxygen enriched air is given, before falling again to appropriate levels as air is given; note that expired nitrous oxide does not fall to zero prior to the next entonox period. 9.5c) and d) show the same set of results at FGFs of 3L/min and 1.0L/min respectively. It can be seen that the inspired gas concentrations increasingly fail to reach either equilibrium or 50% as FGF falls. Comparing figure 9.5 a) and b) to figure 9.1a) and b), it can be seen how the curves exemplified by 9.5 are formed from those in 9.1. Figure 9.5b) shows a time span across entonox breathing periods for all three FGFs; there seems to be a lot of variance in respiratory rate and tidal volume; the data in this set of curves was to check on carbon dioxide handling, and to see what airway pressure was using different breathing systems at different FGFs. The four sets of curves shown in figure 9.5 a) to d) are shown as examples of four sets of data (from one individual) that were analysed; each volunteer who underwent a full set of data collection yielded twenty-seven such data sets.

Table 9.1a), b) and c), shows a collection of all such results for all volunteers, for all three breathing systems, with gas sampling from near the volunteer. Since the curves produced by Bath fp are different to those produced by S5-Collect, a means had to be found to produce variables by which the clinical trial data and the modelling data could be compared, as well as produce a data format by which the breathing systems themselves could be compared to each other. Therefore the following data were tabulated in table 9.1 by hand measurements from all curves, such as the ones in figure 9.5, from all volunteers. Maximum inspired oxygen and nitrous oxide concentrations are recorded, as are ratios between maximum inspiratory and maximum expiratory gas concentrations, noting that these may have occurred at different times in the inspiratory period. Additionally recorded in the inspired / expired ratio of nitrous oxide concentration column, as is the subtraction of the residual nitrous oxide level from the previous run from both inspired and expired measured values, typically 0% in early runs to 2 or 3% in later runs for a given volunteer. In deciding what to do with the residual nitrous oxide concentration [96], it is noted that this is due to continuing excretion of the gas from the lungs, from the blood stream, and from the body tissues, more rapidly from those which are well perfused than those which are less well perfused, and continue to decline throughout the subsequent breathing period. If the kinetics of these processes were mathematically modelled, they would depend on the addition of multiple negative exponential equations, with different amplitude and rate constants; given that the rates of change at this late point in the excretory process (from the previous breathing period) are small, a pragmatic decision was made to assume that the expired residual value remains constant. As the residual value affects subsequent values of both inspiratory and expiratory concentrations, it is subtracted from both, as indicated in the table 9.1. Not all entonox breathing periods were of the same duration, so another variable recorded was the
inspiratory maximum concentration / time to maximum concentration. Calculating this data required a lot of hand measurements (using a ruler) of concentrations and times on all such curves as represented in 9.5 a), c) and d). Another parameter in the table was minimum inspired and expired oxygen concentrations; it would be expected that this would correspond to air breathing, but it was found for some volunteers (with high oxygen consumption) that lower values occurred while breathing entonox with a FGF of 0.5 L/min. From curves such as in 9.5b), end tidal carbon dioxide at different FGFs was recorded, as was respiratory rate, airway pressure and tidal volume at all three FGFs. The data on respiratory rate and tidal volume was less useful than for CO₂ or airway pressure because of the variance. Nevertheless all such data were calculated and presented in table 9.3, and this is as discussed below. In order to make use of gas concentration data from the runs in which gas was sampled, not from near the volunteer (as is clinically common), but also from the proximal end of the inspiratory limb, and from the distal end of the expiratory limb, using measured gas concentrations (rather than analysed data as in table 9.3) these results are shown in table 9.2.

9.2.4. Methods of analysis.
While table 9.1 shows the raw measured data, the question then arises as to how best to analyse this numerical information in order to give data, which is usefully comparable with other measurement and modelling systems, such as Bath fp. Note that in doing such comparisons, units of measurement will have to be aligned, Bath fp gas concentrations being collected as mass fractions, consistent with the calculations carried out by Bath fp, and clinical data being collected as volume percentages, consistent with physiological calculations. The maximum inspiratory rise or change (delta) is the difference between maximum concentration achieved and the initial inspiratory concentration at the start of the breathing period; the initial inspiratory O₂ typically 21%, initial, and residual N₂O typically 0 – 2%; the same applies to maximum rise (delta) in expiratory concentration, where the initial subtracted value for O₂ is 16%, and for N₂O is 0 – 2%. Table 9.3a), b) and c), shows calculated results of this kind, and includes {delta concentration rise / time}, {delta maximum expiratory concentration rise / delta maximum inspiratory concentration rise} (note inversion of ratio from table 9.1), and averaged values of CO₂, respiratory rate, airway pressure, tidal volume, and minimum inspired and expired oxygen concentrations.

Having converted graphical data into numerical data, it was thought useful to show some of the calculated data contained in table 9.3 back into graphical form; this time the graphs combine clusters of data e.g. delta oxygen concentrations plotted against FGF, shown as a series of graphs for all female (or male) volunteers. Some of these are shown for the standard adult circle breathing system in figures 9.6a) and b) for delta inspired oxygen concentration, 9.7a) and b) for delta nitrous oxide concentration, and 9.8a) and b) for airway pressure, plotted against FGF, merely as
Figure 9.6. Graphical representation of change (delta) in maximum inspired oxygen concentrations with FGF on the standard adult circle system. a) female, b) male. Table below shows mean values for male and female at FGF 10 L/min (M,FGF10, F,FGF10) and at FGF 0.5 L/min (M,FGF0.5, F,FGF0.5), their standard deviations, and the p-value associated with a two tailed, unpaired t-test between males and females at each FGF.
Figure 9.7. Excel chart graphical representation of change (delta) in maximum inspired nitrous oxide concentrations with FGF on the standard adult circle system, a) female, b) male. Table below shows mean values for male and female at FGF 10 L/min (M,FGF10, F,FGF10) and at FGF 0.5 L/min (M,FGF0.5, F,FGF0.5), their standard deviations, and the p-value associated with a two tailed, unpaired t-test between males and females at each FGF.
Figure 9.8. Excel chart graphical representation of peak airway pressure with FGF on the standard adult circle system, a) female, b) male. Table below shows mean values for male and female at FGF 10 L/min (M, FGF10, F, FGF10) and at FGF 0.5 L/min (M, FGF0.5, F, FGF0.5), their standard deviations, and the p-value associated with a two tailed, unpaired t-test between males and females at each FGF.

<table>
<thead>
<tr>
<th>M or F; FGF</th>
<th>peak airway pressure cmH20</th>
<th>mean</th>
<th>SD</th>
<th>pvalue</th>
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<tr>
<td></td>
<td></td>
<td>mean</td>
<td>SD</td>
<td></td>
</tr>
<tr>
<td>M, FGF10</td>
<td></td>
<td>6.69</td>
<td>2.83</td>
<td>0.23</td>
</tr>
<tr>
<td>F, FGF10</td>
<td></td>
<td>5.3</td>
<td>1.65</td>
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<td>M, FGF0.5</td>
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<td>2.83</td>
<td>0.98</td>
<td>0.55</td>
</tr>
<tr>
<td>F, FGF0.5</td>
<td></td>
<td>2.52</td>
<td>0.67</td>
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</tbody>
</table>
examples of just some of the data, in which the spread of results at each FGF is indicated. These show that changes in gas concentrations are greater at higher FGFs, with no significant difference between males and females at equivalent FGFs in this regard.

9.2.5 Discussion. See chapter 10.

9.3 Bristol Medical Simulation Centre breathing trial

A second visit was paid to BMSC in October 2013, and all the results that were required were collected on that day. Figure 9.9a) shows the anaesthetic machine available at BMSC, an Ohmeda Excel 210 machine, rather older than the machines in clinical use at RUH Bath, with a pair of monitoring screens, the intubated mannequin Stan D. Ardman, and the gas and electrical connections to the METI control system; 9.9b) is taken from the control room and shows the METI control system and the master monitoring screen. The parameters for the mannequin were set from a control room laptop and programmed into the METI controller, and the mannequin was set to breath spontaneously.

It transpired that the anaesthetic machine available at BMSC did not allow connection of the laptop to download data using S5-Collect, so another means of recording data from it
had to be found. The tabular record of data could not be logged more frequently than every five minutes and the graphical data waveforms produced a set of points (a line connecting inspired and expired gas values) every ten seconds. It was decided that the best way to record data was to take as good a photograph as possible of the monitoring screen at appropriate intervals, using a Canon 700D single lens reflex camera; this was backed up by video footage of the screen for the entire day, which was made available by BMSC staff. The graphs in figures 9.10 to 9.13 show appropriately selected sections of this photographic data, from gas samples taken near the mannequin. In each case, for a given breathing system at a set FGF, a photograph of the graphical record is shown for oxygen and nitrous oxide concentrations and the appropriate section of the tabular data. The numerical data allow a fixed point in time to have numerical values assigned, as a calibration point for the associated graphical data. It can be seen that these data do not have adequate definition to form detailed comparisons with either the clinical data or the Bath fp data, but are adequate to make pictorial comparisons between FGFs and different breathing systems.

![Graph showing oxygen and nitrous oxide concentrations](image)

<table>
<thead>
<tr>
<th>Time</th>
<th>insp/exp %O₂</th>
<th>insp/exp %N₂O</th>
</tr>
</thead>
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<td>12.40</td>
<td>21/15</td>
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</tr>
<tr>
<td>12.45</td>
<td>50/34</td>
<td>50/35</td>
</tr>
<tr>
<td>12.50</td>
<td>22/25</td>
<td>1/18</td>
</tr>
</tbody>
</table>

*Figure 9.10 a). See full legend below*
Looking at figure 9.10a), using 2m long standard circle system at a FGF of 10 L/min, it is difficult to separate the oxygen and nitrous oxide curves, but there is a rapid rise to
maximum inspiratory values of 50% in both cases, with expired values of about 39% at
the end of the breathing period. Note the equally rapid falls in both variables at the end of
the breathing period just after 12.47, when the gas is switched to air. There is an inspired
and an expired value of oxygen concentration consistent with normal respiratory
metabolism. The inspired nitrous oxide falls quickly to zero and the expired value falls
initially to about 20%, and falls further with time.

Figure 9.10b) shows curves for the same system at a FGF of 3 L/min, it can be seen that
inspired oxygen and nitrous oxide concentrations do not reach 50% in the three minutes
or so of the breathing period, reaching perhaps 46% inspired levels, (nitrous oxide
somewhat more slowly), and 32% expired levels.

Figure 9.10c), showing results for a FGF of 0.5 L/min, show little rise in both gas
concentrations, inspired oxygen and nitrous oxide concentrations reaching 25% and 20%
respectively (note the residual nitrous oxide concentration at the beginning of this
period), and expired concentrations of perhaps 15% for both gases, in a breathing period
of nearly five minutes.

Looking at figure 9.11a), using a standard circle system of 1m length at a FGF of 10
L/min, it cannot be concluded that the gas concentrations rise any faster with shorter
tubing, which is consistent with what was found with Bath fp models.

Looking at figure 9.11b), using the same system at a FGF of 3 L/min, the inspired
concentrations of both gases reaches about 45%, nitrous oxide reaching this more slowly;
expired concentration reach about 35%, so very little difference from the 2m tubing.

Looking at figure 9.11c), for a FGF of 0.5 L/min, inspired oxygen concentrations reach
25%, nitrous oxide concentrations reach about 20%, much the same as for the 2m tubing.

Figure 9.12a), shows results using the paediatric circle system, 2m in length, at a FGF of
10 L/min, and shows an almost identical profile to the standard circle. In fact, figures
9.12b) and 9.12c), for the paediatric circle used with FGFs of 3 L/min and 0.5 L/min
respectively, also show results which correspond to the standard circle system
### Figure 9.11 a). See full legend below.

<table>
<thead>
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<td>50/38</td>
</tr>
<tr>
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<td>22/21</td>
<td>0/15</td>
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### Figure 9.11 b). See full legend below.

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<tr>
<td>14.25</td>
<td>21/16</td>
<td>0/9</td>
</tr>
<tr>
<td>14.30</td>
<td>22/17</td>
<td>9/17</td>
</tr>
</tbody>
</table>
Figure 9.13a) shows results for the 2m coaxial circle system, using a FGF of 10 L/min. While the rate at which a maximum inspired oxygen and nitrous oxide concentrations of 50% are reached is as rapid as for the other breathing systems at the same FGF, the expired values of nitrous oxide appear to rise much more slowly than with the other systems, reaching only 28%. This may be because where the gases are sampled, there is a considerable mixing of inspiratory and expiratory gases in such a coaxial system.

Looking at figure 9.13b), from the coaxial system used with a FGF of 3 L/min, in a breathing period of nearly ten minutes it can be seen that maximum inspired values are reached, but at the end of five minutes, similar inspired values are reached as for the other systems at the same FGF. Once again the expired nitrous oxide trail the oxygen expired values. Figure 9.13c) shows another long breathing period of more than seven minutes, at a FGF of 0.5 L/min, when the inspired values of oxygen and nitrous oxide reach 27% and 29% respectively at 09.55, and respective expired values reach 19% and 26%. These are similar to results when the other systems are used at this FGF.
<table>
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</thead>
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<td>0/9</td>
</tr>
<tr>
<td>11.00</td>
<td>50/40</td>
<td>50/41</td>
</tr>
<tr>
<td>11.05</td>
<td>22/23</td>
<td>0/19</td>
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Figure 9.12a). See full legend below.

<table>
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</thead>
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<td>0/10</td>
</tr>
<tr>
<td>11.20</td>
<td>36/20</td>
<td>26/19</td>
</tr>
<tr>
<td>11.25</td>
<td>22/28</td>
<td>1/22</td>
</tr>
</tbody>
</table>

Figure 9.12b). See full legend below.
Whichever breathing system and FGF were being used on the mannequin, the airway pressures recorded throughout the day were between 2 and 4 cm H2O, and the tidal volume was between 310 and 370 ml.

Overall, these results show gas concentration profiles, which are similar to those obtained from clinical breathing trials and Bath fp modelling. In this regard the results show no advantage of one breathing system over another.
### Table

<table>
<thead>
<tr>
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<th>insp/exp %N₂O</th>
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</tr>
<tr>
<td>09.20</td>
<td>22/38</td>
<td>1/16</td>
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</table>

*Figure 9.13a). See full legend below.*

### Table

<table>
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<th>insp/exp %N₂O</th>
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</thead>
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<td>39/25</td>
</tr>
<tr>
<td>09.35</td>
<td>47/46</td>
<td>47/42</td>
</tr>
</tbody>
</table>

*Figure 9.13b). See full legend below.*
This chapter has described the processes involved in studies to clinically validate Bath fp studies. The process of setting up a clinical study, obtaining ethical approval, recruiting volunteers, the software used to record the data, and the methods used in carrying out the trial are described. A few examples of graphical data recorded are shown, and the rationale for recording certain variables is discussed, as well as the rationale for certain initial calculations. Tabulated results are presented, as well as a table of initial calculated data. Many statistical calculations of the data were made for discussion in chapter 10. Another part of the clinical validation is discussed, include the trials carried out at the Bristol Medical Simulation Centre.

It is worth discussing the difficulties and constraints involved in carrying out this type of research, using clinical trials to validate models.

- Bath fp simulation produces mathematical models of different components of the breathing system, and importantly a model of the relevant anatomical and physiological components of a breathing human being undergoing light anaesthesia. Furthermore Bath fp simulates the behavior of all these components linked together. In doing so, the reproducibility of solutions to mathematical equations is the strength of the model.

<table>
<thead>
<tr>
<th>Time</th>
<th>insp/exp % O₂</th>
<th>insp/exp %N₂O</th>
</tr>
</thead>
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<td>09.45</td>
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<td>09.50</td>
<td>24/16</td>
<td>19/17</td>
</tr>
<tr>
<td>09.55</td>
<td>27/19</td>
<td>29/26</td>
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</tbody>
</table>

*Figure 9.13c). FGF 0.5L/min. Coaxial circle system, 2m length. White curves=oxygen; blue curves=nitrous oxide. Dotted horizontal line = 50%.*
Nevertheless, to produce such mathematical models a number of assumptions have been made about the mechanical behaviour of all such components, in particular, human anatomy and physiology, but this is the task biomedical engineers face. In reality, such human systems have a great deal of variance in structure and function at different moments in time. There is no real way of allowing for such variance without extending the research significantly.

- To do clinical trials such as these, in which the participants are professional experts in this field, are knowledgeable colleagues, known to the researcher, a process of ethical approval must be sought, and the application judged by a committee less knowledgeable of the research than the participants themselves. This caused significant delay in starting the trials.
- Even after ethical approval has been obtained, recruiting volunteers is a significant task in itself, and almost proved impossible in this case.
- Components in the clinical breathing systems themselves may not, on different occasions of testing, have identical mechanical characteristics to the ones modelled by Bath fp. Constancy and adherence to the model characteristics must be assumed.
- It needs to be recognised that the breathing pattern of awake or semi-awake participants is irregular, as was shown by the range of tidal volume measurements, and this will widen the variance in the results, reducing their value. It was also noted that some participants talked euphorically during the trials, which added another dimension to the variance of the results.
- It has to be asked what the validity may be of pooled clinical results between, say 50 kg female participants and 100 kg male participants, or between participants under thirty years of age and those over forty. Some attempt was made to use different sets of results, not so much for validating Bath fp as for comparing clinical results between different groups of participants, and participants breathing through different systems.
- A very great deal of data were generated by the clinical trials, and it was necessary to make choices about which data were to be used and which were to be omitted. It is hoped that wise choices were made in this regard, in order to achieve what was required, namely to validate the simulation package, Bath fp.

With regard to the decision to undertake further trials on breathing systems using a computerized mannequin at the Simulation Centre in Bristol, a few comments on the validity of that decision are apposite. It was taken when it was thought that attempts at recruitment of human volunteers were foundering. It was thought to be a back-up plan for providing data from another source which could validate the model from Bath fp. It should be stated that the main purpose of BMSC is to provide training for clinicians in handling emergency situations in anaesthesia and surgery, which can be artificially produced in laboratory-type conditions without endangering the lives of real patients. While the gas measurement devices associated with the breathing systems used produce traces which were used, the mannequin itself is programmed to produce a certain tidal volume, oxygen consumption, carbon dioxide production, and so on. It is in fact, a model...
itself, so does not provide the variance, with the inherent validity that suggests, that human volunteers do. Nor are the measurement systems designed to be particularly accurate. In the event, the traces produced from BMSC results did not provide enough accuracy to be useful as a validation tool. Nevertheless, they are left in this chapter because they support the general conclusions from both Bath fp modelling (that FGF dominates system geometry in determining function) and from the volunteer clinical trials.
Chapter 10.
Analysis and Discussion of Results from Breathing Trial.

10.1. Introduction
It should be recalled that the original purpose of this thesis was to see if the mechanical design of low flow breathing systems could be improved; a mathematical simulation model was developed to try to achieve this, and this model needed clinical validation, having not been previously validated. Having collected a very large amount of clinical data presented in chapter 9, it is necessary to put that data to good use. This can be achieved by using it, not only in a standardized, collectivised form to compare relevant results to Bath fp simulation results, but also to compare the data between population samples, e.g. males and females (for each breathing system), and between breathing systems (for each population sample), each under different conditions of FGF; as a piece of clinical research, these latter two sets of comparisons are useful to contribute to answering the original hypothesis. In comparing the clinical data to Bath fp data, it is necessary to normalize all the clinical data into one group (males and females, all ages), and to convert Bath fp mass fractions into volume percentages, as clinical data is presented. In comparing different sets of clinical data against each other, it was found that most comparisons resulted in no significant differences in, for example inspired oxygen, or increase (delta) in inspired nitrous oxide, so only results with some statistical significance (p=0.1) are presented.

It is thought necessary to have a separate chapter for the analysis and discussion of the results from the breathing trials. Not only will comparisons be made from clinical data between breathing systems and between population samples, but also between the clinical data and the results from Bath fp. Table 10.2 a), b) and c) shows examples of collated trial data in relation to the standard adult circle system, mean maximum inspired oxygen and nitrous oxide concentrations for male and female volunteers with one standard deviation, with the mass fractions calculated from the mean values as indicated in chapter 10; this is compared in the adjacent column to the Bath fp mass fraction, and where given by the Bath fp model, the volumetric percentage in the appropriate tube. In each case the nitrogen mass fractions, as calculated in the way described later in this chapter, are also shown.

As outlined in chapter 9, a number of breathing runs were done, using eight or nine female volunteers, and nine male volunteers, at FGFs of 10, 3, 1 and 0.5 L/min, using three different breathing systems; females and males were considered as samples of two populations for comparison in most cases. Most volunteers breathed at three out of four
FGFs, 10 and 3 L/min, and either 1.0 or 0.5 L/min; some breathed at only 10 and either 1.0 or 0.5 L/min. Means and standard deviations of the different series of measured variables were calculated using the appropriate statistical formulas available with Excel spread sheets. Statistical two tailed, unpaired t-tests were done also using the formula facility on Excel spread sheets, between series of variables being compared. The purpose of doing this is to determine whether different breathing systems behave differently from each other in relation to the measured variables, or whether different samples of volunteers (males and females) place different demands on the breathing systems in relation to different variables. This involved a large number of calculations, and in most cases, there were no differences between variables and samples so that, in general, only those that had significant differences are presented here, with a few exceptions where that information is thought to be useful to the study.

10.2. Analysis of results from breathing trial
In statistically comparing the standard circle, the coaxial circle and the paediatric circle with each other in turn, in relation to maximum inspired oxygen concentrations, and to maximum inspired nitrous oxide concentrations, when sampled from near the volunteer’s mouth, no significant differences were found for either males or females at any FGF. It was originally thought from the raw data collected that some breathing systems more than others, at low FGF, might impose a real hazard of hypoxia or lack of anaesthetic uptake. In general, the statistical data show no differences in key variables such as inspired and expired oxygen and nitrous oxide concentrations or minimum expired oxygen concentrations, between high and low FGFs, except where indicated in the variables discussed below. For delta (increase) in oxygen and nitrous oxide concentrations, as examples, figures 9.6 and 9.7 and their associated tables show that for the standard adult breathing system, there is no significant difference between males and females at each of the two FGFs of 10 L/min and 0.5 L/min. All these results refer to the data sampled near the volunteer (‘patient sampling’), since such differences in data sampled from elsewhere in the breathing systems are of less practical or clinical interest.

It is proposed not to present the bulk of the statistical data for which no significant differences exist. On calculating and reviewing all the statistical data, the following groups of data were found to have statistically significant differences at the p≤0.1 level:

1) Standard circle, FGF 3 L/min:
   Maximum inspired oxygen concentration %:
   females vs males (43.57% ± SD 3.31% vs 39.67% ± SD 3.56%, p≤0.07);
   Maximum expired oxygen concentration:
   females vs males (37.29% ± SD 2.63% vs 34.25% ± SD 2.38%, p≤0.05);
2) Coaxial circle:
   Maximum expired oxygen concentration %:
   FGF 10 L/min, females vs males (46.19% ± SD 1.69% vs 43.50% ± SD 2.36%, p≤0.02);
   FGF 3 L/min, females vs males (37.50% ± SD 3.55% vs 34.42% ± SD 1.69%, p≤0.07);
   Maximum expired nitrous oxide concentration %:
   FGF 10 L/min, females vs males, (49.38% ± SD 2.85% vs 36.94% ± SD 2.62%, p≤0.08).

3) Paediatric circle:
   Maximum expired oxygen concentration %:
   FGF 3 L/min, females vs males (37.14% ± SD 1.80% vs 34.75% ± SD 2.60%, p≤0.09).

4) Rate of increase to maximum inspired oxygen concentration per unit time %/sec:
   FGF 10 L/min:
   Standard circle, females vs males (0.48%/sec ± SD 0.22%/sec vs 0.26%/sec ±SD 0.07%/sec, p≤0.02);
   Female, standard circle vs paediatric circle (0.48%/sec ± SD 0.22%/sec vs 0.30%/sec ± SD 0.16%/sec, p≤0.08).

5) Increase to maximum inspired oxygen concentration %: FGF 10 L/min, female:
   Standard circle vs coaxial circle (48.1% ± SD 1.25% vs 45.8% ± SD 2.42% p≤0.04);
   Standard circle vs paediatric circle (48.1% ± SD 1.25% vs 46.3% ± SD 1.77%, p ≤ 0.03).

6) Rate of increase to maximum inspired nitrous oxide concentration per unit time %/sec:
   FGF 10 L/min:
   Standard circle, female vs male (0.71%/sec ± SD 0.38%/sec vs 0.43%/sec ± SD 0.12%/sec, p≤0.09);
   Male, standard circle vs paediatric circle, (0.43%/sec ± SD 0.12%/sec vs 0.69%/sec ± SD 0.39%/sec, p≤0.08);
   Female, standard circle vs coaxial circle (0.71%/sec ± SD 0.38%/sec vs 0.42%/sec ± SD 0.16%/sec, p≤0.08).

7) Max airway pressure cm H2O: FGF 10 L/min:
   Female, standard vs coaxial circle (5.30 cmH2O ± SD 1.65 cmH2O vs 6.69 cm H2O ± SD 1.62 cm H2O, p≤0.10);
   Female, standard circle vs paediatric circle, (5.30 cmH2O ± SD 1.65 cm H2O vs 6.50 cm H2O ± SD 0.00 cm H2O, p≤0.06);
   Male, standard circle:
   FGF 10 L/min vs FGF 0.5 L/min (6.69 cm H2O ± SD 2.83 cmH2O vs 2.83 cm H2O ± SD 0.98 cm H2O, p≤0.004).
8) Minimum expired oxygen concentrations %: FGF 10 L/min:
   Standard, female vs male (15.8% ± SD 1.6% vs 14.3% ± SD 0.76%, \( p \leq 0.03 \));
   Male, standard circle vs coaxial circle, (14.3% ± SD 0.76% vs 15.6% ± SD 1.67%, \( p \leq 0.06 \));
   Male, standard circle vs paediatric circle, (14.3% ± SD 0.76% vs 15.3% ± SD 1.23%, \( p \leq 0.05 \)).

It is recognised that the samples used to calculate these statistical data are small, and the probability values quoted are not highly significant, even if they are significant at the 10% probability level. Nevertheless, in analysing this data, the following comments can be made about the performance of the breathing systems under different circumstances:

1) For the standard circle breathing system at a FGF of 3 L/min, both inspired and expired oxygen concentrations are higher in female volunteers than in male volunteers. This probably reflects a higher uptake of oxygen in the males, but this difference is not reflected at higher or lower FGFs, which is surprising. While the levels do not represent a clinical hazard, this FGF is commonly used, and therefore some note should be taken of the result.

2) For the coaxial circle system, expired oxygen concentrations are higher in females than in males for FGFs of 10 L/min and of 3 L/min, but not at the 1 L/min or 0.5 L/min. Again this reflects the higher oxygen uptake by males. In addition the expired nitrous oxide concentrations are higher in females than males at a FGF of 10 L/min, once again, probably due to the faster uptake in males due to larger lung volume, higher ventilation and higher cardiac output.

3) For the paediatric circle system at a FGF of 3 L/min, the expired oxygen concentration is higher in females than in males for reasons outlined above. Despite results 2) and 3), note the earlier comment that, when comparing all three breathing systems for males and females individually and at comparable FGFs, no difference was found between them with regard to inspired and expired oxygen and nitrous oxide concentrations.

4) In considering the rate at which oxygen concentration increases to maximum at a FGF of 10 L/min:
   - In the standard circle breathing system, this rate of rise was faster in females than in males, which is consistent with faster oxygen uptake in males.
- For females, the rate of rise is faster in the standard circle breathing system than in the paediatric breathing system. This is surprising, given the smaller volume of the latter.

5) In considering the increase (delta) to maximum inspired oxygen concentration, at a FGF of 10 L/min for females, this is greater for the standard circle breathing system when compared with both the coaxial circle system and the paediatric circle system (consistent with 4) above). This suggests that at this FGF, there is no advantage in using either the coaxial circle or the paediatric circle over the standard circle in terms of increasing oxygen levels. In other words the behaviour of the systems is governed by the FGF rather than their geometry.

6) In considering the rate of increase of inspired nitrous oxide concentrations at a FGF of 10 L/min:
   - For the standard circle breathing system, this rate of rise was faster in females than in males; this is consistent with other results in this category;
   - In males this rate of rise is slower in the standard circle than in the paediatric circle, which is to be expected, given the smaller volume of the latter;
   - In females this rate of rise is faster in the standard circle than in the coaxial circle, which may be a reflection of the higher respiratory work with the coaxial system experienced by some smaller (female) volunteers.

7) In considering the maximum airway pressure in the breathing systems:
   - At a FGF of 10 L/min, for females the airway pressure was higher in both the coaxial circle system and the paediatric circle system, when each was compared to the standard circle system; this is probably because of the simplicity of construction of the breathing tubes in the standard system;
   - For males using the standard system, the airway pressure is significantly higher at a FGF of 10 L/min compared to a FGF of 0.5 L/min; this result is intuitive, but was nevertheless calculated; the same result is true for females and for other breathing systems, as expected.

8) In considering the minimum expired oxygen concentration, rather surprisingly there was only significance at a high FGF of 10 L/min:
   - Using the standard circle system, this was higher in females than in males, once again reflecting the higher oxygen uptake in males;
   - For males it was lower using the standard circle system, than when using either the coaxial or the paediatric circle systems; the trials were always carried out starting with the standard system, so this result might be associated with a lower rate of oxygen consumption later on in each trial, due to familiarity and relaxation with the trial process;
   - It is surprising that no significant difference was found in minimum expired oxygen concentrations, between low FGFs and high FGFs; at one level this is reassuring
for the use of low FGFs, but might also be associated with the small sample sizes. Further testing of the Bath fp models of different types of circle system by looking in more detail at the directions of gas flow on expiration may help to elucidate the reasons behind this.

10.3. Comparison of trial results with Bath fp.

In making comparisons of the results from the breathing trials with Bath fp modelling results, there are two additional steps to carry out. Firstly, all the statistical data from the breathing trials discussed so far treated male and female volunteers as samples of two populations, while the Bath fp model does not distinguish between males and females; the patient model, ABW99, has a fixed tidal volume of 500ml, a fixed respiratory rate of ten breaths per minute, and a fixed respiratory quotient of 0.84. For this discussion it was necessary to treat males and females as one population and calculate means and standard deviations accordingly. Secondly, almost all Bath fp gas concentration data are presented as mass fractions of gases, while all such data from volunteer measurements are presented as volume percentages. Additionally, the gas monitoring available in the volunteer trials, standard clinical monitoring, included only oxygen, nitrous oxide and carbon dioxide; it did not include nitrogen or water vapour, although the latter is accounted in infra-red spectroscopic gas monitors; nitrogen requires a different technology to measure it, such as mass or Raman spectroscopy, not generally available; oxygen is monitored with a fuel cell.

In order to convert volume percentages to mass fractions, a means has to be found to estimate or calculate the other gases in gas mixtures. Therefore some assumptions were made about the carbon dioxide concentration and the water vapour concentration in order to facilitate calculation of nitrogen concentration at various sampling points. The breathing trial confirmed that expired carbon dioxide remained relatively constant at between 4.5 and 5.5%, whatever the FGF and whichever breathing system was used, since the normal physiological response in most awake humans alters ventilation to ensure that is the case; a caveat here is that this ventilatory response in anaesthetised humans is adversely affected, and the carbon dioxide may rise in an anaesthetised, spontaneously breathing patient. In calculating the mass fractions of gases, these assumptions made about carbon dioxide and water vapour concentrations depended on whether the inspiratory or expiratory gas concentrations were being considered, and at which point the gas was being sampled: near the volunteer’s mouth; at the opposite end of the inspiratory tube; or at the opposite end of the expiratory tube. These assumptions were:

1) For sampling near the volunteer’s/patient’s mouth:
Inspiratory gas is dry gas consisting of oxygen, nitrous oxide and residual nitrogen from an initially air filled breathing system; measurement of the first two allows the nitrogen concentration to be calculated.

Expiratory gas consists of oxygen, nitrous oxide, carbon dioxide, water vapour and residual nitrogen; the first two are measured, the carbon dioxide is assumed to be 5% by volume, and the water vapour is assumed to be at 30°C, with a saturated vapour pressure of 4.0 kPa (4% by volume), or a less than fully saturated sample at a somewhat higher temperature [98]; this allows the nitrogen concentration to be calculated.

2) For sampling at the opposite end of the inspiratory tube:
- As the expiratory unidirectional valve is closed during inspiration, and almost entirely prevents reverse flow of expiratory gas, the inspiratory gas is dry, consisting of oxygen, nitrous oxide and nitrogen, as above.
- As the inspiratory unidirectional valve is closed during expiration, preventing expiratory gas from entering the inspiratory tube, it can be asserted that expiratory gas consists of oxygen, nitrous oxide and nitrogen; this is backed up by the fact that almost all measurements in the breathing trial indicated that inspiratory and expiratory gas samples measured at this point were identical.

3) For sampling at the opposite end of the expiratory tube:
- The expiratory unidirectional valve is normally closed during inspiration, but when the FGF is high, a significant amount of FGF will bypass the volunteer, reach the expiratory tube and breach the expiratory valve during inspiration; therefore the inspiratory gas here at one extreme can be considered to consist of oxygen, nitrous oxide and nitrogen; on the other hand when the FGF is low, it all goes to the patient in inspiration, none enters the expiratory tube, and the inspiratory gas here can be considered to consist of oxygen, nitrous oxide, 5% carbon dioxide and water vapour at a temperature of 20°C (lower temperature because of the distance of this point away from the human) with a vapour pressure of 2kPa (2% by volume) [98]; in calculating gas mass fractions using both assumptions, oxygen and nitrous oxide fractions vary by small amounts, and the nitrogen mass fractions vary by an amount more or less equivalent to the presence or absence of water vapour.
- As the inspiratory unidirectional valve is closed during expiration, and almost entirely prevents inspiratory gas being included in the expiratory tube, expiratory gas consists of oxygen, nitrous oxide, 5% carbon dioxide and water vapour at the lower temperature of 20°C, with a vapour pressure of 2kPa (2% by volume) [98].

A means of calculating mass fractions was found in Rogers and Mayhew ([89], pp 276-277). Table 10.1 shows how this calculation was done, using as an example, the standard circle system at a FGF of 1.0 L/min, with mean gas concentrations for both male and
female volunteers, sampled near the mouth. Bracketed variables indicate assumed or calculated values.

Such calculations allowed the calculation of the means and standard deviations of all measured volumetric fractions (average for male and female taken together), for inspiratory and expiratory gases, measured at the three sampling points described throughout this report, for all three breathing systems. As an example, table 10.2 shows this data for inspired and expired gases at the three gas sampling points for the standard adult breathing system. The second column shows the mass fraction, calculated as above. In table 10.2c) for the inspiratory gas concentrations measured at the distal expiratory tube, assuming it consists of FGF, the calculated values given are given without brackets, while those in square brackets are those where the gas is assumed to consist of expiratory gas. The Bath fp mass fraction values are in the adjacent columns of all three tables, measured from the derived graphs in chapters 6.

Where given by the Bath fp model, the Bath fp volumetric fraction is given, though this is only present for comparison purposes; it was not used for any calculations, as the original Bath fp model did not take account of water vapour. The Bath fp mass fractions are taken from all the data collected throughout, the modelling period, using data collected from sites matching the three sampling sites used in the breathing trials. For example:

1) Bath fp data collection sites equivalent to ‘patient sampling’ in the breathing trials:
    - in the standard and paediatric breathing system models, were at port 1 of inspiratory tube AP07(1)
    - in the coaxial system model, were inner port 2 of connector ACA1, or at port 1 of ACA1.

2) Bath fp data collection sites equivalent to ‘inspiratory tube sampling’ in the breathing trials:
    - in the standard and paediatric breathing system models, were at port 2 of tube AP07(1).
    - In the coaxial system model, were at port inner port 2 of the coaxial tube ACP2(1).

3) Bath fp data collection sites equivalent to ‘expiratory tube sampling’ in the breathing trials:
    - in the standard and paediatric breathing system models, were at port 2 of expiratory tube AP07(2).
    - in the coaxial system model, were at outer port 2 of the coaxial tube ACP2(1).
It will be noted from table 10.2 that there are some missing values in the Bath fp column, since not all data for all different FGFs and sampling positions were collected. In the columns of gases sampled at the distal end of the expiratory tube (‘expiratory sampling’) in the breathing trial, two sets of volumetric concentrations and calculated mass fractions are given for inspired gases, one with no brackets, the other with square brackets; in the way described above, the former assumes that the inspired gases at this point of sampling consisted of FGF, the latter that it consisted of expired gases at 20°C.

<table>
<thead>
<tr>
<th>Gas</th>
<th>Mean Volumetric Fraction</th>
<th>Mol Wt</th>
<th>Mass = VolFractn \times\text{MolWt}</th>
<th>Mass fractions = Mass/total mass</th>
</tr>
</thead>
<tbody>
<tr>
<td>O2</td>
<td>0.28</td>
<td>32</td>
<td>9.0</td>
<td>0.28</td>
</tr>
<tr>
<td>N2O</td>
<td>0.22</td>
<td>44</td>
<td>9.7</td>
<td>0.30</td>
</tr>
<tr>
<td>(N2)</td>
<td>(0.5)</td>
<td>28</td>
<td>14.0</td>
<td>0.43</td>
</tr>
<tr>
<td>Totals</td>
<td>1.00</td>
<td>32.7</td>
<td>1.01</td>
<td></td>
</tr>
</tbody>
</table>

*Table 1*
Looking at table 10.2 in detail (standard circle system), it can be seen that, for gas sampling near the patient’s/volunteer’s mouth, for the FGFs at which both Bath fp and trial data were collected, the agreement between collected trial data mean values and Bath fp modelling measurements is rather good, with the recognition that the variance for the mean values exists. This applies to the other two breathing systems as well, although this information is not presented in this thesis. In passing it is worth noting that the variance for trial values for a FGF of 3 L/min is bigger than for the other FGFs. Where there is significant disagreement between Bath fp values and measured values (mostly with ‘inspiratory sampling’ and ‘expiratory sampling’, away from the patient/volunteer’s mouth), it occurred where Bath fp took excessive account or inadequate account of inspiratory or expiratory maxima and minima; this may be an inherent fault in the way values at extremes of the cycle are calculated, or it may be that, where Bath fp assumes complete gas mixing, this does not occur in reality.

For inspired gas mass fractions:
- For patient sampling, the agreement is good for both oxygen and nitrous oxide, across all FGFs and all breathing systems.
- For inspiratory gas sampling, the agreement is better for high FGFs than low FGFs; this may be because of the greater effect that consumption of both gases by the volunteer, with greater variance, has on a low FGF than a high FGF.
For expiratory gas sampling, there is a paucity of Bath fp data, but what there is shows good agreement.

For expired gas mass fractions:
- For patient sampling, the agreement for oxygen expired fractions is better than for nitrous oxide fractions; this difference might be due to fault with the nitrous oxide absorption model used for Bath fp.
- For inspiratory sampling, there is less good agreement for all gas fractions than elsewhere.
- For expiratory gas sampling, there is a paucity of Bath fp data, but what there is shows fair agreement.

Despite the relatively good agreement between mean values from the breathing trial and a number of Bath fp results, it is necessary to show the extent of the individual variance of results between individuals. Figures 10.1a) to d), to 10.3a) to d) show plots of differences between gas concentrations (Bath fp values – measured individual values) plotted against the mean of individual values, (mean of Bathfp result and measured value for each FGF) as proposed by Bland and Altman for assessing the
Table 10.2a). Comparison of inspired gas mass fractions and volumetric percentages in the standard circle system, between breathing trial results and Bath fp results, sampling near the patient/volunteer.
<table>
<thead>
<tr>
<th></th>
<th>O2insp</th>
<th></th>
<th></th>
<th>N2Oinsp</th>
<th></th>
<th></th>
<th>N2insp</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Trial</td>
<td>Trial</td>
<td>Bathfp</td>
<td></td>
<td>Trial</td>
<td>Trial</td>
<td></td>
<td>Calc</td>
</tr>
<tr>
<td></td>
<td>Vol%</td>
<td>massFrctn</td>
<td>massFrctn</td>
<td>Vol%</td>
<td>massFrctn</td>
<td>massFrctn</td>
<td>Vol%</td>
<td>massFrctn</td>
</tr>
<tr>
<td>FGF10</td>
<td>49.88±0.50</td>
<td>0.42</td>
<td>0.42</td>
<td>49.71±0.69</td>
<td>0.58</td>
<td>0.58</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>FGF3</td>
<td>42.77±5.07</td>
<td>0.38</td>
<td>0.41</td>
<td>39.68±6.90</td>
<td>0.49</td>
<td>0.57</td>
<td>0.17</td>
<td>0.13</td>
</tr>
<tr>
<td>FGF1</td>
<td>28.5±2.06</td>
<td>0.28</td>
<td>(0.35)</td>
<td>21.4±2.07</td>
<td>0.28</td>
<td>(0.30)</td>
<td>0.51</td>
<td>0.44</td>
</tr>
<tr>
<td>FGF0.5</td>
<td>19.91±1.45</td>
<td>0.21</td>
<td>0.32</td>
<td>12.91±3.63</td>
<td>0.18</td>
<td>0.30</td>
<td>0.67</td>
<td>0.61</td>
</tr>
</tbody>
</table>

*Table 10.2b). Comparison of inspired gas mass fractions and volumetric percentages in the standard circle system, between breathing trial results and Bath fp results, sampling in inspiratory tube away from patient/volunteer.*
<table>
<thead>
<tr>
<th>O2insp</th>
<th>Trial Vol%</th>
<th>Trial massFrctn</th>
<th>Bathfp massFrctn</th>
<th>Bathfp Vol%</th>
</tr>
</thead>
<tbody>
<tr>
<td>FGF10</td>
<td>49.09±1.09</td>
<td>0.43[0.42]</td>
<td>0.4</td>
<td>47</td>
</tr>
<tr>
<td>FGF3</td>
<td>39.96±7.66</td>
<td>0.37[0.36]</td>
<td>0.35</td>
<td>39</td>
</tr>
<tr>
<td>FGF1</td>
<td>23.00±3.92</td>
<td>0.23[0.22]</td>
<td>0.2</td>
<td>18</td>
</tr>
<tr>
<td>FGF0.5</td>
<td>14.75±4.70</td>
<td>0.16[0.15]</td>
<td>0.02</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>N2Oinsp</th>
<th>Trial Vol%</th>
<th>Trial massFrctn</th>
<th>Bathfp massFrctn</th>
<th>Bathfp Vol%</th>
</tr>
</thead>
<tbody>
<tr>
<td>FGF10</td>
<td>45.00±2.00</td>
<td>0.53[0.51]</td>
<td>0.52</td>
<td>41</td>
</tr>
<tr>
<td>FGF3</td>
<td>32.88±5.44</td>
<td>0.42[0.41]</td>
<td>0.35</td>
<td>29</td>
</tr>
<tr>
<td>FGF1</td>
<td>21.10±3.97</td>
<td>0.28[0.28]</td>
<td>0.2</td>
<td>2</td>
</tr>
<tr>
<td>FGF0.5</td>
<td>13.50±2.12</td>
<td>0.19[0.18]</td>
<td>0.02</td>
<td></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>N2insp</th>
<th>Calc Vol%</th>
<th>Calc massFrctn</th>
<th>Bathfp massFrctn</th>
<th>Bathfp Vol%</th>
</tr>
</thead>
<tbody>
<tr>
<td>FGF10</td>
<td>0.06[0]</td>
<td>0.04[0]</td>
<td>0.05</td>
<td>6</td>
</tr>
<tr>
<td>FGF3</td>
<td>0.27[0.20]</td>
<td>0.22[0.16]</td>
<td>0.22</td>
<td>29</td>
</tr>
<tr>
<td>FGF1</td>
<td>0.56[0.49]</td>
<td>0.49[0.42]</td>
<td>0.72</td>
<td>75</td>
</tr>
<tr>
<td>FGF0.5</td>
<td>0.72[0.65]</td>
<td>0.66[0.58]</td>
<td>0.72</td>
<td></td>
</tr>
</tbody>
</table>

*Table 10.2c). Comparison of inspired gas mass fractions and volumetric percentages in the standard circle system, between breathing trial results and Bath fp results, sampling in expiratory tube away from patient/volunteer.*
Mean difference: -1.04
Standard Deviation of differences 6.17
Limits of Agreement -13.37 to 11.28
Standard Error of Mean: 0.97
N=40
95% confidence interval -3.00 to 0.90

Mean difference -0.43
Standard Deviation of differences 6.32
Limits of agreement -13.07 to 12.21
Standard error of mean 0.96
n=43
95% confidence interval -2.36 to 1.51

Figure 10.1. Bias/precision curves for standard circle system, gas sampling near patient.
a) maximum inspired oxygen concentrations, b) maximum expired oxygen concentrations.
Mean difference -2.09
Standard Deviation of differences 4.11
Limits of agreement -10.31 to 6.14
Standard error of mean 0.64
n=41
95% confidence interval -2.70 to -0.79

Mean difference -0.01
SD 7.71
Limits of agreement -15.43 to 14.27
SEM 1.2
n= 41
95% CI -2.44 to 2.41

Figure 10.1 (cont’d). Bias/precision curves s for standard circle system, gas sampling near patient. c) maximum inspired nitrous oxide concentrations, d) maximum expired nitrous oxide concentrations.
mean difference 1.89
SD 2.64
Limits of agreement -3.39 to 7.17
SEM 0.84
n= 28
95% CI 0.84 to 2.94

Mean difference -0.54
SD 1.35
Limits of agreement -3.24 to 2.16
SEM 0.26
n= 28
95% CI -1.07 to -0.01

*Figure 10.2. Bias/precision curves for coaxial circle system, gas sampling near patient.*
*a) maximum inspired oxygen concentrations, b) maximum expired oxygen concentrations.*
Mean difference -0.09
SD 4.85
Limits of agreement -9.79 to 9.61
SEM 0.92
n= 28
95% CI -1.98 to 1.8

Mean difference -1.23
SD 8.12
Limits of agreement -17.47 to 15.01
SEM 1.56
n= 28
95% CI -4.43 to 1.97

*Figure 10.2 (cont’d). Bias/precision curves for coaxial circle system, gas sampling near patient. c) maximum inspired nitrous oxide concentrations, d) maximum expired nitrous oxide concentrations*
Mean difference 3.05
SD 3.72
Limits of agreement -8.31 to 15.93
SEM 1.12
n= 29
95% CI 1.51 to 6.11

Mean difference 0.72
SD 1.88
Limits of agreement -3.04 to 4.48
SEM 0.35
n= 29
95% CI 0 to 1.44

Figure 10.3. Bias/precision curves for **paediatric circle system**, gas sampling near patient. a) maximum inspired oxygen concentrations, b) maximum expired oxygen concentrations.
Mean difference 1.73
SD 1.89
Limits of agreement -2.05 to 5.51
SEM 0.35
n= 29
95% CI 1.01 to 2.45

Figure 10.3 (cont’d). Bias/precision curves for paediatric circle system, gas sampling near patient. c) maximum inspired nitrous oxide concentrations, d) maximum expired nitrous oxide concentrations.
agreement between two methods of clinical measurement [99]. Correlation and regression is not the preferred way of statistically analysing such data, since it fails to adequately highlight differences between techniques and correlation between methods is not the same as agreement. Wilson et al [100] used a mathematical model, based on Bath fp, to compare respiratory parameters as deduced by the model with those clinically measured, in ventilated patients on an intensive care unit, and they took the same approach in representing these data. Looking at the bias/precision graphs above, in most cases the standard error of the mean is 1.0 or less (except for inspired oxygen concentration differences for the paediatric circle), with values more than 1.0 applying to expired nitrous oxide concentrations; this might be due to either differences in nitrous oxide absorption by volunteers of different body habitus, or due to inaccuracy of the nitrous oxide absorption model used. In general, however, these results suggest reasonably good agreement between Bathfp and measured values, as suggested by the 95% confidence interval for the variables, which justifies its use for modelling such breathing systems. However, the wide limits of agreement in most cases, which may be a reflection of small sample size, reduces confidence somewhat. Furthermore the analysis seems to indicate a similar degree of variance for all three breathing systems. Wilson et al [100], in deciding on the usefulness of such a model, went further, and defined sensitivity gain as the change in a value of a variable predicted by such a model divided by the actual change in that variable value, as indicated by equation 10.1.

\[
\text{sensitivity gain} = \left| \frac{\% \text{ change in predicted value}}{\% \text{ change in parameter value}} \right|
\]

Eqn.10.1.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>(O_2)(_{\text{nap}})</th>
<th>(O_2)(_{\text{exp}})</th>
<th>(N_2O)(_{\text{nap}})</th>
<th>(N_2O)(_{\text{exp}})</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tidal volume</td>
<td>0</td>
<td>0.087</td>
<td>-0.066</td>
<td>0.176</td>
</tr>
<tr>
<td>Breathing rate</td>
<td>0</td>
<td>0.087</td>
<td>-0.066</td>
<td>0.235</td>
</tr>
<tr>
<td>Initial bag volume</td>
<td>0</td>
<td>0</td>
<td>-0.022</td>
<td>-0.0735</td>
</tr>
<tr>
<td>(O_2) consumption</td>
<td>-0.052</td>
<td>-0.231</td>
<td>0.022</td>
<td>0</td>
</tr>
<tr>
<td>(N_2O) consumption</td>
<td>-0.026</td>
<td>0.174</td>
<td>-0.088</td>
<td>-0.353</td>
</tr>
<tr>
<td>Resp. Ex. Ratio</td>
<td>0</td>
<td>-0.058</td>
<td>0</td>
<td>-0.088</td>
</tr>
<tr>
<td>Fresh Gas Supply</td>
<td>0.013</td>
<td>0.202</td>
<td>0.352</td>
<td>0.588</td>
</tr>
</tbody>
</table>

Table 10.3. Sensitivity values obtained from equation 10.1 [99], based on the standard circle breathing system, 1 m length, 22mm diameter, FGF 3 L/min. negative sign indicates reduction in predicted values. Values in red are those greater than 0.1
Using equation 10.1, table 10.3 shows the results obtained when the tidal volume, breathing rate, initial respiratory bag volume, oxygen consumption, nitrous oxide consumption, expiratory exchange ratio and fresh gas supply were increased by 10% in turn. The sensitivity gain values of 0.1 and above are highlighted in red, as it may be regarded as a problem with the Bath fp model if unexpected secondary changes occur due to a primary change in certain variables. The results show that the predicted inspired levels of oxygen and nitrous oxide are relatively insensitive to the parameter changes, except for the following changes, which are to be expected:

- 10% increase in FGF; this shows significant increases in inspired and expired nitrous oxide and expired oxygen; this is to be expected, as the consumption of these gases is unchanged; it is only surprising that the inspired oxygen concentration does not follow suit.

- 10% increase in oxygen consumption or nitrous oxide consumption; these show that increases in the expired results for oxygen and nitrous oxide, respectively, are significant, with the nitrous oxide results showing the largest changes. However, it is entirely expected in these circumstances that the expired oxygen and expired nitrous oxide concentrations respectively, fall.

- 10% increase in either tidal volume or breathing rate (minute ventilation = tidal volume x breathing rate), there is an expected rise in expired nitrous oxide concentration since more nitrous oxide is delivered to the patient, but the consumption rate is unchanged; what looks interesting is that the same changes do not happen to inspired nitrous oxide concentrations or to oxygen concentrations; this may indeed represent an insensitivity of Bath fp.

For those results from table 10.3 which are to be expected, some of these are of the same magnitude as the variance shown in figures 10.1 to 10.3 above. Given the likely human variation in human volunteers in these variables, this provides further support for the use of Bath fp to model breathing systems.

10.4. Conclusion.
Results from the breathing trial described in chapter 10 were statistically analysed, comparing breathing system function between male and female volunteers, and
comparing how different breathing systems behaved with all volunteers. While there was no significant differences between many variables under comparison (and although these results were analysed, they were not produced in this thesis), there were some, and these are presented in this chapter. Most differences in gas concentrations between males and females can be explained by the higher ventilation in males, with higher oxygen and nitrous oxide uptakes. The breathing trial results are also compared to Bath fp results; as Bath fp results are given in mass fractions, these had to be converted into volumetric concentrations to make comparisons with measured results. Bias/precision graphs were drawn for oxygen and nitrous oxide concentrations sampled near the patient, and show reasonable agreement between Bath fp and measured results. This is supported by further data from Bath fp, in which a 10% increase is imposed on certain variables, and some expected variation in other variables is demonstrated.

To summarise some significant aspects of the analyses:

With regard to comparing clinical data between populations and breathing systems:

• Expired oxygen concentrations in the system are higher in females than males, because of the lower oxygen uptake of females. This is true for all three breathing systems at a FGF of 3 L.min⁻¹, but not at higher or lower FGFs for all breathing systems. Further study with Bath fp might elucidate gas composition at key points in each system to determine why this might be the case.

• The rate of rise of inspired oxygen in the system is higher in females than in males at a high FGF, (explained as above), and is higher with the standard circle system compared to the other two; this suggests no advantage is conferred by the other two systems over the standard system in this respect.

• The rate of rise of nitrous oxide in the system at high FGF in males is faster in the paediatric system than in the standard system due to a smaller system volume in the former; in females it is faster in the standard system than the coaxial system, perhaps reflecting a greater level of effort required in th latter.

• The airway pressure in the system at high FGF in females is lower in the standard system than in the other two, perhaps reflecting the simplicity of construction of the former, requiring less effort.

• Collection of this data has formed a useful piece of clinical research, irrespective of the availability of Bath fp.

With regard to comparing clinical data to Bath fp simulation:

• The tabulated comparative results show that for gas sampling close to the patient, the agreement between clinical data and Bath fp is good at high and low FGFs. It may be that at a FGF of 3 L.min⁻¹, the variance in clinical trial gas compositions was not reflected in the Bath fp simulation model.

• The bias precision curves suggest quite good agreement between Bath fp simulation and the clinical results, with a standard error of the mean of less than 1.0.
• The SEM of >1.0 for expired nitrous oxide concentration suggests that there may have been an error in the nitrous oxide absorption model, which could be reviewed further before further Bath fp (or Matlab) simulation. Alternatively this result could reflect that there is wide variation in human nitrous oxide absorption, depending on body habitus.

• Wide limits of agreement reduces confidence in the Bath fp model, but may reflect a relatively small sample size.

• With these provisos, mathematical simulation of respiration and breathing systems is a valuable technique.
Chapter 11.
Summary and recommendations.

For the purposes of informing both engineering and medical readership, the early chapters of this thesis introduced the concept of breathing systems for use in different environments, going into some detail about the structure and function of those systems for the anaesthetic, space and underwater environments. Emphasis was placed on low flow, closed circle systems in these contexts, mainly for reasons of economy of use in all three environments.

Due to the author’s familiarity with anaesthetic systems, the starting hypothesis was that the engineering function of anaesthetic circle breathing systems is not well understood, and that perhaps a detailed study of this system’s structure and function would be of benefit in improving its efficacy and safety. Bath fp was a mathematical modelling technique available in the department of mechanical engineering, and it was decided to use this to analyse the anaesthetic circle breathing system in some detail. The author claims no credit for the development of the Bath fp software used for this project, and is particularly grateful to Dr DG Tilley for this aspect of the work. Nevertheless, an introduction to some of the relevant equations required, as developed by others in the history of the development of Bath fp, is presented in this thesis.

In deciding how this project was to be developed, it was thought best to analyse several versions of the anaesthetic circle system, which are currently used clinically. These were the standard adult system, the paediatric system and the coaxial system. In deciding what model to use for anaesthetic gases, it was decided that a 50% mixture of nitrous oxide in oxygen was the best choice. The reasons for this choice were: because it was easier to see up to a 50% change in gas concentrations than a 1% or 6% change, which is the concentration, at which other anaesthetic agents are used; it was anticipated that a clinical breathing trial would be done to provide some validation for the Bath fp results, and this concentration of nitrous oxide is not a full anaesthetic, making this task logistically easier and ethically more justifiable; 50% nitrous oxide in oxygen is a good surrogate for the other volatile anaesthetic agents in this context. A Bath fp model of nitrous oxide absorption therefore had to be developed.

The rationale for the use of Bath fp is that certain components can be kept the same, and others can be changed at will. The patient model was kept constant throughout the Bath fp studies in order to determine if aspects related to breathing system geometry and FGF altered function. The earlier Bath fp studies provided some very useful information on the
function of the different circle systems at high, medium and low FGFs, some of it unexpected. In the standard system, tubing length and diameter were changed, to see if that would change function. While some change in function was found, it was not as profound as might have been expected, and there were more profound differences in function found with different FGFs, than with different geometry, so this was an unexpected ‘negative’ result. In terms of demonstrating efficacy and safety, the results were very useful in showing that not all clinical users are sufficiently mindful of the impact of, for example, reducing FGFs prematurely, so Bath fp will have provided a useful clinical learning tool. The function of the carbon dioxide absorber was then studied, and it was found that it did an adequate job in controlling CO\textsubscript{2} levels, which we know to be the case in clinical use. In changing the size of the CO\textsubscript{2} absorber it was found that some improvement in its function could be obtained from increasing the volume, or indeed by increasing the length iso-volumetrically; so a useful outcome from the Bath fp studies was to show that absorber geometry does alter carbon dioxide handling. In halving the diameter of the tubes to make the system equivalent to a paediatric circle system, but with the same standard adult sized patient, it was found that function was preserved without increase in respiratory work. This was surprising, considering that clinical dogma teaches that a paediatric sized system is inappropriate for adults; Bath fp once again provides a useful, but surprising result for clinical usage.

In studying the coaxial system, the author was particularly interested to know how function would change with varying coaxial tubing diameters and diameter ratios of inner and outer tubes. It came as some surprise to discover that changing this set of variables did not materially alter function; this was another surprising ‘negative’ result from Bath fp studies. Nevertheless, some useful results were obtained in clarifying the way a coaxial system works.

Although the addition of a venturi to breathing systems is not uncommon in aerospace and underwater environments, this has not found favour with anaesthetic breathing systems. It was thought worthwhile to develop a model for the venturi in such a system, using a coaxial configuration, and discover how it affected function. Given that part of the function of a venturi is to act as a hydraulic valve, to accelerate flow into a low pressure zone and through the venturi, it was thought that the venturi could act in the place of unidirectional valves to ensure predominant unidirectionality of the prevailing gas flow. The unidirectional valves were therefore omitted from the coaxial venturi system in the simulation model; it was recognized that this would put a significant demand on such a breathing system. It was found that similar profiles were produced with the venturi in place, with no unidirectional valves in such a system, as a standard or coaxial system with unidirectional valves, which was another surprising but useful result which requires further exploration. Not surprisingly the efficacy of the venturi system was less good at low FGFs than at high, and this demonstrates that there are limitations in
the use of a venturi in low flow clinical anaesthesia; this contrasts with aerospace and underwater applications, where the FGF is necessarily higher. It was found that bigger coaxial tubing and a smaller FGF nozzle enhanced the function of such a system, these changes increasing the volumetric entrainment ratio and reducing the risk of the venturi stalling. This was in contrast to Bath fp simulations for all other systems, where geometry was found to play an insignificant part in altering system function. This aspect of venturi system design should be explored further.

In the main, therefore, Bath fp performed a useful modelling function, clarifying aspects of breathing system function, demonstrating ways in which breathing system function would and would not be altered. It provided some useful clinical lessons in usage of such breathing systems. It has provided some areas for further research on the mechanics of breathing system design.

The next major part of the work was to undertake a clinical breathing trial, to provide some validation data to support the use of Bath fp. Ethical permission was obtained and nineteen adult, physically fit volunteers were recruited to breathe a gas mixture of 50% nitrous oxide in oxygen. Each volunteer breathed this mixture at several different FGFs, gases being sampled at three different points in the systems; each volunteer breathed through three different systems, a standard adult system, a paediatric system and a coaxial system; there was, of course, no venturi system available to use. A great deal of data were generated with this trial, which allowed comparisons to be made between breathing systems, between volunteers (males vs females), and between FGFs for a large range of variables. While some statistical differences were found, they were fewer in number, and in unexpected situations, compared to what might have been expected. This will provide a lot of additional information for clinical anaesthetists on the function of different breathing systems.

When the clinical data were compared to Bath fp data, a surprising degree of conformity was found between the two sets of data using combined (males and females) mean values, particularly for gas sampling near the volunteers’ mouths, even if there was a degree of variance demonstrated between volunteers. Broadly speaking, Bath fp was found to be a useful mathematical modelling tool for such systems, correlating reasonably well with clinical data. Exceptions to this result were probably due to highly variable nitrous oxide absorption between volunteers, or because the nitrous oxide absorption model used by Bath fp was not accurate for all volunteers.

It is recommended that, while currently used anaesthetic breathing systems may be used in the ways they have been to date, a greater awareness of their function will provide greater safety and efficacy for patients. The volume occupied by such (standard) systems can be reduced considerably, by having tubing which telescopes down to as little as 0.5m
in length, and by using smaller diameter tubing, even in adults. The volume of the CO₂ absorber should not be reduced, but its shape could be changed to make the gas pathway through it longer. A realistic engineering challenge would now be to make the block containing the CO₂ absorber and the unidirectional valves a small item which could be laid on the pillow next to the patient’s head.

It is also recommended that further studies on the use of a venturi in such a system be carried out. In contrast to smaller tubing suggested in the other systems, the venturi performed better with wider coaxial tubing. It would be interesting to know if the addition of unidirectional valves would enhance the performance of such a system by allowing lower FGFs to be used, or whether they would cause a back pressure which would cause the venturi to stall.
## Table 5.2. Physical properties of some gases (from reference [76])

<table>
<thead>
<tr>
<th>Name of Gas</th>
<th>Chemical Formula or Symbol</th>
<th>Approx. Molecular Weight M</th>
<th>Density kg/m³</th>
<th>Specific Gravity Relative to Air</th>
<th>Specific Heat (J/kg K)</th>
<th>Heat Capacity per Cubic Metre (J/m³ K)</th>
<th>Critical Conditions</th>
<th>Thermal Conductivity k W/mK</th>
<th>Coefficient of Viscosity η = 10⁻⁵ Pa s</th>
</tr>
</thead>
<tbody>
<tr>
<td>Acetylene (ethyne)</td>
<td>C₂H₂</td>
<td>26.0</td>
<td>1.0025</td>
<td>9.907</td>
<td>320</td>
<td>1.465, 1.127, 1.601, 1.231</td>
<td>1.30</td>
<td>300</td>
<td>63</td>
</tr>
<tr>
<td>Air</td>
<td></td>
<td>29.0</td>
<td>1.2045</td>
<td>1.000</td>
<td>287</td>
<td>1.609, 1.215, 1.868, 1.40</td>
<td>1.40</td>
<td>133</td>
<td>37.7</td>
</tr>
<tr>
<td>Ammonia</td>
<td>NH₃</td>
<td>17.0</td>
<td>0.7179</td>
<td>0.596</td>
<td>490</td>
<td>2.190, 1.659, 1.572, 1.191</td>
<td>1.32</td>
<td>406</td>
<td>113</td>
</tr>
<tr>
<td>Argon</td>
<td>A</td>
<td>39.9</td>
<td>1.6610</td>
<td>1.379</td>
<td>208</td>
<td>3.19, 2.11, 2.66, 1.57</td>
<td>1.67</td>
<td>131</td>
<td>49</td>
</tr>
<tr>
<td>α-Butane</td>
<td>C₆H₁₀</td>
<td>88.1</td>
<td>1.4897</td>
<td>2.607</td>
<td>143</td>
<td>1.654, 1.460, 3.710, 1.11</td>
<td>1.11</td>
<td>406</td>
<td>38</td>
</tr>
<tr>
<td>Carbon dioxide</td>
<td>CO₂</td>
<td>44.0</td>
<td>1.8417</td>
<td>1.529</td>
<td>189</td>
<td>1.858, 1.680, 1.216, 1.30</td>
<td>1.30</td>
<td>304</td>
<td>74</td>
</tr>
<tr>
<td>Carbon monoxide</td>
<td>CO</td>
<td>28.0</td>
<td>1.1686</td>
<td>0.967</td>
<td>297</td>
<td>1.017, 0.726, 1.185, 0.864</td>
<td>0.864</td>
<td>134</td>
<td>35</td>
</tr>
<tr>
<td>Chlorine</td>
<td>Cl₂</td>
<td>70.9</td>
<td>2.944</td>
<td>2.486</td>
<td>117</td>
<td>1.316, 1.684, 1.084, 1.13</td>
<td>1.13</td>
<td>417</td>
<td>77</td>
</tr>
<tr>
<td>Ethane</td>
<td>C₂H₆</td>
<td>30.0</td>
<td>1.2535</td>
<td>1.649</td>
<td>277</td>
<td>1.616, 1.325, 2.042, 1.674</td>
<td>1.674</td>
<td>306</td>
<td>48.8</td>
</tr>
<tr>
<td>Ethylene</td>
<td>C₂H₄</td>
<td>28.0</td>
<td>1.1744</td>
<td>0.975</td>
<td>296</td>
<td>1.875, 1.373, 1.967, 1.612</td>
<td>1.612</td>
<td>283</td>
<td>51.2</td>
</tr>
<tr>
<td>Helium</td>
<td>He</td>
<td>4.0</td>
<td>1.0663</td>
<td>0.1381</td>
<td>2.078</td>
<td>5.23, 3.153, 8.70, 5.24</td>
<td>5.24</td>
<td>13</td>
<td>10.4</td>
</tr>
<tr>
<td>Hydrogen sulphide</td>
<td>H₂S</td>
<td>34.0</td>
<td>1.4334</td>
<td>1.190</td>
<td>243</td>
<td>1.017, 0.728, 1.458, 1.121</td>
<td>1.121</td>
<td>374</td>
<td>90</td>
</tr>
<tr>
<td>Methane</td>
<td>CH₄</td>
<td>16.0</td>
<td>0.6673</td>
<td>0.554</td>
<td>519</td>
<td>2.483, 1.881, 1.657, 1.255</td>
<td>1.255</td>
<td>191</td>
<td>46.4</td>
</tr>
<tr>
<td>Methyl Chloride</td>
<td>CH₃Cl</td>
<td>35.0</td>
<td>2.2500</td>
<td>1.785</td>
<td>165</td>
<td>1.005, 0.838, 2.161, 0.800</td>
<td>0.800</td>
<td>416</td>
<td>63</td>
</tr>
<tr>
<td>Natural gas</td>
<td></td>
<td>19.5</td>
<td>0.8034</td>
<td>0.667</td>
<td>426</td>
<td>2.345, 1.864, 1.884, 1.483</td>
<td>1.483</td>
<td>11</td>
<td>13</td>
</tr>
<tr>
<td>Nitric oxide</td>
<td>NO</td>
<td>30.0</td>
<td>1.2491</td>
<td>1.057</td>
<td>277</td>
<td>1.675, 1.395, 1.208, 1.493</td>
<td>1.493</td>
<td>179</td>
<td>66</td>
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<tr>
<td>Nitrogen</td>
<td>N₂</td>
<td>28.0</td>
<td>1.1688</td>
<td>0.967</td>
<td>297</td>
<td>1.034, 0.733, 1.204, 0.854</td>
<td>0.854</td>
<td>126</td>
<td>33.9</td>
</tr>
<tr>
<td>Nitrous oxide</td>
<td>N₂O</td>
<td>44.0</td>
<td>1.6249</td>
<td>1.530</td>
<td>189</td>
<td>0.925, 0.706, 1.705, 1.301</td>
<td>1.301</td>
<td>310</td>
<td>73</td>
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<tr>
<td>Oxygen</td>
<td>O₂</td>
<td>32.0</td>
<td>1.3310</td>
<td>1.105</td>
<td>260</td>
<td>0.909, 0.649, 1.210, 0.864</td>
<td>0.864</td>
<td>154</td>
<td>40.5</td>
</tr>
<tr>
<td>Propane</td>
<td>C₃H₈</td>
<td>44.1</td>
<td>1.8814</td>
<td>1.562</td>
<td>188</td>
<td>1.645, 1.430, 3.095, 2.690</td>
<td>2.690</td>
<td>370</td>
<td>42.5</td>
</tr>
<tr>
<td>Propylene</td>
<td>C₃H₆</td>
<td>44.1</td>
<td>1.7477</td>
<td>1.451</td>
<td>198</td>
<td>1.499, 1.315, 2.620, 2.298</td>
<td>2.298</td>
<td>365</td>
<td>46</td>
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<tr>
<td>Sulphur dioxide</td>
<td>SO₂</td>
<td>64.1</td>
<td>2.7270</td>
<td>2.264</td>
<td>129</td>
<td>0.645, 0.512, 1.759, 1.396</td>
<td>1.396</td>
<td>431</td>
<td>78.7</td>
</tr>
</tbody>
</table>
Appendix chapter 8.1
Derivation of equations for function of venturi for use in Bath fp modelling (with permission of Dr D.G. Tilley)
\[ a x^2 + b x + c = 0 \]

Example (10): A quadratic equation is given by
\[ ax^2 + bx + c = 0 \]

Consider the following equation:
\[ a x^2 + b x + c = 0 \]

Case 1:
\[ a x^2 + b x + c = 0 \]

Case 2:
\[ a x^2 + b x + c = 0 \]

Case 3:
\[ a x^2 + b x + c = 0 \]

Case 4:
\[ a x^2 + b x + c = 0 \]

Case 5:
\[ a x^2 + b x + c = 0 \]

Case 6:
\[ a x^2 + b x + c = 0 \]

Case 7:
\[ a x^2 + b x + c = 0 \]

Case 8:
\[ a x^2 + b x + c = 0 \]

Case 9:
\[ a x^2 + b x + c = 0 \]

Case 10:
\[ a x^2 + b x + c = 0 \]

Case 11:
\[ a x^2 + b x + c = 0 \]

Case 12:
\[ a x^2 + b x + c = 0 \]

Case 13:
\[ a x^2 + b x + c = 0 \]

Case 14:
\[ a x^2 + b x + c = 0 \]

Case 15:
\[ a x^2 + b x + c = 0 \]

Case 16:
\[ a x^2 + b x + c = 0 \]

Case 17:
\[ a x^2 + b x + c = 0 \]

Case 18:
\[ a x^2 + b x + c = 0 \]

Case 19:
\[ a x^2 + b x + c = 0 \]

Case 20:
\[ a x^2 + b x + c = 0 \]

Case 21:
\[ a x^2 + b x + c = 0 \]

Case 22:
\[ a x^2 + b x + c = 0 \]

Case 23:
\[ a x^2 + b x + c = 0 \]

Case 24:
\[ a x^2 + b x + c = 0 \]

Case 25:
\[ a x^2 + b x + c = 0 \]

Case 26:
\[ a x^2 + b x + c = 0 \]

Case 27:
\[ a x^2 + b x + c = 0 \]

Case 28:
\[ a x^2 + b x + c = 0 \]

Case 29:
\[ a x^2 + b x + c = 0 \]

Case 30:
\[ a x^2 + b x + c = 0 \]
217

\[ \text{From continuity, } \frac{V_1}{A_1} = \frac{V_2}{A_2} \text{ and } \frac{V_1}{V_2} = \frac{A_2}{A_1}. \]  

\[ (18) \]

The process of which shall occur can be determined from the equation \( \frac{V_1}{V_2} = 0 \) when \( u_i = 0 \). The vertex is assumed to have ended.

\[ \text{Should the actual process increase or decrease pressure reduce, a point without be reached.} \]

\[ p_1 = \frac{\rho_1}{\rho_2} \]

\[ (19) \]

\[ \text{\( u_1 \) = \( \frac{u_1}{u_2} \) \( \frac{\rho_1}{\rho_2} \)} \]

\[ (20) \]

\[ \text{The solution approximate for } u_2 \text{ along } \rho_1 \text{ and } p_2 \text{ can be} \]

\[ \text{from the solution approximate for } u_1 \text{?} \]

\[ \text{If } u_1 \text{ and } p_1 \text{ can be} \]

\[ (16) \]

\[ \text{and} \]

\[ (17) \]

\[ \text{for } \text{eq}(10), \]

\[ \text{A solution to this equation can be found using,} \]
Hence, 

\[ \frac{d^2}{ds^2} \left( \frac{4}{3} \right) + \frac{\partial}{\partial s} \left( -\frac{1}{3} \right) = \frac{1}{3} \left( \frac{d}{ds} \right)^2 - \frac{1}{3} \left( \frac{\partial}{\partial s} \right)^2 \]  

\[ \frac{d}{ds} \left( \frac{4}{3} \right) - \frac{\partial}{\partial s} \left( -\frac{1}{3} \right) = \frac{1}{3} \left( \frac{d}{ds} \right)^2 - \frac{1}{3} \left( \frac{\partial}{\partial s} \right)^2 \]

\( q_n \left( \frac{4}{3} \right) \) 

\[ \frac{d}{ds} \left( \frac{4}{3} \right) - \frac{\partial}{\partial s} \left( -\frac{1}{3} \right) = \frac{1}{3} \left( \frac{d}{ds} \right)^2 - \frac{1}{3} \left( \frac{\partial}{\partial s} \right)^2 \]

\( q_n \left( \frac{4}{3} \right) \)
	  

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\[ u(z) = \frac{1}{2} \left( \frac{\partial}{\partial z} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial x} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial y} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial v} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial u} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial w} \right)^2 \]

Combining like terms and equating to 0,

\[ \frac{1}{2} \left( \frac{\partial}{\partial z} \right)^2 = 0 \]

Substituting \( u(z) = \frac{1}{2} \)

\[ \frac{1}{2} \left( \frac{\partial}{\partial z} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial x} \right)^2 = 0 \]

As \( \left( \frac{\partial}{\partial z} \right)^2 = \frac{1}{2} \), then substituting for \( u(z) = \frac{1}{2} \). Hence

\[ \frac{1}{2} \left( \frac{\partial}{\partial z} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial x} \right)^2 = 0 \]

From which \( u(z) = \frac{1}{2} \)

\[ \frac{1}{2} \left( \frac{\partial}{\partial z} \right)^2 + \frac{1}{2} \left( \frac{\partial}{\partial x} \right)^2 = 0 \]

\[ \text{Therefore, for the flow in case 2, gives} \]

\[ \frac{1}{2} = f(z) \]

\[ \text{Substituting is for} \]

\[ f(z) = 1 - \frac{1}{2} \]

\[ \text{Since} \]

\[ f(z) = 1 - \frac{1}{2} \]

\[ \therefore f(z) = 1 - \frac{1}{2} \]

As before, \( u(z) = \frac{1}{2} \).
act in the opposite direction to the normal flow condition.

including these factors in the analysis has often allowed

\( u_2 = -u_1 \) 
\( u_1 = u_1 \) 
\( u_2 = -u_2 \) 
\( u_1 = u_1 \)

operation, thus a constant for in the model by setting the changing the structure of the

Following the reversed, both vectors to, and \( u_2 \) will be negligible compared to normal.

(where 5 is attempts for the area change, and less factor \( k \))

\[ P = P + \frac{P_f}{z} \]

from another, pressure \( P \) can be found from

\[ \frac{4}{4} \]

and \( c = \frac{-g}{g_0} \)

\[ \frac{4}{4} \]

\[ \frac{4}{4} \]

\[ \frac{4}{4} \]

\[ \frac{4}{4} \]

where in this case.

\[ u_2 = \frac{-b + \sqrt{b^2 - 4ac}}{2a} \]

which case can be employed using the standard form,

Here, a quadratic graph relating \( u_2 \) to \( b \) and the pressure drop \( (P_1 - P_2) \) is obtained.
As the diameter of the orifice is increased, the pressure drop was observed to increase. The ratio of the pressure drop to the orifice diameter squared, for an orifice flow (u = 0), is given by

\[ \frac{A}{A_0} \frac{u}{40A} \]

where \( A \) is the orifice area, \( A_0 \) is the area of the orifice, and \( u \) is the velocity of the fluid.

The outlet velocity in this region is given by

\[ u_x = \frac{u_i}{2} \]

where \( u_i \) is the initial velocity at the orifice.

The pressure drop across the orifice (flow from \( u_i \) to \( u = 0 \)) is given by

\[ \Delta P = \frac{A}{A_0} \frac{u}{40A} \]

where \( A \) is the orifice area, \( A_0 \) is the area of the orifice, and \( u \) is the velocity of the fluid.

In the region where \( u_x \) is the outlet velocity from the orifice, the pressure drop is given by

\[ \Delta P = \frac{A}{A_0} \frac{u}{40A} \]
and reverse flow regions. Ensure that pressures and flows agreed at the junction between the two regions.

The modified pressure drop was then included in the network flow equations to determine the flow rates and pressures across the regions.

The vectors \( z \) and \( u \) were assumed to vary continuously across the regions.

At reverse flow (\( u = c_0 \)),

\[ p_d = \text{static} + \text{reverse flow } p_d. \]

At static (\( u = 0 \)),

\[ p_d = \text{pressure difference} = \text{static } p_d. \]

achieved in the transition region at \( x = 0 \).

converted to a linear characteristic using damping from forward to reverse flow. This was out.
### Table 9.1a) See legend below.
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<th>O2insp/Time %/sec</th>
<th>N2Oinsp max%</th>
<th>N2max insp/exp %/%</th>
<th>N2Oinsp/Time %/sec</th>
<th>etCO2 %</th>
<th>RespRate b/min</th>
<th>AWPpe ak cmH2O</th>
<th>Tvinsp ml</th>
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Table 9.1a) See legend below
Table 9.1a). Data extracted from variables recorded during breathing trials, standard adult anaesthetic circle system, sampling from near volunteer’s mouth. Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3L/min, lo = 1L/min, min = 0.5 L/min. Units: gas concns= %; gas concns/time=%/sec; resp rate=breaths/min; airway pressure (AWP)=cmH2O.
Table 9.1b). See legend below.
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<th>FGF</th>
<th>O2insp max %</th>
<th>O2maxinsp/ exp %</th>
<th>O2insp/ Time %/sec</th>
<th>NZO</th>
<th>NZOinsp/ exp Time %/sec</th>
<th>etcO2 %</th>
<th>RespRate b/min</th>
<th>AWpeak cmH2O</th>
<th>Tinsp ml</th>
<th>O2mininsp/ Exp %</th>
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| M  6 hi | 48 | 48/45 | 48/150 | 46.5 | (46.5-2.5)/(37-2.5) | 46.5/150 | 6 | 4-5 | 5 | 1500-2100 | 22/18 |
| med | 45 | 45/35 | 45/120 | 45 | (45-4.5)/(30-5.4.5) | 45/120 | 6 | 4-6 | 3 | 1500-2000 | 21.5/17 |
| lo | 20 | 20/16.5 | 20/216 | 12.5 | (12.5-2)/(23-2) | 12.5/216 | 4.5-5.5 | 10-15 | 2-3 | 500-1000 | 19/15 |

| M  7 hi | 49 | 49/40 | 49/52 | 48 | (48-1)/(35.5-1) | 48/52 | 5 | 13 | 7 | 700-900 | 20.5/15.5 |
| med | 40 | 40/32 | 40/111 | 35 | (35-2)/(23-2) | 35/143 | 4.5-5.5 | 11-14 | 4 | 800-900 | 20.5/18 |
| lo | 20 | 20/16.5 | 20/216 | 12.5 | (12.5-2)/(23-2) | 12.5/216 | 4.5-5.5 | 10-15 | 2-3 | 500-1000 | 19/15 |

| M  8 hi | 45.5 | 45.5/44 | 45.5/80 | 43 | (43-2)/(37-2) | 43/80 | 4.5-5.5 | 6-7 | 6 | 800-2000 | 20.5/14 |
| med | 42 | 42/37 | 42/141 | 39 | (39-2)/(30-2) | 39/141 | 4.5-5.5 | 7-15 | 3 | 500-1000 | 21/17 |
| lo | 17.5 | 17.5/13 | 17.5/190 | 15 | (15-1)/(12-3) | 15/190 | 5 | 7-13 | 2.5-5 | 1000 | 17.5/13 |

Table 9.1b). Data extracted from variables recorded during breathing trials, coaxial anaesthetic circle system, sampling from near volunteer’s mouth. Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3 L/min, lo = 1L/min, min= 0.5 L/min. Units: gas concns= %; gas concns/time=%/sec; resp rate=breaths/min; airway pressure (AWP)=cmH2O
|   | FGF | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | PAED | O2insp/ | O2maxinsp/ | N2Oinsp | N2Omaxinsp/ | etCO2 | Resp | Rate | b/min | AWPeak | cmH2O | TWinsp | ml | O2mininsp/ | p/exp | %/% |
|---|-----|------|------|------|------|------|------|------|------|------|------|------|------|------|------|------|mark| % | mark| % | % | % | % | % | % | % |
| 1 | hi  | 49   | 49/43| 49/80| 47   | (47-2)/(35-2)| 47   | (47-2)/80| 5    | 12   | 6.5  | 600  | 800  | 20.5/15| | | | | | | | | | |
|    | med | 44   | 44/38| 44/200| 38   | 38/29| 38/200| 5    | 12   | 5    | 0-800| 25/18| | | | | | | | | | |
|    | lo  | 26.5 | 26.5/21| 60   | 22.5 | (22.5-2.5)/(17-2.5)| 22.5 | (22.5-2.5)/160| 5    | 12   | 3.5  | 800  | 21/16| | | | | | | | | | |
| 2 | hi  | 49   | 49/47| 49/120| 49   | (49-2.5)/(42-2.5)| 49   | (49-2.5)/120| 4-5  | 9-13 | 5-10 | 300-| 800-| 21/17.5| | | | | | | | | | |
|    | med | 45   | 45/39| 45/180| 39   | (39-3.5)/(34-3.5)| 39   | (39-3.5)/180| 4-4.5| 10-17| 3-6  | 400-| 100-| 20.5/16.5| | | | | | | | | | |
|    | lo  | 28   | 28/22| 48/202| 27   | (27-2.5)/(21-2.5)| 27   | (27-2.5)/202| 3.5-4| 12-17| 1-8  | 300-| 800-| 20.5/16.5| | | | | | | | | | |
| 3 | hi  | 49   | 49/47| 49/130| 50   | 50/43| 50/130| 5    | 7-11 | 8    | 500-| 1500| 20/15| | | | | | | | | | |
|    | med | 40   | 40/35| 40/150| 35   | (35-2.5)/(28-2.5)| 35   | (35-2.5)/150| 5    | 12-15| 3-5  | 400-| 100-| 20.5/17.5| | | | | | | | | | |
|    | lo  | 27   | 27/21.5| 27/177| 22   | 22/18| 22/177| 4.5  | 13-15| 2    | 700  | 20/15| | | | | | | | | | |
| 4 | hi  | 47   | 47/44| 47/95| 47   | (47-2.5)/(40-2.5)| 47   | (47-2.5)/95| 4    | 13-15| 8-9  | 600-| 1000| 20.5/15| | | | | | | | | | |
|    | med | 40   | 40/35| 40/150| 35   | (35-2.5)/(28-2.5)| 35   | (35-2.5)/150| 5    | 12-15| 3-5  | 400-| 100-| 20.5/17.5| | | | | | | | | | |
|    | lo  | 27   | 27/21.5| 27/177| 22   | 22/18| 22/177| 4.5  | 13-15| 2    | 700  | 20/15| | | | | | | | | | |
| 5 | hi  | 50   | 50/42| 50/115| 48   | (48-2)/(31-2)| 48   | (48-2)/105| 5-6  | 6-10 | 8-10 | 500-| 1500| 21/15| | | | | | | | | | |
|    | med | 40   | 40/36| 40/130| 36.5 | (36-2)/(27-2)| 36   | (36-2)/130| 5.5-6| 8-10 | 4.5-5| 600-| 1300| 20/17| | | | | | | | | | |
|    | lo  | 22.5 | 22.5/17| 22.5/180| 16.5 | (16.5-2)/(12-2)| 16.5 | (16.5-2)/180| 5.5  | 9-11 | 3    | 550-| 1300| 20/16| | | | | | | | | | |
| 6 | hi  | 48   | 48/43| 48/40| 48   | (48-2)/(38-2)| 48   | (48-2)/110| 4.5-5| 9-11.5| 4.5-5| 450-| 700-| 20.5/18| | | | | | | | | | |
|    | med | 42   | 42/38.5| 42/120| 40   | (40-2)/(32.5-2)| 42   | (40-2)/120| 4.5-5| 10-12| 3-3.5| 550-| 700-| 21.5/17.5| | | | | | | | | | |
|    | lo  | 22   | 22/17| 22/210| 18   | (18-2)/(14.5-2)| 18-2| (18-2)/210| 5    | 13-14| 2    | 500-| 750-| 19.5/16| | | | | | | | | | |

*Table 9.1c*. See legend below.
| FGF | F7 | hi | 50/48 | 50/125 | 48 | (48-0.5)/(42-0.5) | 48/125 | 5.5 | 11-14 | 6-8 | 600 | 20/15 |
|     | med | 44/38 | 44/138 | 41 | (41-2)/(33-2) | 41/138 | 5.5 | 11-14 | 5 | 600 | 20/16 |
|     | lo  | min  | 19.5/14.5 | 19.5/155 | | | | | |

| F8 | hi | 48 | 48/44 | 48/180 | 47 | (47-2)/(35-2) | 47/180 | 6 | 4-5 | 10 | 1800-2800 | 20/15 |
|     | med | 38/35.5 | 38/120 | 31.5 | (31.5-2.5)/(26.5-2.5) | 31.5/120 | 5.5 | 6-8 | 5-6 | 1000-1500 | 21/16 |
|     | lo  | min  | 20 | 20/16 | 20/167 | 12.5 | (12.5-2.5)/(10.5-2.5) | 12.5/167 | 5.5 | 6-7 | 2-2.5 | 1000-1700 | 20/16 |

| M1 | hi | 48/46 | 48/52 | 48 | (48-1)/(40-1) | 48/72 | 8 | 6-7.5 | 8 | 950-1800 | 21/17 |
|     | med | 29/22 | 29/180 | 26 | (26-4)/(21-4) | 26/180 | 5.3-5.7 | 8-10 | 3 | 950 | 20/15 |
|     | lo  | min  | 27.5 | 27.5/22 | 27.5/190 | 25 | (25-2.5)/(21-2.5) | 25/190 | 4.5-5 | 11-16 | 3.5-5 | 600-900 | 20/15 |

| M2 | hi | 49.5 | 49.5/48.5 | 49.5/100 | 48.5 | 48.5/39.5 | 48.5/73 | 4-5 | 10-14 | 6-7.5 | 600-900 | 21/15 |
|     | med | 27.5/25.5 | 27.5/190 | 25 | (25-2.5)/(21-2.5) | 25/190 | 4.5-5 | 11-16 | 3.5-5 | 600-900 | 20/15 |
|     | lo  | min  | 15 | 15/13 | 15/240 | 8 | (8-2)/(8-2) | 8/240 | 4-5.5 | 5-10 | 5-10 | 800-2500 | 15/13 |

| M3 | hi | 49.5 | 49.5/45 | 49.5/125 | 47.5 | (47.5-1.5)/(37.5-1.5) | 47.5/125 | 5.5 | 8-10 | 11-18 | 1000-2000 | 20/14 |
|     | med | 43/34 | 43/240 | 42.5 | (42.5-1)/(31-1) | 42.5/240 | 5.5-6 | 5-10 | 8-14 | 1000-2800 | 19.5/13 |
|     | lo  | min  | 15 | 15/13 | 15/240 | 8 | (8-2)/(8-2) | 8/240 | 4-5.5 | 5-10 | 5-10 | 800-2500 | 15/13 |

| M4 | hi | 50 | 50/45 | 50/42 | 50 | (50-2)/(38-2) | 50/45 | 4.5-5 | 16 | 6-8 | 500-800 | 20/17 |
|     | med | 40/36 | 40/80 | 37 | (37-2)/(26.5-2) | 37/80 | 4-4.5 | 15 | 4.5-5 | 600-1000 | 20/17 |
|     | lo  | min  | 19 | 19/16 | 19/150 | 13.5 | (13.5-2)/(10-2) | 13.5/150 | 4.5 | 15-16 | 3-5 | 600-1000 | 19/16 |

Table 9.1c). See legend below.
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Table 9.1c). Data extracted from variables recorded during breathing trials, paediatric anaesthetic circle system, sampling from near volunteer’s mouth. Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3 L/min, lo = 1L/min, min= 0.5 L/min. Units: gas concns= %; gas concns/time=%/sec; resp rate=breaths/min; airwaypressure (AWP)=cmH₂O.
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Table 9.2. See legend below.
Table 9.2. Data extracted from variables recorded during breathing trials, all three circle systems, sampling from inspiratory tube away from volunteer (‘insp sample’), and from expiratory tube away from volunteer (‘exp sample’). Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3 L/min, lo = 1L/min, min= 0.5 L/min. Units: gas concns= %;
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<th>ΔN2O max exp/insp %</th>
<th>ΔN2O max insp/time %/sec</th>
<th>etCO2 %</th>
<th>RespRate bpm</th>
<th>AWP Peak cmH2O</th>
<th>Tvinsp ml</th>
<th>O2min insp/exp %/%</th>
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**Table 9.3a).** Data calculated from data in table 9.1a), standard adult anaesthetic circle system, sampling from near volunteer’s mouth. Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3 L/min, lo = 1L/min, min= 0.5 L/min. Units: gas concns= %; gas concns/time= %/sec; resp rate=breaths/min; airway pressure (AWP)=cmH2O.
Table 9.3b). See legend below.
### Table 9.3b)

See legend below.
Table 9.3b). Data calculated from data in table 9.1a), coaxial anaesthetic circle system, sampling from near volunteer’s mouth. Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3 L/min, lo = 1L/min, min= 0.5 L/min. Units: gas concns= %; gas concns/time= %/sec; resp rate=breaths/min; airway pressure (AWP)=cmH2O
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<th>ΔN₂O InsP/max time %/sec</th>
<th>etCO₂ %</th>
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Table 9.3c). Data calculated from data in table 9.1a), paediatric anaesthetic circle system, sampling from near volunteer’s mouth. Female volunteers F1 to F9, male volunteers M1 to M9. FGF: hi=10L/min, med=3 L/min, lo = 1L/min, min= 0.5 L/min. Units: gas concns= %; gas concns/time=%/sec; resp rate=breaths/min; airway pressure (AWP)=cmH2O.
The first part of a PhD project in engineering has been to model the function of anaesthetic circle breathing systems using software called ‘Bathfp’. This has characterised individual gas flows, pressures and concentrations at numerous points in the system using a breathing mixture of 50% nitrous oxide in oxygen, a spontaneously breathing patient model, and a range of fresh gas flows. The system geometry was then altered by changing tubing diameter and length, and the models were run again to determine if system efficiency and safety can be improved by design changes. There is now a requirement to validate Bathfp software clinically by running a small clinical trial using volunteers. Calibration of Bathfp against human breathing volunteers, rather than against an artificial lung, will provide the validation required by the designers of Bathfp and other medical engineers.

The proposal is to recruit up to twenty anaesthetic, surgical and nursing colleague volunteers from the RUH, Bath, and from the neighbouring hospitals of Frenchay, Southmead and BRI. The age range of volunteers will be 25 to 55 years, and the unfit, pregnant and non-consenting will be excluded in the recruitment process. The volunteers will therefore be clinical professionals themselves, capable of understanding the process in the trial they are consenting to. The process will be explained, as will risks and benefits. Written consent will be obtained on the explicit understanding that volunteers may withdraw from the study at any time. Managerial approval will have been obtained to use the standard operating theatre suite at RUH Bath, using standard anaesthetic equipment.

The process will involve asking the volunteers to breathe through a standard anaesthetic facemask and circle breathing system, a mixture of 50% nitrous oxide in oxygen. This mixture is not a full anaesthetic, and will only cause a mild feeling of elation, but forms a useful basis for comparison to the Bathfp model and acts as a surrogate for all other anaesthetic agents in this respect. Some initial training in standardising the breathing pattern to match the Bathfp patient model will be required. During a normal anaesthetic, measurements are made of gas pressures, flow rates and concentrations from one point in the breathing system; during this trial, these measurements will be made from three points in the system. The volunteers’ physiological well-being will also be monitored according to normal standards set out by the Association of Anaesthetists of Great Britain and Ireland (1). The chief investigator, who is an anaesthetist with thirty years experience, will be clinically responsible for the volunteers’ clinical safety and well-being. Nine runs of breathing, lasting fifteen minutes each (total 135 minutes) on four separate occasions to study different types of circle breathing system, will be sufficient to
obtain the required data. The volunteers will be looked after by the chief investigator, according to normal clinical standards in the recovery area of the operating theatres, until they are fully recovered and physiologically autonomous.

The data obtained from this study will be coded so that individual volunteers cannot be identified, and will be kept secure in a password locked information storage system at the Chief Investigator’s home and at University of Bath. It will only be used to validate Bathfp software and will be destroyed three months after successful thesis submission.


**Participant Information sheet:** Version 2 October 2012
I would like to invite you to take part in a clinical trial. This trial is designed to validate some software modelling I have carried out over recent years on circle anaesthetic breathing systems. Before you decide I would like you to understand why the trial is being done and what it would involve for you. **I will go through the information sheet with you personally and answer any questions you have.** This should take about fifteen minutes.
Talk to others about the study if you wish.
(Part 1 tells you the purpose of this study and what will happen to you if you take part.
Part 2 gives you more detailed information about the conduct of the study).
Ask me if there is anything that is not clear.

**Part 1**

**Introduction**
The first part of my PhD project in engineering has been to model the function of anaesthetic circle breathing systems using software called ‘Bathfp’, originally written by my colleagues in the Department of Mechanical Engineering at the University of Bath to model breathing systems used for diving. I have used Bathfp to characterise individual gas flows, pressures and concentrations at numerous points in the anaesthetic circle system using a breathing mixture of 50% nitrous oxide in oxygen, a spontaneously breathing patient model, and a range of fresh gas flows. The system geometry was then altered by changing tubing length and diameter and absorber volume, and by the addition of a venturi; the models were run again to determine if system efficiency and safety can be improved by these design changes. As part of the PhD work there is now a requirement to validate Bathfp software clinically by running a clinical trial using volunteers, in order to calibrate the software model against human breathing. While it would be possible to calibrate the software model against a hardware model in the form of an artificial lung, from a medical engineering point of view validation of an artificial
model comes from calibration, not against another model, but against living, breathing humans.

My hypothesis is that the anaesthetic circle system has been poorly designed and constructed, and I believe its efficacy and safety can be improved. As much as medical engineers have faith in their modelling calculations, they are wise enough to look for ways to validate them against human structure and function, and this clinical trial will provide that opportunity for this project.

The site for the trial will be the main operating theatre suite at the RUH, chosen with familiarity and maximum safety in mind. I would like to recruit up to twenty participants of reasonable fitness, between the ages of 25 and 55, to breathe a gas mixture of 50% oxygen in nitrous oxide (‘entonox’) for a few minutes at a time, with air breathing recovery periods. A single trial may take up to two and a quarter hours. This gas mixture is not an anaesthetic, but may cause mild sedation and euphoria (‘laughing gas’; ‘gas and air’ for use in obstetric labour); in addition, this mixture does not significantly impair the heart or the breathing in people of reasonable health and fitness, and is quickly reversible. Participation will be invited from anaesthetic, surgical and nursing colleagues at RUH, Frenchay, Southmead and BRI. To avoid the possibility of potential coercion of such colleagues for whom I might be considered to be a line manager, the initial point of contact for recruitment will be Professor Mark Tooley, Director of Medical Physics and Bioengineering at RUH, who is not a stakeholder in this project, and is not a line manager of any potential participant. It is up to you to decide whether to join the trial or not, and no attempt will be made to coerce you into doing so. If you indicate a willingness to participate, I will describe the study and go through this information sheet verbally. Then, on a subsequent occasion, and only if you have decided to take part, I will ask you to sign a consent form. You are free to withdraw at any time, without giving a reason. This would not affect the standard of care you receive from me, or any subsequent working relationship between us.

**Conduct of the trial**

A non-working day for your participation will be chosen, and I will collect you by car to bring you to the RUH and take you home afterwards. It is recommended that you avoid driving or operating hazardous machinery, or returning to work for 24 hours afterwards. Initially a confidential, short, pre-trial conversation will take place with you to exclude co-morbidities, pregnancy, or incompatible medicines. After a short training period to standardize breathing patterns of volunteers, I propose initially giving you three ‘entonox’ breathing periods, each lasting about five minutes, delivered through a tight fitting standard anaesthetic facemask, using a standard circle anaesthetic breathing system. Each of the three runs will correspond to low, medium and high fresh gas flows. This gas mixture is not an anaesthetic, merely an analgesic, and you will just feel a bit
euphoric. You will breathe oxygen enriched air for up to ten minutes between runs to clear your system and the breathing system, going to air breathing before the start of the next run. At the end you will breathe oxygen enriched air to facilitate your recovery.

I will be looking after your physiological well-being by close personal observation, and by monitoring you according to internationally approved standards (Association of Anaesthetists of Great Britain and Ireland) which anaesthetists use every day on surgical patients; these include: ecg; non-invasive blood pressure; pulse oximetry; inspired and expired oxygen, nitrous oxide and carbon dioxide; volumetric flowrates and pressures. The gas variables being monitored will have resulted from sampling at the standard point in the breathing system close to the face mask. It is proposed to repeat the same anaesthetic runs with the same gas data being collected from two other points in the breathing system: at the entrance to, and exit from the carbon dioxide absorber, since these are easily accessible and correspond to points in the Bath fp model at which a lot of data was recorded in the model. This will make a total of nine anaesthetic runs, each lasting fifteen minutes, a total of up to two and a quarter hours. There will be a trained operating department assistant available to help me look after you, should the need arise, and I will then be responsible for your full recovery to fitness for going home, and I will take you home myself.

Ideally, I would like to keep the same volunteers to repeat the same runs using shorter tubing, narrower tubing (e.g. a paediatric circle system), and a coaxial circle system. These would take place on separate occasions.

I will be recording your monitoring data on a standard anaesthetic chart, identified by a number, which I will have allocated to you, so that the chart itself is not directly identifiable to you. All clinical information relating to this trial will be kept securely on site at the RUH.

**Risks and Benefits**

If you kindly volunteer to take part in this trial, it is clearly a significant proportion of a day of your own time, which might be morning, afternoon or evening, on up to four occasions that I am asking for. I am very grateful for this, and am mindful that it may cause you some inconvenience. As you know the side effects of breathing an entonox mixture may include dizziness and possible nausea and vomiting; if these symptoms persist into the ‘postoperative’ period, I will treat you with oral or intravenous fluids and with oral or intravenous anti-emetic drugs. It is unlikely that there will be any other sequelae, but I will remain clinically responsible to you for dealing with them. As this trial is taking place at the RUH, NHS indemnity applies for any liability issues which may occur. Nevertheless, before agreeing to participate, you may wish to consider
whether participation in this trial will affect any other insurance issues you have, and seek advice if necessary.

Although a number of studies in the use of nitrous oxide have failed to elicit any association with harm to the unborn child, I would prefer not to recruit female colleagues for whom there is even the slightest chance that they may be pregnant. It would also be better not to recruit anyone who is breastfeeding within twelve hours of the trial. Although there is no evidence that nitrous oxide causes sperm damage, I would also caution male colleague participants against such a risk for a period of twenty four hours following participation.

Any complaint about the way you have been dealt with during the study or any possible harm you might suffer will be addressed. The facts relating to your participation in the trial will be confidential and will follow ethical and legal practice; all information about you will be handled in confidence.

The benefit of this study is that you will have materially helped in an in depth study of the function of the circle breathing system, which you and I use on patients every day, and in improving its design for the benefit of future patients.

The results I obtain from this trial will be compared to those obtained during the Bath fp modelling process. This will provide evidence, both for the value of the Bath fp model for circle anaesthetic breathing systems, and to show ways in which modelling processes based on fluid mechanics equations may be improved. It is hoped that conclusions can be drawn from both clinical and modelling results to improve the mechanical design of the circle system. I will present to you, my professional colleagues, the outcome of such results, and I would hope to publish in the anaesthetic and engineering literature, as well as write up the whole of the work as a PhD thesis.

This completes Part 1. If the information in Part 1 has interested you and you are considering participation, please read the additional information in Part 2 before making any decision.

**Part 2**

Sometimes new information about the system being studied becomes available. If this happens, I will tell you and discuss whether you should continue in the trial. It will be perfectly alright if you decide not to carry on, but if you decide to continue I may ask you to sign an agreement of the discussion which has taken place. Alternatively, I might consider that you should withdraw from the study, and I will explain the reasons.
If you withdraw from the study, I will destroy all your identifiable information, but we will need to use the data collected up to your withdrawal.

If you have a concern about any aspect of this study, you should speak to me and I will do my best to answer your questions [07891 544 632]. If you remain unhappy and wish to complain formally, you can do through the NHS Complaints Procedure. Details can be obtained from Elspeth Alexander in the Department of Anaesthesia, RUH [01225 825057].

In the event that something does go wrong and you are harmed during the research, and this is due to someone’s negligence, then you may have grounds for a legal action for compensation against the RUH Bath NHS Trust, but you may have to pay your legal costs. The normal National Health Service complaints mechanisms will still be available to you.

If you join the trial, some parts of the monitoring data collected from you will be looked at by persons authorized by me, such as my clinical line manager (clinical director, Dr Alex Goodwin) or a statistical expert within the department (Dr Andy Padkin), or by a statistical expert within the University of Bath (Dr Gordon Taylor), or by my research supervisor at the University of Bath (Dr Roger Ngwompo). They may also look at these data to check that the study is being carried out correctly. All will have a duty of confidentiality to you as a research participant and we will do our best to meet this duty.

All research in the NHS is looked at by an independent Research Ethics Committee, to protect your interests. This study has been reviewed and given favourable opinion by the South West Research Ethics Committee.

You may wish to access other information before deciding whether to volunteer for this study. This may include general information about research, or specific information about this research project. It may also include advice as to whether you should participate or whom you should approach if you are unhappy with the study. I am happy to give you advice on any of these matters. For more independent advice, senior consultant colleagues in the department who are or have been involved in research may be approached, in particular Dr Tim Cook, Director of Research within the Department of Anaesthesia at RUH, or Professor Mark Tooley, Director of Medical Physics and Bioengineering, and of Research and Development at RUH.

Thank you for your time in looking at this information sheet.

7.x.12
Patrick Magee
Consultant in Anaesthesia, RUH Bath
Consent form

Patient Identification Number for this trial:

Title of Project: **Clinical Validation of ‘Bathfp’ software simulation of anaesthetic circle systems**

Name of Researcher: Dr Patrick Magee

Please initial each numbered item as well as signing at the bottom of the page.

1. I confirm that I have read and understand the information sheet dated................ (version2 October 2012) for the above study. I have had the opportunity to consider the information, ask questions and have had these answered satisfactorily.

2. I understand that my participation is voluntary and that I am free to withdraw at any time without giving any reason, without my medical care or legal rights being affected.

3. I understand that relevant sections of my medical data collected during the study, may be looked at by individuals from the Department of Anaesthesia, RUH Bath, from regulatory authorities or from other individuals at RUH Bath where it is relevant to my taking part in this research. I give permission for these individuals to have access to my records.

4. I agree to take part in the above study.

Name of Participant
Date
Signature

Name of Person taking consent
Date
Signature

When completed: 1 for participant; 1 (original) for researcher site file, to be kept securely on RUH premises
References


7) Personal communication, Amy Ross, Johnson Space Center, Houston TX, 2002.


76) Lo JKW, Tilley DG. The Digital Simulation of a 75m Surface Demand Diving System. School of Mechanical Engineering Report no. 005/1994, University of Bath 1994.


93) www.datex-ohmeda.com


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97) http://www.e-lfh.org.uk/e-learning-sessions/rcoa-novice/content/sessions/01_28_01/jpg/ana_01_070_4_3_med.jpg


Author’s list of Journal Publications

1) Lanyon LE, Magee PT, Baggott DG: Relationship of functional stress and strain to bone remodelling; an experimental study on the sheep radius. J.Biomech. 12, 1979, 593-600.


14) Magee PT. Gas flow between coaxial tubes: impedance to gas flow in an
endotracheal tube increases with a catheter within. Proc IMechE, Part H: J Engineering in Medicine, 226 (6), June 2012, 491-494.

Presentations done, and proposed publication in relation to this research.

Posters.


Verbal

Publication.

Proposed publications.
1) The many uses of low flow circle systems in anaesthesia, underwater, mountaineering and aerospace activities.

2) Mathematical simulation modelling to elucidate the function of the currently used low flow circle system in anaesthesia.

3) The mathematical modeling of coaxial low flow circle systems.

4) The mathematical modeling of a venturi in a coaxial circle system

5) Clinical testing of low flow circle systems for use in anaesthesia.

6) The clinical validation of mathematical modeling of low flow circle system