Estimation of tidal volume in a pediatric setting using depth camera

by

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Évaluation de la détresse respiratoire des patients aux soins intensifs en temps réel

Grâce HOUNKANRIN

RÉSUMÉ

L'insuffisance respiratoire fait partie des maladies attachées à la race humaine et ce indépendamment de l'âge. Elle peut surgir à n'importe quel moment, et peut être détectée par l'étude de mesures respiratoires telles le volume d'air inspiré et la fréquence de respiration par le clinicien. Mais l'analyse de ces paramètres est très difficile à l'œil nu spécifiquement du volume et spécialement quand il s'agit des enfants.

Dans le but de palier à ce problème rencontré lors des diagnostics, ce projet de recherche a pour objectif de mettre en place une approche plus efficiente et efficace pour l'analyse des mesures respiratoires. Principalement, la conception et développement d'un système informatisé et autonome basé sur des images 3D, ce qui permettrait d'étudier le volume d'air inspiré et expiré en se basant sur le mouvement thoraco-abdominal ainsi que l'étude de la fréquence respiratoire dans le but de détecter une insuffisance respiratoire. Une méthodologie a été proposée pour arriver à cette fin dans le département pédiatrique.

Cette méthodologie consiste en l'acquisition d'image 3D du patient pendant la respiration avec une caméra KINECT Azure positionnée au-dessus du patient, suivi d'une reconstruction de l'image obtenu puis de l'analyse de cette image en créant des graphes sinusoïdaux pour représenter les cycles respiratoires dans l'acquisition utilisant Matlab et CloudCompare. La détection des cycles respiratoires permet de calculer par soustraction le volume d'air inspiré. Les cycles respiratoires sont détectés grâce à la position de l'abdomen et du thorax à chaque instant (abaissé en expiration et remonté en inspiration).

La fiabilité et l'exactitude de la méthode ont été examiné en réalisant plusieurs tests sur des patients réels et des mannequins, comparant les résultats obtenus par la méthode aux résultats réels.

Les tests de fins d'inspirations et d'expirations ont été comparé aux résultats détectés visuellement par un médecin ce qui a donné des erreurs de 1-3 frames. Les tests ont été faits sur mannequin et sur patient réel. Un test sur patient réel a donné le frame 1 pour fin d'inspiration premier cycle, frame 27 pour second cycle et le frame 60 pour troisième cycle. Pour la fin de l'expiration, nous avons obtenu les images 13, 43 et 70 pour les premier, deuxième et troisième cycles respectivement.

Après l'analyse, le clinicien a détecté les frames 3, 27, 57 pour fin d'inspiration du premier, deuxième et troisième cycle respectivement. Pour la fin de l'expiration, les trames 12, 45 et 70 pour le premier, le deuxième et le troisième cycle.

Par rapport au volume respiratoire, les tests ont été comparés aux valeurs d'un ventilateur ce qui a donné une erreur en moyenne de 14%.

Les tests montrent l'efficacité du système et prouvent également que c'est un système prometteur. Il sera d'une grande aide en situation de manques d'experts, de pandémie et surtout en pédiatrie ou les patients sont de bas âge avec des volumes respiratoires très minimes et système immunitaire plus fragile.

Mots-clés : détresse du patient, nuage de points, volume respiratoire, segmentation, reconstruction de surface.

Patient respiratory assessment

Grâce HOUNKANRIN

ABSTRACT

Respiratory failure is one of the diseases of humans, regardless of age. It can occur at any time and can be detected by the study of respiratory measurements such as inhaled air volume and breathing frequency by the clinician. But the analysis of these parameters is very difficult with the naked eye, especially when it comes to children.

To overcome this problem encountered during diagnosis, this research project aims to develop a more efficient and effective approach to the analysis of respiratory measurements. Mainly, the design and development of a computerized and autonomous system based on 3D images, which would allow the study of the volume of inspired and expired air based on the thoraco-abdominal movement as well as the study of the respiratory frequency to detect a respiratory insufficiency. A methodology has been proposed to achieve this in the pediatric department.

This methodology consists of the acquisition of 3D images of the patient during breathing with a KINECT Azure camera positioned above the patient, followed by a reconstruction of the image obtained and then the analysis of this image by creating sinusoidal graphs to represent the respiratory cycles in the acquisition using Matlab and CloudCompare. The detection of the respiratory cycles makes it possible to calculate by subtraction the volume of inspired air. The respiratory cycles are detected thanks to the position of the abdomen and the thorax at each instant (lowered in exhale and raised in inhale).

The reliability and accuracy of the method has been examined by performing several tests on real patients and mannequins, comparing the results obtained by the method with the real results.
The end of inhale and exhale tests were compared to the results detected visually by a physician, resulting in errors of 1-3 frames. The tests were performed on a mannequin and on a real patient. A real patient test gave frame 1 for end of inhale first cycle, frame 27 for second cycle and frame 60 for third cycle. For the end of exhale, we got the frames 13, 43 and 70 for first, second and third cycle respectively.
After the analysis, the clinician detected frames 3, 27, 57 for end of inhale of the first, second and third cycle respectively. For the end of exhale the frames 12, 45 and 70 for first, second and third cycle.
The tests were compared to ventilator values for respiratory volume, resulting in an average error of 14%.

The tests show the effectiveness of the system and prove that it is a promising system. It will be of great help in situations of lack of experts, pandemic and especially in pediatrics where the patients are young with very small respiratory volumes and fragile immune system.

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Keywords: patient's distress, point cloud, respiration assessment, segmentation, surface reconstruction.

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LIST OF ABREVIATIONS

CC	CloudCompare
PC	Point Cloud
RGB-D	Red Green Blue-Depth
PCD	Point Cloud Data
PLY	Polygon
LAS	Laser
OBJ	Object
AI	Artificial Intelligence
PTX	Plain Text
FBX	FilmBox
ASCII	American Standard Code for Information Interchange

LIST OF SYMBOLS

L	litre
ml	millilitre
mm	millimeter

INTRODUCTION

0.1 Clinical context

One of the important physiological tasks in living organisms is breathing, this important task comes with various respiratory diseases that requires attentive care and intensive training. Understanding these abnormalities in breathing through volume measurement can help describe the mechanisms required in detecting breathing during daily life.
To identify respiratory abnormalities, respiratory parameters such as tidal volume and respiratory rate must be precisely calculated. The tidal volume represents the difference between the volume of inhaled air and volume of exhaled air. As for the respiratory rate, it represents the number of respiratory cycles per unit time. To quantify the tidal volume of a patient, the moment of end of inhale and end of exhale must be detected. Unfortunately, do it visually all the time is a long-term work. Moreover, calculating the tidal volume is not possible visually but only when a patient is under respiratory help. Hence the need of a system or mechanism to dynamically quantify the respiratory parameters of any patient whether under ventilator or not.

Spirometry and pneumotachograph are the most two commonly used techniques to calculate the respiratory volume. Spirometry is the mostly used method for dynamically measuring full respiratory volume changes, also the inhale and exhale time. It helps in detecting respiratory illness such as asthma, pulmonary fibrosis, chronic bronchitis etc. As a result, charts that plot the volume and flow of air called spirogram are created. With the spirometer parameters such as total lung capacity, tidal volume, residual volume can be measured.
The patient is instructed to breath through a device called a spirometer. Additionally, a nose clip is used to guarantee that the breath would flow only through the mouth. Breathing exclusively through a mouthpiece requires cooperation from the subject. Therefore, the spirometer is often not suitable for infant or children, heavily sedated or unconscious patients, or infants with breathing difficulty, limiting the usefulness of the spirometry as a tool to monitor and diagnose pulmonary diseases. For such medical assessment to be effective, it is

important that the measurement techniques do not interfere with the free movement of the chest, nor influence subject's respiration pattern.

Pneumotachograph is an equipment for measuring gas flow which is composed of a flow head with a passageway, multiple layers of screens stationed in the passageway to deliver a difference in pressure on diverse sides of the screens, a pressure transducer to transfer the pressures into a flow value, and a presentation or recorder for transferring the pressure readings to an operator to develop the precision of the flow measurements taken by the device. In this case as well, cooperation of the subject is needed, hence it cannot be used for unconscious or sedated patients and only used for subjects of 6years or older.

0.2 Research objectives

As explained above, those two methods are based on contact technics which may not be suitable in every condition. Nowadays there are no method of measuring tidal with children breathing spontaneously. Only when the patient is under ventilator can we measure the tidal volume with spirometer. Therefore, the need of non-contact methods to compute the tidal volume of a patient whether he is under respiratory help, has respiratory deficiency or not.

Based on that, this project consists in developing methods to accurately compute the tidal volume of a patient using 3D image acquisitions which is a non-contact approach. The method to compute the tidal volume is based on the analysis of a point cloud (PC). The analysis allows the detection of end of inhale and exhale through the variation of the thoracic surface. The measurement of that variation is proportional to the tidal volume value.

The quantification of this parameter will allow not only to detect any respiratory deficiency but also to characterize the severity of the deficiency (normal, mild or severe).

The measurement of respiratory parameters is essential to detect breathing disorders such as pulmonary disorders, COVID-19, etc...

Another aspect of this work is to appreciate the impact of the methods used on the result of volume. This will help determine how to minimize the error. The system will be validated using the mechanical ventilator, the reference standard used in intensive care.

The quantitative respiratory measurement interface will be a support tool for the remote and objective detection of respiratory anomalies, especially when health care resources are limited.

The system can be used to screen patients with symptoms of congenital diseases such as COVID-19. Detecting symptoms without direct patient contact will reduce the high risk of contagion among healthcare professionals.

0.3 Limitations

As expected, there are some limitations to the methods. The patients concerned by this research being children, it makes it complicated to keep them still while taking the images. The motions have impacts on the volume calculated, which opens some questions such as finding a way to detect the motions in an acquisition to drop those cycles in the calculation of the final volume.

0.4 Achievements

This research has resulted in demonstrating the impact of existing methods and libraries in the calculation of tidal volume, detecting the respiratory cycles using 3D images and computing the tidal volume to get a result as close as possible to the real tidal volume of a patient. This research has also demonstrated the challenges in working with children as to diagnose a disease especially respiratory deficiency.

At the end of this research a method to detect the end of inhale and exhale has been put together and tested. The test of the method has been approved by result from a clinician. For example, on a patient, with the method the end of inhale obtained were the frames: 1, 27, 60, 92, 122 and the result from the clinician was the frames: 3, 27, 57, 95, 123. Which gives a difference of 1-3 frames between our estimation and clinician result.

Also, a method for the calculation of tidal volume has been put together and tested. The tests have been compared to real values from a ventilator giving errors of 10 % to 18%

0.5 Thesis outline

The present report consists of 5 chapters. The first chapter presents the literature review. The second chapter is an introduction to the article and presents the organization of the document. The third chapter is the actual article. The fourth chapter present the evaluation of the results followed by discussions in the final chapter. Encompassing all of this is the introduction and the conclusion.

CHAPTER 1

LITERATURE REVIEW

1.1 Introduction

This research is based on the study of the phenomenon of breathing in children. Different organs such as the nose, the mouth, the pharynx, the lungs, the diaphragm come into play during breathing with specific functions. These different organs and their roles will be explained in this chapter, but it is noted that the main organ whose movement is studied in this work is the diaphragm.

Also, several works have been done on the same subject by different groups of people with different methodologies such as the use of point cloud with deth camera, machine learning, reconstructed 3D point etc. In this chapter these articles will be summarized as well as the work that has been done previously by Haythem on the subject treated in this research.

In the case of our study, we studied point clouds obtained from an Azure kinect DK camera and this study was mainly done using CloudCompare. What is a point cloud? What are the characteristics of the Azure Kinect DK camera, and what are the features of the CloudCompare software? These are the questions that will be answered in this chapter.

Finally, based on Haythem's work, some contributions have been made namely the detection of breathing cycles, the calculation of the breathing volume, the study of the impact of the different parameters used on the result of the volume, how to make the acquisitions etc. These contributions will be briefly summarized in this chapter.

1.2 Literature review

1.2.1 Physiology and mechanic of breathing

Breathing is the process by which oxygen from the air enters your lungs and is carried through the body. The lungs take oxygen from the air and deliver it to the bloodstream, which carries it to the tissues and organs that allow us to walk, talk and move.

The lungs also remove carbon dioxide from our blood, through the air we exhale.

1.2.1.1 Physiology of breathing

The two movements of breathing, inhale, and exhale are controlled by the autonomic nervous system (it is therefore an unconscious function) and adapt to the needs of the organism, according to its effort.

The lungs are the main organs that are involved in respiration but there are also other organs like: nose, mouth, pharynx, larynx, diaphragm as shown in figure 1.1.

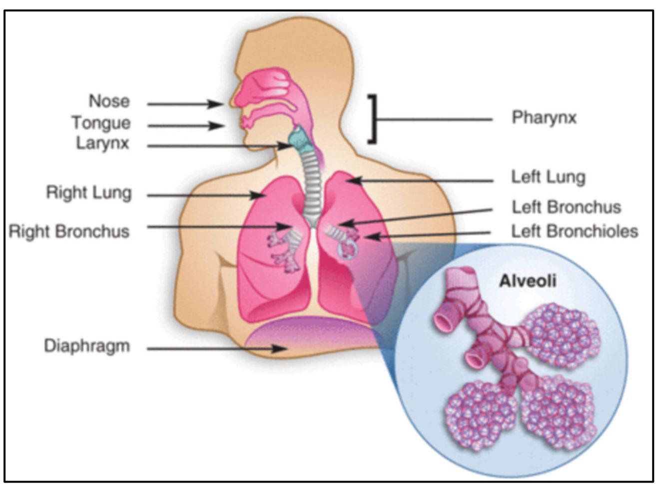

Figure 1.1 Respiratory organs
Taken from CK-12 CBSE Biology
Class 10: Respiration in Humans (2022)

Each of the organs showed abode has a specific role in the breathing process of a human as explained below:

The nasal cavity (nose) is the best way for outside air to enter the respiratory system. The hairs that line its inner walls are part of an air cleaning system.

Air also enters through the oral cavity (mouth), especially for people that are used to mouth breathing or with temporal blockage of the nose.

The pharynx (throat) collects air from the nose and channels it down into the trachea.

The larynx contains the vocal cords. The incoming and outgoing air creates the sound of the voice.

The trachea is the tube that connects the pharynx to the lungs.

The ribs are the bones that support and protect the chest cavity. Slightly mobile, they help the lungs to inflate and contract.

The trachea divides into two bronchi (tubes), one for each lung. The bronchi in turn divide into bronchioles. The right lung has three lobes or sections. The left lung has two lobes.

The diaphragm is the strong muscular membrane that separates the chest cavity from the abdominal cavity. It contracts when the lungs are taking in air and the chest volume gets larger. In the other hand it relaxes to let the lungs push out air while the chest volume gets smaller. That movement is the one studied in this research.

1.2.1.2 Mechanic of breathing

Breathing has two movements, inhale and exhale. The following two paragraphs explain each of these movements.

Inhale is the respiratory movement during which air enters the lungs. Inhale occurs when the intercostal muscles contract, which lifts the rib cage and increases the volume of the lungs. The contraction of the intercostal muscles allows the ribs and sternum to be pulled up. The diaphragm then contracts, becomes flat while lowering and stiffens which makes the rib cage thus takes on volume. The lungs then also take on volume. Indeed, since the pleura is stuck to the inner wall of the rib cage, it forces the lungs to stretch hence the pressure inside the lungs lowering than that outside. The air goes into the lungs. Each inhale allows the entry of approximately 0.5 L of air (BBC, Cardio-respiratory system, p. 2).

Exhale is the respiratory movement during which air exits the lungs. It is associated with the relaxation of the intercostal muscles which will promote the compression of the lungs by lowering the rib cage. Relaxing the intercostal muscles allows the ribs and sternum to descend. The diaphragm relaxes, curves and rises then the rib cage loses volume. The result is the lungs having a smaller volume. The pressure inside the lungs is then greater than that outside which makes the air go out. When exhaling, the lungs do not empty completely. There will always remain a small amount of air called the residual volume. It is estimated at about 1.2 L (John J. Lofrese, Connor Tupper, Deanna Denault & Sarah L. Lappin, 2023).

1.2.2 Existing works

In 2018 KyeongTaek Oh; Cheung Soo Shin; Jeongmin Kim and Sun K. Yoo presented research called Level-Set Segmentation-Based Respiratory Volume Estimation Using a Depth Camera. The respiratory volume estimation is done by computing the difference between two depth images. But before that, segmentation is done on the images to detect the regions involve in respiration. Each image is segmented based on two principles: region involve in respiration which is the chest wall and location where depth change occurs on previous image.
Level-set method is the method used to distinguish the region of interest which method was combined with spatial and temporal information. The spatial information is the shape of the human chest wall which is predefined. The temporal information is the respiratory related region segmented from previous images in a time-aligned depth image.
Depth changes can occur in other parts of the body, but those parts are not considered. They got an error of 8.41% by calculating the average of tidal volume compared to the actual volume. The actual volume was obtained from a ventilator measurement.

Meng-Chieh Yu; Jia-Ling Liou; Shuenn-Wen Kuo; Ming-Sui Lee and Yi-Ping Hung published in 2012 a noncontact method for respiratory volume measurement. In this study a depth camera was used to take images of people and those images have been segmented to get the region of interest. This helps to study chest wall motion for respiratory volume estimation.

Morphological changes of the chest wall are measured to estimate the volume. A study was also done on regional pulmonary measurement and the results showed difference of pulmonary functional between diseased and the contralateral sides of the thorax after thoracotomy. After the chest wall mask is segmented, the volume estimation itself is calculated by computing the difference of the current depth image and the reference depth image. Finally, the tidal volume in each breath cycle could be estimated by detecting the peak/valley points of the volume changes. Other parameters can be calculated like the inhalation volumes, exhalation volumes, respiratory rate, and even the respiratory method, such as abdominal breathing.

Real-Time Tidal Volume Estimation Using Iso-surface Reconstruction published in 2016 by Shane Transue; Phuc Nguyen; Tam Vu and Min-Hyung Choi treats of the same subject.

The research objective was to present a method to monitor the tidal volume of a patient by using a reconstructed 3D chest surface image obtained by a single depth camera. They have put together an algorithm to reconstruct in real-time a surface that helps generating omni-direction deformation states of a patient's torso during breathing process. That deformation is used in calculating the tidal volume based on a per-patient correlation metric from Bayesian-network learning process.

Instead of using an orthogonal deformation which is the up and down movement of the chest, they used omni-directional deformation because it incorporates a better vision of the displacements on the patient's chest due to breathing. The movement on the chest comes from the movements in the left and right lungs which is modelled as balloons.

After image acquisition, a segmentation is down to acquire the chest, then the holes in the image are filled followed by normal estimation. The surface is then reconstructed using triangulation to obtain a mesh. The volume of the mesh is calculated using signed tetrahedral volume algorithm. The tidal volume is then computed by the difference between the mesh volume and a volume initially recorded during monitoring process called base volume.

During the tests of the method, the patient is kept mobilised to make sure that the chest movement is proportional to the tidal volume.

After implementation they have achieved 92.2% to 94.19% accuracy in the tidal volume estimations.

The last project presented is the work of Haythem Rehouma, Rita Noumeir, Philippe Jouvet, Sandrine Essouri published in 2020 using depth camera to estimate respiratory parameters. This project is focusing on capturing inhale and exhale images from pediatric patients to compute the volume of air that the patient takes in. Hence if the volume is lower that the normal expected volume, the doctor can detect that the patient has respiratory deficiency. Before now the only way a doctor could know the volume of air being taken in by a patient is when a patient is using the respiratory aid.

In the study 2 depth cameras were used (kinect v2 and v1) to make the image acquisition. then using CloudCompare (CC), manually segment, filter, reconstruct and finally compute the volume, hence this approach took so much time, and it was too technical for a doctor to use. The image acquisitions are done between 2 to 5 seconds with the patient laying down on the bed and creates up to 1000-point clouds (3D images). The reconstruction was done by adding a plane surface to the back of the image to close it. Manually studying each of those images that have been divided to get the volume took so much time.

In order to get the volume of each frame, each of them was divided into small cubes called octrees. Depending on the parameter that are put in CC, those cubes can be too small or too big. It can be so small that no part of the image is found in that cube. Hence more than required leading to more volume because the volume at the end is computed by multiplying the number of cubes by the unit volume of each cube.

To compute the tidal volume, a graph was created out of the volume per image. The peak of the graph which is a sine graph represent the end of inhale while the valley represents the end of exhale.

Furthermore, the volume is computed by taking the difference between those two moments. The test on mannequins could not be confirmed because the manufacturer of those mannequins is not able to tell exactly how much volume of air this mannequin was supposed to take in.

1.3 Contributions

Based on Haythem work, it has been concluded that the reconstruction algorithm is not adequate because of the impact the parameters have on the volume. The number of octrees created on the image depends on the parameters entered. The number can increase or decrease and there is no exact value of parameter that can be applied to all the images. It is more of a trial and error.

Also, the plane that is added to the image in order to close the back of the image doesn't fit totally, which leaves some uncovered spaces or cut some spaces out.

Finally, the detection of the end of inhale and end of exhale, which was done using the volume per frame graph, trying to replicate doesn't give the expected result. There was a difference of 3-7 frames sometimes between the real frame of end of inhale or exhale and the one obtained from the graph. The real frame is gotten by a clinician analysis.

Moving forward, the process to reconstruction of Haythem's research has been used without the parameters. Having access to this work helped in the foundation of my research.

The contributions of this research are seen first in detection of end of inhale and end of exhale. The detection is not only based on the volume per frames but also on the coordinates of each frame. Combining both technics gives a more efficient and stable result that has been proved trustworthy by test with a clinician.

Then the reconstruction of the image has also been done differently. Instead of introducing another element which is the plane, the image is reconstructed by triangulation.

According to Delauney, a triangulation is the creation of a triangle network such that the triangle abc obtained by connecting three points of the point cloud can be considered as a legal triangulation if the circumscribed circle of this triangle abc does not contain any other point of this point cloud. This definition applies to a plane but by replacing circles by circumscribed spheres, the same method is applied to 3D surfaces.

By doing this we ensure not only that we connect all the points but also that we have triangles with the smallest possible area. This reduces the errors of adding surfaces that could increase the volume of the 3D image.

This process of reconstruction doesn't have an impact on the volume depending on the parameters. Avoiding an additional element in the image reduces the risk of additional or reduced volume.

Lastly the tidal volume calculation is done by calculating the difference between the volume of the frame of end of inhale and the volume of the frame of end of exhale.

1.4 Materials used

Through out the work different materials and apps has been used. To do the images acquisition, the camera used is an Azure Kinect DK which gives as result PC and those PC are treated in CloudCompare (CC) for volume calculation. This section describes each material.

1.4.1 Camera

RGB-D sensor is a depth sensing device which works in association with RGB camera. The purpose is to add images depth information that is the distance to the sensor in a per-pixel basis. Microsoft released 3 RGB-D (Red Green Blue-Depth) sensors: Microsoft's Kinect (Xbox Kinect V1 & V2) and the azure Kinect DK which is the one we use in this study.
The Azure kinect DK (figure 1.2) is a sophisticated camera with computer vision and speech models, advanced AI (artificial intelligence) sensors and a range of performant SDKs that can be connected to azure cognitive services. These are the elements inside the azure Kinect DK:

- 1-MPixel depth sensor with wide and narrow field-of-view (FOV) options that help you optimize for your application
- 7-microphone array for far-field speech and sound capture
- 12-MP RGB video camera for an additional color stream that's aligned to the depth stream
- Accelerometer and gyroscope (IMU) for sensor orientation and spatial tracking
- External sync pins to easily synchronize sensor streams from multiple Kinect devices.

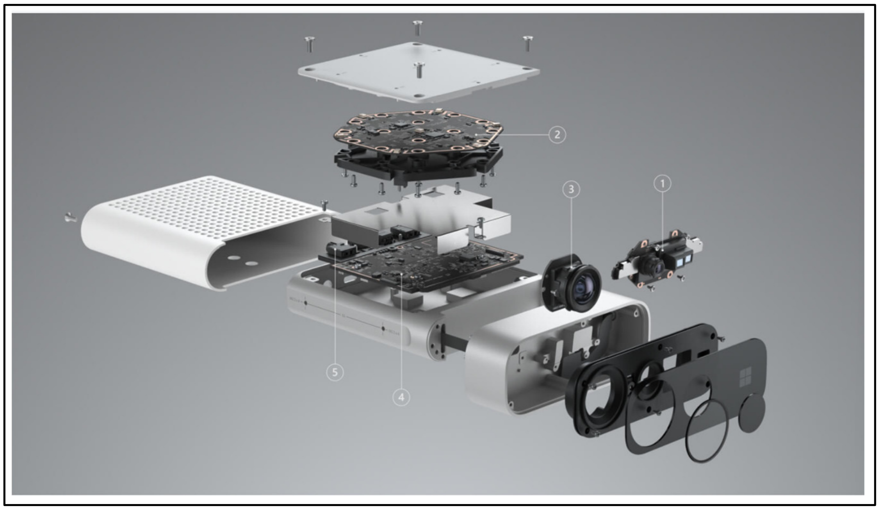

Figure 1.2 Kinect Azure DK

Taken from Azure Kinect DK, Microsoft.com (2023)

1.4.2 Point Cloud description

Also known as 3D visualisation, a 3D PC is the source of an accurate 3D model of the real world. It is used to represent or process 3D model of any kind of object. It is a collection of points plotted in 3D space which points are generally captured using a 3D laser scanner. The scanner or camera used to acquire the PC combines the vertical and horizontal angles gotten by the laser to calculate 3D X, Y, Z coordinate for each of the points in the object to produce measurements which include the color stored in RGB and intensity. The more the points are dense, the more detailed the image is which allows a more accurate result while processing it. There are different file extensions for PC such as PCD (Point Cloud Data), LAS(LASer), XYZ, OBJ (object), PLY (polygone). The difference between those types is the use of ASCII and binary. ASCII (American Standard Code for Information Interchange) is rooted in binary but conveys information using text which is readable for humans. The ASCII file extensions are XYZ, OBJ, PTX (Plain text) and ASC. In the other hand binary systems store information directly in binary. The binary file extensions are usually PCD and LAS. But some other file types can be stored both in ASCII and a binary equivalent for more compact storage and more rapid saving and loading, such as PLY, FBX (filmbox). The PLY extension is the one most used in this work. It is a format used to describe an object as a polygonal model. The first time this format was used was a 3D laser triangulation scanner model.

The image below shows the PC obtained after acquisition. As seen in the image, the whole of the room is taken, hence the need for segmentation. Each line in the file especially ASCII format represents the coordinates of a specific point in x,y,z and the color in RGB. (Annexe 3 PC file content).

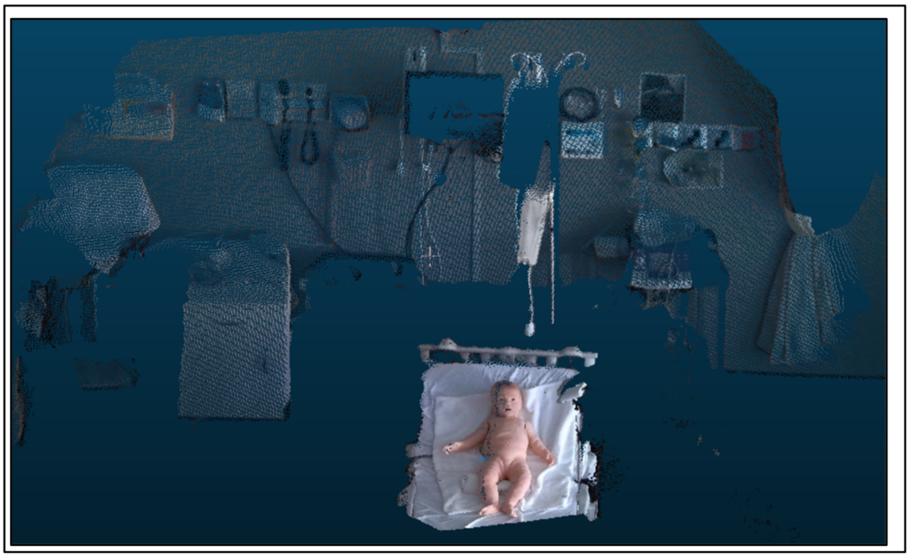

Figure 1.3 Image obtained after acquisition

1.5 CloudCompare

CloudCompare is the software used for almost all the calculations. It is a 3D PC and triangular mesh processing software and designed based on PCL library. The purpose behind the development of this software is to perform comparison between two 3D PCs obtained with laser scanner or between a PC and a triangular mesh. But also, to perform processing of many PC (10 million points and up to 120 million with 2GB of memory). It can be used on Windows, Linux and macOS. It works based on an octree structure that is highly optimized for the purpose of the process.

1.6 Conclusion

The study of movement and respiratory volume has been a hot topic for a number of years. methodologies have been put in place to ensure a better assessment of respiratory volume using common elements such as point clouds, depth camera as well as CloudCompare and MATLAB software.

The developed methodology which consists in acquiring point clouds followed by segmentation because as seen in the image above the camera takes the whole room, filtering to remove noise, reconstruction to close the back side of the patient, detection of respiratory cycles and finally volume calculation will be presented in detail in the next chapter.

The main question that the following chapter will answer is how the different equipment presented intervene in the volume calculation.

CHAPTER 2

RESEARCH APPROACH AND ORGANIZATION OF DOCUMENT

As said earlier in the chapter above (Theory), the acquisitions are done using Azure Kinect DK and the resulting point clouds are constructed.

As per Haythem's work (Rehouma et al., 2020), there are 4 steps to reconstruct an image: segmentation, filtering, computing and orienting normal then the reconstruction itself. (Shown in the Diamond Shaped representation in Figure 3.8)

- Segmentation: to be able to get the part of the body involve in respiration, i.e abdomen and torso because the camera will take a picture of the whole patient and the whole room. The obtained image gives more data than needed hence having to cut out off some part of the image to get the torso and the abdomen.
- Filtering: this helps remove all the noise in the image. When some point of the image captured is out of place it creates the noise. Noise is also created when a part of the image is not complete i.e., the camera focuses on a side of the patient and the other side is not full (missing some points).
- Computing the Normals: The normals are axis that are perpendicular to each point per image. There must be created and point to the same direction which is orientation to get the reconstruction right.
- Reconstruction: during reconstruction, a plane is created to cover part of the image, this then appears as a second image being put in the original image, the reason is that when the camera takes the image, it does not take the back of the patient, hence the image is an open image at the back, and it needs to be closed giving the introduction of a plane.

The next chapter present the article submitted to IEEE Access, covering the work done in this research. The chapter is divided as followed: (1) methods applied which are data collection, segmentation, filtering, reconstruction, and volume calculation, (2) results, (3) discussion.

The chapter after will present an evaluation of the work, other approaches, and some limitations. The final chapter will discuss the results.

CHAPTER 3

POINT CLOUD-BASED TIDAL VOLUME ESTIMATION FOR PEDIATRIC CARE

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Abstract

Respiration is crucial to sustain life and can be clinically assessed by measuring inhaled/exhaled air volume (tidal volume) and breathing frequency. However, tidal volume measurement is only roughly estimated with the caregivers naked eye in spontaneous breathing children. Then, a more efficient and effective approach to respiratory measurement and analysis is needed. We present a method of tidal volume quantification based on 3D video analysis. An Azure Kinect camera is used to acquire 3D images of the thorax and abdomen of patients. The acquired images are used to reconstruct the volume of the patient's body, and respiratory cycles are analyzed after the detection of the thoraco-abdominal end of inspiration and end of expiration. We tested the reliability of the proposed method on humans and mannequins. The proposed method obtained an error of one to three frames to detect the ends of inspiration and expiration compared to the clinician assessment. We then measured the tidal volume (difference between inspiratory and expiratory thoraco-abdominal volume) with an average error of +/- 23 mL when compared to values from a spirometer of a ventilator in mechanically ventilated children. While refining our tidal volume calculation method, we will be able to provide accurate measurement to caregivers even in spontaneous breathing children.

Index terms patient's distress, point cloud, respiration assessment, segmentation, surface reconstruction.

3.1 Introduction

Monitoring the breathing of critically ill children requires attentive care and intensive training. Assessing respiration includes calculation of respiratory parameters such as the tidal volume (i.e., volume of air inhaled/exhaled during a respiratory cycle) and respiratory frequency. The measurement of tidal volume can only be roughly estimated by visual inspection alone. Hence, a reliable measurement of tidal volume and respiratory rate is needed to dynamically monitor these respiratory parameters in critically ill children.

The two most common techniques for measuring tidal volume in clinical practice are spirometers placed at the mouth (spirometry measurement) or in the respiratory circuit of a ventilator when patients are mechanically ventilated. Spirometry is used by pulmonologist in children older than 6 years and adults to follow-up respiratory condition in asthma, pulmonary fibrosis, and chronic bronchitis. Spirometry consists in measuring changes in the pulmonary volumes and flows. The patient is taught to breathe through a spirometer, which has a nose clip to guarantee that the patient only breathes through the mouth, so it requires cooperation from the subject. Therefore, spirometry is often unsuitable for children, heavily sedated or unconscious patients, and patients with breathing difficulty, which has limited its usefulness for monitoring and diagnosing pulmonary diseases. An effective measurement tool must not interfere with the free movement of the chest or influence the respiration pattern of the patient. A spirometer on a mechanical ventilator is a medical device that measures gas flow and volume in real time either at the T piece or at the expiratory port of the ventilator. Both spirometry and spirometer included in mechanical ventilators are contact-based techniques, which may not be suitable for all patient conditions. Hence, a non-contact method is needed to measure the tidal volume of patients with respiratory failure and who are spontaneously breathing.

Transue et al. estimated the tidal volume by using a depth camera to acquire patient images and performing segmentation to acquire the region of interest (i.e., chest). They applied a filter and calculated the normal vectors, after which the surface was reconstructed to obtain a complete image of the patient’s upper body. The volume of each reconstructed surface was calculated and was used to detect the ends of inhalation and exhalation. The tidal volume was defined as the difference in volume between those two moments. In their tests, the patient was not allowed to move to ensure that the chest movement was proportional to the tidal volume.

Oh et al. proposed a level-set segmentation-based method for estimating the tidal volume from the difference between two depth images. For each image, the surface was segmented based on two principles. The level-set method was used to distinguish the region of interest regarding spatial and temporal information. The spatial information was the shape of the human chest wall, which is predefined, and the temporal information was the respiratory-related region segmented from previous acquisitions in a time-aligned depth image. Changes in depth that can occur in other parts of the body were not considered. With their method, they were able to calculate the average tidal volume in healthy adults with an error of 8.4% compared to the actual tidal volume.

Massagram et al. used non-contact Doppler radar to monitor breathing and measure the tidal volume. The Doppler effect is a change in frequency when a radio wave reflects off a moving object. Doppler radar involves transmitting a radio signal toward an object, receiving the reflected signal, and comparing the two signals. They suggested that the movement of the chest due to lung displacement could be detected by changes in the sensor output, which is a sine graph that is reconstructed to remove offset due to internal leakage and clutter reflections. They calculated the tidal volume from the difference between the peak and valley amplitudes of the signal. They compared their results with reference values from a spirometer. The comparison between the radar volume and the real volume showed an average mean difference of 39 mL in seated patients and 24 mL for supine patients.

We propose a method for accurately calculating the tidal volume of a patient. Our method is a non-contact approach that uses 3D image acquisition for point cloud (PC) analysis to detect the ends of inhalation and exhalation based on the variation of the thoracic surface. This variation is proportional to the tidal volume. Similar to Massagram et al., we created a sine graph of the thoraco-abdominal movement and defined the peaks and valleys as the ends of inhalation and exhalation, respectively. We also defined the tidal volume as the difference in volume between the two moments. We analyzed the effects of various parameters involved in the volume calculation. We applied our proposed method to calculating the tidal volumes of children mechanically ventilated and compared the results to those obtained from the respirator. The effects of segmentation and filtering were also analyzed.

This work is an extension of our previous study, which we tested on mannequins. In contrast, the present study used human subjects. Our previous tests revealed that the previous method needed to be improved for implementation in clinical practice, especially regarding the detection of the respiratory cycle. We changed our previous approach of analyzing the volume per frame to estimating the distance between the camera and abdomen. In our previous study, we assigned the largest volume to the end of inhalation and the smallest volume to the end of exhalation. However, a comparison of the results to the analysis by a clinician revealed a lot of discrepancies. In the present study, we based the detection of the respiratory cycle not only on the volume per frame but also on the coordinates of each frame (distance between the camera and the subject). Combining both techniques should yield more reliable and stable results. The reconstruction of the surface was also changed from introducing a plane surface to Delaunay triangulation. A triangle obtained by connecting three points in a PC is considered legal if the circumscribed circle of the triangle does not contain any other point of the PC. This definition is applicable to a plane, but by replacing circles with circumscribed spheres, the same method can be applied to 3D surfaces. With this method, we ensured that the triangles not only connected all points but also covered the smallest possible area. This reduced errors from surface augmentation that could increase the calculated volume.

Our contributions are as follows: 1) we present the respiratory volume as calculated for critically ill children in real condition from depth images; 2) we show that the respiratory volume cannot be used to determine the end of inhale and the end of exhale because of noise, and we therefore propose a new method to determine these two instants based on the distance of the chest to the camera; 3) we analyze and discuss the impact of filtering, segmentation, and normal computation on tidal volume estimation; and 4) we propose a new method to estimate the tidal volume of critically ill children and compare the estimated values to those obtained by spirometers.

3.2 Methods

Our proposed method for calculating the tidal volume has four steps: segmentation, filtering, computing the normal vectors, and reconstruction. The four steps are done in CloudCompare. Segmentation consists of cutting out some part of the surface to obtain the parts of the body involved in respiration, (i.e., abdomen and torso). Segmentation is required because the camera takes a picture of the whole patient and provides more information than needed. Filtering removes noise in the PC data that is produced when part of the surface is not complete. During image acquisition, the camera focuses on one side of the patient, and the other side is partially covered. Other elements such as wires and duvets also create noise in the PC data. Computing the normal vectors consists of calculating the vectors perpendicular to each point in the PC. They must be created and pointed in the same direction for accurate reconstruction. The directions and positions of the normal vectors determine the direction of the back during reconstruction. The surface of the back must be parallel to the chest wall and abdomen. Reconstruction consists of creating the body surface and a plane to close the back of that surface. Because the camera does not take the back of the patient, the surface is open at the back, and it needs to be closed to have an object with a volume.

3.2.1 Data collection

The dataset consisted of PCs obtained by video recording critically ill children in the pediatric intensive care unit of Ste_justine Hospital (Montreal, Canada), with the consent of their parents, using an RGB-D sensor. Data were collected from children spontaneously breathing and mechanically ventilated using RETRACT system, which is a system developed by our team for calibrating the camera, starting the acquisition, and saving the PC data from one or two cameras directly on a computer (Rehouma et al., 2020). RETRACT also combines the point clouds obtained when 2 cameras are used for acquisition. In this study, the camera used is the Azure Kinect DK, which is a Microsoft RGB-D (Red Green Blue-Depth) sensor. The camera was mounted so that its field of view covered the region of interest (i.e., the abdomen and torso of the patient). This resulted in PC data (.ply) that captured the whole room with 500 images in 30 s. We ensured that no materials were present that would make segmentation

difficult (e.g., duvet or wires) and that the patients did not move during the image acquisition. Figure 3.1 shows the acquired image of a mannequin as a reference. The acquired images of the patient should be similar with no diaper, duvet, or wires, and the patient should be lying on their back.
We obtained the hospital ethical board's approval for this research (Ste-Justine review ethical board number: 2020-2276).

Figure 3.1 Image of a good acquisition (no diaper, wire, or duvet)

3.2.2 Segmentation

Segmentation was performed manually using CloudCompare, which is a 3D PC and triangular mesh processing software that is based on Point Cloud Library. We used CloudCompare for almost all calculations in this study. During segmentation, the bed and room in the original PC were removed, and the region of interest (i.e., the chest, abdomen) was kept. The region of interest started from the clavicle and went through the armpits to where the patient's body touched the bed. It continued under the abdomen where the thighs are connected to the body, as shown in Figure 3.2 Example of acquisition with diaper. For a patient with a diaper or duvet on (e.g., Figure 3.3), we cut the top of the diaper or duvet because it would affect the volume. When the patient was breathing, the abdomen moved up and down. However, when a diaper was covering the abdomen, the difference in volume would not be clearly observed in the PC data, so this obstacle had to be addressed. In cases where the patient was covered with a blanket, the blanket was avoided as much as possible during the segmentation.

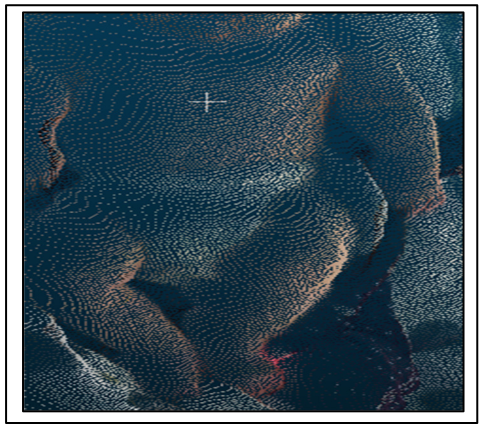


Figure 3.2 Example of acquisition with diaper

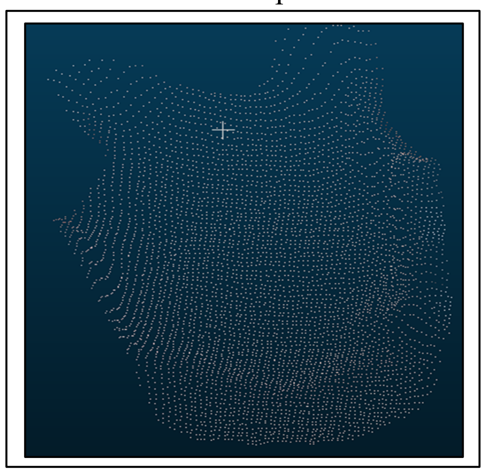


Figure 3.3 Example of adequate segmentation

3.2.3 Statistical outlier removal

An outlier is any piece of data that is at abnormal distance from other points in the dataset. It is a non-informative data point that tampers with the predictive modelling performance The PC datasets typically have different point densities and measurement errors create sparse outliers. These impact the estimation of surface normal and curvature changes leading to erroneous values. To solve some of these irregularities, a statistical analysis on each point's neighborhood can be performed followed by trimming of the neighbors which do not meet a certain criterion. The sparse outlier removal of CC is based on the computation of the distribution of point to neighbors' distances in the input dataset. For each point, the mean distance from it to all its neighbors is computed. By assuming that the resulted distribution is Gaussian with a mean and a standard deviation, all points whose mean distances are outside an interval defined by the global distances mean and standard deviation can be considered as outliers and trimmed from the dataset.

To align with the language of CloudCompare we will call it filtering in the rest of the paper. Hence after segmentation, the resulting PC is filtered using the statistical outlier removal filter of the Point Cloud Library (PCL).

Two parameters are entered into CC tool for filtering. We have the mean distance estimation (MDE) set at 12 and the standard deviation (STD) set at 5 (Figure 3.8).

Removing too many points may divide the original image into too many parts, especially around the ribs, or create holes in the image. Figure 3.4 shows a portion of the shoulder that was separated from the main surface due to filtering. The left ribs show a hole that was not present in the original image. The parameters need to be calibrated by trial and error depending on the camera used for image acquisition to ensure that no holes are created in the segmented and filtered image.

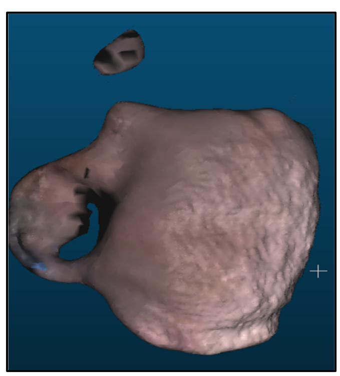


Figure 3.4 Reconstruction of an image with wrong parameters (hole created in left ribs)

3.2.4 Computing the normal vectors and reconstruction

After filtering, normal vectors were created with CC tool and oriented in the same direction to define the position of the back during reconstruction. The normal vectors were then used to reconstruct the surface. The reconstructed back needed to be perpendicular to the normal vectors.

In the previous work (Rehouma et al., 2020), a plane surface created in CC was introduced to the image to be reconstructed. During this reconstruction, the plane surface was found to create an additional volume in the original frame volume calculation, or the volume was reduced. This is because the surface either cuts out the original image or does not close the back hermetically as observed in Figure 3.4. Nevertheless, other reconstruction methods, including Poisson surface reconstruction with Neumann boundary and Poisson surface reconstruction with Dirichlet boundary, can be considered. In this study, we chose the latter reconstruction method because the other method reconstructed the chest without closing the back. The boundary represents a constraint envelope that specifies the reconstruction function to be zero in regions devoid of samples. This is achieved by ensuring that the reconstruction does not overlap the exterior leaf node by the envelope rasterization into the octree (M. Kazhdan, M. Chuang, S. Rusinkiewicz & H. Hoppe, 2020). Such an approach avoids an additional surface in the base image, reducing the errors from the surface augmentation and increasing the calculated volume. For the reconstruction, the following parameters were used: octree depth = 11, samples per node = 1.5, and point weight = 2 (Figure 3.8). These values are chosen using the trial-and-error method on different frames.

The parameter sample per node can be described as a floating-point value denoting the minimum number of sample points required within an octree node as the octree construction is adapted to sampling density. For noise-free samples, small values in the range of 1.0 to 5.0 can be used. For additional noisy samples, larger values in the range of 15.0 to 20.0 may be needed to obtain a smoother and noise-reduced reconstruction. The default value of sample per node is 1.0 (M. Kazhdan, 2022).

Similarly, the parameter point weight is described as a floating-point value that represents the interpolation of point samples in the formulation of the screened Poisson equation. The results of the original (unscreened) Poisson Reconstruction can be obtained by setting this value to 0. The default value of point weight is 4.

The parameter depth is denoted as an integer representing the maximum depth of the tree used for surface reconstruction. Note that the specified reconstruction depth is only an upper bound since the reconstructor adapts the octree to the sampling density. The parameter depth has a default value of 8.

The chosen reconstruction method is the closest to Delauney triangulation triangulation (Delaunay triangulation, Wikipedia, 2023). According to Delauney triangulation, a triangle obtained by connecting three points on a PC is considered legal if the circumscribed circle of the triangle does not contain any other point on the PC. This definition is applicable to a plane and can be also applied to 3D surfaces by replacing the circles with circumscribed spheres. This method ensures that the triangles not only connected all points but also covered the smallest possible area. A previous study by A. Akdim et al., has compared the quality criteria between Poisson surface reconstruction and Delauney triangulation (A. Akdim, A. Mahdaoui, H. Roukhe, A. M. Hseini & A. Bouazi, 2022).

We then obtained a reconstructed surface with a curved torso and abdomen, and a planar back (Figure 3.6 (a)). The dark surface represents the back of the patient, which is not present in the original image Figure 3.6 (b)). The reconstructed images did not contain any holes. A hole indicates that the normal vectors are not pointing in the same direction.

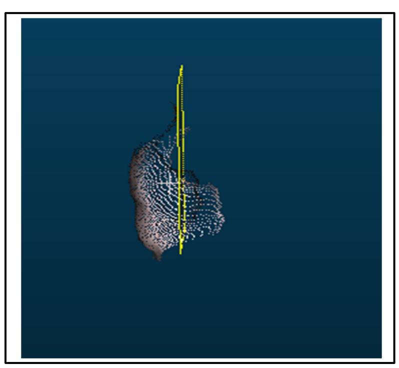


Figure 3.5 Reconstruction with
the introduction of plane surface

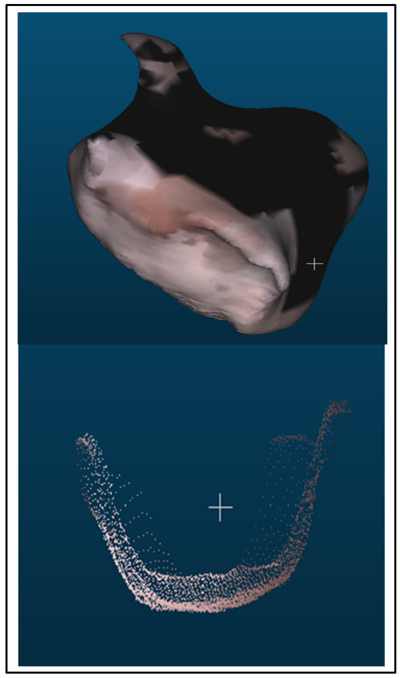


Figure 3.6 Top represents the patient's back after reconstruction and the bottom is the non-reconstructed image.

3.2.5 Tidal volume calculation

To detect respiratory cycles, we calculated the average distance between the patient and camera along the Z-axis. The average distance was then used to create a sine graph as shown in Figure 3.7, where the peaks represent the end of inhalation (green), and the valleys represent the end of exhalation (red). The horizontal axis represents the frame index (e.g., 500 images in Figure 3.7). The vertical axis represents the average distance along the Z-axis for each PC. The labels above and below the curve give the frame indices for the ends of exhalation and inhalation. For instance, Figure 3.7 shows that the ends of inhalation and exhalation were at frames 78 and 110, respectively. A range of seven frames before and seven frames after these positions were taken, which correspond to frames 71–85 and frames 103–117, respectively. The frame with the maximum volume in the inhalation range was considered to represent the end of inhalation (frame 81 in Figure 3.7). The frame with the minimum volume in the exhalation range was considered to represent the end of exhalation (frame 115 in Figure 3.7). The difference between the end of inhalation and the end of exhalation volumes (maximum − minimum) is the tidal volume.

Figure 3.8 presents the steps and parameters used for reconstruction. First, the PC data were segmented to output a .pcd or .ply file. The segmented data were filtered according to two parameters: MDE and STD. The filtered data were used to estimate normal vectors according to the surface model, octree radius, and minimum spanning tree. The normal vectors were oriented based on the parameters of the neighbors. Finally, a new PC was reconstructed according to the octree depth, advanced boundary (reconstruction type, e.g., octree cubes or triangulation), samples per node, and point weight.

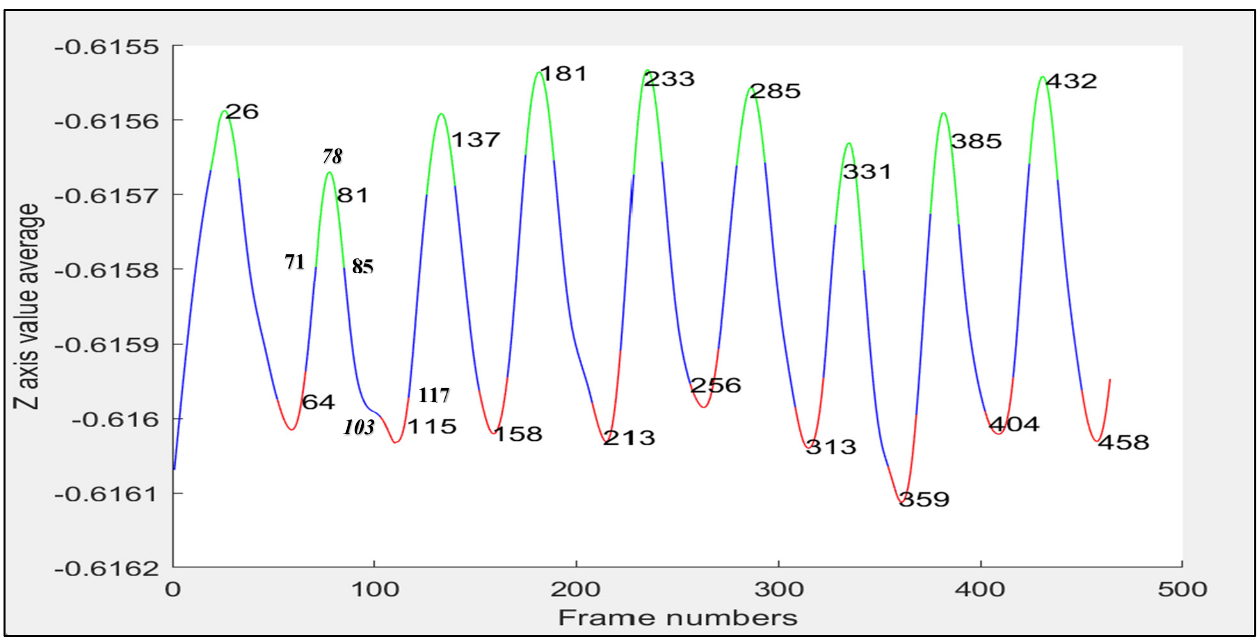


Figure 3.7 End of inhale in green and exhale in red

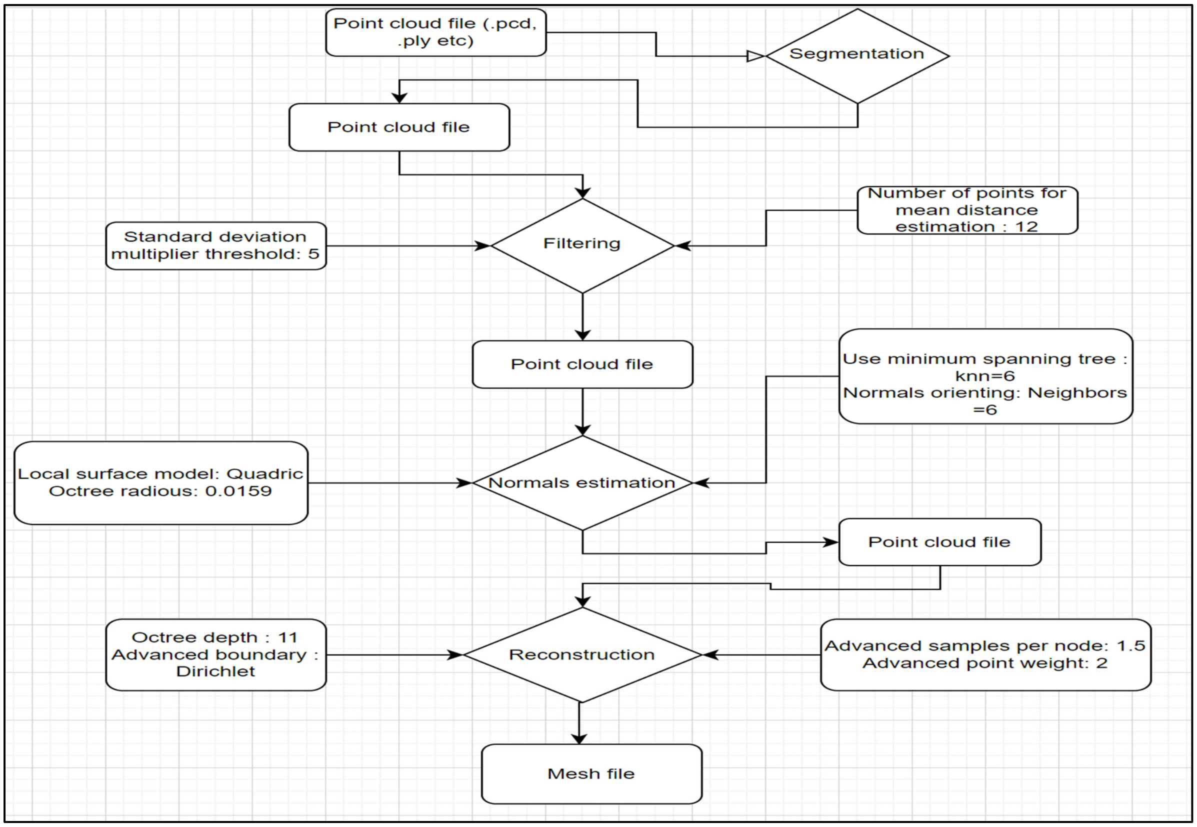


Figure 3.8 Steps to reconstruction

3.3 Results

3.3.1 Estimation of end of inhalation and exhalation

Table 3.1 compares the ends of inhalation and exhalation acquired from a mannequin as estimated by our proposed method and visually estimated by a clinician. The first column represents the index of the respiratory cycle. The second combined column displays the frame indices of the end of inhalation as estimated by the clinician (left) and by our method (right). The third combined column indicates the frame indices of the end of exhalation as estimated by the clinician (left) and by our method (right). The clinician estimation was done manually by visualising the video of the patient during breathing. The last combined column provides the error between the estimations of the clinician and our method in the number of frames. The difference was between one and three frames. For cycle 1, the clinician estimated the end of inhalation at frame 7, which is the same as that estimated by our method. The clinician estimated the end of exhalation at frame 21, whereas our method estimated the end of exhalation at frame 22, which corresponds to an error of one frame. For respiratory cycle number 3, the clinician estimated the ends of inhalation and exhalation as frames 58 and 77, respectively. Our method estimated the ends of inhalation and exhalation as frames 60 and 74, respectively. These correspond to a difference of two frames for the end of inhalation and three frames for the end of exhalation. The test was done on 4 acquisitions, both mannequin and patient acquisition. The average difference observed was 0-3 frames.

Table 3.1 Frame indices of ends of inhalation and exhalation

Cycle	Frame index: End of inhale		Frame index: End of exhale		Errors in number of frames	
	Clinician results	Research results	Clinician results	Research results	End of inhale	End of exhale
1	7	7	21	22	0	1
2	33	32	45	47	1	2
3	58	60	77	74	2	3
4	88	88	104	101	0	3
5	118	115	132	129	3	3

Table 3.2 Tidal volumes as calculated for patient A0085

Cycles	Tidal Volume (ml)
1	93
2	162
3	185
4	165
5	158
6	122
7	146

Table 3.3 Tidal volumes as calculated for patient A0113

Cycles	Tidal Volume (ml)
1	182
2	118
3	180
4	129
5	147
6	155
7	221
8	299
9	87
10	122
11	188
12	89
13	108
14	195
15	109
16	170
17	131
18	118
19	157
20	127
21	126
22	155
23	104
24	152
25	154
26	160
27	100
28	69
29	81

3.3.2 Calculation of the tidal volume

Table 3.2 presents the calculated tidal volumes for patient 1 (9.5 years old, bodyweight of 28 kg and reason for PICU admission: seizures). The first column represents the respiratory cycle, and the second column represents the calculated tidal volume by our proposed method. Tidal volumes of 93 and 162 mL were calculated for the first and second respiratory cycles, respectively. The mean and median of the calculated tidal volume over seven respiratory cycles were 147 and 158 mL, respectively. The error between the average of our method and the value obtained by spirometer was 22 mL (12%). The median tidal volume measured by the ventilator spirometer over the same period was 180 mL. The median value was defined as the calculated tidal volume for a specific period.

Table 3.3 presents the calculated tidal volumes for patient 2 (4 years old, bodyweight of 12.6 kg and reason for PICU admission: of hypovolemic shock). The first column represents the respiratory cycle index, and the second column represents the tidal volume calculated by the proposed method. The tidal volumes for the first and second cycles were 182 and 118 mL, respectively. The mean and median values of the tidal volume over all respiratory cycles were 143 and 131 mL, respectively. The median tidal value measured by the ventilator spirometer over the same period was 111 mL. The error between the median and the spirometer values was 20 mL (18%). The median value was defined as the calculated tidal volume over the specified period.

3.3.3 Effect of segmentation

The effect of segmentation on the tidal volume estimation was observed during surface reconstruction. Because the reconstructed surface was large, a combination of errors affected the tidal volume estimation. For example, a reconstructed surface representing only the chest would be expected to have an error in the estimated volume. Adding the abdomen to the chest surface would accumulate errors from both the chest and abdominal reconstructions. During reconstruction, all points in the image are linked together by Delaunay triangulation, which ensures that the surface is closed. However, linking these points produces a surface that does not effectively simulate the body of the patient. For example, the reconstructed back is not as flat as a human back. Some curves are created when points from the back are linked to points in the front. These curves may increase the actual volume and result in error even with good segmentation. Those curves increase the surface area, which in turn increases the volume. Even with excellent segmentation and filtering, errors can occur and reduce the accuracy of the estimated tidal volume.

We examined the effect of segmentation by comparing three surfaces. The first surface comprised the torso and abdomen (Figure 3.9), and the second and third surfaces partially contained the arm (Figure 3.10(a) and leg (Figure 3.10(b)). For the first respiratory cycle, Figure 3.10 indicates that the volume was 4364 mL at the end of inhalation (circle) and 4278 mL at the end of exhalation (star) for the first surface. When the surface was greater than the region of interest, Figure 3.11 indicates that the first respiratory cycle had a volume of 5213 mL at the end of inhalation (circle in the first red range) and 5113 mL at the end of exhalation (star in the first green range). Overall, the volumes per frame (values beside the circles and stars) were greater in Figure 3.12 than in Figure 3.11. This is because Figure 3.12 shows the segmentation of a bigger surface than in Figure 3.11. Because the tidal volume is the difference in volume between two PCs created by thoraco-abdominal movement, it should be the same value independent of the segmentation. However, this was not the case because Figure 3.12 indicates that the tidal volume was 100 mL for the first respiratory cycle, but Figure 3.11 indicates that the tidal volume was 86 mL. The differences in tidal volume with different segmentations can be attributed to errors from reconstruction. For two consecutive reconstructed surfaces, movement can be observed not only in the region of interest (i.e., the torso and abdomen) but also in the leg. This movement is caused by imperfect reconstruction, which may vary slightly from one surface to the next because it depends on the location of each point inside the PC and how close the points are to each other.

Table 3.4 presents the estimated tidal volumes of the three segmented images of the same patient over five respiratory cycles. The first column presents the respiratory cycles. The second column presents the estimated tidal volume for the torso and abdomen segmentation. The third column presents the tidal volume for the segmentation containing a leg. The last column presents the tidal volume for the segmentation containing an arm. The mean and median tidal volumes for the first, second, and third segmentations were 89 and 86 mL, 114 and 112 mL, and 96 and 100 mL, respectively. The error between the first and second segmentations was almost 26 mL (23%), and the error between the first and third segmentations was 14 mL (14%). These results indicate that a larger reconstructed surface produces greater errors.

Table 3.4 Volume comparison for different size of segmentation

Respiratory Cycles	Seg1(n ormal)	Seg2(with leg)	Seg3(with arm)
1	87	100	81
2	86	125	99
3	121	143	107
4	88	99	106
5	80	118	102

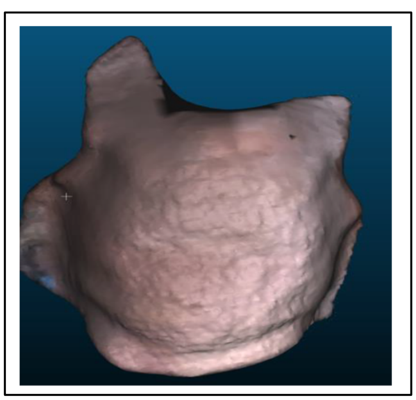


Figure 3.9 Torso and abdomen segmentation

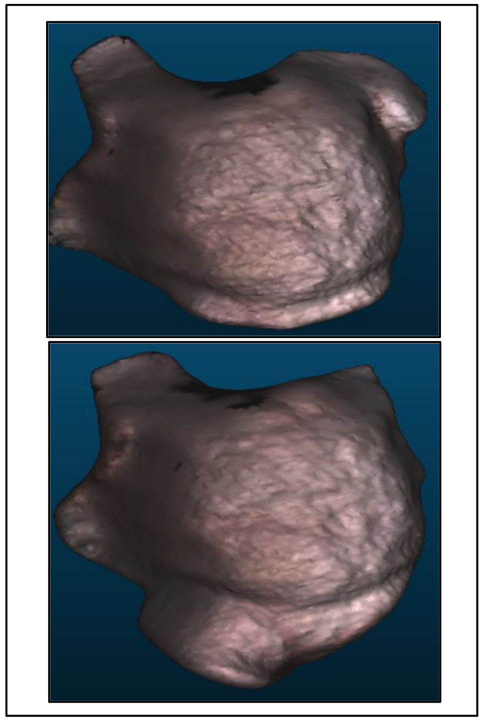


Figure 3.10 The top image represents segmentation with arm added and the bottom image represents the segmentation with leg added

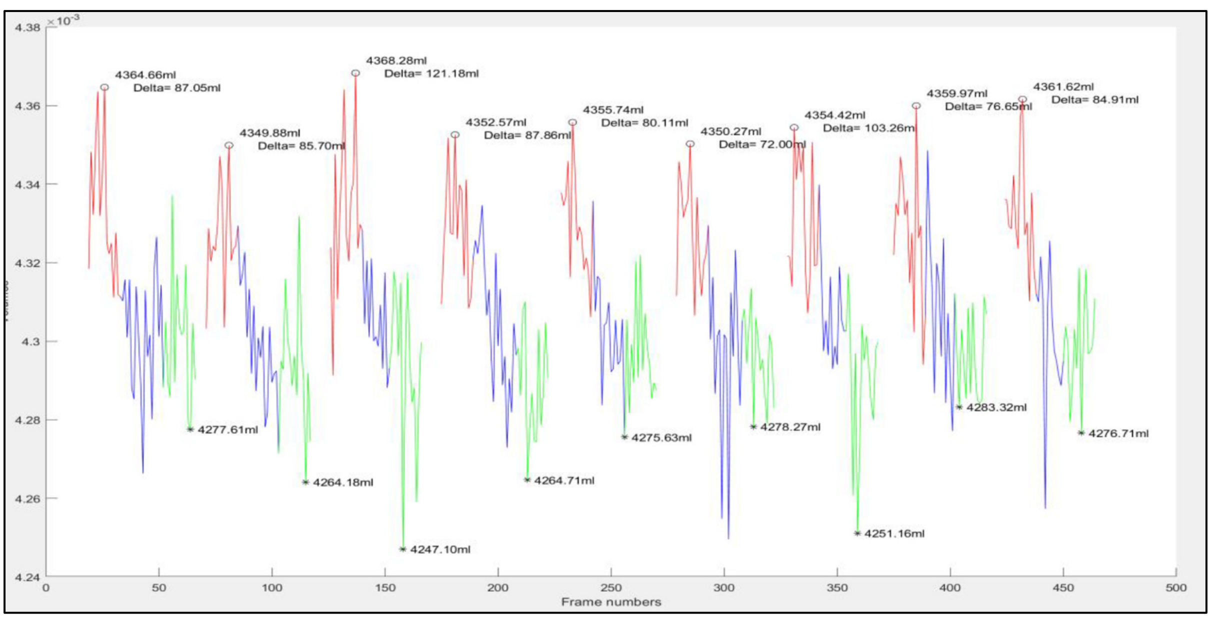


Figure 3.11 Impact of segmentation (torso and abdomen segmentation)

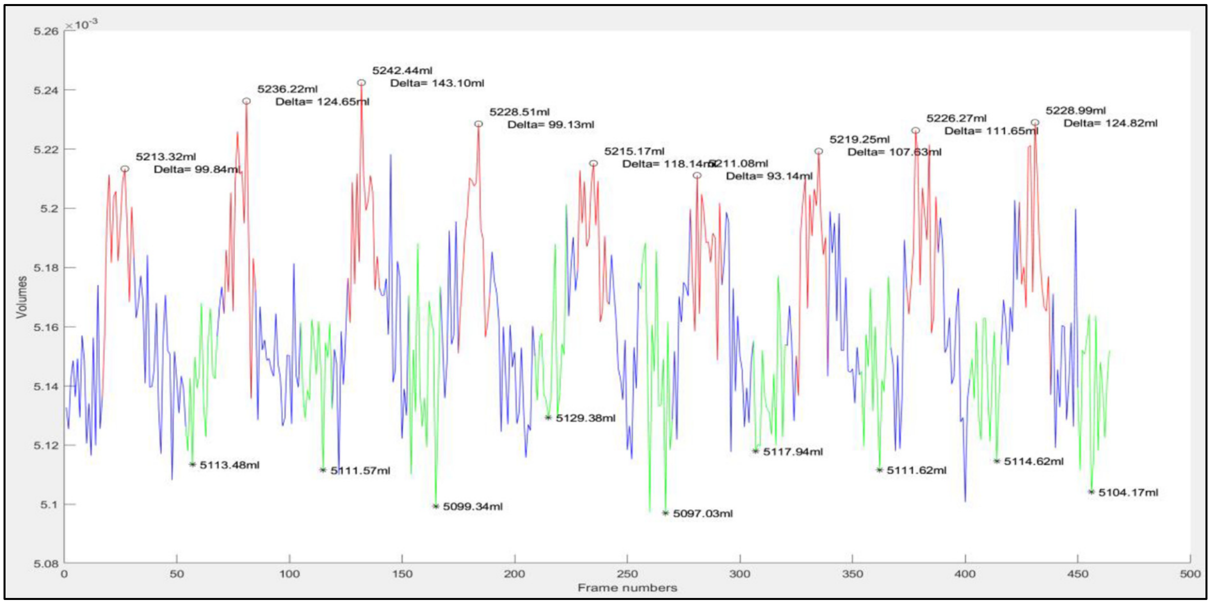


Figure 3.12 Impact of segmentation (segmentation larger than ROI)

3.3.4 Filtering

To investigate the effects of filtering on the tidal volume estimation, three different sets of filtering parameters were applied to the data acquired from the mannequin. The filtering process involved two parameters that are multiplied with each other: MDE and STD. Smaller values for the parameters remove more points from the images and vice versa. Figure 3.13, Figure 3.14, and Figure 3.15 show the estimated volumes per frame calculated with each set of parameters. The red, green, and blue colors indicate the ranges of inhalation, exhalation, and in between, respectively. The circles and stars indicate the frames defined as the ends of inhalation and exhalation, respectively. The delta indicates the tidal volume per cycle. The volumes at the ends of inhalation and exhalation are given in milliliters next to the circles and stars. The graphs in these figures have almost identical shapes and oscillations, which demonstrates that the volume estimation is quite robust against changes in the filtering parameters.

Table 3.5 compares the calculated tidal volumes with three sets of filtering parameters. The first column represents the respiratory cycles. The remaining columns present the estimated tidal volumes with MDE = 8, 12, 15 and STD = 5. For each pair of parameters, we calculated the median value of the estimated tidal volume, which is given in the bottom row. For the first respiratory cycle, MDE = 8, 12, and 15 resulted in tidal volumes of 85.47, 87.05, and 85.46 mL, respectively. The difference between these median values being relatively small, we can say that MDE doesn't have a huge impact on the tidal volume estimation. Thus, the most important element of filtering is to ensure that the region of interest is complete and that no parts are missing.

Table 3.5 Volume comparison for different filtering parameters

Cycle	MDE: 8 STD: 5	MDE: 12 STD: 5	MDE: 15 STD: 5
1	85	87	85
2	90	86	86
3	121	121	121
4	93	88	88
5	70	80	67
	Median = 90	Median = 87	Median = 86

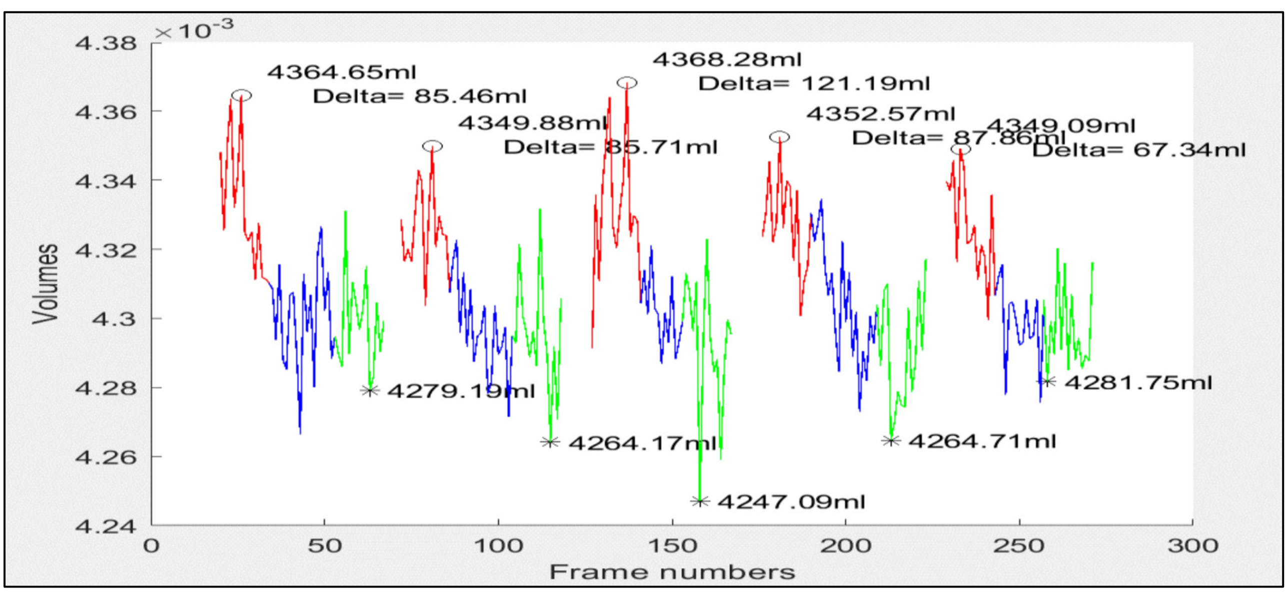


Figure 3.13 Volume graph with MDE = 15 and STD = 5

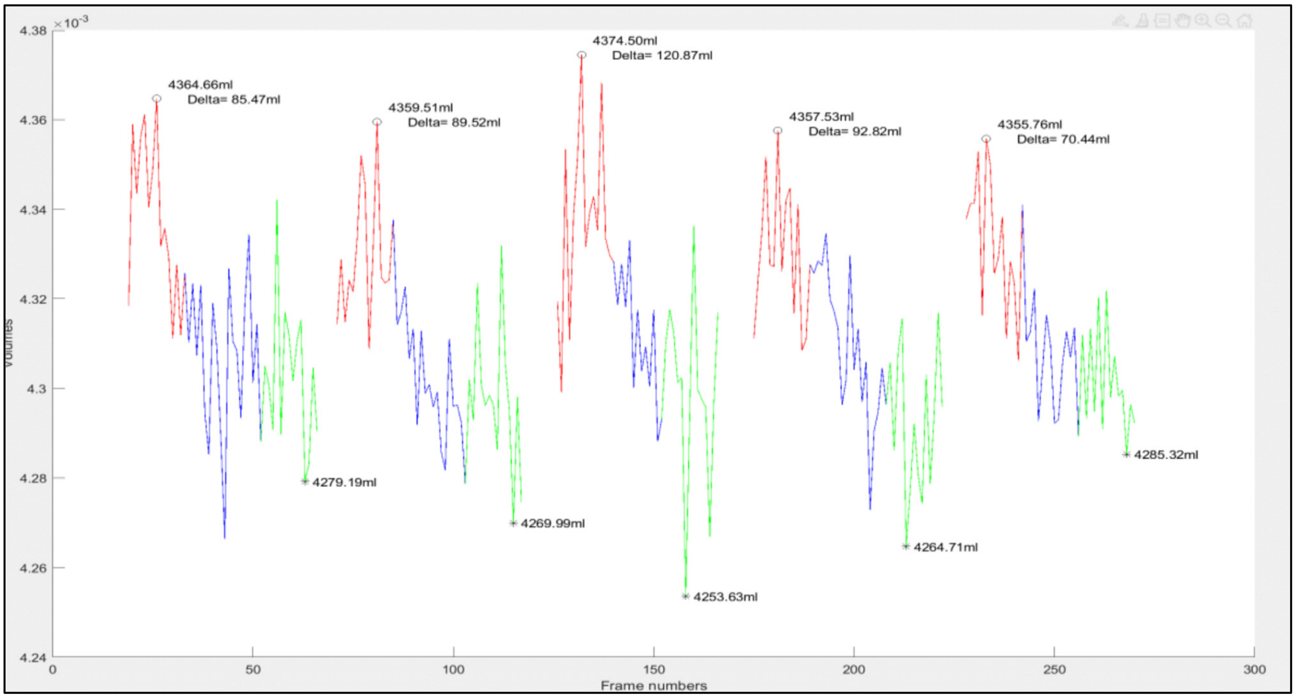


Figure 3.14 Volume graph with MDE = 8 and STD = 5

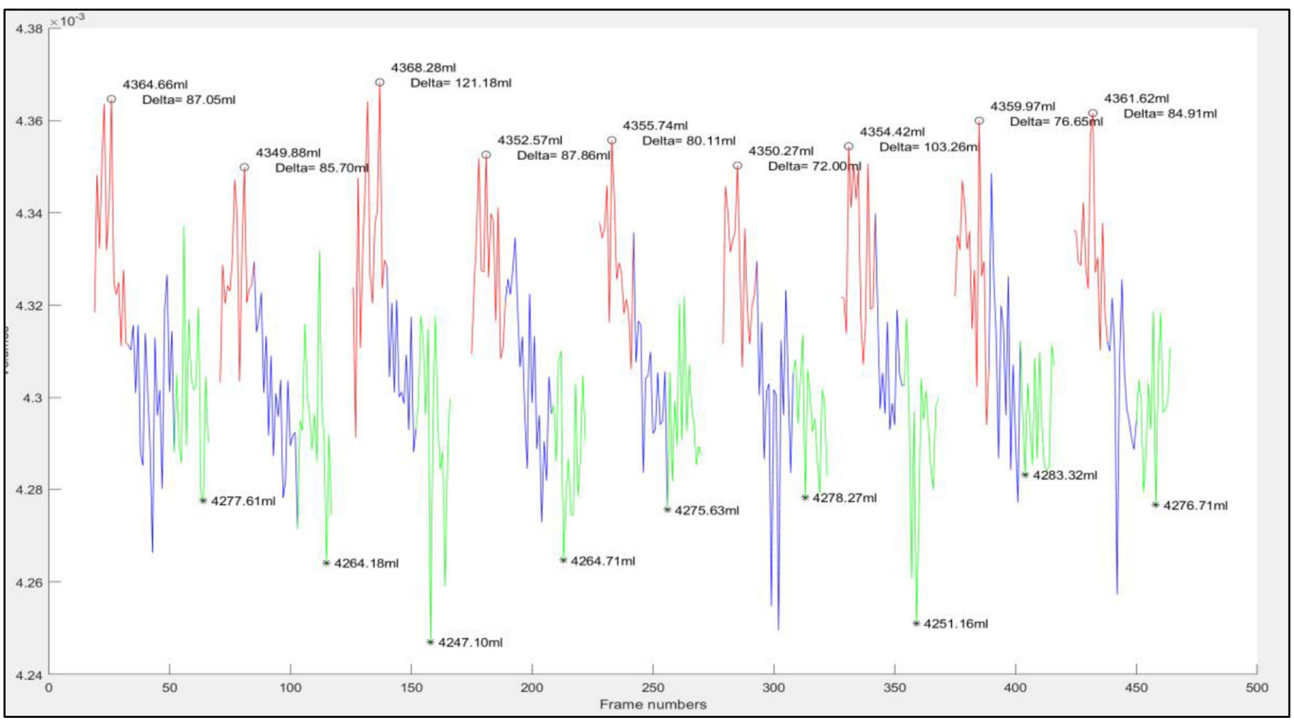


Figure 3.15 Volume graph with MDE = 12 and STD =5

3.4 Discussion

Goals for this study were to detect the ends of inhalation and exhalation from a set of images for a patient. A dataset of PCs was created by recording patients' breathing movements with an RGB-D camera, and the ends of inhalation and exhalation were estimated according to changes in the Z-axis distance. Our results showed that the volume per image could not be used to accurately determine the ends of inhalation and exhalation. Graphing the volume per frame (Figure 3.14) showed noisy oscillations, which may have been caused by the patient or camera moving and the presence of occluding objects such as diapers, duvet, or cables. Another objective was to estimate the tidal volume from the difference between the maximum volume around the end of inhalation and the minimum volume around the end of exhalation. Our results showed that the detected ends of inhalation and exhalation do not necessarily match the maximum and minimum volumes. Variations between cycles confirmed that the tidal volume is not constant across cycles even for a healthy patient. One explanation is that the lungs may retain a residual volume of air during exhalation or exhaled more air. A third objective was to clarify the effects of PC filtering and segmentation on the tidal volume estimation. Our results showed that filtering may distort the image and have a considerable effect on the volume estimation, but the error is generally negligible. Meanwhile, the effect of segmentation depends on the surface reconstruction, and the error increases with the size of the reconstructed surface. While the effect of surface reconstruction could not be quantified, it clearly cannot be ignored because it affects the quality of the points before and after filtering, which affects the volume estimation, generally by increasing the estimated value. Our current limitations being due to the one imposed by surface reconstruction. However, this can be mitigated by adapting a template fitting method.

PS Adison et al. also measured respiratory rate and tidal volume using depth sensing camera. The ROI detection after acquisition was done using a flood fill method. The tidal volume was estimated from the peak-to-through changes in the volume calculated from respiratory volume signals. After doing ten separate runs on a single volunteer, they obtained a bias of – 210 mL and a root mean squared difference of 230 mL. Compared to our study, their measure was not accurate enough to be used in children.

The difference in tidal volume measurement that can be acceptable in clinical practice is around 10%. As we observed, an absolute difference of 20 ml, assuming a tidal volume of 10 ml/kg, the estimation of tidal volume is acceptable with this video method for patients over 20 kg i.e., older than 6 years old.

3.5 Conclusion

The detection of respiratory disorders still faces many challenges, one of which is the inability to measure the air intake in spontaneously breathing patients. While most research has focused on adults, we focused on pediatric patients who move a lot and have low tidal volumes. Our objective was to detect the ends of inhalation and exhalation, estimate the tidal volume, and assess the reliability of the proposed method. This work was motivated by the goal of developing decision support systems to help clinicians assess the respiratory function in a pediatric acute care environment.

Breathing results in thoraco-abdominal movement: the height of the torso and abdomen increases during inhalation and decreases during exhalation. The difference in volume between the ends of inhalation and exhalation determines the tidal volume for a given respiratory cycle. In this study, we successfully determined the ends of inhalation and exhalation. However, the estimation of the tidal volume was not very accurate and depended on the quality of the torso segmentation and surface reconstruction. We obtained an error of 12%–18% for the tidal volume estimation. It is not possible otherwise to assess the tidal volume if the patient is breathing spontaneously. This study showed that PCs can be used to assess respiratory function. Based on our results, we were able to evaluate the efficiency of the proposed method and determine what needs improvement. Our team is investigating other methods to segment the torso and achieve a more accurate surface reconstruction.

CHAPTER 4

EVALUATION

4.1 Introduction

The errors present in the volume calculation come from the fact that the acquired images are not complete, i.e. there are missing points at the ribs. This lack is filled by the algorithm during the reconstruction but the result is not exact, it creates excess surfaces that increase the value of the volume. This led us to introduce the idea of acquisition with 2 cameras. In this chapter we will present the shortcomings of the 2-camera acquisition and talk about the precision and reliability of our method.

4.2 Two cameras acquisition issues

While analysing the difference between using two cameras or one camera, the first finding was the impact that it has on the volume. During the reconstruction, while using one camera, it has been explained earlier that there are some parts of the patient or the mannequin that are not totally in the field of the camera. Those sides or those parts are less complete (they look shallow) and the filtering has an impact on those sides. Reason why the right parameters of filtering must be found. But then we try to investigate two cameras' acquisitions, to see if there is still a big impact on the volume calculation due to the filtering. While using an acquisition from two cameras, after reconstruction it has been realized that the result is not smooth. Looking at the chest of the patient in the first image (Figure 4.1), on one side there seems to be two paths, one on top of the other. But looking at the second image (Figure 4.2), which is one camera construction, the image is smooth from the abdomen to the torso. From one set of ribs to the other everything connects well together. Meanwhile, the other image from one side to the other, it gives impression that there is a layer underneath that makes the connection from side to the other side. But another layer on top that stops in the middle.
With two cameras, one stays at one side, maybe the bottom or the head of the patient the other one stay at the opposite side. They are both placed at left side and right side of the patient. In

that way they both connect at some points. The fields of the two cameras overlap. It explains the fact that the resulting image gotten as a result that is supposed to be the fusion of both images comes with those common paths. There are common parts that both cameras are seeing at the same times. And those common parts when they are fused together it creates that impression of an image on top of another (Figure 4.1) after reconstruction so that has an impact on the volume. Thus, the resulting volume's graph does not give any type of form that can be recognized (Figure 4.3).

Instead of having a sine graph, it gives a graph which form cannot be explained. And many volumes are too high or too low. As result of that, we decided that we want to stay with one camera and clinically speaking, putting two cameras together in the same room to take acquisitions from a patient is not possible.

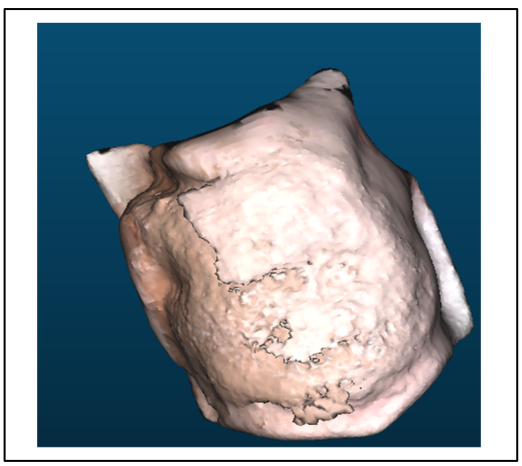


Figure 4.1 Overlap image (2 cameras)

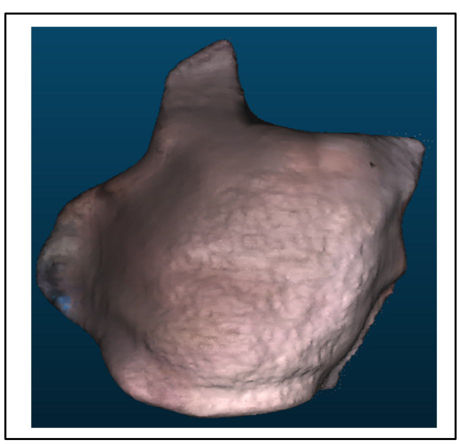


Figure 4.2 Image without overlap(1camera)

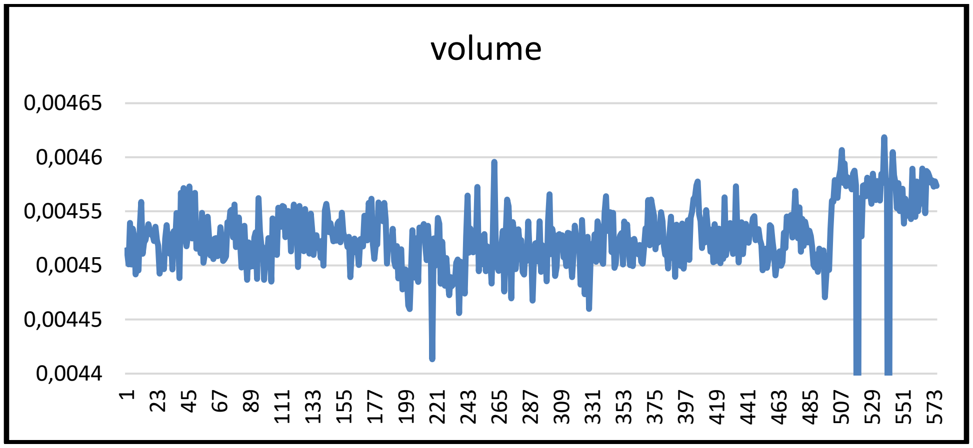


Figure 4.3 Two cameras volume graph

4.3 Cross examination with previous work

Some tests were also done using the mannequin to cross examine the present results with the result of Haythem when he was working on the same project. The first test was made on an acquisition with one camera at the frequency of 35 (gold_standard_35_az1). Using the method presented in this study, the range of volume gotten is from 72 ml to 88 ml and the median of those values is 85 ml.

Looking at the frequency which represents here the length of one cycle in number of frames, the length is between 46 and 55 number of frames for one cycle and comparing those results to the result that they got earlier in the project they had a minimum of volume of 38 ml, a maximum of 66 ml and an average of 52 ml. Talking about the number of frames per cycle they had between 50 to 53. Not forgetting that they had one cycle that was 34 frames. Let's not forget the fact that there is no gold standard value i.e the expected value of the volume for the mannequin is not known. There is no proof that the results gotten in the previous research are the right ones. But looking at the results in the present research, there is not too much gap between the values which already is an encouraging situation. The table below (table 3.6) shows the comparison of tidal volume on nine cycles for the acquisition gold_standard_35_az1.

Another test was made from the same type of acquisition but this time around with two cameras (gold_standard_35_az2).

Some results are always out of range, for example, on this test some volumes were 500ml. Of course, those cycles will not be considered in the rest of the process. But the rest is within a range of 43 to 84 millimeters. And the median of those values is 62 millimeters. Talking about the number of frames per cycle, the range is 29 to 38 frames per cycle. Now comparing those results to what they had earlier for this project, the minimum value of volume that they had was 62 millimeters, the maximum was 64 and the average was 62. About the number of frames per cycle they had 24 to 25 frames per cycle, not forgetting that one cycle was 12 frames. Conclusion, our results match well.

The results are satisfying, although there are still some cycles that are giving wrong results of tidal volume. Some cycle gives values that are really too high or too low, but that's normal. That's expected because there are always errors in calculations due to filtering and segmentation.

Table 4.1 Volume cross examination with previous research

Cycles	Haythem values (volumes in ml)	Research values (volumes in ml)
1	44.87	87.05
2	48.73	85.70
3	49.55	121.18
4	42.59	87.86
5	52.90	80.11
6	51.71	72
7	64.27	103.26
8	56.18	76.65
9	58.84	84.91
Median	51.71	85.7
Average	52.18	88.75

4.4 Balloon test

Balloon data would have been a great set of data to use to test our method because we know exactly the quantity of volume that has been pumped into that balloon. And the balloon does not move. It is not like a real person that moves all around. But the problem is that a balloon is supposed to be round on every side. But when the balloon is put down, and the image of that balloon is taken with the camera up looking towards the balloon, the camera does not see the other side i.e the under side of the balloon that is covered or that is not seen by the camera. That's why in all the images, turning them upside down, there are openings and that's why the reconstruction is required as explained earlier. The algorithm that is doing the reconstruction is not able to detect if whether it is a balloon or not. It only reconstructs the image trying to put a plane path under it to cover it up. With the balloon it will put a plane path instead of a round path. That means half of the balloon is lost, half of the volume, half of the movement that is put in a respiratory cycle.

While using the balloon to calculate the volume and do the tests, it has been found out that the volume was either half of the expected volume or even inferior to the half and it was not stable at all. If we want to use a balloon, we have to make sure that the balloon is not round on every side, maybe the side that will be touching the bed or maybe put one camera under the balloon and one camera over then make a fusion of those two images to create the whole image.

4.5 Precision of the proposed method

There are still some improvements to do to get more stable and precise values. For the detection of end of inhale and exhale for example, the process to make the average graph smooth before being able to get the peaks and valleys, is not constant. The smoothing algorithm takes a range as an input, calculate the average of the points in that range and replace those points by the resulting average. It is the range that needs to be readjusted depending on the situation. Depending on the original PC, if the distance between the points is large a higher range is needed and lower range in the opposite case. When a small range is chosen on the wrong PC, it gives the result seen in the figure 4.4. It can clearly be seen that around the position 100 on the X axis, 2 peaks back-to-back. On the other side, looking at the figure 4.5, those peaks have been merged with a wider smoothing range. It is to be noted that there is no specific range that works for each type of image. Rather the ranges are chosen by trial and error.

Another situation talking about precision will be the choice of the different ranges (smoothing, volume calculation). As explained earlier, the tidal volume is calculated by using a range of frames around the end of inhale and exhale detected. The values of those ranges don't have any specific or precise explanation. Those values were chosen out of trial and error, that means there is no real stability about the values. Depending on the type of acquisition the values will need to be adjusted. It is also important to notice that the percentage of error caused by those values is not none now.

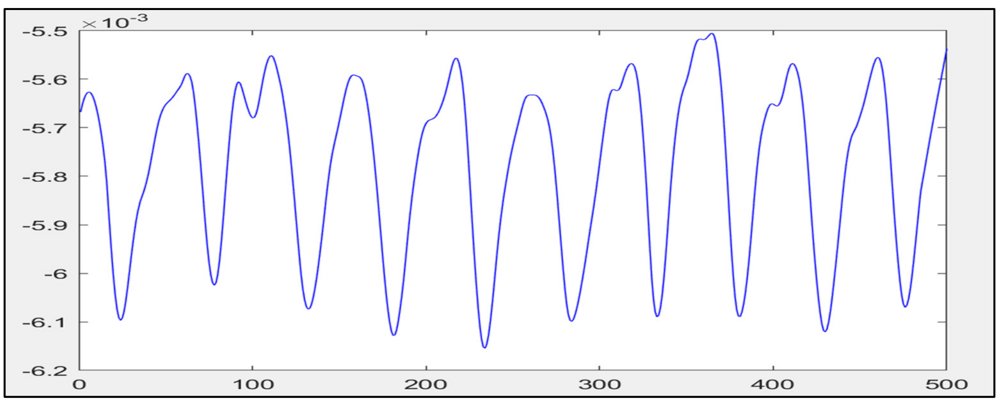


Figure 4.4 Image obtained for wrong range of smoothing

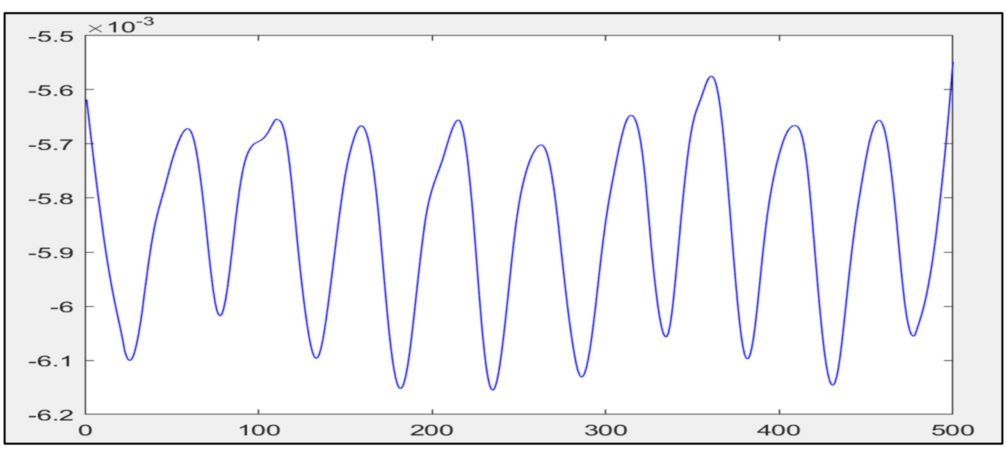


Figure 4.5 Image obtained for good range of smoothing

4.6 Conclusion

In conclusion, the acquisition with two cameras is not a viable idea for the moment due to the fact that after reconstruction we see a surface superposition. This superposition is created by the fields of the two cameras which intermingle.

In terms of precision, there is still room for improvement because several values are applied by trial and error such as the smoothing of the averaging curves and the volume calculation intervals. The smoothing values change depending on the camera used, the position of the points, the quality of the image etc.

The present research is therefore a springboard for the next people who will work on this project in order to create a final volume calculation application.

CHAPTER 5

DISCUSSION

The purpose of this work was first be able to detect end of inhale and end of exhale using set of patients' images. A data set of point clouds was created by recording patients' breathing movements with RGB-D camera. Using that dataset, the end of inhale and exhale was obtained by focusing on the Z axis. It helped find out that the volume per image could not be used to detect end of inhale and exhale due to large variation in values. The graph obtained using the volume per frame was noisy due to motions in the original image created by patients moving, camera moving or presence of foreign elements like diapers, duvet, cables etc.

Another part of this work was to compute the tidal volume per cycle, which was done by using ranges around the end of inhale and exhale detected earlier. The maximum and minimum were taken respectively in those ranges then the difference of those two values represents the tidal volume. The inconsistency in volume from one cycle to another also confirms the fact that there is no equality in respiratory volume across cycles even for a person in good conditions. This is explained by the fact that during exhale sometimes there is residual volume of air left in the lungs.

The final part was to explain the impact of filtering and segmentation on the volume calculation. The study on that part revealed that the impact of filtering can be enormous if the resulting image is distorted but other than that the error can be negligible. There is a certain range of value that should be applied during filtering to ensure that the result is right. The purpose of the filtering being to remove noise it shouldn't be changing the integrity of the original image which is what the method is exposed to when filtering parameters are not well defined.

Then about the study on the impact of segmentation, the findings are quite uncertain or stable. The segmentation doesn't have a quantifiable impact but can just be approximated (or guessed) based on the impact of reconstruction. The larger the surface to reconstruct the more the error which inspire the question: is the actual reconstruction made in CC the best way to get the desired result? This question must be investigated because as it is, the reconstruction is not as perfect that it should be since it most of the times increases the surface hence the increase of volume. The impact of the reconstruction is not quantified in this work but in this work, it is seen that it is not consistent because it depends on the size of the image to be reconstructed, the quality of the image before and after filtering.

Overall, the methods discussed in this study helped to understand the degree of evolution of the project. It is still needed to make the methods more consistent and stable especially in this condition which is hospital context since it is about human health. The takeout is that for a start this method can be used in nonlethal conditions meaning noncritical disease diagnosis.

CONCLUSION AND RECOMMANDATIONS

There are still many challenges in detecting respiratory diseases with automation. This start by being able to compute the volume of air that the patient's breath in. Most of the works that have been done was focused on adult while this project is focusing on pediatric patients where we are faced with patients that can move a lot with little value of tidal volume. The goal of this work is to detect the moment of end of inhale and end of exhale, compute the tidal volume more effectively and finally study the accuracy of the applied method. The purpose is to enable future work to program an app which can automatically compute the tidal without too much input from human and be user-friendly for doctors.

The set of data used, called point clouds was obtained by recording the patients with an RGB-D sensor. The purpose being to study the thoraco-abdominal movement because of breathing. During inhaling the torso and abdomen increase in height and during exhaling the contrary happens. Difference between those moments in respiratory cycle gives the tidal volume for that cycle.

This study showed that the methods used need improvement because of the level of error shown and uncertainty. In order of the volume to be computed effectively, the methods need to be a bit corrected to reduce the percentage of error which is about 15% presently. Most specifically the segmentation, filtering and reconstruction. As explained in this document, by the time the tidal volume calculation is done, there is already an accumulation of error which explains the average of 15% error. Which of course can be critical if used in a critical and value sensitive disease diagnosis.

Finally, this study has demonstrated that we are one step closer to the goal as it shed light on the elements to improve about the methods.

The first recommendation is to find a way to make the reconstruction fit better the original image specifically the back. Be able to get a plane and smooth reconstructed back.

Secondly convert the MATLAB code into an executable app and combine it with the program that will do the CC process automatically. Achieving will be achieving the first purpose on this work which is to have a script or an app who can compute the tidal volume without any manual work from the user.

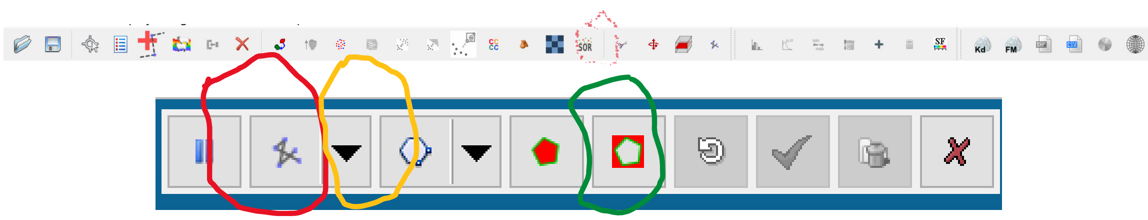
Lastly it will be interesting to investigate a way to get two cameras acquisition done without overlap. This could help reduce the error created by the filtering because a more complete image will be acquired.

ANNEX I

CC USER GUIDE

I.1 Segmentation

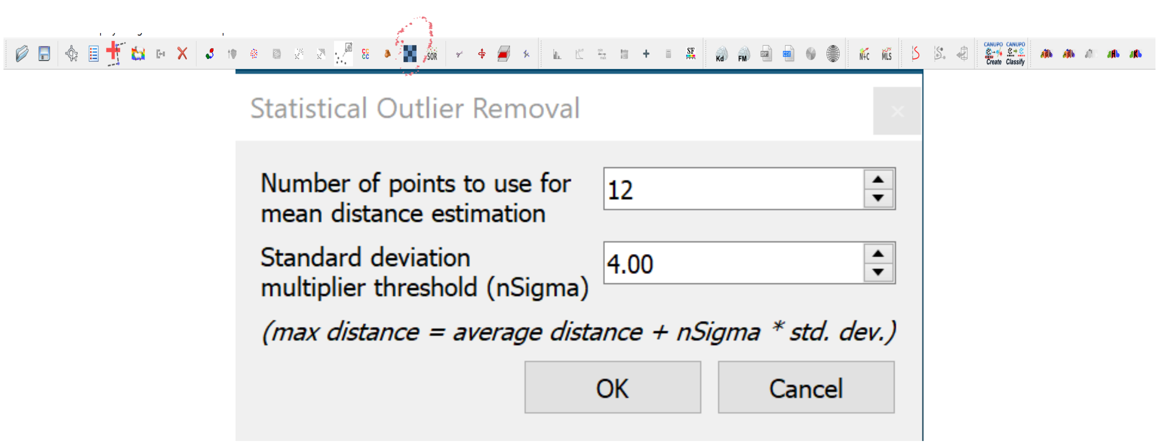
To segment it is necessary to select all the point clouds at the same time, then in the top menu of cloud compare we click on the scissor symbol, in the menu that is displayed we choose the type of selection (rectangular or polygonal circled in red). Just click on the relevant sections of the image, then the 'segment In' icon (full red polygon circled in yellow) in the displayed menu and finally the hook (circled in green) for the segmentation to be done.



I.2 Filtering

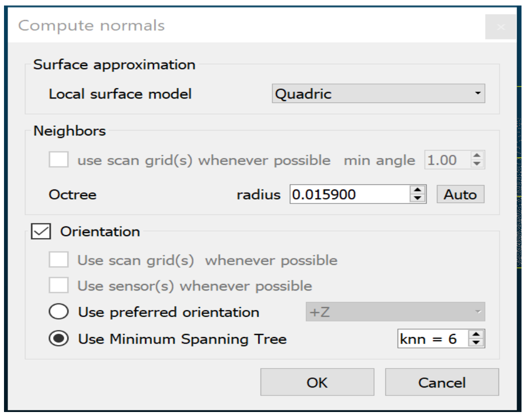
After the segmentation we filter all the frames to remove the noises. We use SOR (Statistical Outlier Removal) method. The filtering parameters are:

1. Select all frames and use the SOR tab of CC,
2. Fill in the values then ok.

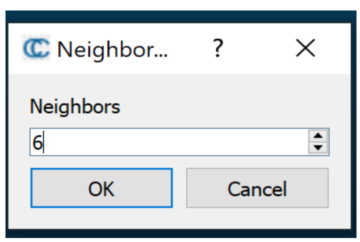


I.3 Normals computing

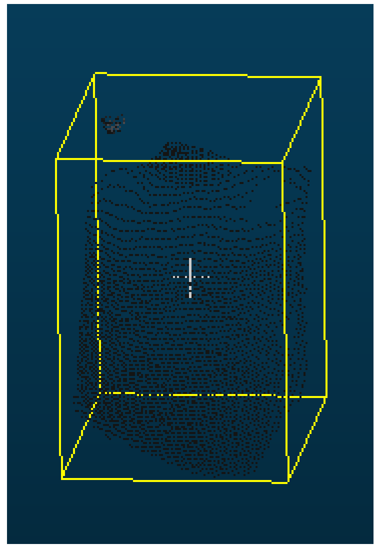
1. Select the frames resulting from the filtering,
2. Go to the Edit->Normals->Compute tab,
3. Enter the values as follows:



4. Orient the normals like this: Edit->Normals->Orient the normals->With the minimum spanning tree.

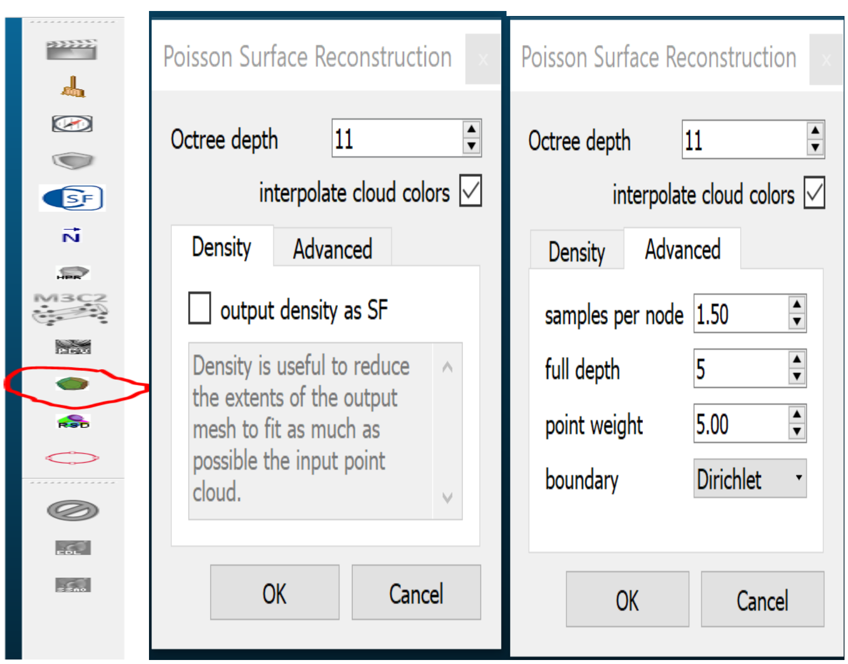


5. Do this only if you no longer see the frames or some of them in color: Select the frames concerned then Edit->Normals->Invert.

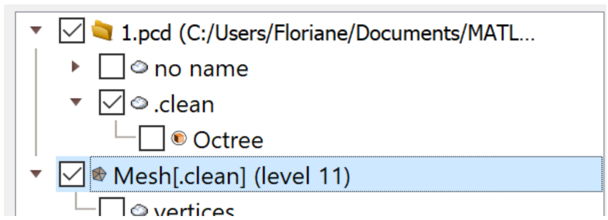


I.4 Reconstruction

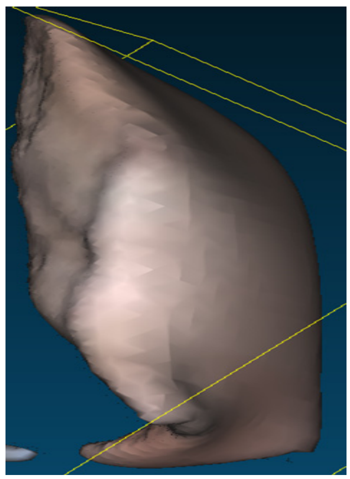
1. Select the frames one at a time
2. Go to the fish reconstruction tab on the right, fill in the values then ok



A Mesh object will then be created below the frames on the left.

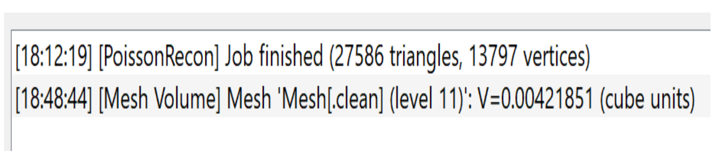


Visually it should look like this:



I.5 Volume calculation

1. Select the Mesh 1 at a time and do: Edit->Mesh->Measure Volume
2. The volume is displayed in the console.



ANNEX II

HOW TO DO THE SEGMENTATION

For the segmentation, I start under the clavicle, and I go right in the armpits to make sure I take in all the torso or the chest. For the sides where the ribs are, I make sure that I go right through the place where the patient's body is touching the bed. That way I'm sure that I'm taking the whole patient's body in it. And at the bottom, I tried to go under the abdomen. Sometimes some patients have on diapers because they are babies.

When the baby has diaper on, it can cover part of the abdomen. And when I get to that point, I kind of cut it blindly because I really want to try to get the abdomen in it, but if the diaper takes too much of the abdomen, I just caught right at the top of the diaper because it has an impact on the volume. When the patient is breathing, the abdomen is supposed to go up and down. But if the diaper is covering the abdomen, that difference of volume will not be seen clearly in images in our calculations. That's why the diaper has an impact on the volume calculation. And sometimes some babies are covered with a blanket. In that situation, I try as much as possible to avoid the blanket.

The perfect segmentation will be cutting right under the clavicle or on the clavicle, passing through the armpits, getting in the sides by cutting where the patient is contacting the bed and then go right below the abdomen that means that you're cutting where the thigh is connected to the body.

ANNEX III

PC FILE CONTENT

```
ply
format ascii 1.0
comment Created by CloudCompare v2.12 alpha
comment Created 2022-01-05T19:14:16
obj_info Generated by CloudCompare!
element vertex 4790
property float x
property float y
property float z
property uchar red
property uchar green
property uchar blue
end_header
0.324000 0.080000 -0.779000 84 66 70
0.326000 0.080000 -0.781000 85 65 70
0.330000 0.077000 -0.783000 77 59 64
0.330000 0.080000 -0.781000 81 62 68
0.333000 0.074000 -0.782000 74 56 63
0.333000 0.077000 -0.783000 76 58 64
0.333000 0.080000 -0.782000 79 61 66
0.336000 0.071000 -0.781000 76 58 64
0.336000 0.074000 -0.782000 74 56 62
0.336000 0.077000 -0.784000 74 56 62
0.336000 0.080000 -0.783000 77 59 64
0.339000 0.067000 -0.780000 80 62 66
0.339000 0.070000 -0.782000 78 61 65
0.339000 0.074000 -0.782000 76 59 63
0.339000 0.077000 -0.782000 77 60 64
0.339000 0.080000 -0.782000 76 59 63
0.340000 0.082000 -0.780000 79 62 66
0.342000 0.064000 -0.780000 84 66 68
0.342000 0.067000 -0.779000 84 64 67
0.342000 0.070000 -0.781000 82 68 71
0.342000 0.073000 -0.781000 82 67 70
0.342000 0.076000 -0.780000 84 67 71
0.342000 0.079000 -0.780000 82 65 69
0.343000 0.082000 -0.779000 81 64 68
0.345000 0.058000 -0.776000 85 69 75
0.345000 0.061000 -0.778000 84 67 71
```

0.345000 0.064000 -0.780000 84 66 68
0.345000 0.067000 -0.779000 85 67 70
0.345000 0.070000 -0.780000 85 71 74
0.345000 0.073000 -0.780000 86 72 75
0.345000 0.076000 -0.779000 84 70 73
0.346000 0.079000 -0.779000 83 69 72
0.346000 0.082000 -0.778000 82 65 69
0.348000 0.054000 -0.776000 85 71 74
0.348000 0.057000 -0.779000 85 71 74
0.348000 0.061000 -0.778000 85 70 73
0.348000 0.064000 -0.780000 85 68 72
0.348000 0.067000 -0.780000 88 71 75
0.348000 0.070000 -0.779000 91 74 78
0.348000 0.073000 -0.779000 89 72 76
0.349000 0.076000 -0.778000 86 69 73
0.349000 0.079000 -0.777000 84 67 71
0.349000 0.083000 -0.776000 83 66 70

ANNEX IV

INHALE AND EXHALE DETECTION, TIDAL VOLUME CALCULATION

```
for count = 1:frames
     ptCloud = pcread(fnames(count,:));
     moy(count,1)= count;
     moy(count,2)= mean(ptCloud.Location(:,3));

 end
 yy2 = smooth(moy(:,1),moy(:,2),45,'rloess');
[val, insp] = findpeaks(yy2);   %end of inhale

 yy2Inv = 1.01*max(yy2) - yy2;
 [val_peaks,exp] = findpeaks(yy2Inv);   %end of exhale

if size(insp,1)< size(exp,1)
    limit= size(insp,1);
else
    limit= size(exp,1);
end
tvolume = zeros(limit,1);
max_pos = zeros(limit,1);
min_pos = zeros(limit,1);

for i = 1: limit
    maxi = 0;

    if insp(i) -7 >= 1 &&  insp(i)+7 <= frames
        for j = insp(i)-7 : insp(i)+7
            if volume(j,2)> maxi
             maxi = volume(j,2);
             max_pos(i)= j;

            end
        % take the volume of the corresponding frames and calculate
the max
        end

    else

        max_pos(i) = insp(i);
        maxi = volume(insp(i),2);
     end


      if exp(i) -7 >= 1 &&  exp(i)+7 <= frames
        minimum = volume(exp(i) -7, 2);
```

```
        min_pos(i) = exp(i) -7;
        for k = exp(i) - 7 : exp(i)+7
        % take the volume of the corresponding frames and calculate
the min
            if volume(k,2) < minimum
                minimum = volume(k,2);
                min_pos(i) = k;
            end
        end

     else

        minimum = volume(exp(i),2);
        min_pos(i) = exp(i);
     end
    tvolume(i)= abs(maxi-minimum);
    disp(i);

    disp(find(volume(:,2)==maxi,1,'first'));
    disp(".......................");

    I = find(volume(:,2)==minimum);
```

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