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MENG PROJECT PHASE 1 REPORT
<A PORTABLE DEVICE DESIGN, USING FORCED
OSCILLATION TECHNIQUE, FOR RESPIRATORY
IMPEDANCE MEASUREMENT OF PATIENTS IN
CHRONIC OBSTRUCTIVE PULMONARY DISEASE >
3 AUGUST 2022

MEng Project Mission Statement

Biosensor system design: measuring the respiratory impedance for patients who have Chronic obstructive pulmonary disease (COPD)

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Subject Area: Biomedical application/ Biosensor/ Micro-control system

Project Definition:

Chronic obstructive pulmonary disease(COPD) is the name for a group of lung conditions that cause breathing difficulties [1]. Patients suffering from this disease need regular monitoring on their breathing condition. The device based on spirometry remains significant for wide clinical use in this area. However, massive inconvenience has risen due to its inflexibility of gaint size and high requirement of cooperation from patients. Therefore, a portable biosensor device using the **Forced Oscillation Technique (FOT)** would be developed in this project as an approachable daily measurement for COPD patients. Generally, it will contain a sensor system to capture information provided by patients, a micro-control system to manage data feedback, and a communication system that can send information to the PC and preserve it for doctors and patients to check.

Task Clarification:

No.	Stage	Task
1	Micro-control system design (Phase 1)	Get familiar with STM32 MCU board and software. Design a micro-control system and communication system Run simulation
2	Sensing system design (Phase 1)	Figure out dynamics of airflow Design and fabricate a pneumotachometer Collect signals from sensing system(P/F)
3	Fabrication (Phase 2)	Combine systems together A model to imitate lung functions may be fabricated at this stage
4	Optimisation (Phase 2)	Additional functions as LED display will be extended here to improve the integrity of this biosensor system. Operate further signal processing, improve accuracy of the biosensor system.



Scope for Extension:

1. Thermal effect:

Previous research shows the accuracy of a pneumotachometer is mostly related to the temperature.[2] Therefore, a thermal sensor will be added to the control system in order to perform BTPS (Body temperature and pressure, saturated) corrections.

2. Signal processing:

Further biomedical signal processing should be added at phase-two stage to improve the accuracy for whole biosensor system.

3. Lung mimic:

An Extra model to intimate lung function can be developed and fabricated to prove the integrity of entire system.

Background Knowledge:

- Control and instrumentation
- Analogue/ digital signal processing
- Sensor and instrumentation
- Flow dynamics
- C-programming

Resources:

- STM32 Nucleo Board
- STM32 CUBEIDE
- Flow Head ML311
- Breadboard and jumper cables
- Digital potentiometer
- Temperature sensor
- Interface circuits
- Heating cables
- 3D printer
- Electromagnetic signal generator

Reference:

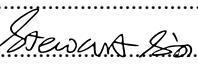
[1] [Chronic obstructive pulmonary disease \(COPD\) - NHS \(www.nhs.uk\)](https://www.nhs.uk) , cited on 14th Feb, 2022

[2] <https://www.pftforum.com/blog/pneumotach-accuracy/> ,cited on 8th Feb, 2022

The supervisor and student are satisfied that this project is suitable for performance and assessment in accordance with the guidelines of the course documentation.

Signed

Student:

Supervisor: 

Date: ..14/02/2022....

Abstract

A portable device, aiming for measuring the respiratory impedance of patients with **chronic obstructive pulmonary disease (COPD)**, is developed in this project in two phases. The whole system functioning as a biosensor is established based on **the forced oscillation technique (FOT)**, which can provide detailed information on lung mechanisms, especially the peripheral airways. In phase 1, the physiological background of COPD and the technical background of spirometry and FOT/IOS were both explicitly clarified. The system was considered to be divided into 6 subsystems, with different functions as sensing, actuating, computing, powering, wireless communicating and heating. The processing methodology for each functional module was carefully illustrated. Preliminary design has been delivered in electronics architecture and software logic. **The augment RIC model(aRIC)** was built to verify and simulate the whole strategy. Simulation results from aRIC model prove the reliability of FOT and the preliminary structure of the project. A detailed plan for tasks in phase two and an overall Gatt chart were both given, more practical works, including prototyping and integrating on the bench, will be brought at the next stage.

Declaration of Originality

I declare that this thesis is my
original work except where stated.

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Statement of Achievement

As an electronics and electrical engineering student, this biosensor project rewards me with a great amount of biomedical knowledge, broadening my horizons in both the technical field and physiological fields. I learned how to deliver the system design for a product, build up a robust electronics architecture as well as manage the relationships between different functional modules. More specially, I handled better on many software like LTSpice, MATLAB, STM CUBEIDE, Target PCB, etc. My capability of selecting components, customising circuits, testing and debugging has also been further developed. This valuable biotech project kicks off my interest in the bio-engineering field as a brilliant start.

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List of Symbols

Pound sign; used with the meaning “number of”.

θ Angle between the radial line and the z-plane.

Glossary

aRIC The augmented RIC model .

COPD Chronic Obstruct Pulmonary Disease.

eRIC The extended RIC model .

FEV1 Forced expiatory volume.

FOT Forced Oscillation Technique.

GOLD Global Initiative for Chronic Obstructive Lung Disease.

IOS Impulse Oscillatory System.

LDO Low dropout regulator.

LS The least square, as an error criterion .

LSA Linear Solenoid Actuator.

MCU Micro controller unit .

Chapter 1

Introduction

1.1 Purposes of project

A portable device, aiming to measure the respiratory impedance of **COPD** patients, would be developed as my dissertation project. It is chosen mainly for the two reasons given below.

On the one hand, there is a market harvest of small and portable devices for respiratory measurement. Most devices in the market are giant and expensive, requiring complex operations under the supervision of professionals. For COPD patients, chronic disease needs regular monitoring. It is inconvenient and costly if COPD patients have to go to the hospital frequently. Therefore, a portable device, which can be operated by a single person, is decided to be developed in this project, satisfying patients' needs for their daily use.

On the other hand, this inter-disciplinary project requires knowledge in multiple fields, not only electronics but also mechanical, software, and physiology. It is always a great dream for me that I could be involved in biotechnology design, applying personal knowledge to research and develop authentically helpful products. Although challenges have been foreseen, I still believe that the whole project would be reliably finished with my great passion.

1.2 Challenges of project

Electronics Challenge

As a complicated biosensor system, the portable device is embedded with multiple subsystems, giving different functions. Therefore, the first challenge is to build up a robust electronics architecture, carefully considering demand function blocks. What's more, the core of this portable device is flow sensing and data computation. In this case, various data types need to be collected and transported. Thus, the design of different interfaces comes as another challenge. Each system shall have an interface circuit to

communicate with **MCU** which includes the voltage regulator, switching circuit and etc.

Software Challenge

MCU needs to be programmed properly for impedance computation and subsystem control. A detailed explanation of MCU is given in the following section 4.2. An STM32 Nucleo-32 board with part number L432KC is chosen, which is required to be programmed by C/C++ through its specified software platform STM32CUBIDE.

Mechanical Challenge

The mechanical design significantly affects the process of respiratory measurement. For example, the length from mouthpiece to pneumotachograph has an impact on the precision of flow sensing. Besides, the space inside of the device must be well-organized as this is a small portable device but required it to be equipped with various functional blocks. There is a great challenge for an electronics student to finish a complicated mechanical design, using professional software in the mechanical field. Ideally, a mechanical frame is expected to be printed out at the end of Phase two.

1.3 Expectations from project

Phase 1

- Read and get familiar with large amounts of background materials required.
- Understand and appreciate the principle of the biosensor system.
- Build up a robust electronics architecture of the portable device.
- Preliminary simulations shall be run for verifying the reliability of the preliminary design.
- Give a detailed plan and consideration for the project at Phase 2.

Phase 2

- Managed to design and verify each functional block of electronics architecture with detailed circuits.
- Manage to collect and compute data by programming a reliable control system.
- Managed to design and print out the mechanical frame of the portable device.
- Manage to give an overall prototype of the portable device, working properly with all functions.

Chapter 2

Background

2.1 COPD - Chronic Obstruct Pulmonary Disease

2.1.1 What is COPD?

Chronic obstructive pulmonary disease, known as COPD, is a chronic inflammatory lung disease that is characterised by typical symptoms like coughing, shortness of breath, mucus production, wheezing, etc [1] ... Objectively speaking, smoking and air pollution, containing irritating gases or particulate matter, are the two main factors that cause COPD. Research in 2006 [2] pointed out that around 70.5% of COPD patients have ever smoked. But why do irritants cause lung disease? How does the lung get affected?

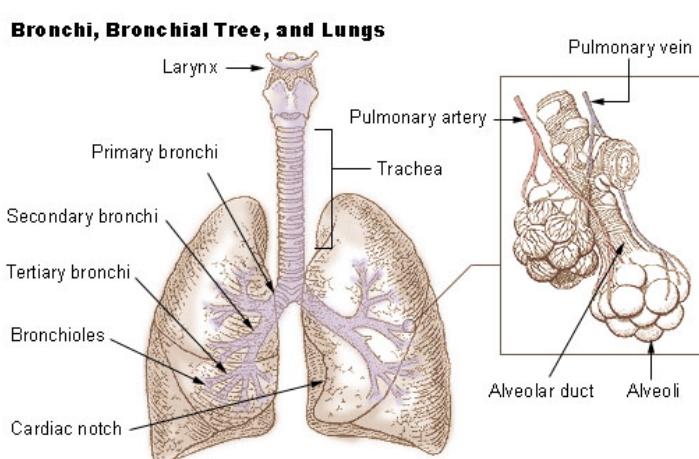


Figure 2.1: Diaphragm - Human respiratory system

A simple diaphragm illustrating the human respiratory system is shown in the figure 2.1 Air travels inside of the human body and gets into the lung through two large tubes called **Bronchi**. Inside of human lung, the bronchi get divided into hundreds of smaller branches named **Bronchioles** and all of them eventually end in shelters of small air sacs, known as **Alveoli**. [1] Alveoli contain skinny wall that is full of tiny **capillaries**, through which oxygen inhaled and carbon dioxide exhaled can pass and be exchanged with the human bloodstream.

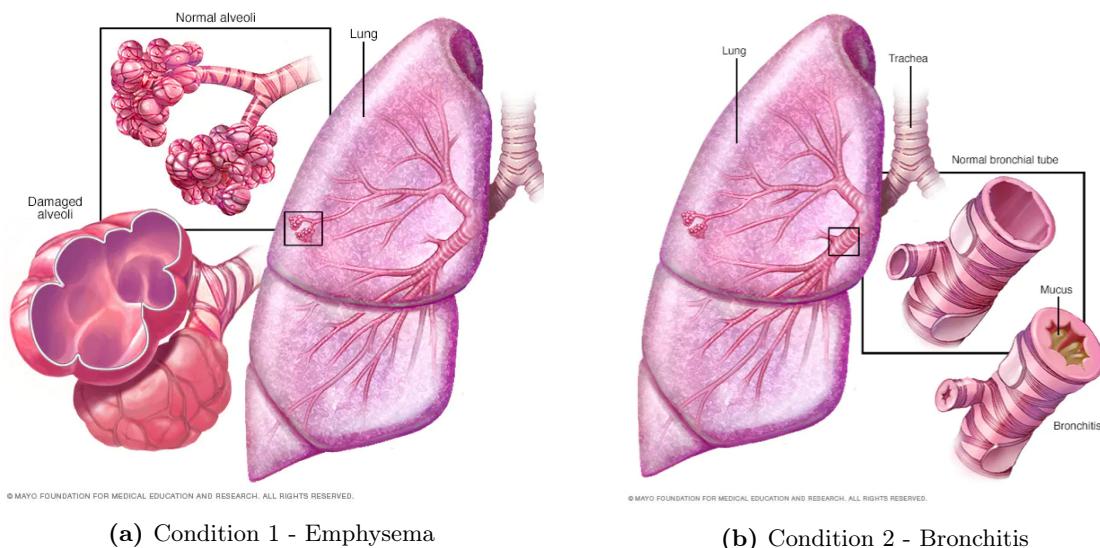


Figure 2.2: Two causes of airway obstruction in human lung [1]

This gas exchange process, with inspiration and expiration, is called breath, known as the foundation of the human life cycle. The natural elasticity of the bronchial tubes and alveoli determines how well the lung can force the air out of the body. Those two are less elastic and over-expanded inside COPD patients, exactly due to irritating gases and matters. Physiologically, COPD can be categorised as two typical conditions [1], Emphysema 2.2a and Bronchitis 2.2b, separately shown in the figure 2.2.

Emphysema

Emphysema indicates the damage to alveoli, resulting in the collapse of small airways when humans expire.

Bronchitis

Bronchitis is the condition when bronchial tubes get inflamed and tumescent. This brings more mucus secreted by the lung, which causes further blocking inside of narrowed bronchial tubes.

2.1.2 How often should a COPD patient get monitored?

According to Global Initiative for Chronic Obstructive Lung Disease (GOLD) [3], it is crucial that have a routine follow-up of COPD patients. Lung function varies with time, regular monitors on symptoms, exacerbation and objective measures of airflow limitation are vital to helping modify treatments and identify complications further.

GOLD clarifies that regular evaluation of COPD patients shall depend on the severity of the disease [4]:

- For mild and stable COPD patients, monitoring intervals shall be within 6 months.

- For severe COPD patients who are recently hospitalised, monitoring intervals shall be within 1 month.
- Patients who deliver non-pharmacological treatments are better to be monitored or give comprehensive self-management on a daily basis, in order to avoid exacerbation due to aggravating factors.

2.1.3 How to monitor a COPD patient?

Six main factors shall be monitored according to principles set by GOLD 2022 report, which are summarised in the table 2.1.

Monitoring Factor	Methodology	Explanation
Measurement	Spirometry; Impulse oscillatory system(IOS); Forced oscillatory techniques(FOT)	Objective data show that lung function has to be monitored regularly. Lung function can be evaluated using measurements of different technologies, given at Methodology on the left side. For example, FEV1 from spirometry or the respiratory impedance from IOS and FOT .
Functional capacity	Timed walking test	This is a measurement of oxygenation from an arterial blood gas sample [3], which proves whether supplemental oxygen would work when COPD patients are in severe resting hypoxemia.
Symptoms	Questionnaires like the COPD Assessment Test(<i>CAT</i> TM)	Continuous recording of symptoms from COPD patients are more valuable compared to a single measurement.
Exacerbation	Document recorded	All exacerbation happening has to be documented, including the frequency, severity, causes, hospitalisation, etc
Imaging	Medical imaging techniques	Delivered when the condition of COPD patients worsens.
Smoking status	Document recorded	The current status of smoking shall be documented at every visit, proper action related needs to be followed up.

Table 2.1: Monitoring for COPD patients

Accurate measures for lung function are the most valuable approach, indicating the progression of COPD patients' condition. It is also the field where this portable device is applied. A detailed comparison of different techniques will be given in the section2.2.

2.2 Technique

2.2.1 Spirometry

Spirometry is the most common technology as a lung function test, directly measuring the air during a patient's inspiration and expiration, which can be simply performed by individuals after training. Inherent characteristics of spirometry, like noninvasive, sensitive to early change and reproducible [5], make it dominate the lung measurement market. Results delivered by the spirometry are capable to detect the progression of lung diseases, quantify lung damnification, monitor the impacts of the rehabilitation environment and medication and carrying forward the next step in treatments.

Measures

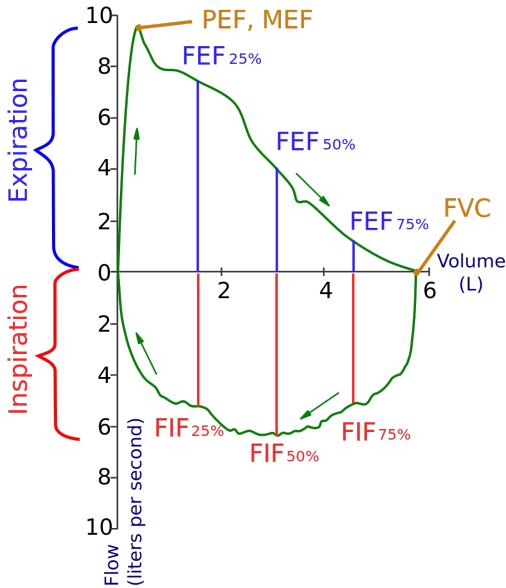


Figure 2.3: Spirometry diagram [6]

Measures of spirometry are labelled in the figure 2.3 and listed as follows[7] [5]:

- **FEV₁** , known as Forced Expiratory Volume, calculates the amount of air that a person can force out of their lungs in 1 second.
- **FVC**, known as Forced Vital Capacity, calculates the maximum amount of air that can be blown out by patients as fast as possible.
- **IVC** , known as Inspiratory Vital Capacity, calculates the maximum amount of air that can be breathed in after a full expiration.
- **PEF**, known as Peak Expiratory Flow, give the peak flow measured during an expiration process with a steady rate.
- **FIF%/FEF%** , known as Forced Inspiratory/ Expiratory Flow with the breathing rates at 25%, 50% and 75%.

Spirometry in COPD

Research [8] shows that obstructive pattern presents a smaller **FEV1** and thus **FVC** reduces accordingly as shown in figure 2.4a. **FEF_{75%}** is crucial to determine as it is a representative of the small airways. COPD normally presents an identifiable flow loop pattern compared to this health condition and restrictive diseases as shown in figure 2.4b.

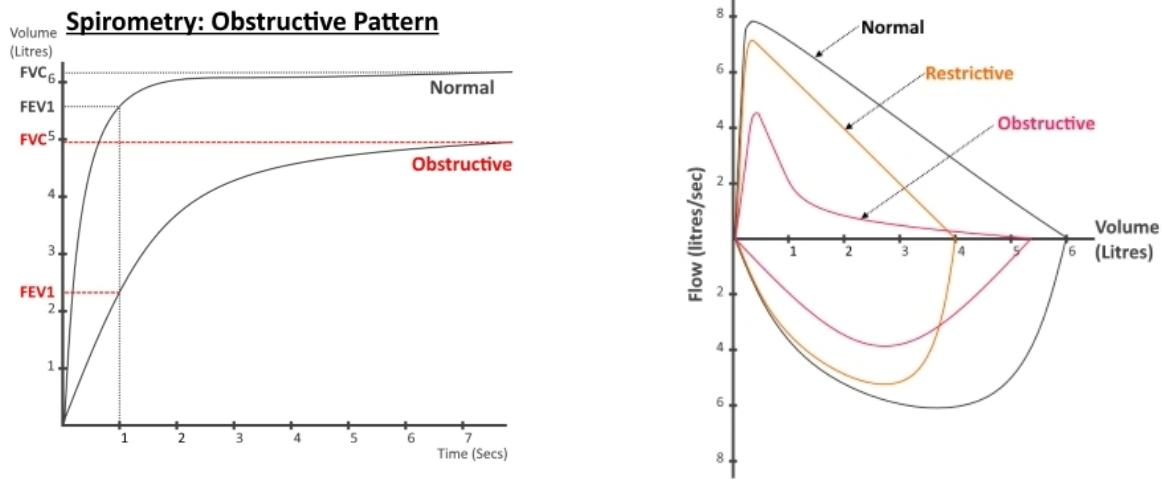


Figure 2.4: The flow pattern of COPD patients in spirometry [8]

However, the forced expiratory and inspiratory process from spirometry requires massive cooperation from patients, which makes it hard to be operated among specific groups like the children, the elderly and patients with physical and cognitive limitations. More advanced technology like FOT and IOS will be introduced later, which can provide a higher sensitivity as well as an easier interpretation.

2.2.2 FOT/IOS

The Forced Oscillation Technique(FOT) is another highly-accurate technology to evaluate lung function.

Principle

FOT was firstly described as a lung measurement in 1956 by DuBois et al. [9]. A basic FOT device shall be equipped with an actuator, several sensors and a micro-control system. The actuating impulses of single frequencies are generated by the actuator on the FOT device, injected through the patient's mouthpiece and travel through their large and small airways, which can be regarded as they are superimposed on the patient's tidal breathing as shown in the figure 2.5. The respiratory impedance, containing the respiratory resistance(R_{rs}) and the respiratory reactance(X_{rs}) over a range of frequencies (5-30Hz), is computed according to the flow dynamics with the equation:

$$Z_{rs} = \frac{P}{Q} \quad (2.1)$$

Where pressure(P) and flow rate (Q) are measured through differential pressure sensors attached to the device. Details of parameters and interpretation will be further explained in the chapter3.4.

The impulses oscillation system (IOS) was developed in 1975 by Michaelson et al [10] and can be regarded

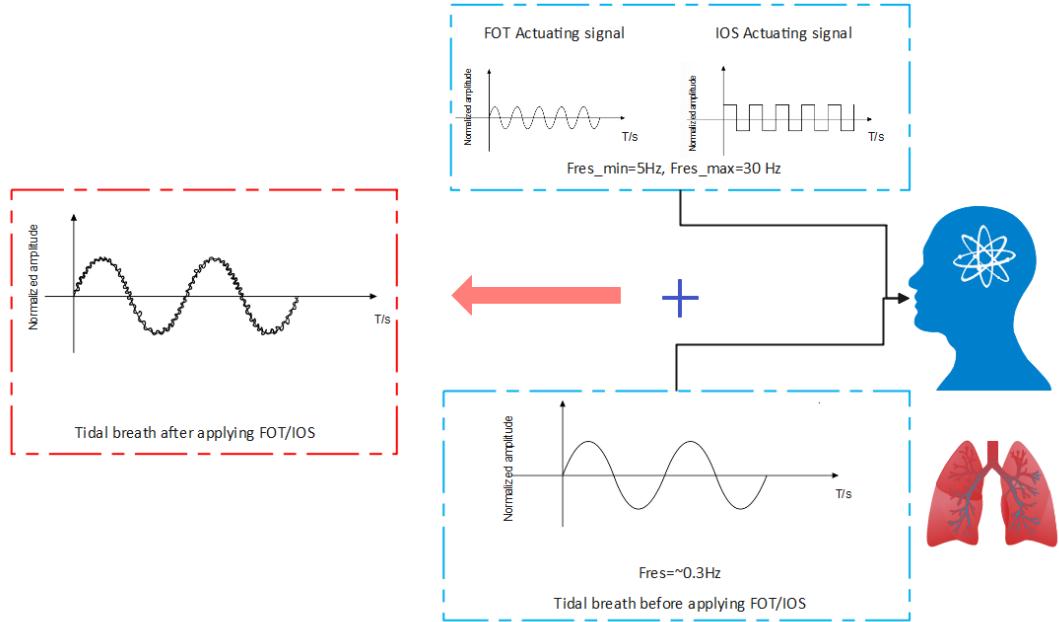


Figure 2.5: Explanation of FOT/IOS process

as an improvement of FOT. The main difference between FOT and IOS is that IOS sends out actuating waves of multiple frequencies at one time, where FOT only sends out a single signal per time, repeated by sweeping its frequency from 5 Hz and 30 Hz. Though both the FOT and IOS devices measure the respiratory impedance at multiple frequencies, similar results might not be attained due to their inherent characteristics.

This information given by FOT/IOS is completely different from spirometry measurement, which is respiratory impedance, providing better time resolution. It is highly valuable as it provides the mechanical properties of the human airways, from both central and peripheral. It gives a possibility for doctors to diagnose pathology more specifically and reliably. Besides, the procedure of manoeuvring the FOT devices is neat and easy, not like spirometry which is highly strict on co-operations from patients. Tidal breath from the patients under FOT/IOS shall provide information accurately enough for diagnosis and daily monitoring.

Careful consideration of FOT and IOS has also been made. Although multiple forced pulses of pressure waves generated by IOS devices at each time shorten the measured time, they might also bring less comfort compared to FOT devices.[10] As a portable device aiming for daily measurements, it is not reasonable to trade smaller time scales with sacrificing the comfort of patients.

Therefore, the FOT, standing out for its higher accuracy, simpler manoeuvre and greater user experience compared to spirometry and IOS, is chosen as the core technology to develop our portable device. The differences between spirometry, FOT and IOS are summarised in the table 2.2.

Table 2.2: The differences between technologies for lung measurements

Parameter	Spirometry	FOT	IOS
Measuring principle	Flow rate sensing and lung volumes are measured to diagnose lung diseases.	A single actuating signal is pushed into the patient's lung as a pressure wave to measure the respiratory impedance each time	Actuating signals at multiple frequencies are pushed into the patient's lung as pressure waves to measure the respiratory impedance each time.
Main measures	(Volume) FEV1,FVC (Flow rate) PEFR, FEF	Zrs, Rrs, Xrs, Fres, AX	
Patients co-operation required	○ ○ ○	○	
Breathing manoeuvre	Forced expiration	Tidal breathing	
Sensitivity - Central airways	○	○ ○ ○	
Sensitivity - Peripheral airways	○○	○ ○ ○	
Insight into mechanics	○	○ ○ ○	
Measuring time	○ ○ ○	○ ○ ○	○
User experience	○	○ ○ ○	○○

Chapter 3

Methodology

3.1 Measurement

3.1.1 Measure the Pressure

The pressure sensor converts physical signals into electrical signals, which are widely used for pressure measurements. Piezoelectric and piezoresistive sensors are two main types of the pressure sensor that are explicitly explained below

Piezoelectric sensors

Piezoelectric sensors are based on the piezoelectric effect. Piezoelectricity was found by two French scientists Jacques Curie and Pierre Curie in 1880. They discovered that compressing piezoelectric material would generate electric charges, this electricity created is known as piezoelectricity^{3.1a}. Alternatively, the piezoelectric effect is also reversible, which can be used as a transducer to deliver acoustic signals as shown in the figure 3.1b Piezoelectric materials are non-conductive, allowing the piezoelectric effect to work. They are categorized as crystals and ceramics. Special biological matter can also be used as piezoelectric sensors [11].

Piezoresistive sensors

Piezoresistive sensors are different from piezoelectric sensors, as they cannot generate electricity. Instead, the inherent resistance of piezoresistive sensors will change when mechanical pressure is applied, accordingly, changing the electrical response.[12] Piezoresistive sensors are most commonly used in pressure measurement as they can provide stable characteristics and are fairly easy to be controlled and measure.

Differential pressure sensors

Differential pressure sensors are equipped with piezoelectric or piezoresistive components. They have a different structure compared to sensors with a single port as shown in figure 3.2.

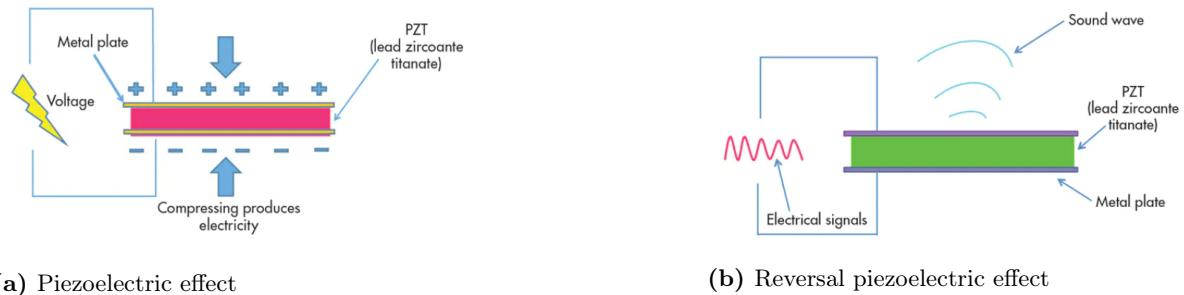


Figure 3.1: Diagram of the piezoelectric effect [11]

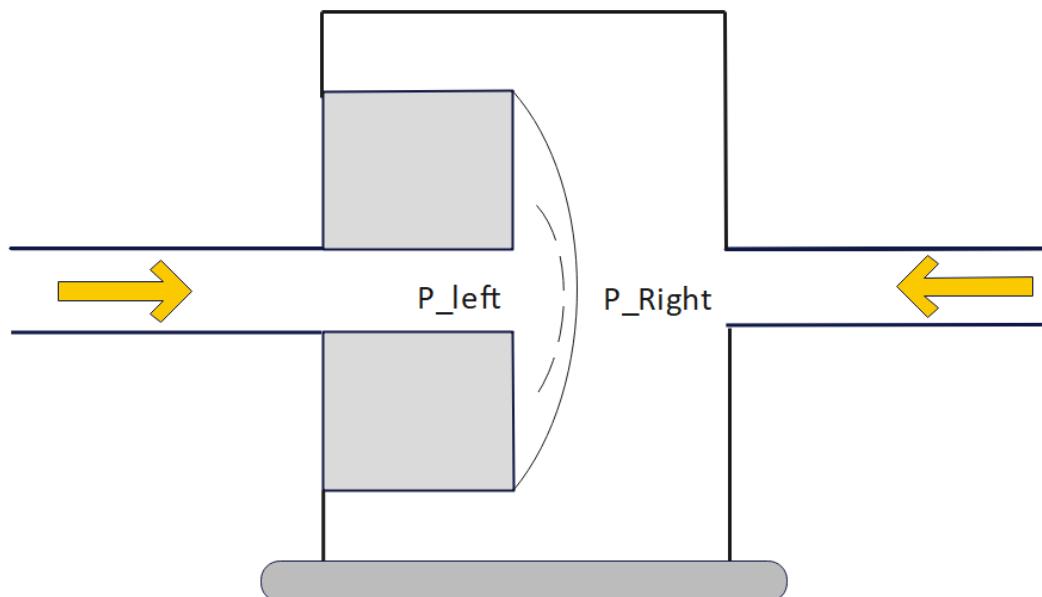


Figure 3.2: Structure of a differential pressure sensor

3.1.2 Measure the flow

Pneumotachograph can be used to measure the flow rate (Q) based on the Venturi effect. Typical categories of pneumotachograph are Fleisch type and Lilly type, which are shown in the figure 3.3 and 3.4, respectively. They both measure differential pressure between their sensing element. However, the sensor of the Lilly type is a mesh with known resistance while the Fleisch type uses parallel capillaries as its sensor. Fleisch type is the most precise flow measurement technology as there is no moving part in the whole device. Therefore, the Fleisch type is preferred and also chosen to illustrate the working principle of a pneumotachograph below.

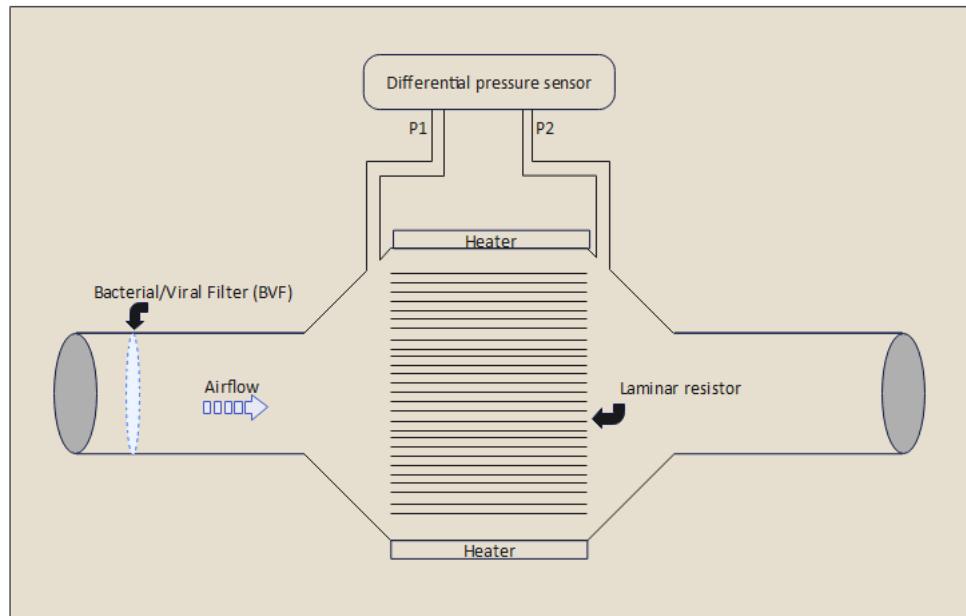


Figure 3.3: Fleisch Pneumotachograph

The Venturi effect states that there is a reduction in pressure and an increase in velocity of the fluid if it is forced through a constricted section. As shown in figure 3.3, when a patient's breath passes through the Bacterial/Viral filter, the series of small parallel tubes force it to be through the area with a narrower cross-section. The volumetric flow rate Q is determined by velocity (v) and area of the cross-section (A) as shown in the equation:

$$Q = v_1 A_1 = v_2 A_2 \quad (3.1)$$

Therefore, a higher velocity of the fluid is attained, simultaneously with a reduction of the static pressure (potential energy of the fluid), which is stated as Bernoulli's principle. It derives from the principle of conservation of energy. The increasing velocity, indicating larger kinetic energy, thus, results in a decrease

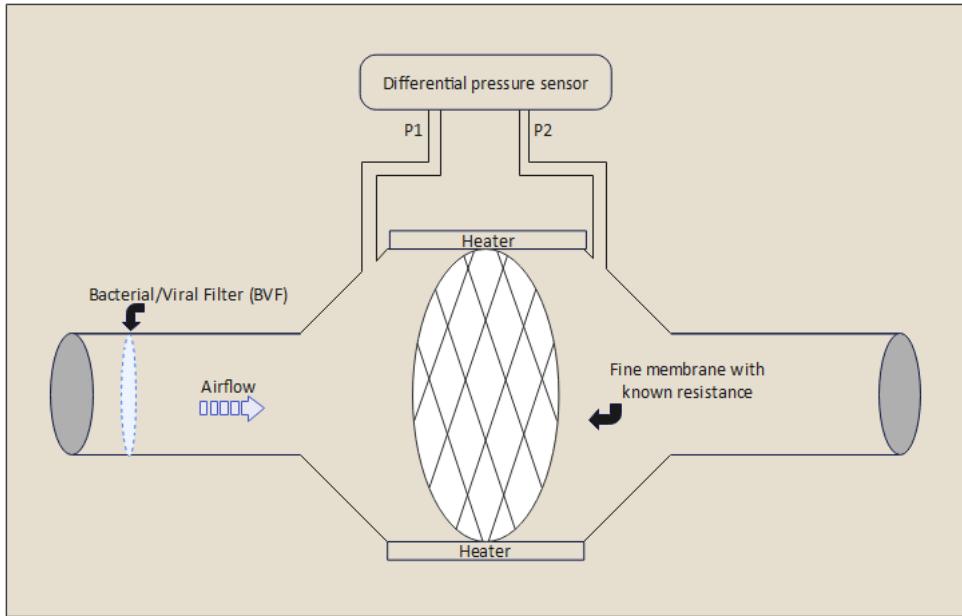


Figure 3.4: Lilly pneumotachograph

in the sum of potential energy that includes static pressure. Bernoulli's principle can be expressed as shown in the equation:

$$P_1 - P_2 = \frac{\rho}{2} \times (v_2^2 - v_1^2) \quad (3.2)$$

By combining these equations above, the volumetric flow rate(Q) can be achieved as this equation:

$$Q = A_1 \times \sqrt{\frac{2}{\rho} \times \frac{P_1 - P_2}{(\frac{A_1}{A_2})^2 - 1}} = A_2 \times \sqrt{\frac{2}{\rho} \times \frac{P_1 - P_2}{(1 - \frac{A_2}{A_1})^2}} \quad (3.3)$$

Differential pressure sensor is used to measure $P_1 - P_2$. With known A_1 and A_2 , the flow rate then can be simply calculated.

Temperature, as a crucial factor, mostly affects the accuracy of pneumotachograph. On the one hand, the volume of inhaled air increases 13% at body temperature (37 °C) compared to room temperature(20 °C) [13], which means a greater volume of exhaled air is captured by pneumotachograph than that of inhaled air. On the other hand, the temperature difference between in-body and out-of-body brings vapour condensation. The condensation of water vapour alters the resistivity of the pneumotachograph, impacting measurement. Therefore, heat elements often need to be coped with pneumotachograph.

3.2 Actuator of FOT

3.2.1 Loudspeaker

The dynamic loudspeaker is widely used as a tool to convert electrical signals to acoustic waves, which makes it possible to generate the signals that are required for FOT during respiratory measurement. But how does a loudspeaker actually work? What type of signal a loudspeaker can generate? Is that compatible with FOT requirements? What factors drive the limitation if a loudspeaker is chosen as the actuator in this case? Research related to those questions has been done and would be explicitly illustrated below.

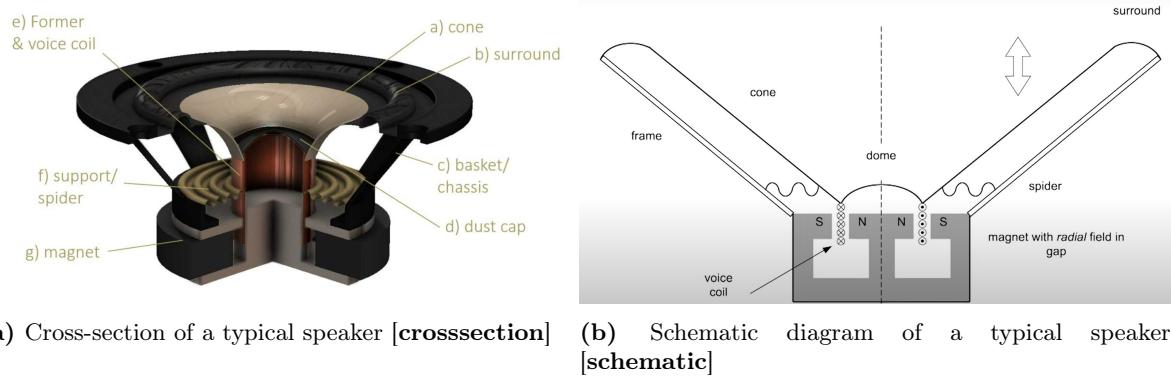


Figure 3.5: Loudspeaker composition

Generally speaking, a loudspeaker is composed of six fundamental elements, a cone, a voice coil, a permanent magnet, a dusk cap, spiders and the surround. Voice coil (e) in figure 3.5a, attaching to the cone (a), is carefully centred between the middle of the permanent magnet (g). When an electric current is applied, the magnetic field causes a force on the coil and then the cone. The current mostly oscillates in a loudspeaker, with an outgoing current on the right and an in-going current on the left. There will be an upward force on the cone according to the 'Right-Hand Rule' as shown in figure 3.5b. Similarly, when the current is reversed, a downward force is induced. Therefore, the cone oscillated, generating compression and expansion of the air outwards, which is known as sound.

As for FOT, the actuating signal is expected to be continuous, with small amplitude and frequencies sweeping from 5 Hz to 30 Hz. Loudspeaker stands out for it has a low cost, less power consumption and a relatively simple structure to be integrated. However, a speaker is still oversimplified as the acoustic actuator, providing a low accuracy and imitating in many aspects. For example, a simple loudspeaker, which has $4 \text{ to } 8 \Omega$ as its impedance, can barely generate signals at low frequency. Normally, the actuating signal at 5 Hz, which contains significant information given by the patient's respiratory system, cannot be generated properly due to the limited power capacity and low signal-to-noise ratio of a loudspeaker. Meanwhile, a loudspeaker often requires an additional sound amplifier. Extra converter circuits need to be developed for a loudspeaker actuator of FOT.

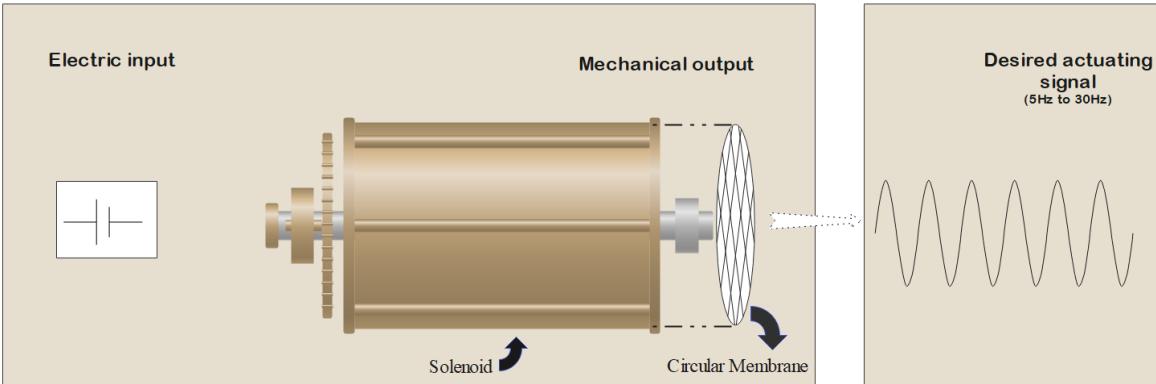


Figure 3.6: Scheme of LSA

3.2.2 Linear Solenoid Actuator

A linear solenoid actuator (**LSA**) is a device that can convert electrical signals into mechanical waves, which allows it to be used as an actuating signal generator of FOT. It is basically composed of two parts, a solenoid and a circular membrane. The overall scheme of an LSA device is shown in the figure 3.6.

Solenoid

As shown in figure 3.7, the coil of the solenoid, which is made by copper wires, generates an electromagnetic field once an electric current is supplied. According to the right-hand grip law, the south and north poles of this pole can be identified as figure 3.7 shows. It is well-known that different poles attract and the same poles repulse. Therefore, an electromagnetic force is generated, which drives the oscillation of a metal piston inside. An extra spring is also used, as it helps to relocate the piston when there is no current applied through the coil. By adding a membrane on one side of the solenoid, vibration waves then can be generated when the piston hits the membrane at a specific frequency.

The vibration of a circular membrane

The principle behind this part is similar to how a drum produces sounds. A rubber membrane is stretched and tightly tied up with the solenoid. When the piston hits the membrane, it vibrates, which makes the air particle around vibrates in time. This forms sound waves that are required. By switching the applied voltage on and off, the fundamental frequency of waves can be controlled by sending different PWM signals from the micro-controller as FOT requires. However, signals generated by LSA would be easily distorted by multiple factors like elasticity of membrane, a reflection of the tube wall and so on. Careful calibration and correction need to be done at phase 2 if LSA is chosen.

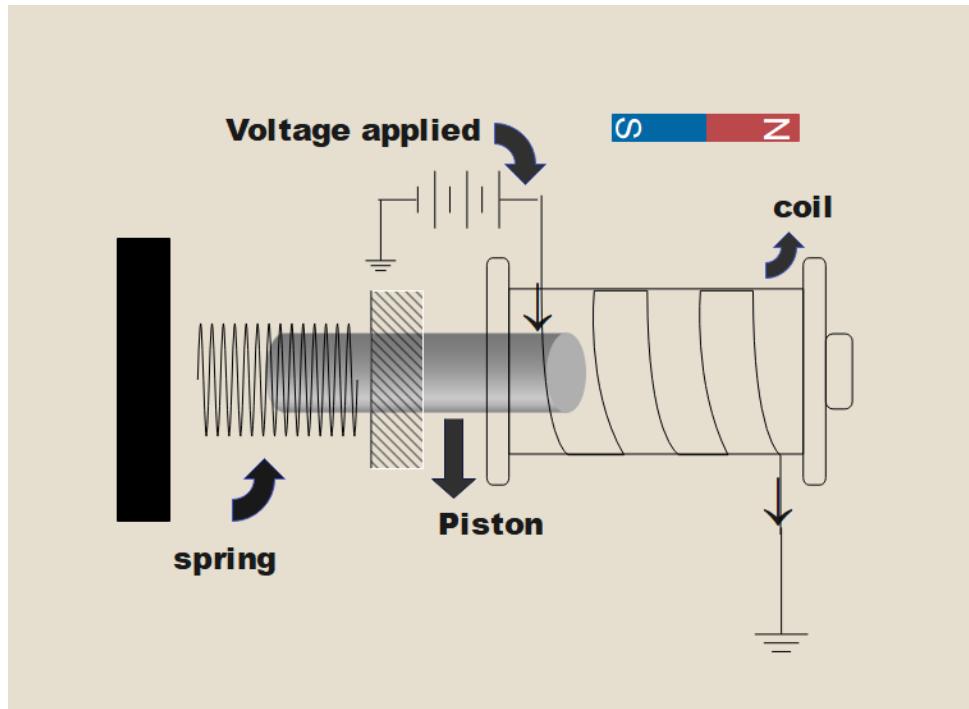


Figure 3.7: Scheme of the solenoid

The final choice for AC generator

There are many other actuators, like fans or electromagnetic actuators, that can be used, but FOT requires waves at extremely low frequency, therefore, massively limiting the final choice. LSA is chosen as the ac actuator of FOT for now due to its relatively high accuracy and flexibility in this case. However, mechanical linear actuator like LSA is hard to control. The amplitude of this small actuating signal cannot be controlled by changing the voltage applied through a micro-controller, which is mainly an inherent characteristic carried by the solenoid itself. There is also little previous work using LSA to generate signals at varying frequencies, limited reference makes this choice unpredictable. But detailed illustration will still be given if the choice of AC generator is changed in phase two.

3.3 Impedance Computation

Frequency domain analysis is used for respiratory impedance computation operated in the microprocessor.

Based on flow dynamics, with an oscillating pressure $P(t)$ given and correlated flow rate $v(t)$ measured, respiratory impedance can be calculated by the equation

$$Z_{rs} = \frac{P(t)}{Q(t)} \quad (3.4)$$

The respiratory impedance is always a complex value. It could also be expressed as an equation

$$Z_{rs} = R_{rs} + jX_{rs} \quad (3.5)$$

R_{rs} indicates that pressure and flow rate are in phase, reflecting viscous resistance of large to small airway tissue, which is defined as respiratory resistance. On the contrary, respiratory reactance X_{rs} is obtained when pressure and flow rate are out of phase. In this case, it represents the elastic resistance caused by small lung airway tissues like bronchi and the inertial resistance induced by human chest wall tissues.

Both pressure and flow rate are time-varying functions, the Fourier transform needs to be applied if respiratory impedance at specific frequencies ($f = \frac{\omega}{2\pi}$) would like to be achieved as shown in the equation:

$$Z(\omega) = \frac{P(\omega)}{Q(\omega)} \quad (3.6)$$

Respiratory impedance contains useful information at frequencies sweeping from 5 Hz to 30 Hz can then be simply attained, which should also be calibrated as impedance-frequency curves for purpose of diagnosis and further detection.

3.4 Interpretation

3.4.1 Parameters

Impedance response of FOT in different subjects has been summarised from a previous study [14] and shown in the figure 3.8 With various combinations of different parameters, doctors can determine the location and nature of the pathology. Key parameters as respiratory resistance R (in $\text{cm H}_2\text{O.L}^{-1}\cdot\text{s}^{-1}$), respiratory reactance X (in $\text{cm H}_2\text{O.L}^{-1}\cdot\text{s}^{-1}$), Resonant frequency F_{res} (in Hz) and Reactance area AX , are given details below.

Respiratory Resistance(R)

As shown in the equation3.5, respiratory resistance R_{rs} is a part of respiratory impedance, indicating viscous resistance of tissues of different sizes. Specifically, R_5 is a representative of the total airway resistance, while $R_{19/20}$ shows the large/central airway resistance.[14] The difference between R_5 and $R_{19/20}$ is usually massive among children but negligible for adults, this is because the lung tissues of an adult are more robust and developed whereas peripheral tissues play a more significant role of a kid. For COPD patients with peripheral airway obstruction, $R_5 - R_{19/20}$, reflecting resistance in the small airways, is frequency-dependent and higher than in normal subjects as shown in figure 3.8. In large airway obstruction, both R_5 and $R_{19/20}$ increase, therefore, the respiratory resistance is independent of actuating frequency. Restrictive lung diseases bring no effect on the change in respiratory resistance.

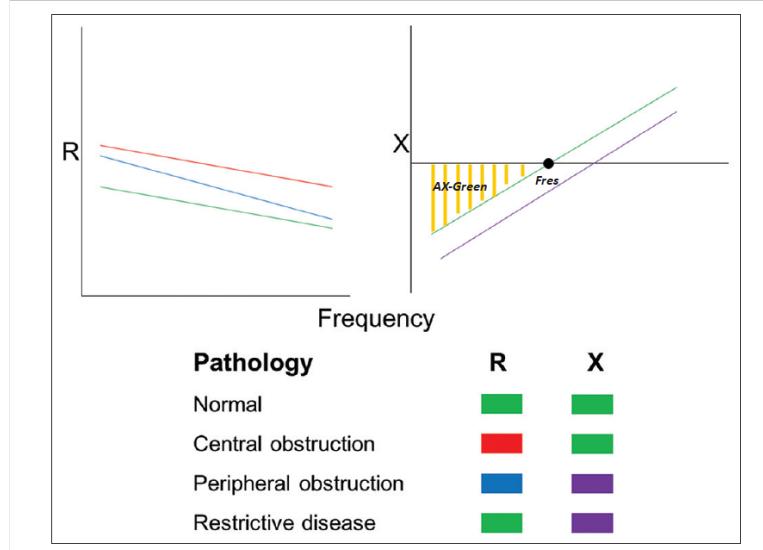


Figure 3.8: Impedance response of FOT in health and disease [14]

Respiratory Reactance(X)

Respiratory reactance X is the imaginary part of the respiratory impedance. It is a crucial parameter that indicates elastic resistance caused by small lung tissues. More specifically, X is composed of the mass-inertive forces of the moving air column shown as inertance(I) and the elasticity of the lung periphery expressed as compliance(C) [10]. Both C and I depend on the oscillation frequency. At low frequency, the capacitative pressure loss is greater than the inertive pressure loss, therefore, compliance C dominates. On the contrary, high frequency determines a smaller capacitative pressure dissipation, thus I , the inertive properties of large airways, play a more significant role. Respiratory reactance is rebounding and frequency-dependent. This would be continuously discussed later in the chapter 5.2.1 Patients with either peripheral airway obstruction(COPD) or restrictive diseases have a more negative X compared to healthy subjects. Central obstruction does not affect the response of respiratory reactance.

Resonant Frequency(F_{res})

Resonant frequency shifts from left to right as shown in the figure 3.8in both peripheral airway obstruction and restrictive disease. Large airway obstruction brings no effect on the change of F_{res} .

Reactance Area(AX)

AX , also called "Goldman Triangle" [10], is the area between 5 Hz to F_{res} where the respiratory reactance is purely negative. Both peripheral airway obstruction and restrictive disease have a bigger

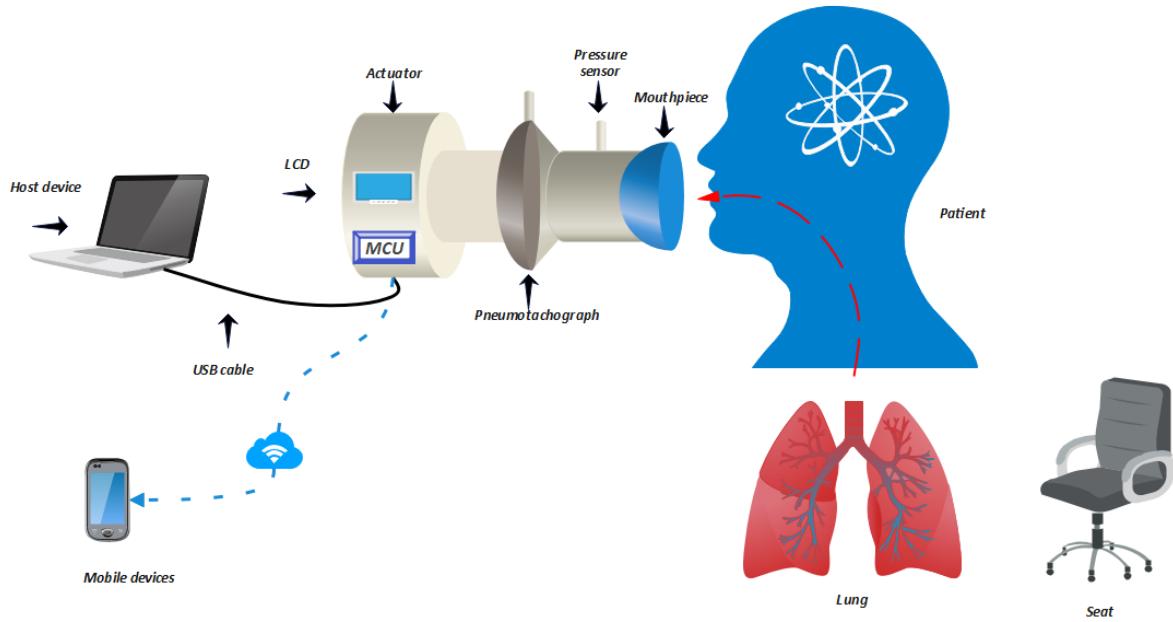


Figure 3.9: Diagram of interpretation of portable device

AX , and there is no change of AX in large airway obstruction.

Predictive values

Previous studies have published amounts of reference data of the respiratory impedance measured by FOT. [14] R and X always varies from different ethnicity, determined by factors like gender, age(A), height(H/cm) and weight(W/kg). The normalised value of the respiratory resistance and reactance can be estimated by equations[14] 3.7:

$$\begin{aligned}
 R_{men} &= -0.2454H + 0.001564W - 0.00055A + 0.5919 \\
 X_{men} &= 0.1479H - 0.000402W - 0.00022A - 0.1721 \\
 R_{women} &= -0.4300H + 0.00165W - 0.00070A + 0.9312 \\
 X_{women} &= 0.2487H - 0.001700W - 0.00053A - 0.2158
 \end{aligned} \tag{3.7}$$

specific subject was chosen for preliminary simulation and detailed reference values will be given at chapter 5.2.2.

3.4.2 Operation

Sets of operating criteria have been set during FOT measurement with this portable device as shown in table 3.1 3.2. The basic configuration is set up for the overall process shown in the figure 3.9and would

Interface	Criterion
Patient-self	Sit still at a proper height; neck extended, spine straight; nose clipped on; cheek held firmly.
Patient-device	Mouthpiece shall be attached without gaps; tidal breathing without any effort is required; no speaking, coughing, swallowing, or tough obstruction during measurement.

Table 3.1: Position requirements for the patient

Step	Procedure
1	Connect the device with PC through USB cable, LED is on showing that the device can be powered properly
2	Confirm the patient's position, push down the button once everything is ready, and the measurement is turned on.
3	Patients shall breathe normally and wait for 38-40 seconds while actuating signal is delivered every 20 cycles sweeping from 5 Hz to 30 Hz and the corresponding impedance is computed and stored.
4	After 40 seconds, the LED starts blinking, indicating that a single measurement has been done. Verify the validity of collected data.
5	If data is not valid, repeat procedures 2-4. If the data is valid, push and hold the button for 3 seconds, and turn off the device.
6	Disconnect device and host device, abandon disposable mouthpiece.

Table 3.2: Procedures of FOT measurement device

be explicitly listed below:

- **Record personal information of the patient who is confirmed with the capability of using the FOT measurement device.** Note that patients with cognitive limitations shall not use this device; children and the elderly shall not use this device without supervision; patients during their sleep shall not passively use this device.[14]
- **Check the availability and calibration of the portable device.** The portable device should be powered by the host device through a USB cable. The default LED is on, showing the device is properly powered.
- **Confirm the position of the patient according to the criterion set in the table3.1.**
- **Perform FOT procedure according to the criterion set in the table 3.2.**
- **Confirm the validity of data according to the criterion set in figure 3.8**
- **Finish measurement.**

Chapter 4

Preliminary Design

4.1 Functional Block diagram

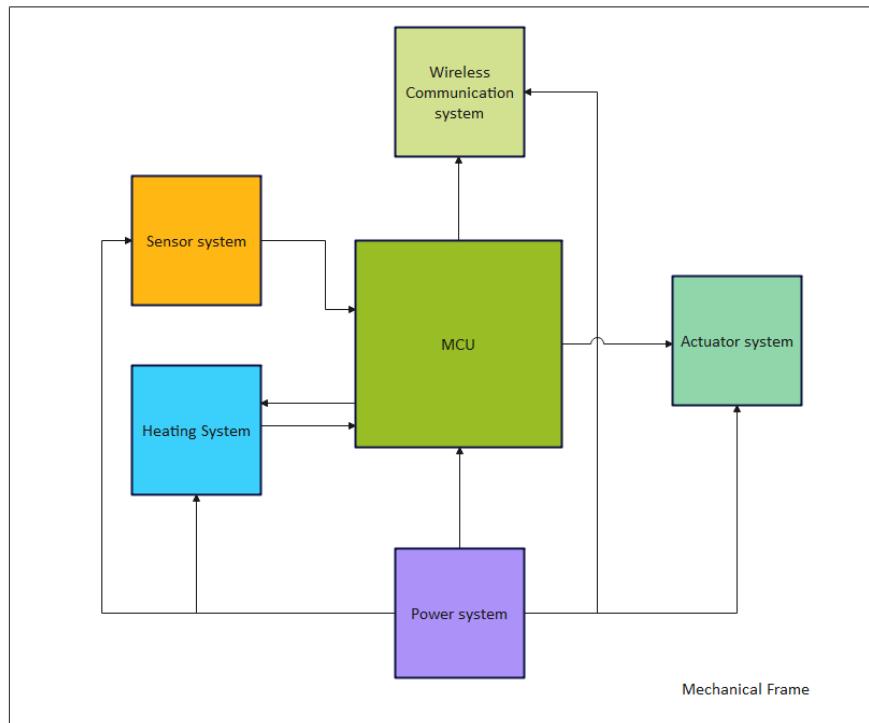


Figure 4.1: Functional block diagram

A functional block diagram has been developed based on previous research, embedded with different functional systems, which is shown in the figure4.1. Detailed explanation for each system would be explicitly illustrated in subsections, including microcontroller unit, sensor system, actuator system, heating system, wireless communication system and power system.

MCU

MCU, known as the microcontroller unit, is an integrated circuit which acts as a computer. It is the centre where all information gets gathered, computed and sent. The information input of MCU is mainly from the sensor system. Once data from sensors, pressure and flow rate of patient's tidal breath, in this case, gets collected, they would be processed and calculated in MCU. If a mismatch is detected, MCU would send instructions to the heating system and actuator system, correcting through whole negative feedback. Finally, MCU sends the final evaluated results through the wireless communication system. Respiratory impedance measured for COPD patients could be checked by either patient themselves or doctors on their mobile devices.

Sensor system

There are two types of sensors that would be used, differential pressure sensor and temperature sensor. Differential sensors are used for measuring both the pressure and flow rate of a patient's tidal breath. The information they captured at different frequencies, varying from 5Hz to 30Hz, would be sent to MCU for computing the respiratory impedance. The temperature sensor is used for monitoring the temperature at a pneumotachograph. The great temperature difference causes water vapour condensation, increasing inherent resistance inside of the device, which massively affects the accuracy of measurement through pneumotachograph. Therefore, this temperature sensor will be a monitor to control temperature differences with the cooperation of heating elements.

Actuator system

The actuator system contains a series of components acting as an AC generator. LSA is initially chosen. The natural frequency of solenoids will be controlled by MCU. MCU chooses to switch ON and OFF of the solenoid, which is able to provide actuating signals at required frequencies.

Heating system

Heating elements would be used to control the pneumotachograph within a reasonable temperature range. It doesn't have to be big, as errors mainly happen in the flow rate detection region. Small heating mats with a relatively small power input are considered to be used for both their availability and flexibility.

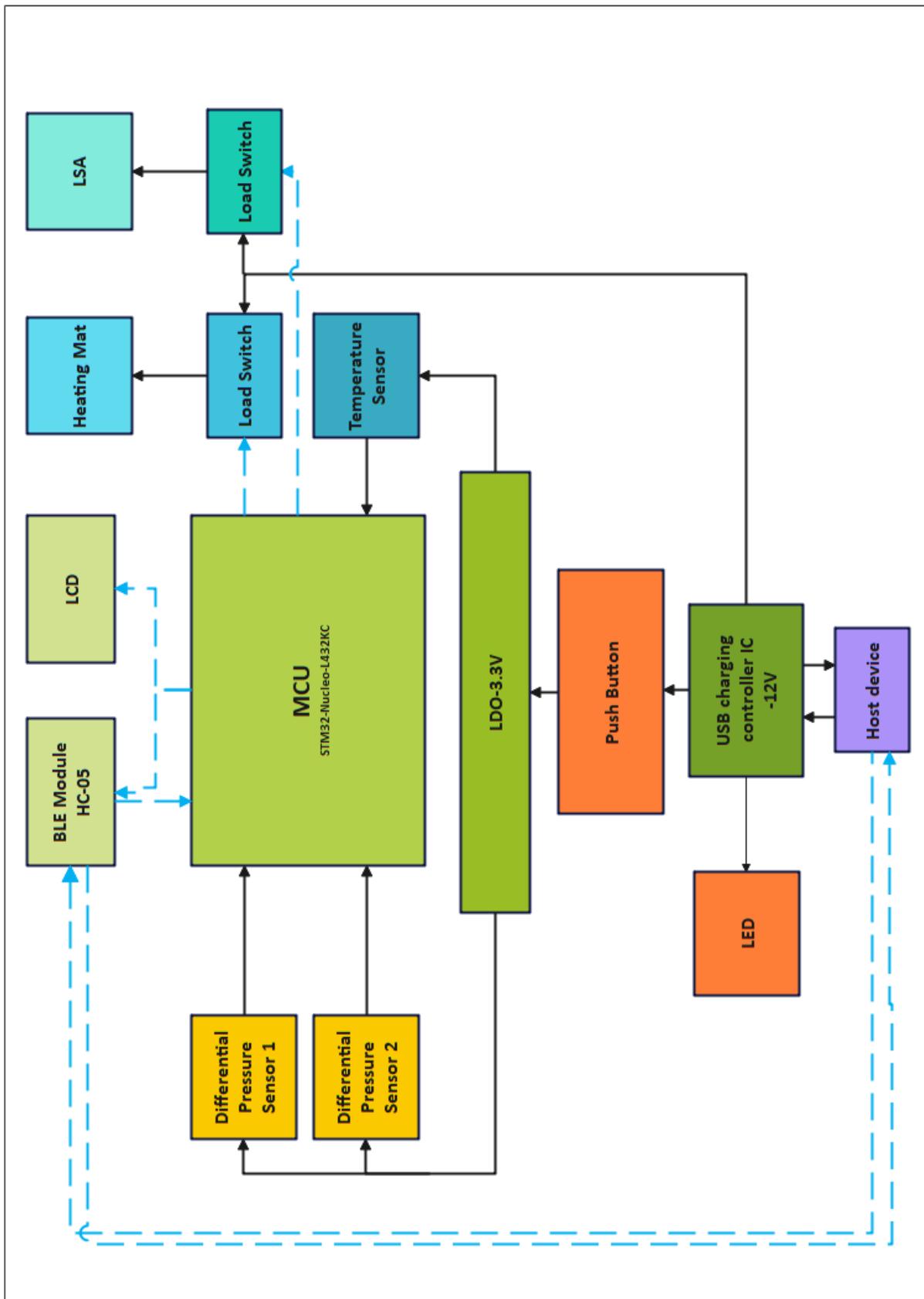


Figure 4.2: Electronics Architecture

Wireless communication system

A wireless communication system would be embedded in this portable device, aiming for delivering data from MCU to mobile devices, so that patients and doctors can check them without limitations of time and locations.

Power system

As a portable device, a flexible and relatively small-size power system is required. Therefore, a battery pack or USB charger is the prime choice. The whole power system is equipped with different converting circuits in order to provide power to different functional blocks. Charging protection IC would be alternatively set to deal with low-drive or over-drive scenarios.

4.2 Electronics Architecture

Both mechanical design and electronics design have to be done for this respiratory measuring system. However, electronics architecture is carefully considered in phase 1, as shown in the figure4.2, whereas mechanical design will be delivered in phase 2 after some practical work getting finished.

USB charging controller IC

Multiple voltages are required for different functional blocks of this system. Most electronics components have a low voltage input, normally within the range of 3.3v - 5v. However, heating elements and the mechanical actuator require a relatively high voltage input that is around 12V. General lithium battery packs supply voltage only up to 7.4V. This means an extra boost converter needs to be added, sacrificing current to boost voltage, which brings insatiability and also a limitation for operating heating and mechanical elements. The USB charging protocol is preferred in this case for its low cost, constant power supply, and flexibility in customizing the input voltage. A STUSB4500 reference design board with part number EVAL-SCS001V1[15] would be used as shown in the figure4.3. It can sink power up to 100w(20V,5A). Therefore, power just needs to be sunk by **LDO** instead of using unstable boost converters.

LED

LED is a sign. When LED is on, it shows that the power system of the device is on and works properly. On the contrary, the off light indicates that the device has not been powered yet and is not ready to work.

Push button control

Push-button IC is a mechanical interface, that allows patients or doctors to choose the timing to start actuating measurement procedures. Besides, this methodology can also reduce power consumption when

NO.	Component	Key parameters	Cost/£/each	Amount
1	USB power sink IC	Up to 20V, 5A	33.37	1
2	Microcontroller	3.3V input- STM32-Nucleo-L432KC	11.09	1
3	BLE Module	3.3V input- HC-05	4.00	1
4	Temperature sensor	3V3 input-(-10°C- 125°C)-19.5mV/°C	1.11	1
5	Silicon heat mat	12V input-2w	17.68	2
6	Linear drop voltage regulator	Vout=3V3	3.10	2
7	Boost Converter	7V4 to 12V-1.6A	2.20	1
8	Differential Pressure sensor	3V3 input-(6.89kPa 6.89kPa-)	21.68	2
9	Load Switch	12V input	1.02	2
10	Linear solenoid actuator	12V input	14.01	1
11	Dentinal rubber	-	10.58	1
12	Push Button control IC	LTC2955CTS8-1-TRMPBF	5.07	1
13	LED	-	0.3	1

Estimated Total Cost: £174.3

Table 4.1: Resources required for preliminary design

the device is on 'Stand-by' as top modules would not be powered if the push button is off. To start the measurement, simply push the button down once. To end measurement, push and hold the button for three seconds.

LDO-Low dropout regulator

LDO is often used to convert high DC voltage to low DC voltage. It is crucial that use LDO as a connector to different power subsystems for a system design. In this case, the requirements for LDO are :

1. Minimum input voltage shall be 12V.
2. Output voltage shall reasonably power MCU at 3.3V.
3. Output voltage shall reasonably power Sensors within the range from 1.8v to 5.5V.

LM1084[16] is chosen which has a wide range of input voltage with 25V as maximum, besides, it also has



EVAL-SCS001V1 Reference Design

Figure 4.3: USB Charge IC Design Board

a large current allowance and adjustable voltage output, bringing high flexibility to following tasks.

MCU-Micro Controller Unit

Desired MCU for this respiratory measuring system should be capable to compute, process and send out data. An STM32 Nucleo-32 board with part number L432KC[17] is chosen. It is equipped with 32 GPIO pins and also DACs embedded, which means analogue data from the sensor system can be directly sent into MCU without any other converter circuit. Besides, STM32 has abundant COM resources, supporting this MCU being easily merged with other functional ICs from the STM32 family, such as the BLE module and USB PD controller IC. STM32CubeIDE is an integrated development environment for STM32s that can be used to program L432KC, which is also free and graphic.

Differential pressure sensor

The differential pressure sensor will be used to measure both the pressure and flow rate of a human's tidal breathing. Therefore, the operating pressure of it shall be relatively small to precisely capture the tidal breathing that is within $\pm 10\text{cmHg}$ [18]. What's more, both inhaled and exhaled airflow need to be measured, thus the differential pressure shall bidirectionally work. Finally, the voltage supply of the differential pressure sensor shall be able to cope with the output of LDO, neither too high nor too low. Above all, a sensor from Honeywell Sensing with part number ABP2DRRT001[19] is chosen, providing

$\pm 0.25\%$ accuracy under operating pressure from -6.89kPa to 6.89kPa (-5.17cmHg to 5.17cmHg). It can also be powered from regulated LDO output at 3.3V.

BLE Module

BLE module is used for wireless communication, sending out computed respiratory impedance at different frequencies to a mobile device. HC-05 Bluetooth Serial Transceiver[20] is chosen as the master Bluetooth device that can be simply connected with STM32 MCU, interfacing with other slave Bluetooth devices.

Load switch

Load switch is a smart switch IC, containing multiple MOSFETs. The difference between a load switch and a traditional MOSFET circuit is mainly that a load switch is carefully integrated and packaged. It also provides an adjustable soft start and other advanced protection characteristics. However, the stock of load switches in the current market is not stable. A small PCB circuit of MOSFETs can still be an alternative choice.

LSA-Linear solenoid actuator

An LSA equipped with a spring is preferred. Solenoid only has two states, either fully on or off. A spring can force the piston of a solenoid back to the original state, which allows the solenoid to be controlled by PWM signals sent by MCU. Besides, FOT also requires actuating signals with a small amplitude. Therefore, A LSA, operating under low voltage at 12V DC and providing a relatively low force, is chosen.

Temperature sensor

A temperature sensor is used to monitor the temperature at the pneumotachograph. It needs to detect temperature changes between 20°C-37°C and keep the temperature maintained at 35°C inside of the device with the cooperation of MCU and heating mats.

Heating Mat

Heating mats are designed to be attached at the entrance of the pneumotachograph, controlled by MCU, maintaining the temperature at 35°C. Small heating mats can be easily assembled and also requires a small amount of input power, which is more ideal to be used here compared to heating cables.

All resources that have been chosen to be used are listed in table 4.1. Note that they are only for initial design, specific components may be replaced during or after practical work in phase two.

Chapter 5

Preliminary Simulation- The aRIC Model

Practical work is not required in the Phase one stage. However, simulations that can verify the reliability and stability of the preliminary design have been developed, giving more references and confidence to the next series of work. The augment RIC model of this biosensor system will be explained explicitly below in particular.

5.1 Introduction and background

5.1.1 Why is the RIC model built?

Nowadays, FOT has become increasingly popular for measuring respiratory impedance, which not only requires fewer efforts from patients but also brings higher accuracy to acquire data from the low-frequency areas during the respiratory process. By applying an external actuating signal during the patient's breathing, the responding airflow is measured, giving results as the frequency-dependent impedance curves, which could demonstrate reversible airflow obstruction reliably and appropriately.

Those frequency-dependent impedance curves are achieved by simply measuring pressure (P) and flow (Q) next to the mouthpiece of patients during their respiratory process and computed according to to flow dynamics in a vessel:

$$R = \frac{P}{Q} \quad (5.1)$$

This is exactly similar to Ohm's law that resistance can be determined by its crossing voltage (V) and the current flowing through (I):

$$R = \frac{V}{I} \quad (5.2)$$

Impedance in respiratory system can be simply expressed by the combination of resistances (R -units of $(\text{in } \text{cm } \text{H}_2\text{O} \cdot \text{L}^{-1} \cdot \text{s}^{-1})$, or $\text{Kpa} \cdot \text{L}^{-1} \cdot \text{s}^{-1}$), inertances (I —units of $\text{cm } \text{H}_2\text{O} \cdot \text{L}^{-1} \cdot \text{s}^{-2}$ or $\text{Kpa} \cdot \text{L}^{-1} \cdot \text{s}^{-2}$) and compliance (C —units of $\text{cm } \text{H}_2\text{O}^{-1}$ or Kpa^{-1}) by building an electrical model correlated. Therefore, multiple RIC models have been built to serve as supplementary quantitative approaches to diagnosing diseases like COPD and asthma.

5.1.2 Why is the aRIC model used instead of others?

Previous studies have compared and verified 6 impedance-fitting respiratory models that were widely used, known as the basic RIC(RIC) model, extended RIC(eRIC) model, DuBois model, Mead1969 model, Mead model and augmented RIC(aRIC) model. Equivalent electrical circuits of them each are shown in the figure5.1

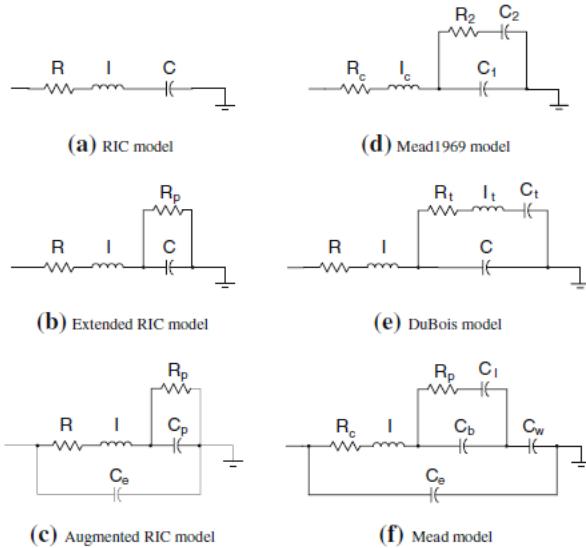


Figure 5.1: 6 Electrical models of the respiratory system [21]

In 2006, researchers at Texas Christian University did a study[21] to evaluate these 6 models of measuring children's respiratory impedance based on impulse oscillometry data gained from both normal subjects and asthmatic patients. The method they used for model parameter estimation is to compare distributed errors between ideal model data and experimental data by using the least squares(**LS**) criterion. Consequently, they found that both DuBois and Mead models performed lower mean total errors compared to the aRIC model, however, they also produced far more unrealistic estimations for compliances in the human respiratory system. On the contrary, the parameter estimation of aRIC model well conformed to their expectation in healthy subjects and patients. In 2009, they did a further study to evaluate the aRIC model, expanding their ill subjects from asthmatic children to adults with either asthma or COPD. Results[22] showed that the model was 13.1 – 66% more accurate than the eRIC model, as well as highly

realistic for estimating compliance compared to the Mead1969, DuBois and Mead models. Therefore, the aRIC model stood out for both its reliability and reality.

5.2 Method

5.2.1 Principle

The aRIC) model is widely used to effectively measure respiratory impedance and diagnose respiratory disease. But how does it work? What does each parameter mean in this model? Would it be appropriate to convert physical parameters to electrical components?

5 components are contained in the aRIC model, which is R (large airway resistance), I (large airway inertance), Rp (peripheral airway resistance), Cp (peripheral airway compliance) and Ce (extra-thoracic compliance). However, the aRIC model was not built inherently in the first place. In fact, it was developed as an improvement of previous RIC models.

The initial RIC model only had 3 components, resistance, inertance and compliance in series. It included a basic structure for the human respiratory system, however, showing an extremely low accuracy during a mount of experimental investigation which means the doctor would not be able to diagnose the lesion in detail. The eRIC model was then proposed by adding another peripheral resistance in parallel with the compliance. This allowed the frequency-dependent observation of typical real impedance data. But modulation of eRIC model was still not accurate enough because it ignored the upper airways shunt effects in reality. At this stage, the aRIC model was finally proposed. Cp, modelling the extra-thoracic compliance, was added and gave a better performance of the entire model by increasing the real part of respiratory impedance to compensate for upper airways shunt effects. The impedance of aRIC model was given by a study[21], shown as:

$$z = \frac{A(RA + Rp)}{[A(1 - \omega^2 ICe) + \omega^2 Rp^2]^2 + [\omega Ce(RA + Rp)]^2} + j \frac{\omega(IA - Rp^2 C)[A - \omega^2 Ce(IA - Rp^2 C)] - \omega Ce(RA + Rp)^2}{[A(1 - \omega^2 ICe) + (\omega^2 Rp^2 CCe)]^2 + [\omega Ce(RA + Rp)]^2} \quad (5.3)$$

Where $A = 1 + (\omega RpC)^2$.

As we discussed previously, respiratory impedance is measured and calculated in the same way as an impedance is obtained in an electrical circuit. Pressure can be regarded as electrical voltage and flow rate can be seen as the electrical current flowing through. Therefore, physical units of respiratory resistance, inertance and compliance can be converted to electrical units of resistance, inductance and capacitance correspondingly, which makes it easier for us to simulate the respiratory systems using electrical characteristics.

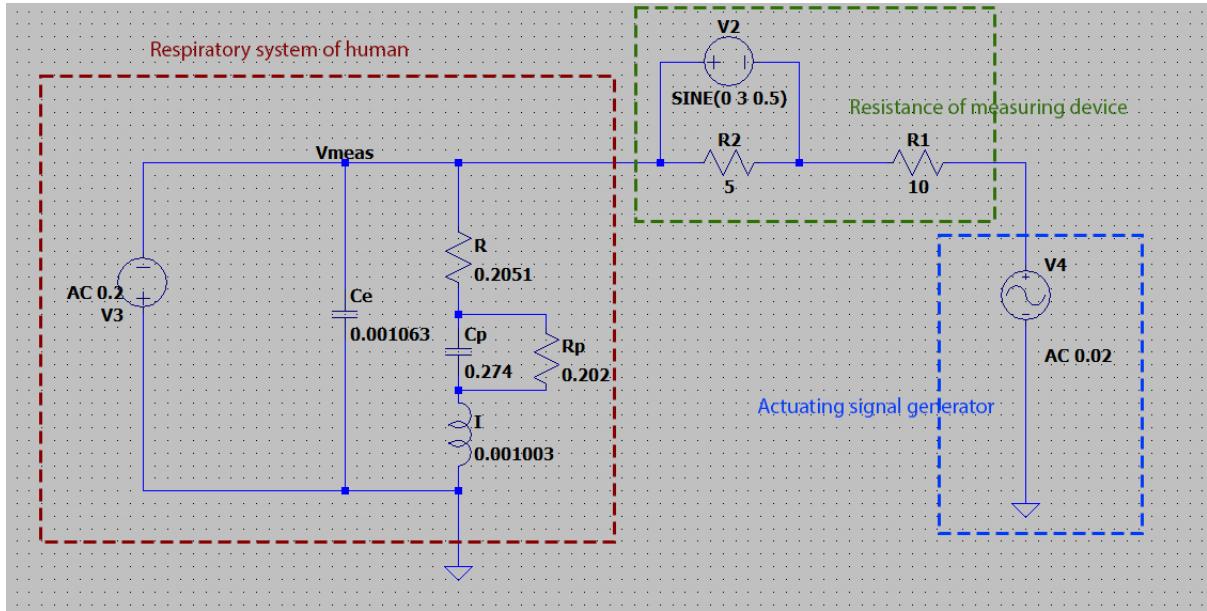


Figure 5.2: Equivalent circuit for integral biosensor system built in LTSpice

5.2.2 Strategy

LT Spice was used to run all analogue simulations in this part. An equivalent circuit with a proper structure, representing the overall biosensor system, has been built in Spice and shown in figure 5.2. All data were chosen from the previous study.[22] Two types of simulations were run, called the confirmatory experiment and the control experiment. The former one aimed to verify the availability of aRIC model, while the latter one did a further simulation to show how impedance would change by modifying 2 peripheral parameters Rp and Cp, which could be used to verify the difference in respiratory impedance between COPD patients and healthy subjects. The results of both two experiments were given as the impedance-frequency curves. For clearer demonstration, resistance and reactance were separately presented in curves.

Data Selection

According to Diona and Rajagiri's study in 2009[22], they figured out the relationship between an adult's height and the parameters of aRIC model as shown in the figure 5.3. A female with a height of 173cm as a healthy subject(A) was assumed for running simulations below. Required data have been calculated and shown in the table 5.1

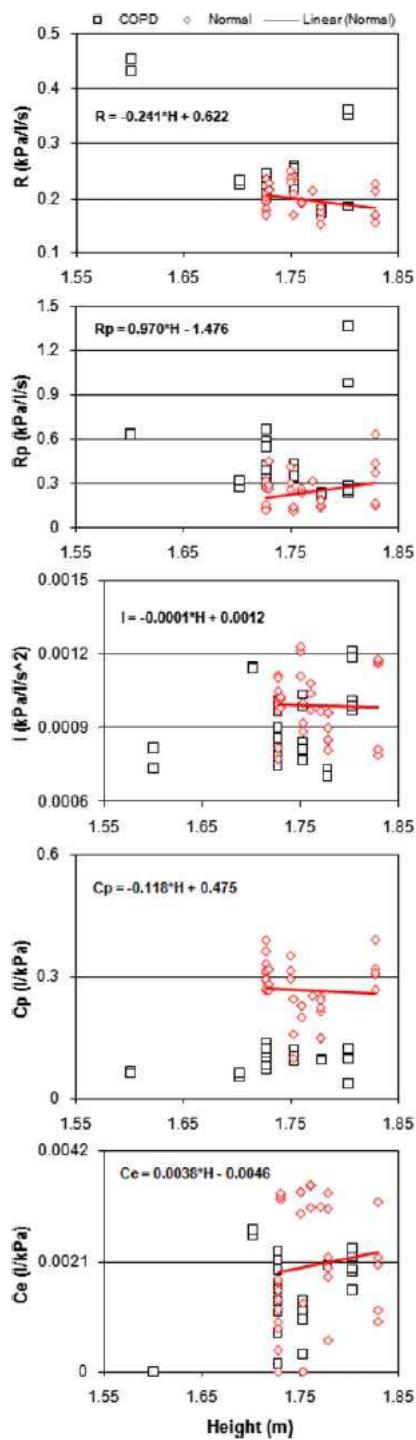


Figure 5.3: Adult's height Versus each parameter of aRIC in the previous study[22]

Simulation data of the healthy subject in aRIC model			
Electrical parameter	Physical meaning	value	unit
V1	Pressure	Amp=0.2	Kpa
R	Large airway resistance	0.2051	Kpa/l/s
I	Large airway inertance	0.00103	Kpa/l/s ²
Rp	Peripheral airway resistance	0.202	Kpa/l/s
Cp	Peripheral airway compliance	0.274	l/Kpa
Ce	Extra-thoracic compliance	0.00163	l/Kpa
V2	Flow rate measurement	Amp=3,f= 0.5	Kpa, Hz
R2	Resistance of pneumotachometer	5	Kpa/l/s
V3	Actuator	Amp=0.02	kPa
R3	Other resistance in device	0.04	Kpa/l/s

Table 5.1: Simulation data gave by a healthy female subject(A) with a height of 173cm

Confirmatory Experiment

The confirmatory experiment was used to verify the availability of the aRIC model. Theoretical data gained from the table5.1 was fitted into the model, shown in the figure5.2. AC analysis was then run for this circuit in LTSpice, sweeping frequency from 5Hz to 30Hz, which represents the actuating signal was added to the respiratory system of the subject. Finally, the respiratory impedance of this subject was automatically computed by LTSpice. By comparing the impedance-frequency curves obtained from the simulation with practical results directly measured in the previous study[22][21], we could simply conclude whether aRIC model is proper to be used for diagnosing and detecting diseases with the abnormality of patient's respiratory impedance.

The Impedance-frequency curves of subject A are shown in the figure5.6, and calibrated curves separately represented by resistance and reactance are given in the figure5.4.

Previously, a schematic illustration of impedance-frequency developed by E. OOSTVEEN ET AL study[23] is shown in the figure5.5. Researchers pointed out that FOT was often set under a middle range of frequency to effectively detect human respiratory disease. Under this range, the healthy respiratory system would show characteristics as a resistance(Rrs) that was mainly composed of the airway resistance and reactance(Xrs) either an elastic reactance or an inertia reactance dominated at each frequency. Both those two components were highly frequency-dependent.

By comparing our simulation result5.4 with the schematic5.5 from E.OOSTVEEN's study, obvious uniformity was found. With the increasing frequency, resistance was always represented as the airway resistance, while reactance turned from initially negative to positive as the inertia reactance gradually

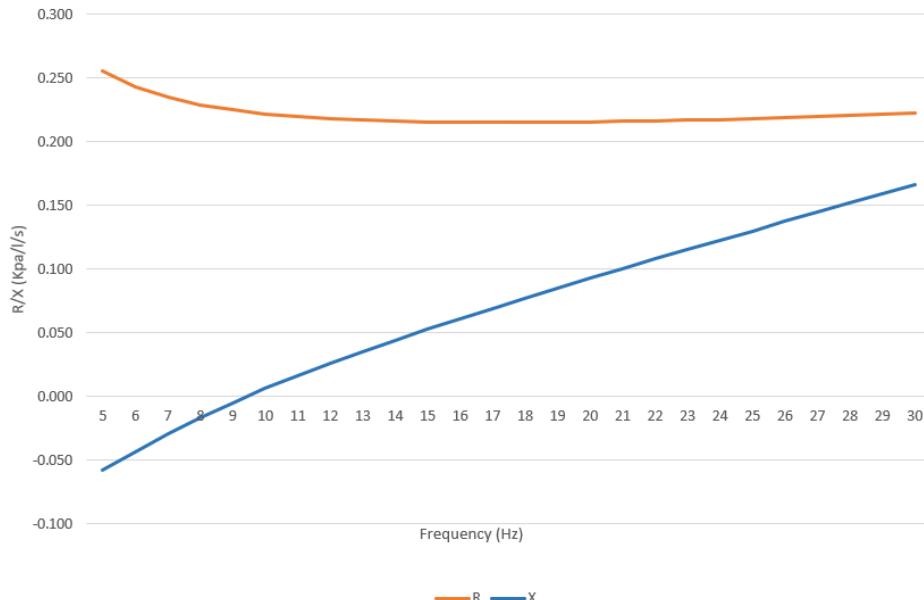


Figure 5.4: Calibrated RX plot of subject A

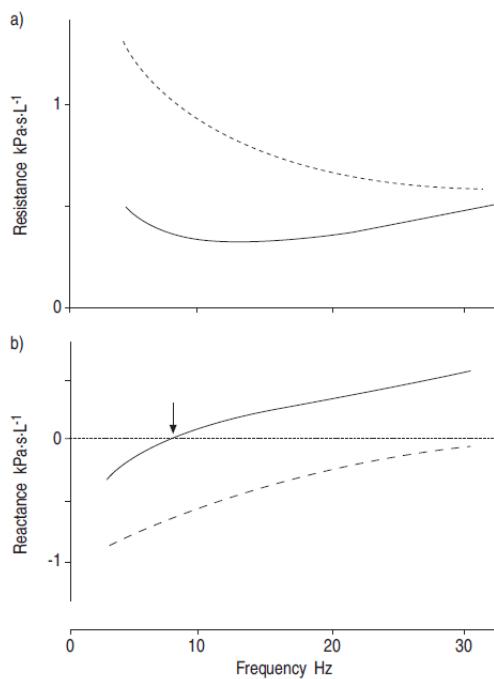


Figure 5.5: [23] study: Schematic illustration of the frequency dependence of respiratory impedance of adults in the medium frequency range, in health and disease. Compared to the normal impedance data (—), in airway obstruction, respiratory resistance (- - -) is higher (a) and negatively frequency-dependent, whereas respiratory reactance is lower (b). Arrow indicates resonant frequency.

took over the elastic reactance at higher frequencies. Arrows 5.5 showed the resonant frequency ($X_{rs}=0$) at which the elastic force equally counteracted the inertia force. Our results perfectly proved it as the corresponding resonant frequency appeared at around 9.2Hz, within a reasonable range. Therefore, the availability of this aRIC model designed for a specific biosensor system has been strongly verified.

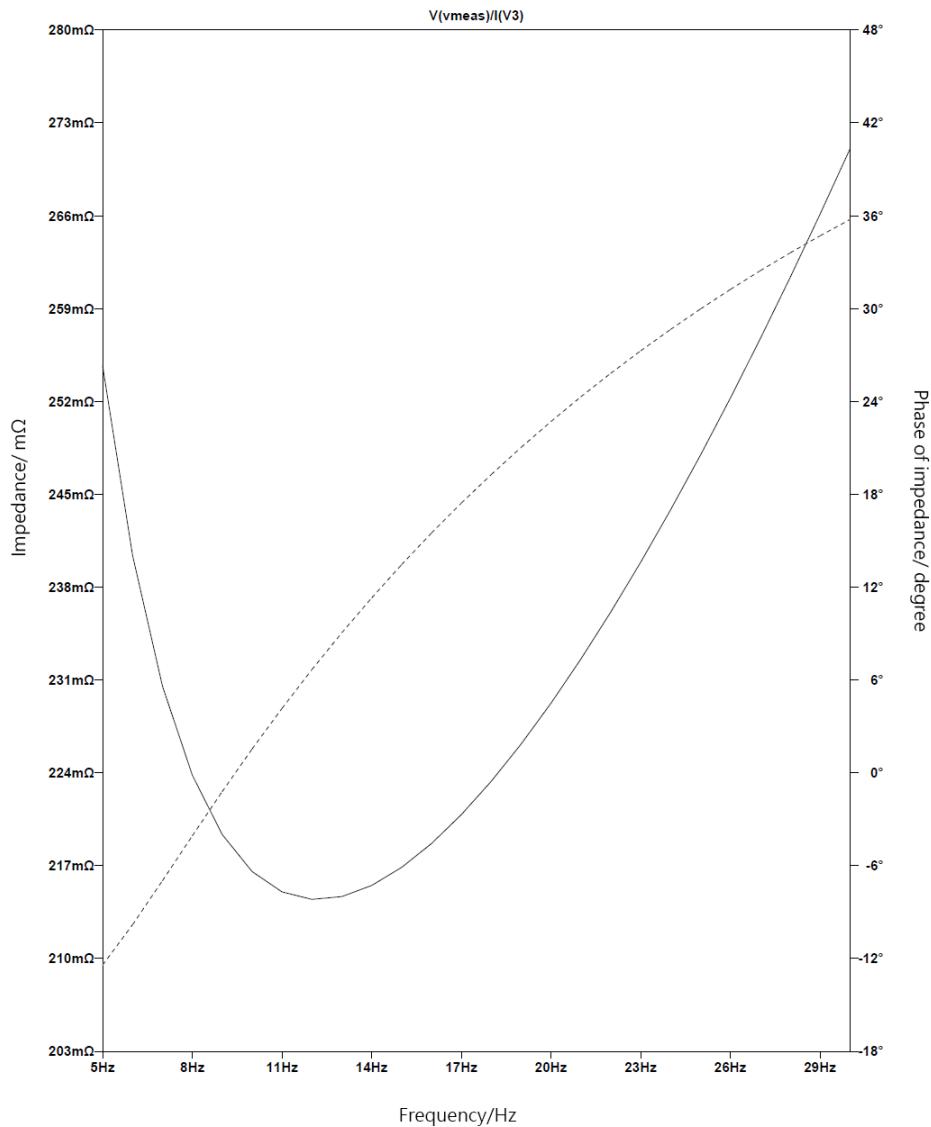


Figure 5.6: Impedance-Frequency curves for subject A

No.group	Decrease	Constant	Increase
1	0.5Rp	Rp	5R
2	0.5Cp	Cp	5Cp

Table 5.2: The setting of parameters in the control experiment

Control Experiment

In this experiment, effects induced by parameters of aRIC model were expected to be interpreted. However, instead of comparing all parameters in this model, only peripheral resistance(R_p) and compliance(C_p) were taken into account. This is because our biosensor system was designed for daily measurement and monitoring of patients who have a small airway disease, specified in this case, COPD. A control experiment was then developed. 3 groups of simulations were run for parameters R_p and C_p as shown in the table5.2.

By keeping other parameters stationary, the resonant frequency and AX region were changed accordingly when R_p or C_p was solely decreased or increased each time. $X_{rs}(5)$, representing the small airway reactance at 5HZ was specifically examined here. In the end, the difference in peripheral parameters between healthy subjects and COPD patients was figured out by comparing simulation results with practical data obtained from the previous studies[23], which would be a crucial standard for doctors to diagnose in reality.

Parameter		Resonant frequency/Hz	$X_{rs}(5)/Kpa/1/s$
R_p	C_p	9.5	-0.0580
0.5 R_p	C_p	8.1	-0.0208
5 R_p	C_p	9.8	-0.0845
R_p	0.5 C_p	11.5	-0.0718
R_p	5 C_p	6.5	-0.0225

Table 5.3: Numerical data from the control experiment

Practically, respiratory resistances would not be easily changed due to a stable physiological structure of the human respiratory system as well as the low-frequency region initially defined. Therefore, researchers and doctors were more interested in the change in respiratory reactance during FOT measurement. Practical data given by the study [23] suggested that COPD patients had a bigger resonant frequency as well as a significantly large $X_{rs}(5)$. Besides, [22] research on aRIC model clarified that they found Normal adult C_p was statistically significantly greater than C_p in adults with COPD after amounts of study tests.

Back to our simulation result, it is obvious that both resonant frequency and $X_{rs}(5)$ would be amplified as shown in the figure5.7 and table5.3 by increasing R_p and decreasing C_p . This result is consistent with known physiological knowledge that obstruction of small airways makes tissues in the respiratory tract more resistive and less elastic. Our observation here is perfectly in line with physiology and conclusions obtained from previous studies[22][21]. Doctors could make an initial diagnosis that the subject

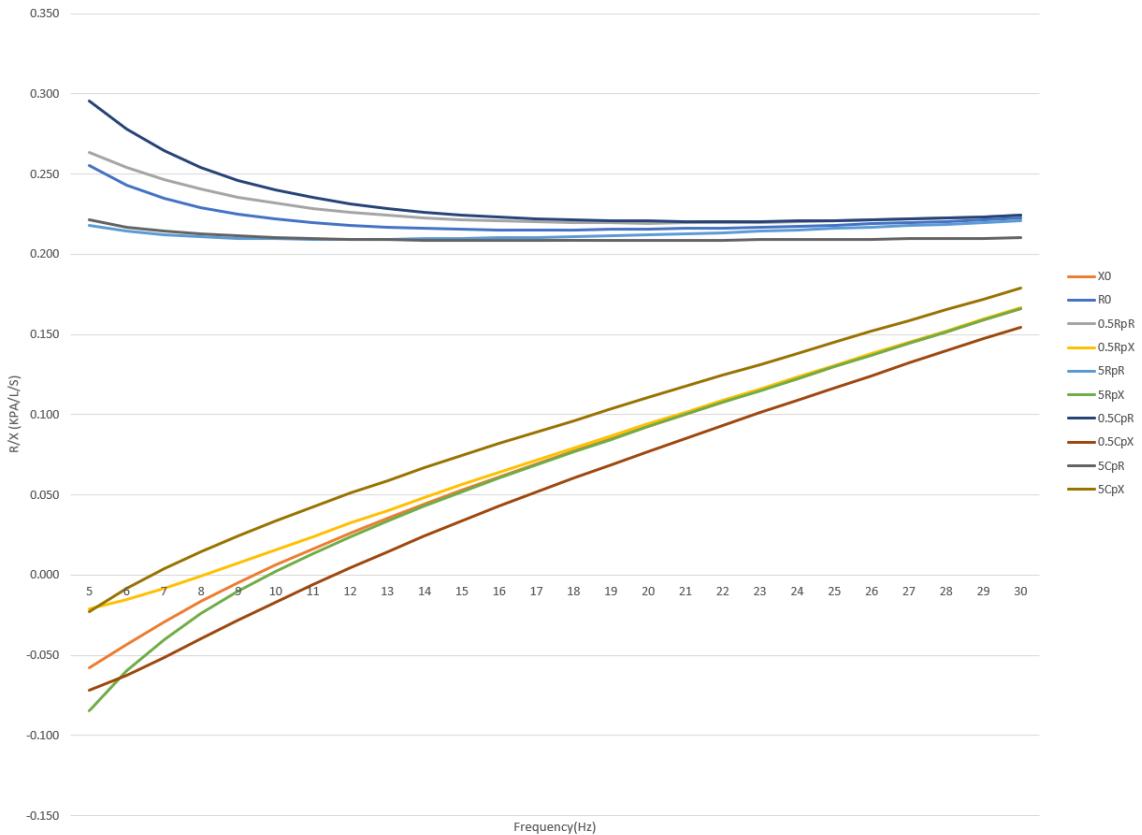


Figure 5.7: Impedance-frequency curves from the control experiment

might have COPD if the measuring result from the aRIC model non-ideally reflects a far small C_p and relatively large R_p .

5.3 Conclusion

This aRIC model simulation verified that it is reasonable to fit our biosensor system into an electrical model that could be easily simulated and quantified. Parameter changes of aRIC model are consistent with known physiological variations caused by obstructive airway diseases. However, a previous study[22] also pointed out that aRIC model was still an 'oversimplified' model as it could only detect obstructive diseases without being capable to give a diagnosis focally. For example, asthma and COPD cannot be distinguished from the results given by aRIC model. Nevertheless, it is no doubt that aRIC model is still worthy to be built as a low-cost prediction toolkit, offering reliable and appropriate diagnosis and treatment for many respiratory diseases.

Chapter 6

Summary

6.1 Summary of Phase-one

A great amount of work has been done in phase one:

- Overall principle has been understood and appreciated.
- Project has been refined into different sub-tasks, and careful consideration of functional blocks has been given.
- Preliminary design of electronics architecture has been delivered.
- The verification simulation of aRIC model has been worked through, testifying to the feasibility of measuring methodology.

6.2 Plan of Phase-two

Work for Phase two is divided into two main stages: functional design test and integration.

For the first stage, different functional blocks will be practically designed and tested on the bench. This stage is also refined into three sub-stages according to different fields: electronics, software and mechanical. The basic principle is that simulation from software shall be processed before any practical work. For instance, in electronics, LTSpice simulation will be run for each block first, and then verified schematics will be applied on a breadboard. If all functionalities can be implemented, PCB boards will be finally designed and printed out. Integration is the last stage with a great challenge, which aims to bring every functional module together. The performance of the portable device will be analysed and evaluated with the data collected at this stage. A detailed plan is summarised in the table 6.1.

6.3 Gantt chart

The Gantt chart, indicating the progression and plan of the entire MEng project, is given in the figure 6.1

Table 6.1: A detailed plan for each stage in Phase-two

Stage	Sub-stage	Task	Sub-task	Estimated time/day
		LTSPICE Simulation	Push-button IC USB charge IC Load switch IC LDO IC	7
Functional Design Test	Electronics test on the bench	Tests on the breadboard	Push-button IC USB charge IC Load switch IC LDO IC	7
		PCB fabrication	Push-button IC USB charge IC Load switch IC LDO IC	7
	Software development of MCU	Programming on STM32 CUBEIDE	Sensor data management Impedance computation BLE Module connection	3
	Mechanical test on the bench	CAD in 3D printing software	Software exploration Mechanical frame design	3
Integration	Electronics-software	Verify the overall control logic is correct		10
	Mechanical- Electronics	Verify the sensing data sensed is correct under current mechanical design		4
	Mechanical - Software	Verify all parts can be assembled as CAD output		7

Date of Phase-Two: 29/09/2022 to 01/03/2023



Figure 6.1: Gantt Chart for Peggy's MEng project

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