CZECH TECHNICAL UNIVERSITY IN PRAGUE

FACULTY OF BIOMEDICAL ENGINEERING

Department of Biomedical Technology

Analyses of the connection between a high-frequency jet and a CPAP ventilators

Bachelor thesis

Study programme: Biomedical and Clinical Technology

Author of Bachelor thesis: Luka Ivaneishvili

Supervisor of Bachelor thesis: Ing. Petr Kudrna, Ph.D.

BACHELOR‘S THESIS ASSIGNMENT

I. PERSONAL AND STUDY DETAILS

Student's name:	**Ivaneishvili Luka**
Personal ID number:	**491342**
Faculty:	**Faculty of Biomedical Engineering**
Department:	**Department of Biomedical Technology**
Study program:	**Biomedical and Clinical Technology**
Branch of study:	**Biomedical Technician**

II. BACHELOR’S THESIS DETAILS

Bachelor’s thesis title in English:

Analysis of the connection between a high-frequency jet and a CPAP ventilator

Bachelor’s thesis title in Czech:

Analýza propojení vysokofrekvenčního tryskového a CPAP ventilátoru

Guidelines:

Analyse the possibility of combining a high-frequency jet ventilator and a CPAP ventilator for neonatal needs. Design and implement accessories to connect the high frequency jet ventilator to the CPAP generator. Use 3D printing methods. Test the proposed solution using a simulator in terms of work of breathing (WOB) and tolerance of spontaneous breathing activity.

Bibliography / sources:

[1] Moon, Joon Ho et al., Moon, Joon Ho et al. “Validation of a wearable cuff-less wristwatch-type blood pressure monitoring device.” Scientific reports vol. 10,1 19015. 4 Nov. 2020, doi:10.1038/s41598-020-75892-y, Scientific reports, ročník 10, číslo 1, 2020
[2] Tricia L. Gomella, Neonatology: management, procedures, on-call problems, diseases and drugs, ed. Sixth Edition, McGraw Hill Professional, 2009, ISBN 78-0-07-154431-3
[3] Tack, P., Victor, J., Gemmel, P. et al., 3D-printing techniques in a medical setting: a systematic literature review, BioMed Eng OnLine, ročník 115, číslo 15, 2016

Name of bachelor’s thesis supervisor:

doc. Ing. Petr Kudrna, Ph.D.

Name of bachelor’s thesis consultant:

Date of bachelor’s thesis assignment: **14.02.2022**
Assignment valid until: **22.09.2023**

....

doc. Ing. Martin Rožánek, Ph.D.
Head of department’s signature

prof. MUDr. Jozef Rosina, Ph.D., MBA
Dean’s signature

DECLARATION

I hereby declare that I have completed this Bachelor thesis with the topic "Analyses of the connection between a high-frequency jet and a CPAP ventilator" independently, and that I have attached an exhaustive list of citations of the employed sources to the Bachelor thesis.

I do not have a compelling reason against the use of the Bachelor thesis within the meaning of Section 60 of the Act No.121 / 2000 Coll., on copyright, rights related to copyright and amending some laws (Copyright Act).

In Kladno 12.05.2022 …..……………..…………………...

Luka Ivaneishvili

ACKNOWLEDGEMENTS

I would like to use this opportunity to express my gratitude to my supervisor, Ing. Petr Kudrna, Ph. D, for all the professional guidance he has provided throughout my bachelor thesis work.

Title of thesis:

Analyses of connection between High-frequency jet and CPAP ventilators.

Abstract:

Premature birth can be a cause for different diseases, to face this problem there are already existing ventilation supports in the clinical practice. For example, nowadays one of the approaches to treat neonates with respiratory diseases is using high-frequency jet ventilation in tandem with conventional ventilator. HFJV delivers high frequency breaths with the tidal volumes compared to the anatomical dead space, while expiration is passive due to chest recoil. Meanwhile conventional ventilator provides continuous pressure of air inside the lungs, keeping the respiratory system from collapsing. Continuous positive airway pressure ventilator can provide everything CV does in this ventilation circuit. Both respiratory support systems require a connector for the LifePort adapter of the HFJV to function. Since already existing versions of this connector for CPAP or CV and HFJV have some disadvantages concerning the high internal volume, it adds up to increasing the total dead space of the ventilation circuit. Increased total dead space consecutively increases the work of breathing of the patient. In case of prematurely born neonates, high work of breathing can cause a respiratory stress and develop into more severe complications. Therefore, the main aim of the thesis was to design the connector that would tolerate spontaneous breathing of the patient and have the least dead space possible, to reduce the total dead space of the ventilation circuit. This would lead to the reduction of the work of breathing of the patient. Multiple solutions of the problem were introduced, printed, and tested on the lung simulator. Work of breathing of the patient was evaluated and compared to already existing CV circuit to prove the efficiency of offered designs.

Key words:

Continuous Positive Airway Pressure, CPAP, Imposed work of breathing, IWOB, High frequency jet ventilation, HFJV, Infant flow generator

Table of Contents

List of abbreviations

Abbreviation	Importance
HFJV	High frequency jet ventilation
CPAP	Continuous positive airway pressure
NCPAP	Nasal continuous positive airway pressure
CV	Conventional ventilator
HFOV	High frequency oscillatory ventilation
MAP	Mean airway pressure
PEEP	Positive end expiratory pressure
PIP	Peak inspiratory pressure
WOB	Work of breathing
WOB_{TOT}	Total work of breathing
WOB_{P}	Physiological work of breathing
IWOB	Imposed work of breathing
CDP	continuous descending pressure

List of symbols

Symbol	Unit	Importance
W	Joule	work
P	Watt	pressure
V		Volume
P_{ETT}	cmH_2O	Pressure at endotracheal tube

1 Introduction

Developing baby undergoes important phases throughout the pregnancy, including final weeks. Premature birth is when an infant is born before the due date. This is caused by different reasons: multiple pregnancies, infections, diabetes, chronic diseases etc. One of the most common health complications caused by premature birth is incomplete development of respiratory system, leading to respiratory distress syndrome (RDS) [1] [2]. In the human, if spontaneous breathing is disrupted by lung diseases like respiratory distress syndrome, there is a need for respiratory a support for the patient.

Nowadays, mechanical artificial lung ventilation is used for the treatment of the respiratory failures for all type of patients. However, when this method fails, it is possible to use alternative methods of ventilation, so called High frequency ventilation (HFV). This ventilation system applies to all weight groups of the patients, especially the smallest weight group, neonates. Two of high frequency ventilation techniques - High frequency oscillatory ventilation (HFOV) and High frequency jet ventilation (HFJV) dominates the clinical practice in neonatal care. Nevertheless, high frequency jet ventilator happens to be more suitable choice in terms of treating neonates [3]. HFJV provides breaths of high frequency with the tidal volumes compared to the anatomical dead space, while expiration is passive. Supplying lung with small volumes of air reduces risk of lung injuries [4]. HFJV cannot operate without another device providing positive end expiratory pressure (PEEP) to keep the respiratory system from collapsing. For providing continuous pressure in the lungs, Conventional ventilator (CV) or Continuous positive airway pressure device (CPAP) is used in the ventilation circuit. CPAP device can do all the things that conventional ventilator can. Moreover, CPAP happens to be more convenient and affordable choice [4].

Therefore, to address the problem of safe ventilation of prematurely born infants, there needs to be a connection of the high frequency jet and CPAP ventilator. For the respiratory support system to work there needs to be connector of the CPAP ventilator to the Life port adapter of the HFJV. The connector needs to provide continuous flow of air to the patient during inspiration and redirect the flow of air to the outlet to atmosphere, during expiration. Offered solutions for the design of the connector needs to be tested on the lung simulator and work of breathing needs to be evaluated to observe the efficiency of the offered solution.

2 Overview of the current state

Various studies have been done about the ventilation support for newborns with respiratory diseases. Nowadays, we can see the progress we have made throughout last decades, decreasing the amount of death in newborns from 5.0 million in 1990 to 2.4 million in 2019 and to 1.2 million in 2022. This is a considerable progress, but more lives can be saved with optimized less costly support if available [5].

There are different of reasons for premature birth of children, some related to the health of the mother. For example, pregnant women with diabetes, heart diseases, high blood pressures and infections are more likely to experience premature birth. Factors like poor nutrition or smoking before and during pregnancy plays a role in this as well. The normal due date of the child is 40 weeks, the earlier the birth the more potential health complications can occur. Usually, it happens before the 37 weeks of infant's development in mother's body, the time when the lungs and brains are supposed to finish maturing [6]. This leads to the main problem with premature babies, underdeveloped respiratory system. This poorly developed respiratory system leads to common respiratory diseases including apnea of prematurity, respiratory distress syndrome (RDS), transient tachypnea (TTN) of the newborn [7]. For example, RDS is the condition when there is not enough surfactant in the lungs of the infant to maintain proper oxygenation of the body. Surfactant also helps increasing pulmonary compliance, recruitment of the collapsed airways and avoiding atelectasis by decreasing the surface tension of the alveoli. Dealing with not RDS in the infant can be dealt by delivering surfactant through a catheter placed in the trachea, however this kind of procedure for an infant can be risky, since it can cause further respiratory system damage [8]. Potential complications caused by the administration of the surfactant inside the neonate lungs can be, ETT obstruction, transient bradycardia, pulmonary hemorrhage, or pneumothorax. To avoid these complications research has been done on ventilation systems that can provide safe and efficient respiratory support for the neonates with respiratory diseases [8].

Traditionally, mechanical ventilation (MV) is used to treat patients with respiratory diseases to decrease their work of breathing until, they no longer need ventilation. MV can be characterized by the tidal volume and frequency close to the spontaneous breathing. However, some clinical studies shows that the mechanical properties of the respiratory system of the patient, effects the parameters of the ventilation therefore effect the efficiency of the MV as well [9]. Airflow resistance inside respiratory system or lung and chest compliance varies during different respiratory diseases. For example, RDS can directly affect the efficiency of mechanical ventilators. For example, ventilating less compliant lung of an infant on mechanical ventilation can be risky, due to the less compliant parts of lungs not being able to expand normally during inspiration, leading to lung fractures [9].

Therefore, when MV fails to support the patient, we can use alternative method of ventilation called high frequency ventilation (HFV). HFV is different from all methods of mechanical ventilation. HFV belongs to nonconventional way of ventilation, characterized by smaller tidal volumes than MV together with higher frequencies than spontaneous breathing. Tidal volumes ranges are comparable to the anatomical dead space of the infant's respiratory system and frequencies can be described as multiple times higher than normal breathing frequency of the human [9]. From High frequency ventilators two of them high frequency jet ventilator (HFJV) and high frequency oscillatory ventilator (HFOV) dominates in the clinical care [3]. Main difference between the two is that during HFOV both inspiration and expiration is active, meaning air is pushed in and sucked out of the lungs. While, during HFJV only inspiration is active, and expiration is passive caused by the elastic recoil of the chest [4].

When you are using the HFV you need to provide the continuous positive pressure inside the lungs. Positive-end expiratory pressure (PEEP) is a positive pressure of air inside the lungs at the end of the expiration. Typically, PEEP is used to keep alveoli from collapsing, when applied alveolar volume and pressures increased resulting in reopening or stabilizing the collapsed or unstable alveoli. There are different ways to provide PEEP inside the body. Conventional ventilator has been used in tandem with high frequency ventilators to provide the continuous pressure of air inside the lungs [4]. However, Continuous Positive Airway Pressure or CPAP ventilator can fulfill all the requirements for this experiment as well. It can deliver continues PEEP to the lungs, while being more convenient choice.

2.1 High frequency ventilation

High frequency ventilation (HFV) is a therapeutic technique, using the fast supply of small breaths around the size of the normal anatomical dead space of the patient. High frequency ventilation is different from all other types of traditional mechanical ventilators. Traditional ventilators try to mimic the normal breathing of the patient, while HFVs instead of doing the same, it delivers air in a rather different manner [10]. In more detail, HFV is to reduce the amount of volume of the air sent inside the patient's lungs, in order to reduce the risk of lung tissue fractions or injuries and increases the frequency of breaths to mantain the normal oxygenation inside the body. This is achieved by first assuming that the relationship between the CO_2 elimination(V_{CO2}) is proportional to the minute Volume(V_{min}) or in another words Tidal Volume(V_T) times the frequency [9]:

$$\dot{V}_{CO2} \propto \dot{V}_{min} = f * V_T \qquad (2.1)$$

During HFV this relationship has to be modified to Formula 2, to reflect the increased contribution of the tidal volume. Here CO_2 elimination is defined by the equation [9]:

$$\dot{V}_{CO2} \propto f^a * V_T^b \qquad (2.2)$$

Where the exponent b is greater than the exponent a. Here a has been estimated to be between 0.75 and 1.24 and b is between 1.5 and 2.2 [11]. For practical reasons this relationship is accepted as Formula 3 [9]:

$$\mathrm{V}_{CO2} = f * V_T^2 \qquad (2.3)$$

Therefore by exploiting this relationship, it is possible to provide high frequency ventilation breaths to the patient, and have adequate alveolar ventilation by using tidal volumes close to the anatomical dead space of the lungs. A few advantages of using HFV can be the reduction of risk of volutrauma and prevention of ventilator-induced lung injuries. Also, maintaining constant level of alveolar inflation, preventing the inflate-deflate cycle and improving oxygenation. Nowadays, there are couple of different HFV that is used in the clinical practice [10]:

- High-frequency oscillatory ventilation (HFOV)
- High-frequency positive pressure ventilation (HPPV)
- High-frequency jet ventilation (HFJV)
- High-frequency percussive ventilation (HFPV)

As mentioned before HFV can be used to treat all weight groups of patients. While treating neonates there are couple of indicators that can be used to decide when using HFV is adequate, For example [10]

- Persistent pulmonary hypertension
- Acute respiratory distress syndrome
- Pulmonary interstitial emphysema
- Meconium aspiration
- Pulmonary hypoplasia

2.1.1 HFOV and HFJV

HFOV is one of the most common methods of HFV. It is most often used as a rescue strategy when conventional ventilation fails in severe ARDS. In neonatal patients, HFOV can be used in premature lungs as the first line to prevent lung injury by conventional ventilation. In this technique, the tidal volume set is less than dead space, and respiratory rates are very high, ranging from 3-15 Hz [10]. During HFOV pressure oscillations inside the patients breathing circuit are caused by oscillating piston or diaphagm. During HFOV primary setting is mean arterial pressure (MAP) and flow oscillates around the set MAP. During HFOV both inspiration and expiration is active, air is first pushed inside the lungs and then sucked out. The basic configuration of the HFOV is shown in Figure 2.1 [9].

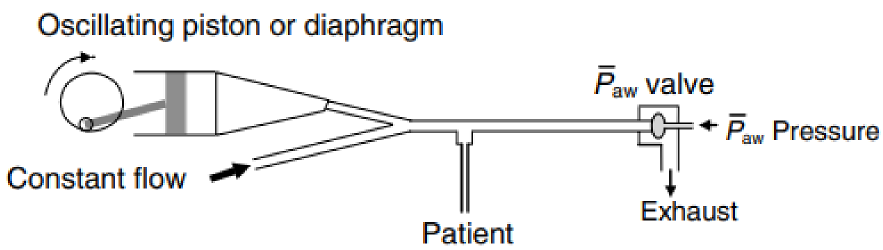


Figure 2.1: Basic design of HFOV [9].

Figure 2.2, can be observed to compare the breaths of conventional and high frequency ventilators. Here, it is possible to observe the shapes of HFOV and HFJV signals compared to conventional ventilator breath. For HFOV the pulse off the signal is a sinusoudal wave, this allows to move the air in and out of the lungs, therefore both inhalation and exhalation is active. HFJV on the other hand has completely different waveform, it can be described as the peaks of pressures, only used to send the air inside the body and leave the exhalation to the patient (passive exhalation, due to the elastic recoil of the chest). It is also visible how signals from high frequency ventilators dampen as they reach the distal parts of the respiratory system. Furthermore, frequency difference between ventilators is observed, it is visible that for only one wave of the conventional ventilator there are several number of waves from HFJV and HFOV present at the same time [9].

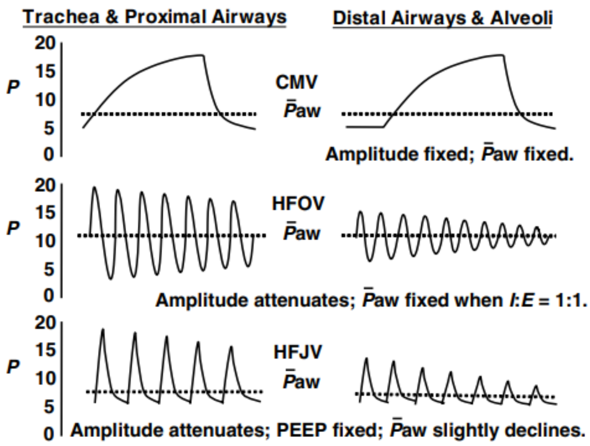


Figure 2.2: HFJV, HFOV and conventional ventilator pressure waveforms [9].

HFOV is often used with high mean airway pressures. Higher mean airway pressures can stabilize surfacant-deficient alveoli but also be counterproductive when airleaks and hemodynamic problems are present. As discussed before, during HFOV gas enters and leaves the lungs more symetrically than during HFJV. Inspired gas flows slowly, but expired gas comes out faster in HFOV compared to the ‘‘jetted‘‘ inspirations and passive exhalations of HFJV. The ‘‘active‘‘ expirations enables faster rates to be used, without encountering gas trapping. However this is only possible when using high mean arway pressure. If lower MAP is used actve exhalation can lead to airway collapse or to ‘‘choke points‘‘, since intraluminar pressure drops faster than the pressure in the alveoli surrounding the airways [4].

During HFJV gas trapping is avoided by using shortest possible inspiratory times of 0.020 seonds, even during the lowest frequency rate 240bpm. Short Inspiratory time minimizes the inpirattory gas flow and I:E ratio, leaving more time for exhalation. If the gas traping will still occure during HFJV, monitored PEEP value on high frequency jet ventilator will be higher than the set PEEP value on CPAP device [4].

Table 2.1: Charachreristics of HFOV and HFJV.

HFOV	**HFJV**
Active exhalation (ventilator pulls the air back out)	Passive exhalation (Chest recoul pushes air out)
Fixed I:E 1:2.5-1:1	Variable I:E 1:12-1:3
Tv dependent on frequency	Tv independent on frequency
Frequency 3-15 Hz	Frequency 4-11 Hz

In the Table 2.1, It is possible to observe the basic charachteristics of HFJ and HFO ventilators. HFJV unlike HFOV has frequency range of 4-11 Hz. Also I:E ratio is changing from 1:12 till 1:3 effecting the time patient has for the expiration. Furthermore in the HFJV, Tidal volume is independent from the frequency while in HFOV tidal volume depends on the frequency of the provided breaths [9].

2.1.2 High frequency jet ventilation

As discussed above, high frequency ventilation (HFV) is a therapeutic technique, using the fast supply of small breaths. This kind of device can be used in case of newborns respiratory support because of the less stress on the alveoli during the breathing.

Nowadays one of the only used HFJV is Bunnell life pulse fan (figure 2.3). This device sends air with small tidal volumes of 2 to 5 ml/kg, with frequency of frequency of 240-660bpm (4-11 Hz) to the life port and to the lungs of the patient, while exhalation is passive. PIP (peak inspiratory pressure) range is from 0.8-5kpa, and inspiratory time range varies from 0.02-0.034 s [12].

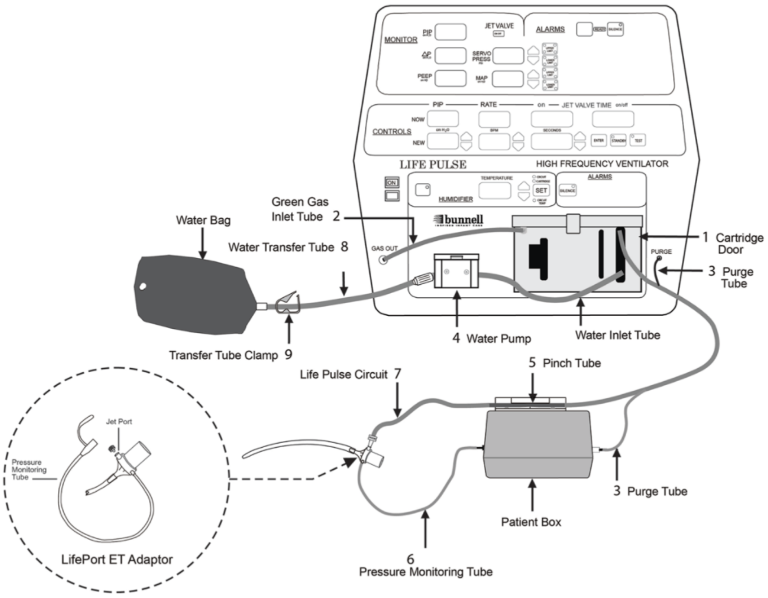


Figure 2.3: Bunnell life pulse high frequency ventilator full arrangement taken from [13].

In the Figure 2.4, you can see the LifePort adapter for HFJV. It consists of pressure measuring tube, Jet port for inspired air and the 15mm head to provide constant PEEP inside the lungs by the help of conventional or CPAP ventilator. LifePort adapter connects to the endotracheal tube and ventilates the patient invasively.

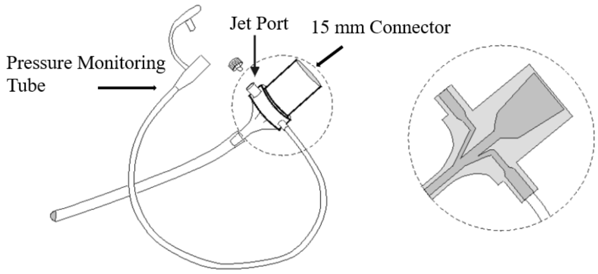


Figure 2.4: LifePort Endotracheal tube adapter taken and modified from [4].

Life pulse is pressure-limited and time-cycled like most simple conventional ventilators. It becomes Jet ventilator only when it reaches the life Port ETT adapter (Figure 2.4). In the life Port adapter, we have a 15mm connector, jet port and pressure monitoring tube for PIP (peak inspiratory pressure). This pressure monitoring tube connects to small plastic "Patient box" which also creates high frequency ventilation breaths. The placement of the box is as near to patient's airway to eliminate the dampening of the signal. When the air is injected to the jet port, it swirls down the center core of the airways, where the resistance is lowest. During the expiration gas flows out passively, also seeking the path of low resistance, this case it moves out along the airway walls around the highly accelerated inspired gas. You can observe this phenomenon in Figure 2.5 [4] [14] .

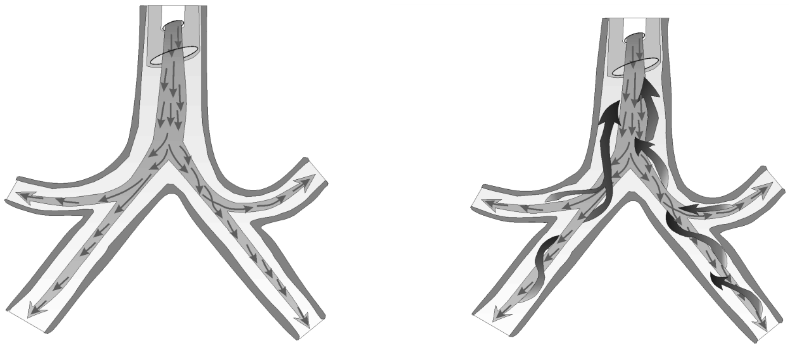


Figure 2.5: Picture on the left is Jet sending the gas into the lungs. Picture on the right shows the exhaled gas moving up along the airway walls [4].

HFJV provides more effective ventilation and comparable oxygenation at mean airway pressures that are considerably less than those required during conventional ventilation. However, since smaller tidal volumes are effective with HFJV, higher PEEPs can be used safely. Small tidal volumes also mean that lung compliance has little influence on gas distribution within the lungs [4].

2.1.3 Gas exchange in high frequency jet ventilation

When talking about HFJV, several studies have shown that the CO_2 elimination during this type of ventilation is proportional to frequency times the tidal volume squared (V_T^2). It is important to note that during high frequency jet ventilation real determinant of the tidal volume is ΔP or the difference between the set PIP and set PEEP. For example, there can be a case where PEEP can be raised in order to reduce the tidal volume to increase the low level of partial pressure inside the arterial blood ($PaCO_2$) [4].

$$\mathrm{V}_{CO2} \approx f * V_T^2 \qquad (2.3)$$

$$\Delta P = (PIP - PEEP) => V_T \qquad (2.4)$$

2.2 Continuous Positive Airway Pressure

Continuous positive airway pressure is a form of positive airway pressure ventilation in which a constant level of pressure greater than atmospheric pressure is continuously applied to the upper respiratory tract of a person [15]. Nasal continuous positive airway pressure (CPAP) is a noninvasive form of respiratory assistance that has been used to support spontaneously breathing infants with lung disease for nearly 40 years. Modern NCPAP devices differ from one another by the manner they provide the air to the patients. CPAP devices usually contain: a flow source of heated and humidified ventilation mixture, a device to generate the CPAP pressure (exhalation valve, fluidic chamber, fluid generators) and interface allowing patient connection (Nasal or orofacial) [16]. Currently, in the clinical practice you meet following NCPAP devices: Variable flow, Constant flow, and Demand flow. According to this classification you can discuss all types of CPAP devices in use nowadays [17]. Demand flow NCPAP devices are special group, which can sense the onset of inspiration and allows increased airflow towards the patient. With the end of inspiration flow of air lowers [18]. In variable flow NCPAP devices, when resistance occurs, the flow of air changes to keep the constant pressure inside the lungs. Constant flow CPAP is a form of positive airway pressure ventilation in which there is a constant flow of air, pressure is created by adjusting the exhalation valve orifice. This method of pressure generation is used mainly by neonatal pulmonary ventilators, which control pressure by the valve with variable resistance [16].

Figure 2.6: Inspire rPAP (left) and infant flow SIPAP (right) [19] [20].

Two very good options of devices to use as CPAP device to provide continuous pressure of air in the lungs of the infant can be SiPAP (Cardinal Health, Dublin, Ohio) and rPAP (Inspiration Healthcare, Leicester, England), you can see in Figure 2.6. The first mentioned option is a noninvasive form of CPAP breath support for neonatal intensive care units. It is possible to set two separate levels of CPAP on this device. One for the inspiration and one for the expiration. Second option is to use Inspire rPAP, which also has two levels PIP and CPAP [21] [22].

2.3 ASL simulator

ASL 5000 (IngMar Medical, Pittsburgh, PA, USA) is a computer controlled active lung simulator popular for many applications including design and testing of ventilators and ventilatory monitors, testing of ventilatory modes and interaction of a ventilator with the patient's lungs, education, training of medical staff and other purposes. ASL 500 is known for its ability to simulate a wide range of patients - from newborns to adults. The ASL 5000 can breathe spontaneously even with artificial lung ventilation and can be used with any ventilator in different types of ventilation modes. This highly accurate and versatile device is a first-class choice for product development and testing of ventilators, CPAP devices, aerosol drugs and other respiratory devices. ASL 5000 Can be connected to any ventilator, just like a real patient [23]. ASL 5000 can be seen in Figure 2.7.

Figure 2.7: ASL 5000 lung simulator taken from [24]

2.4 Work of breathing

WOB or work of breathing is widely used term and it represents the energy spent to inhale and exhale the breathing gas. Several components contribute to the total work of breathing [25]:

Elastic work

- Work done to overcome elastic recoil of the lung.
  - This increases with increasing inspiratory volume
- Work done to overcome elastic recoil of the chest

Resistive work

- Work done to overcome tissue resistance otherwise referred to as viscous resistance:
  - Chest wall resistance
  - Lung resistance
  - Displacement of abdominal organs

Work done to overcome airway resistance:

- Airway resistance
- Resistance of the airway devices and ventilation circuits

Figure 2.8 is the graphical representation of the total wok of breathing experienced by the patient while breathing with the ventilation circuit.

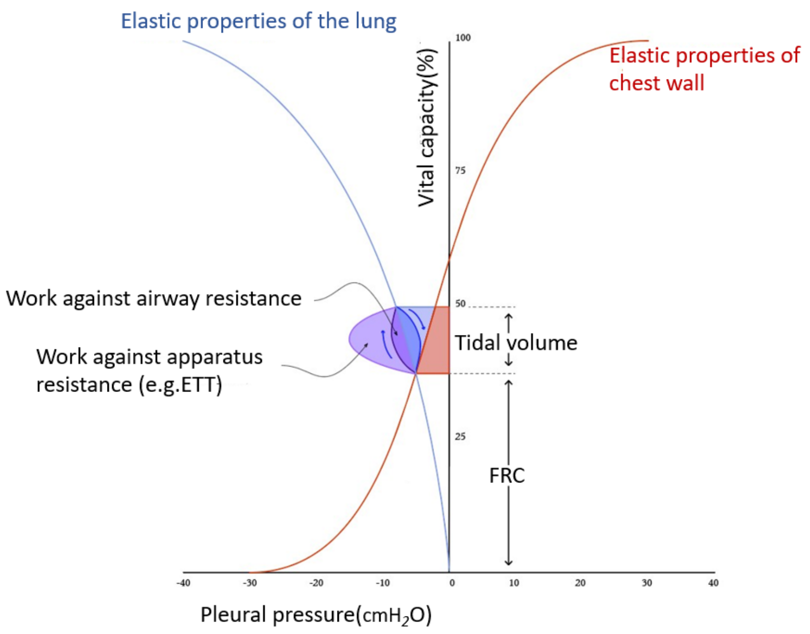


Figure 2.8: Graphical representation of total WOB of normal patient on ventilation system taken from [25] .

In the Figure 2.8, it is visible that total work of breathing is equal to elastic work together with the resistive work of inspiration and expiration together with the work to overcome the airway resistance of the ventilation circuit. Total respiratory muscle work of a patient breathing with a respiratory support is equal to imposed and physiological work together (Formula 2.4). Imposed work of breathing is the additional resistive workload from the breathing circuit and the ventilator's flow delivery system during the ventilation. [26]

$$WOB_{TOT} = WOB_P + IWOB \quad (2.4)$$

In the physical sense, work is performed when transmural pressure (P) changes the volume (v) of a distensible structure (Formula 5). Most often expressed as Joules per liter (J/l) [26].

$$W = \int P \times dV \qquad (5)$$

Imposed Work of breathing during the ventilation mode of continuous fresh air supply is directly related to the integral of pressure measured at the end of the endotracheal tube (P_{ETT}) times the volume change (Formula 2.5) [26].

$$IWOB = \int P_{ETT} \times dV \qquad (2.5)$$

The imposed WOB is directly related to the breathing-related difference between the set mean airway pressure (MAP) and the $P_{ETT.}$ the greater the difference, the greater the imposed WOB and thus patient effort.

2.4.1 Work of breathing during HFV

When talking about the work of breathing during high frequency ventilation, there are only couple of studies done. Bench study [27] that will be discussed here is about the imposed work of breathing during HFOV. Here computer-controlled piston driven test lung was used to simulate a spontaneously breathing patient. The test lung was connected to a HFOV by Endotracheal tube. The inspiratory and expiratory airway flows and pressures at various places were sampled. The spontaneous breath rate and volume, tube size and ventilator settings were simulated as representative of the newborn to adult range. The fresh gas flow was set at low and high level and Imposed WOB was calculated using the Campbell diagram (Figure 2.9). Here you can see the start of inspiration A and end of inspiration B. The grey area represents the imposed inspiratory work of breathing. [26]

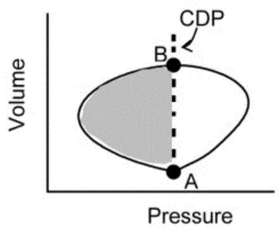


Figure 2.9: Modified Campbell diagram where A is the start of inspiration and B is the end of inpiration [26].

$$\mathrm{IWOB} = \sum(CDP - MAP_{ETT}) \cdot \Delta V \quad (2.7)$$

Using Formula 2.7, it is possible to calculate imposed work of breathing of the patient. Here CDP is continuous distending pressure or set MAP level on SensorMedics oscillator, and MAP_{ETT} is mean airway pressure in the test lung [26].

2.5 Problem definition

As mentioned before, HFJV is traditionally used in tandem with conventional ventilator in the clinical practice. In this bachelor work, this existing version of respiratory support circuit will be modified. Instead of conventional ventilator, CPAP device will be used to deliver the constant PEEP value to the lungs. This requires the connection of the CPAP device to the LifePort adapter of HFJV. LifePort adapter, consists of three things: pressure monitoring tube, Jet port for the inspired air from the HFJV and the 15mm connector. The main goals of this thesis were to design the connector that would fit on top of the LifePort 15mm head, provide the continuous flow of air from the CPAP or device to the lungs, allow the passive exhalation of the patient to the atmosphere and have least amount of inner dead space. Least amount of total dead space in the ventilation system and ability for passive exhalation not against but with the continuous flow of air from the CPAP device can reduce the expiratory work of breathing of the patient. Infant flow geometry was used to design the following connector for the connection of CPAP and HFJV ventilators. In More about the design can be seen in the following chapter.

Figure 2.10 represents the full arrangement of the HFJV connection with CPAP device with the infant flow generator. Pressure monitoring tube from the LifePort adapter is split in two and send to both devices. Infant flow generator connected to the LifePort has two connectors, one for the continuous flow of air towards the lung from the CPAP machine and second open to the atmosphere for the exhalation of the patient.

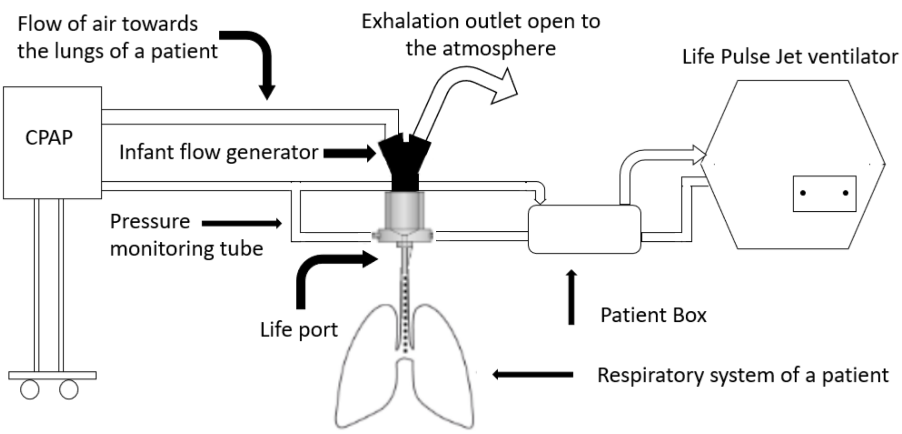


Figure 2.10: Connection of CPAP and HFJV ventilators.

2.6 Science behind the design

The design of the infant flow generator was done based on the “fluidic-flip” mechanism. The principle of the technology is that according to Bernoulli effect gas gets directed to each nostril of the valve and according to Canada effect the flow of the air can flip during the patient's exhalation and exit the generator via expiratory outlet. This approach allows the constant flow from the CPAP device to the patient, as well as the passive exhalation, during which the patient does not have to exhale against the constant flow of air, since it’s redirected to the outlet (Figure 2.11). This can consecutively lead to less work of breathing required from the infant. In the Figure 5 you can see (a) Inhalation: Air flows towards the inspiratory branch, (b) Exhalation: Jet flow diverted down expiratory branch [28].

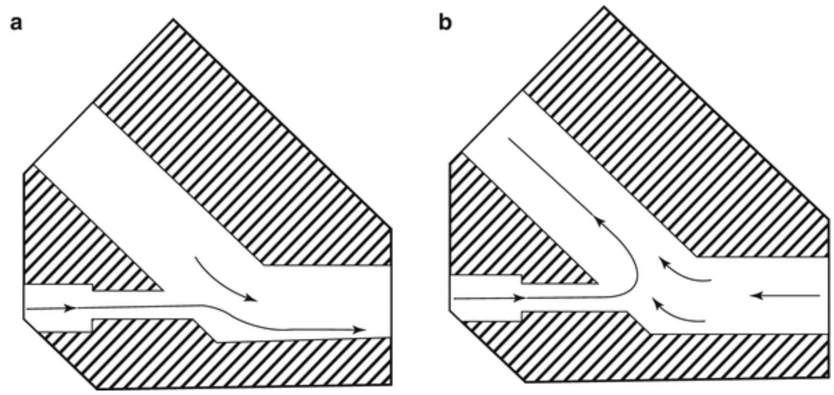


Figure 2.11: Schematic representation of the "fluidic flip' technology in the infant flow geometry. Inspiration(a) and Expiration (b) [29].

One of the studies used to design the connector for the Life port adapter, showed the functionality of abovementioned Infant Flow generator. The original Infant Flow geometry was used with simulated infant breathing at three different CPAP levels. Levels used for a constant driving flow were 1.4L 2.7 L/min and 4.1 L/min) driver flow of dry air at 25°C [30].

These were selected to generate CPAP at a range of approximately 1.5 to 6.5 cm H2O. These flows also include driver flows below, similar to and above inspiratory flow maximum 3.0 L/min. The patient breathing pattern was recorded on a 3.4 kg healthy infant with 30.8 mL tidal volume, inspiratory/expiratory ratio of 0.63 and an inspiratory volume flow max of 6.0 L/min. The volume flow to one generator was set to half of this recorded breath [30].

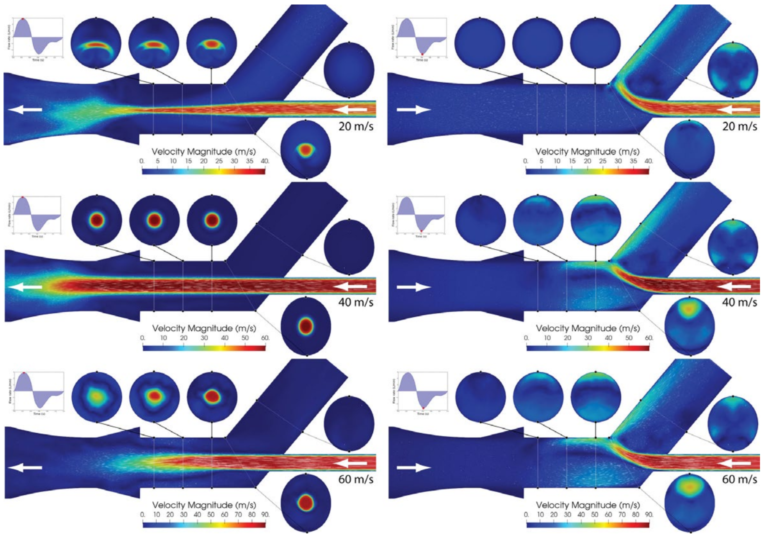


Figure 2.12: Inhalation (left) and exhalation (right) at three levels of driving flow [30].

The experiment demonstrated and resolved the phenomena in the infant flow geometry. The main continious flow of the gas was redirected to the outlet during expiration on all three levels of the flow (Figure 2.12). It had to be mentioned that only 40m/s and 60m/s were most relevant clinical simulations, since CPAP generated pressure and 20m/s is lower than what is normally used [30].

The mentioned geometry was used to create the multiple versions of the connector for the LifePort adapter for HFJV. In the chapters below, it is possible to observe the metods of designing the infant flow generator in the 3D modeling software, 3D printing of offered solutions and their verification with the lung simulator.

3 Aims of the diploma thesis

As mentioned before the ventilation circuit has downfall, because of its large size. The further increase of the dead space inside the connector leads to increased amount of total dead space. Increased dead space can affect the work of breathing of the patient during the expiration. Therefore, the aim of this bachelor thesis will be split in following parts.

- Designing the 3D model of the connector for HFJ and CPAP ventilators in Fusion 360 software using the Infant flow geometry.
- Connector needs to connect to the LifePort adapter for HFJV.
- Connector needs to provide the continues flow of air from the CPAP device (constant PEEP value) and allow passive exhalation of the patient to the atmosphere.
- Designed connector needs to have the least dead space, to reduce the total dead space of the ventilation support system.
- 3D printing of the offered solutions for the connector.
- Verification of the printed connector with a lung simulator and a ventilator.
- Evaluation of Imposed work of breathing of the patient with offered design solutions.

4 Methods

4.1 Methods of design inside the 3D modeling software

The design of the infant flow generator was designed in the Fusion 360 3D modeling software. Fusion 360 is widely used 2D and 3D modeling software for different purposes. In this case software was used to create the connector for LifePort adapter for HFJ and CPAP ventilators. Connector needed to fulfill following requirements

- Connect to the Life Port ETT adapter 15mm head.
- Allow continuous flow from the CPAP device.
- Provide passive exhalation through the outlet open to the atmosphere.

To design the mentioned connector, we used 3D modeling setting from the software and start building the 2d wireframe for the design. At first the circles of different sizes were drawn. With the “Extrusion” setting inside the Fusion 360 software it was possible to create the 3D object from the drawn circles. By extruding space between the two circles of 15.5mm and 17.5mm it was possible to create the cylinder that would connect to the life port adapter. Next step was to create the certain angle of 55^{o} between the inspiration and expiration tubes. This was required for the “Fluidic flip” mechanism to function. This was created by using “Plane at an angle” and “Offset plane” settings inside the software. For expiration tube there had to be two circles drawn out and extruded. For the inspiration tube the same applied here, circles of different sizes were used to create the outer cover and the connector for the CPAP ventilator. For the inner geometry of the inspiration tube, the size of the inner diameter was decreasing as it was getting closer to the patient. This geometry was created to accelerate the continuous flow of air from the CPAP machine towards the patient. The "Loft” setting was used to create this geometry and to connect the cylinders of different sizes.

4.2 Design Solutions Version 1

Fist set of proposed solutions for the connector of the HFJ and CPAP ventilator can be observed in the Figure 4.1 and 4.2. In the Figure 4.1, design A and B has the inner diameter of 15.2mm and outer diameter of 17.2mm for the connection to the LifePort adapter. Design A has the 7mm exhalation tube, with the wall size of 1mm. For the inspiration, it has 8 mm outer diameter tube where the inner diameter is decreasing in size from 7mm till 1.5mm. Design B has the expiration tube of 11mm outer and 9.5mm inner diameter. For inspiration tube, design B has 9 mm of outer diameter with the inner diameter decreasing in size from 7mm till 2mm.

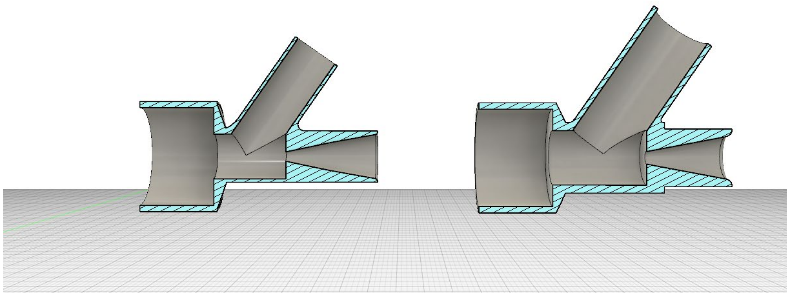


Figure 4.1: Version 1 offered design solutions A (left) and B (right), for infant flow generator (sectional view).

In the design C and D we have the same size of all dimentions (Figure 4.2). Only difference is the cut near the inspiration tube nozzle. This was created to direct the air more towards the patient. Connector for the LifePort adapter has 15.3mm inner and 17.3 outer diameter. Exhalation tube has 11mm outer and 9mm inner diameter. Inspiration tube has the outer diameter of 8mm and the inner diameter is decreaasing from 6mm till 2.6mm.

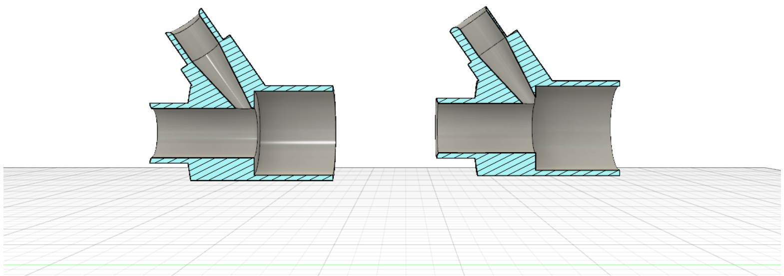


Figure 4.2: Version 1 offered design solutions C (left) and D (right), for infant flow generator(sectional view).

In the Figure 4.3 and 4.4, it is possible to observe the connection of the designed solutions with the LifePort adapter for HFJV. The lifePort adapter was designed in the Fusion 360 software as well, to demonstrate the functionality of designed connectors.

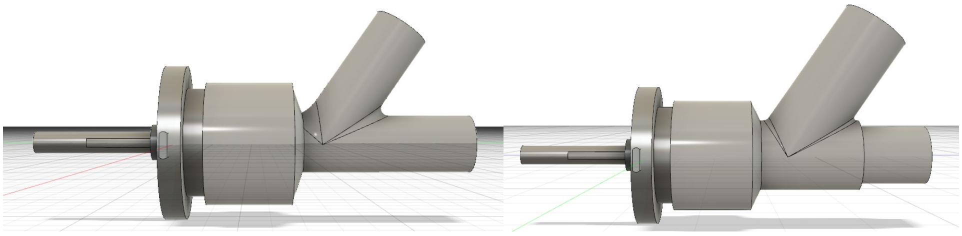


Figure 4.3: Version 1 design solutions A and B, connected to the life port.

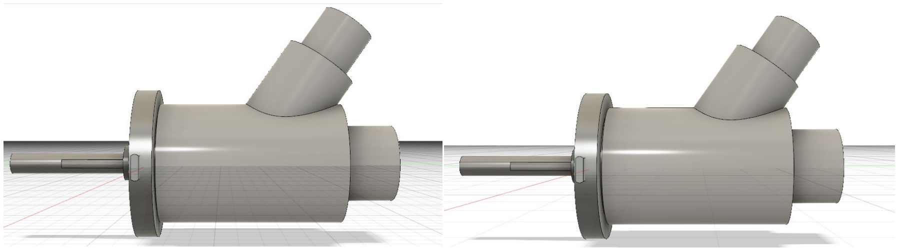


Figure 4.4: Version 1 offered design solutions C and D, connected to the life port.

4.3 Design Solutions Version 2

Second set of proposed solutions for the connector of the HFJ and CPAP ventilator can be observed in the Figure 4.5, 4.6 and 4.7. In the Figure 4.5, design A and B has the inner diameter of 15.5mm and outer diameter of 17.4mm for the connection to the LifePort adapter. Design A has the 3mm exhalation tube, with the outer diameter of 5mm. For the inspiration, it has 5mm outer diameter tube, where the inner diameter is decreasing in size from 4mm till 1.45mm. Design B has same dimensions only difference is the exhalation tube inner diameter which is 3.5mm.

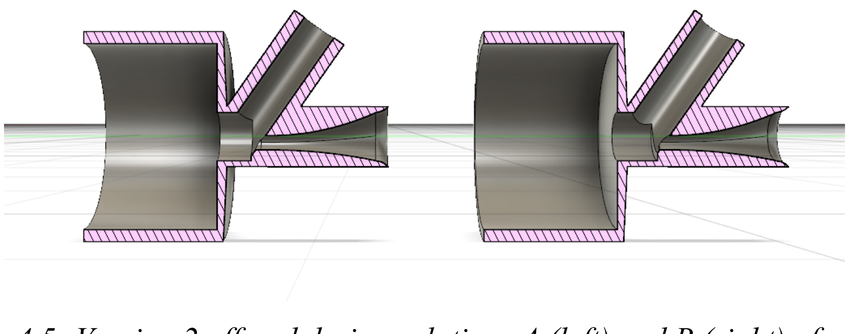


Figure 4.5: Version 2 offered design solutions A (left) and B (right), for infant flow generator (sectional view).

In the Figure 4.6, design C and D has same connector with the inner diameter of 15.5mm and outer diameter of 17.4mm for the connection to the LifePort adapter. Design C has the 4mm exhalation tube, with the outer diameter of 6mm. For the inspiration, it has 4mm outer diameter connector, where the inner diameter is 1.2mm. Design D has the exhalation tube of 3.5mm with the outer diameter of 5mm. For the inspiration tube there is a connector with the outer diameter of 6mm, inner diameter is decreasing in size from 3mm till 1.45mm.

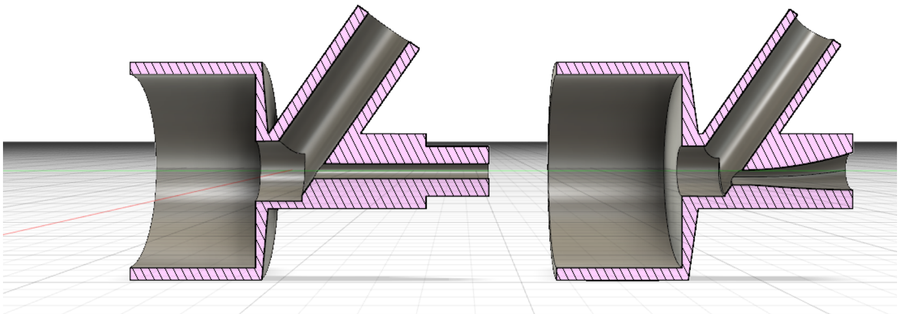


Figure 4.6: Version 2 offered design solutions C (left) and D(right), for infant flow generator (sectional view).

In the Figure 4.7: design E has the iner diameter of 15.5mm and outer diameter of 17.4mm for the life port adapter connection. The exhalation tube has outer 6mm and inner 3mm diameter. Inspiration tube has a 5mm connector with the inner diameter decreasing from 4mm till 1.13mm.

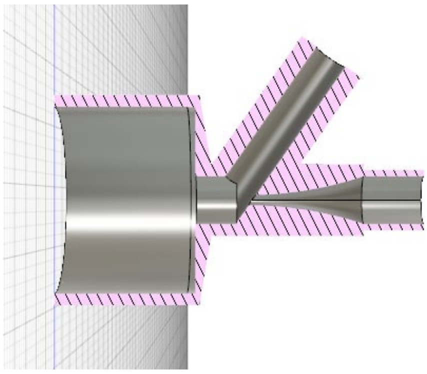


Figure 4.7: Version 2 offered design solution E, for infant flow generator (sectional view).

In the Figures 4.8, 4.9 and 4.10, it is shown how the offered solutions of the infant flow generator will connect to the LifePort adapter for High frequency jet ventilation. LifePort model was designed inside the Fusion 360 software, in order to demonstrate the functionality of the designed connectors.

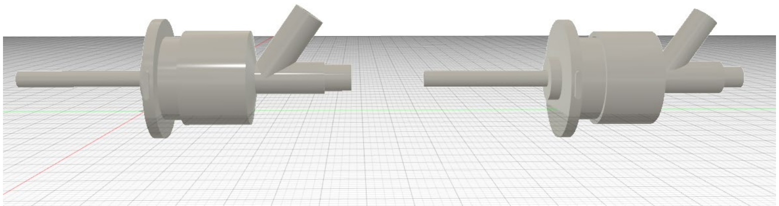


Figure 4.8: Version 2 offered design solutions A and B, connected to the lifeport.

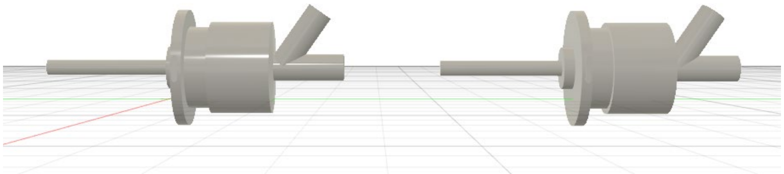


Figure 4.9: Version 2 offered design solutions C and D, connected to the life port adapter.

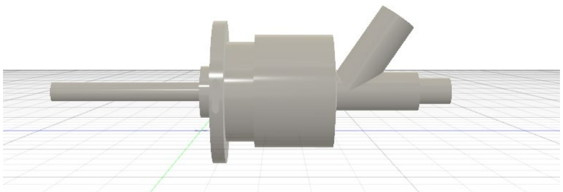


Figure 4.10: Version 2 offered design solution E, connected to the life port.

4.4 Design Solutions Version 3

In the Version 3 of design solutions, there was used the geometry from the Version 2 designs. Most of the details stayed same. The major change in this version was the thickening of the designs to provide more support and strength of the connector. In the Version 3 design A and B there was no major change in the inner geometry, only the outer part was reinforced.

In the Figure 4.11, you can see the design A and B, with the inner diameter of 15.5mm and outer diameter 17.4mm for the LifePort adapter. Exhalation tubes have same outer diameter of 6mm, inner diameter for design A is 3mm and for design B 3.5mm. Both designs have 9mm outer diameter before the connection for the inspiration tube for the reinforcment purposes. The connectors for the inspiration are the size of 5mm for the design A and 4mm for the design B. Inside diameter is decreasing from 4mm till 1.13mm for design A and from 3mm till 1.45mm for design B.

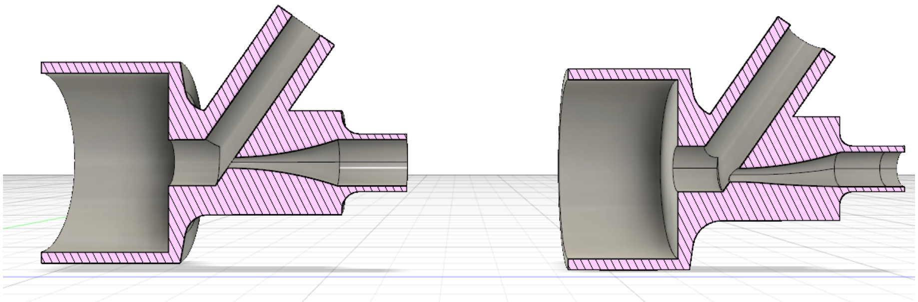


Figure 4.11: Version 3 offered design solutions A (left) and B (right), for infant flow generator (sectional view).

In the Figure 4.12, you can see the design C and D, with the inner diameter of 15.5mm and outer diameter 17.4mm for the LifePort adapter. Exhalation tubes has outer diameter of 6mm and inner diameter 3.5mm for design C. For design D exhalation tube us 7mm outer and 4mm inner diameter. Both designs have 9mm outer diameter before the connection for the inspiration tube for the reinforcment. The connectors for the inspiration is the size of 5mmand the inside diameter decereases from 4mm till 1.38mm. For the design D the connector for inspiration has 4mm diameter and the inner diameter stays same 1.5mm as it gets closer to the patient.

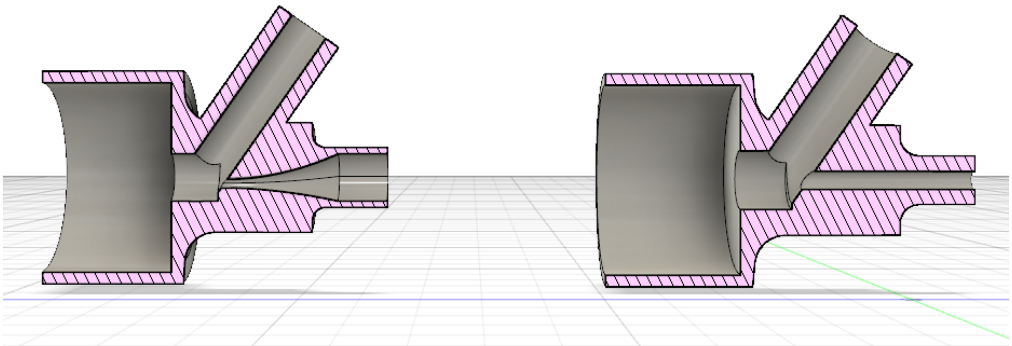


Figure 4.12: Version 3 offered design solutions C (left) and D (right), for infant flow generator (sectional view).

In the Figure 4.13, design E has the inner diameter of 15.5mm and outer diameter 17.4mm for the LifePort Connector. The exhalation tube has outer diameter of 6mm and inner diameter of 3mm. Inspiration tube has a connector with the outer diameter of 6mm. Inner diameter is decreasing from 3mm till 1.83mm. 9mm outer coverage was created here as well to strengthen the design.

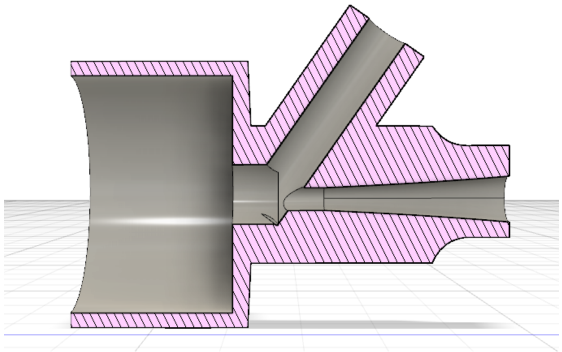


Figure 4.13: Version 3 offered design solution E, for infant flow generator (sectional view).

In the Figures 4.14, 4.15 and 4.16, it is possible to observe the connection of the Version 3 designs in connection with the LifePort adapter.

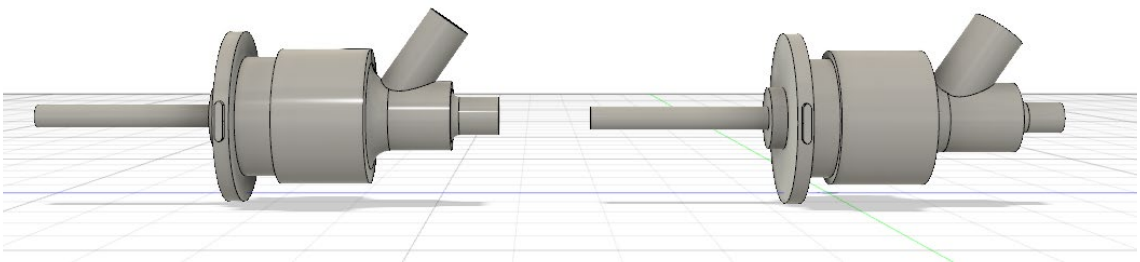


Figure 4.14: Version 3 offered design solutions A (right) and B (left), connected to the life port.

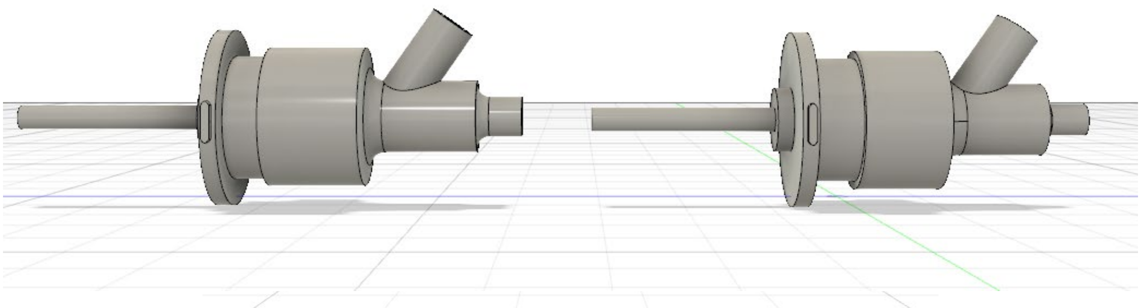


Figure 4.15: Version 3 offered design solutions C (right) and D (left), connected to the life port.

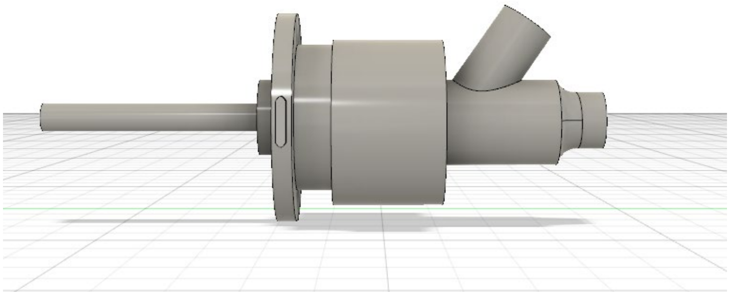


Figure 4.16: Version 3 offered design solution E, connected to the life port.

All the above-mentioned designes were exported in .stl format to the Prusa i3 MK3 printer software environment (Prusa Research, Prague, Czech Republic). An entry for the printing itself was created in the PrusaControl companion program PrusaSlicer, with the print accuracy of 0.1 mm. Supports for the design had to be created inside the Slicer software for each design to make sure that angled parts would not hider the the printing process (Figure 4.17). Same type of supports were created for the Version 2 and Version 3 design solutions as well. The PLA type of material was chosen for printing, which is suitable for printing objects containing small details. Subsequently, the so-called gcode was generated and transferred to the printer using a memory card.

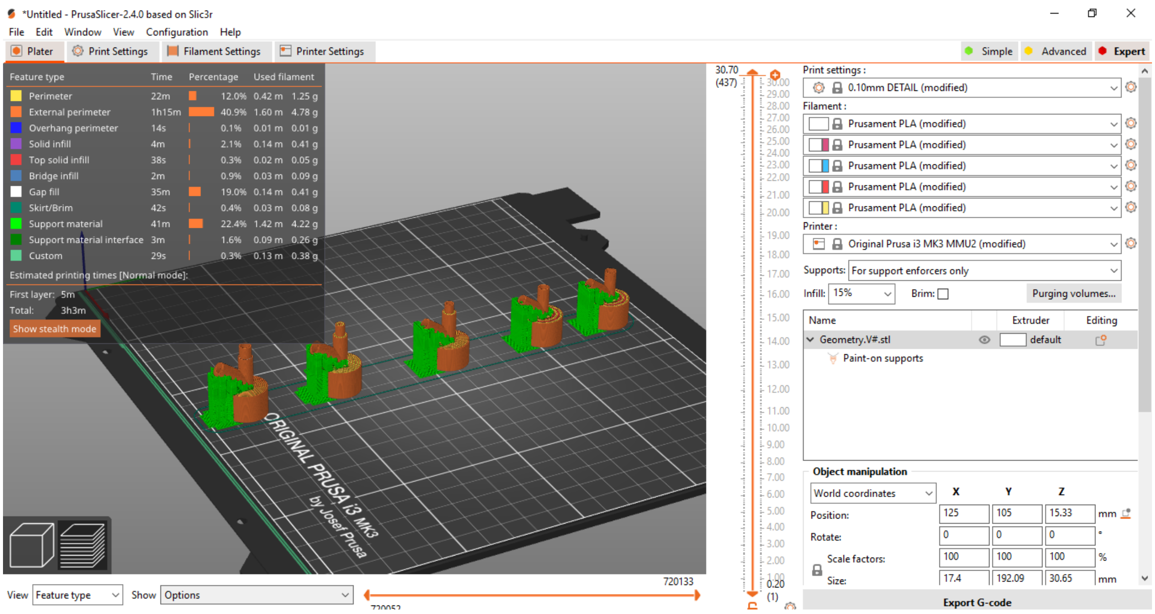


Figure 4.17: Created supports for designs in the PrusaSlicer software.

4.1 3D printing results

In this chapter, it is possible to observe the 3d printing process of offered design solutions. Figure 4.18 and 4.19 covers the Version 1 design printing process and results.

Figure 4.18: 3D printing process of Version 1 designs.

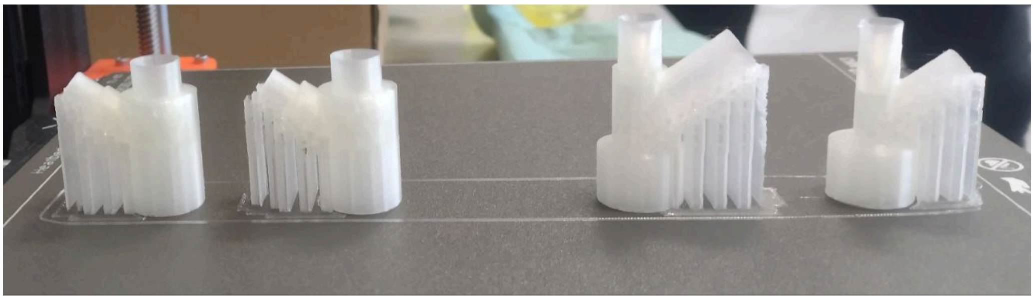


Figure 4.19: 3D printed Version 1 Designs.

In the Figure 4.20, It is possible to observe the printing process of the Version 2 design solutions. All the designs were printed but, the material was fragile since the details were small. Therefore, after removing the supports only one design was not damaged and ready to be tested.

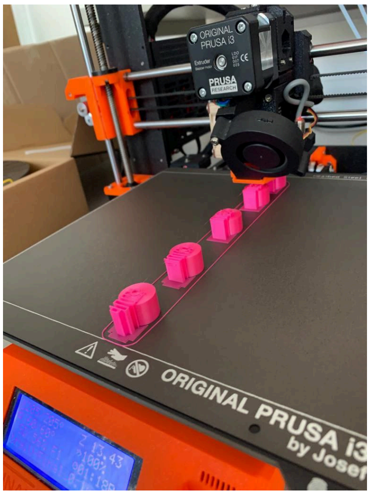


Figure 4.20: 3D printing process of Version 2 designs.

Finally, Figure 4.21 represents the Version 3 design printing results. Only three out of five designs managed to print without any complications.

Figure 4.21: 3D printed Version 3 designs.

4.2 Laboratory work for verification of offered solutions

Measurement of breathing work and verification of the effectiveness of all solutions of the design took place in a laboratory with the sources of medical gases and all the necessary apparatus at the Faculty of Biomedical Engineering of CTU in Prague.

The definition of iWOB implies that it is the work done by the patient to overcome the resistance of the components of the ventilation circuit, therefore the HFJV was not started at the time of the test, which provides a fast and very short inspiration that the patient cannot interact with. Only during exhalation can the patient experience higher work of breathing, because not only patient has to breathe against the resistance of the ventilaton crcuit but also the continious flow of air from the CPAP device.

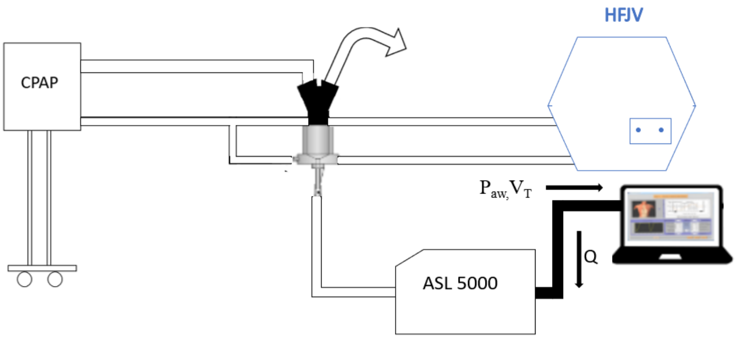


Figure 4.22:Arrangment of experimental measurement of HFJV and CPAP

Therefore, the laboratory experiment was modified to account for the patient's spontaneous breathing process. In the Figure 4.22, it is possible to observe the arrangment of the circuit made during the experiment. HFJV was excluded from the circuit and the measurement was conducted only with CPAP ventilator and ASL5000 lung mechanic simulator.

4.3 Setting up ASL 5000 lung simulator

The ASL 5000 lung mechanics simulator was set up in the same way for all measurements. After running the control software version 3.6, the neonatal-normal script was run. In the Edit section, the FlowPump operating mode was selected and the breathing curve parameters were set, see Figure 4.23. Subsequently, the data storage file was confirmed. In the Table 4.1, You can see the basic parameters of the healthy neonate inside the ASL 5000 lung simualtor.

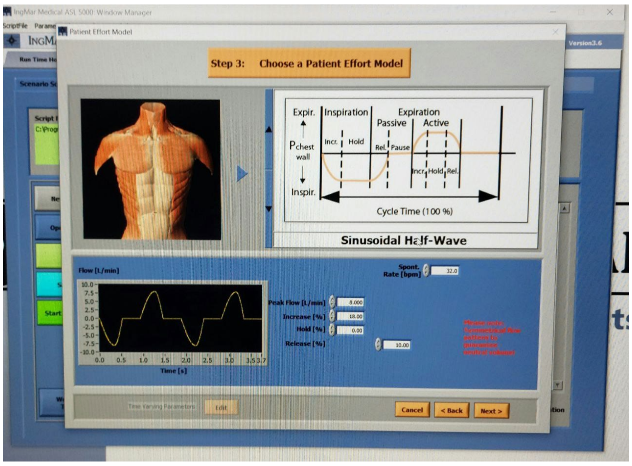


Figure 4.23: Setup parameters of ASL 5000.

Table 4.1. Healthy neonate parameters

Name	Neonate_normal	-
Curve Shape	Sinusoidal	-
Respiratory rate	32	breath/min
I:E Ratio	0.31	-
Breathing volume	36.55	mL
Maximum flow	7.14	L/min
Simulator dead space compensation	200	mL
System dead space compensation	No compensation	-

4.4 Course of the experimental measurement

First, all devices were started - conventional ventilator or CPAP unit and ASL 5000 lung simulator. Subsequently, all the above parameters were set on the ASL 5000 simulator. The experiment consisted of recording a total of 35 breathing cycles each time for a given setup at a set pressure of 10 cmH2O. The data were sampled at a standard frequency of 512 Hz and the parameters of tidal volumes and airway pressures were stored each time for each breath. After the 35th echo, the measurement was terminated. For the analysis of the acquired data, already prepared scripts for Matlab computing software were used.

4.5 Data processing and IWOB calculation

The processing was performed in Matlab software (MathWorks, Natick, Massacgusetts,USA).Using the script data_nacteni.mat, the data.dtb and data.rwb files were read as bits and converted to double data format. The time course of tidal volume VT was stored as dtb-Vtot, and the airway pressure vector Paw was called dtb_Paw. The sample serial number information is stored in the dtb_breathnum file.

The second script, data_adjustment.mat, analyzed the data noise in the dtb_Vtot file, i.e., values less than 10-38. The noise values were replaced with the nearest true volume values.

In the next step, the dtb_Vtot and dtb_Paw data were Butterworth filtered with a cutoff frequency of 2 Hz.

Subsequently, the VT breath volume data were sorted into vectors according to breath number, and the end of the inspiration as well as the beginning of the expiration, i.e. the maximum in each breath, was found in each group of data.

Afterwards, matrices were pre-prepared to fit the positions of the maxima and minima in each breath, which were subsequently assembled into bulk matrices with the corresponding names breathmin and breathmax.

In the next script, data_vyhodnoceni.mat, the breath work iWOBexp, iWOBinsp is calculated. iWOBtot is calculated using the function vypocet_iWOB, which is based on the numerical integration of the trapezoidal trapz method. In addition to the breath work values, the script also displays P-V diagrams graphically. All three breath work iWO- Bexp, iWOBinsp, iWOBtot are normalized to the value of the total breath volume VT for the sake of relevant comparison of all solutions. The unit of normalized iWOB is J/L.

5 Results

5.1 Results of IWOB Calculation for all design versions.

In the Figure 5.1, you can see the chart for the calculated imposed work of breathing from different design versions compared to the reference iWOB from the CV circuit. The chart shows that designed components based on the infant flow geometry, can achieve lower iWOB than the standard conventional ventilator in the standard circuit (Y-piece, inspiratory and expiratory limb).

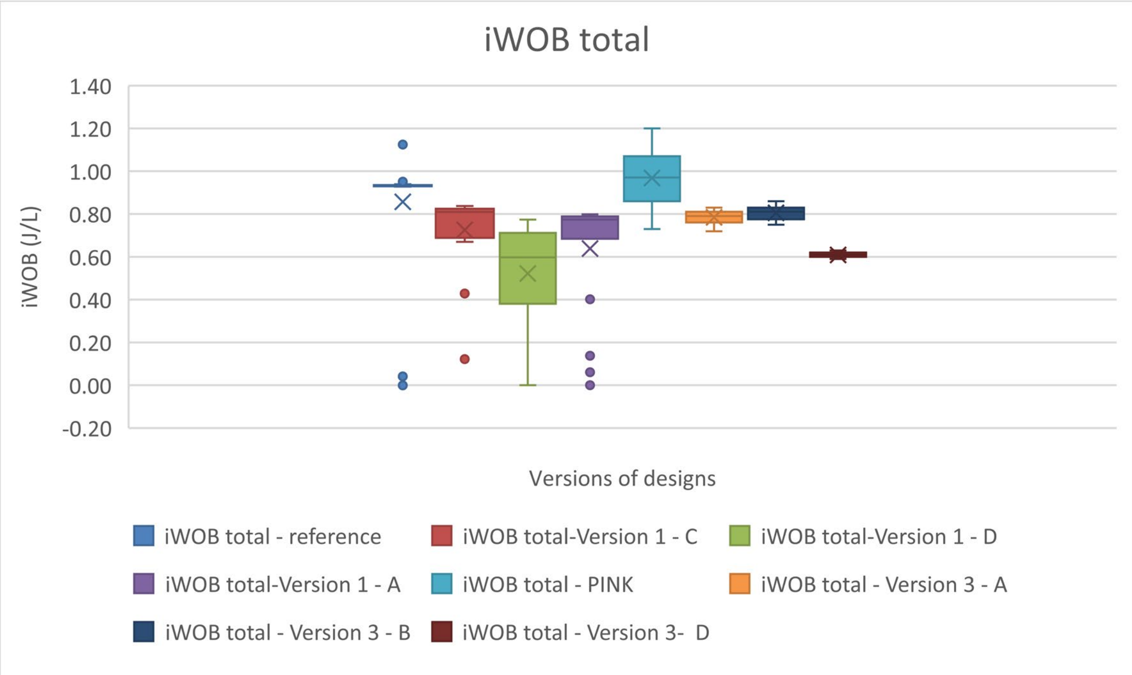


Figure 5.1: Chart for the comparison of iWOB reference to the all the offered design solutions for the connector.

In the Figures 5.2, 5.3 and 5.4, you can observe the pressure to volume graphs for the Version 1 design solution A, C and D at PEEP level of $10cmH_2O$. Pink part on the left, describes the inpsiration and grey on the right corresponds to the expiration of the patient. The black line in between is the end of inspiration or expiration. This represents the flexibility of the simulated neonatal lungs and seperates individual parts of breathing work.

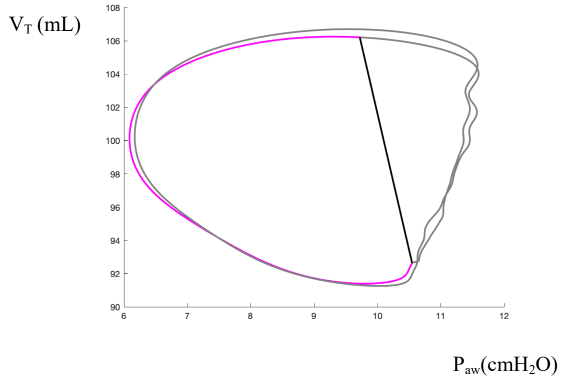


Figure 5.2: Pressure to Volume graph for Version 1 design solution A, at 10cmH₂O PEEP.

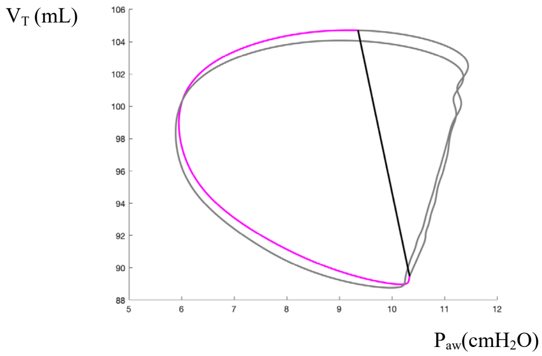


Figure 5.3: Pressure to Volume graph for Version 1 design solution C, at 10cmH₂O PEEP.

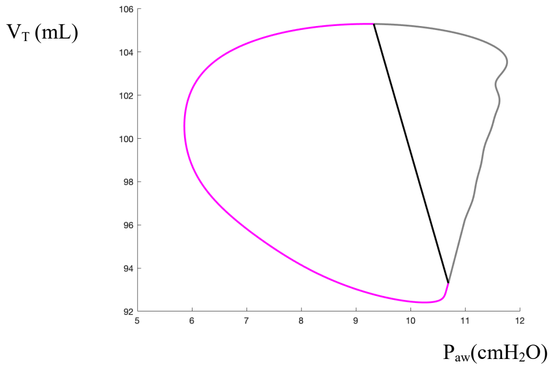


Figure 5.4: Pressure to Volume graph for Version 1 design solution D, at 10cmH₂O PEEP.

In the Table 5.1, you can see the results of iWOB calculation for the Version 1 design solutions A, C and D. In the first column, it is possible to observe the conventional ventilator iWOB values as the reference material.

Table 5.1: Results of total iWOB calculations for Version 1 designs.

IWOB Total - Reference (J/L)	0.04	0.97	1.12	0.95	0.93	0.93	0.93	0.93	0.93	0.94	0.94
IWOB Total-Version 1-A (J/L)	0.40	0.77	0.68	0.79	0.80	0.77	0.79	0.73	0.73	0.78	0.79
IWOB Total-Version 1-C (J/L)	0.72	0.67	0.68	0.80	0.83	0.83	0.81	0.83	0.84	0.72	0.82
IWOB Total-Version 1- D (J/L)	0.74	0.61	0.60	0.57	0.54	0.29	0.65	0.69	0.47	0.74	0.77

In the Figures 5.5, 5.6, 5.7 and 5.8, you can observe the pressure to volume graphs for the Version 2-Pink and Version 3 design solution A, B and D at PEEP level of 10cmH2O. Pink part on the left represents the inpsiration and grey on the right corresponds to the expiration of the patient. The black line in between is the end of inspiration or expiration.

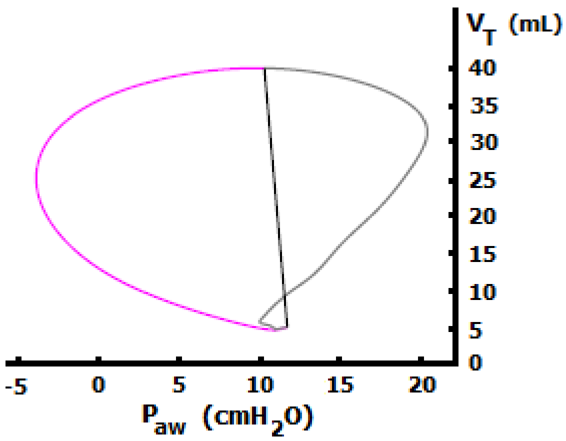


Figure 5.5: Pressure to Volume graph for Version 2 design PINK, at 10cmH_2O PEEP.

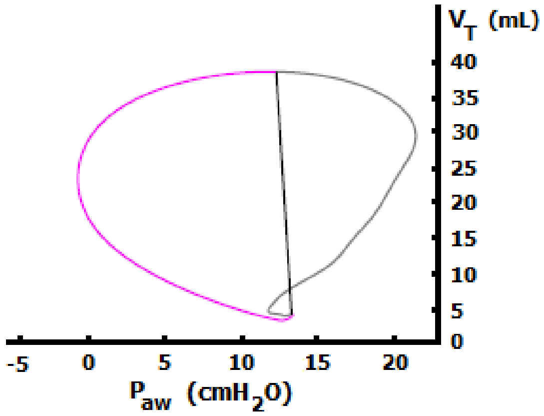


Figure 5.6: Pressure to Volume graph for Version 3 design solution A, at 10cmH2O PEEP.

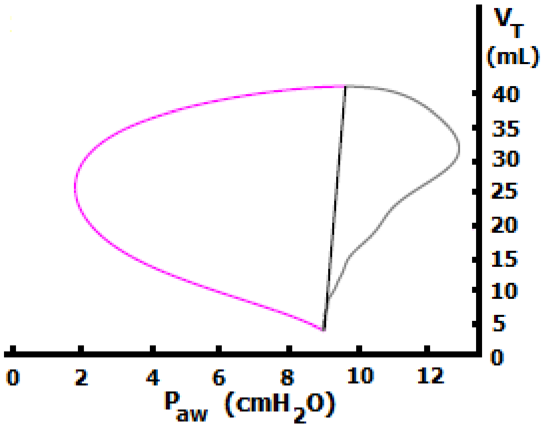


Figure 5.7: Pressure to Volume graph for Version 3 design solution B, at 10cmH2O PEEP.

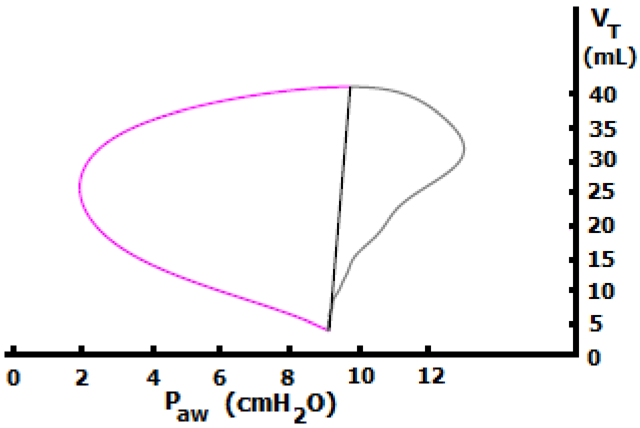


Figure 5.8: Pressure to Volume graph for Version 3 design solution D at 10cmH2O PEEP.

In the Table 5.2, 5.3 and 5.4, you can see the results of iWOB calculation for the Version 2 design solutions PINK and Version 3 solution A, B and D. In the first column, it is possible to observe the conventional ventilator iWOB values as the reference material.

Table 5.2: Results of total iWOB For Version 2 and 3 designs.

iWOB total-reference (J/L)	0.04	0.97	1.12	0.95	0.93	0.93	0.93	0.93	0.93	0.94	0.94
iWOB total-PINK (J/L)	0.8	1.12	0.81	1.13	0.80	1.12	0.92	0.97	0.92	1.16	1.13
iWOB total-Version 3-A (J/L)	0.81	0.78	0.83	0.76	0.80	0.79	0.77	0.80	0.83	0.76	0.83
iWOB total-Version 3-B (J/L)	0.80	0.82	0.75	0.76	0.85	0.83	0.083	0.83	0.77	0.82	0.85
iWOB total-Version 3-D (J/L)	0.61	0.60	0.63	0.61	0.63	0.63	0.60	0.61	0.63	0.62	0.61

Table 5.3: Continuation of results of total iWOB For Version 2 and 3 designs.

iWOB total-reference (J/L)	0.93	0.93	0.93	0.94	0.93	0.93	0.93	0.93	0.94	0.94	0.94
iWOB total-PINK (J/L)	1.06	1.08	0.99	1.06	0.74	0.95	0.97	0.96	1.09	1.20	0.90
iWOB total-Version 3-A (J/L)	0.76	0.72	0.76	0.78	0.72	0.74	0.83	0.83	0.78	0.79	0.73
iWOB total-Version 3-B (J/L)	0.84	0.84	0.80	0.81	0.80	0.77	0.81	0.75	0.82	0.82	0.76
iWOB total-Version 3-D (J/L)	0.63	0.61	0.60	0.61	0.60	0.59	0.61	0.61	0.63	0.59	0.60

Table 5.4: Final part of results of total iWOB For Version 2 and 3 designs.

iWOB total-reference (J/L)	0.93	0.93	0.93	0.94	0.93	0.93	0.93	0.93	0.94	0.93	0.00
iWOB total-PINK (J/L)	0.8	1.02	1.01	0.89	1.05	0.73	0.82	0.85	0.96	1.03	0.86
iWOB total-Version 3-A (J/L)	0.79	0.81	0.74	0.80	0.82	0.75	0.79	0.81	0.78	0.77	0.81
iWOB total-Version 3-B (J/L)	0.77	0.83	0.81	0.79	0.76	0.78	0.83	0.79	0.79	0.86	0.83
iWOB total-Version 3-D (J/L)	0.60	0.62	0.59	0.63	0.59	0.61	0.60	0.59	0.62	0.59	0.61

6 Discussion

The main aim of this diploma thesis was to reduce the work of breathing of the patient during High frequency jet ventilation. The assumption was confirmed by designing the connector for CPAP and HFJV ventilators for the respiratory support system for neonates, based on infant flow geometry. The main finding of this diploma thesis is that the reduction of work of breathing during High frequency jet ventilation is possible by exploiting the "fluidic flip" technology. The offered design solution Version 3-D demonstrated the greatest reduction of work of breathing, while measured at CPAP pressure level 10cmH_2O. All desgned connectors had to be tested on the ASL 5000 lung simulator. Afterwards, for the calculation of iWOB values there was used already prepared script in the MATLAB environment [31].

Three versions of solutions for the connector design were offered, Version 1, 2 and 3. All the offered design solutions had the connector for the LifePort adapter, CPAP ventilator and the exhalation valve open to the atmosphere. Conventional ventilator circuit iWOB values were used for the reference material, to compare to the findings and prove the efficiency of the offered solution designs.

The reference iWOB achieved standard levels and minimal deflections. For the proposed solution of the design Version 1, the variance of the values is higher, which is due to the flow inside the valve, which had a high internal volume. High internal space leads to a high iWOB. In the Figure 5.1, it is possible to observe that Version 1 designs resulted in minor reduction of the work of breathing, due to the high internal space. However, Version 1 designs confirmed the functionality of the infant flow geometry for the connector.

This justifies the changes that led to the PINK Version 2 design solutions. Version 2 design's geometry was done with the sole purpose of reducing the inner space of the connector to the minimum. The five solutions were introduced and different internal geometries were implemented. Main difference was that both expiration and inspiration tubes were reduced in size. Version 1 design A and B was taken and modified to create the Version 2 designs. Compared to Version 1 designs, exhalation tubes in the Version 2 decreased in size from inner radius of 7mm and 9mm to the 3mm, 3.5mm and 4mm. For the inspiration tube, the inner diameter was decreasing in size from the 4mm or 3mm till 1.5 instead of 7mm to 2mm in the Version 1. This lead to decreased internal volume of the connector.

However, after the process of printing of Version 2 designs with the least amount of space possible, the complication was detecter. The minimalistic geometry connector led to designs being fragile and another variant of the designs had to be offered to solve this problem.

Version 3 connectors were designed with the purpose of eliminating the fragility factor. The internal designs were kept same as in Version 2, but with the main difference of having the stronger and wider outer covarage for the connectors. Reinforcment of outer cover of the designs led to the optimal solutions for printng and testing the infant flow geometry for the reduction of brething work during HFJ ventilation.

In the Figure 5.1, it is possible to observe that for Version 3 designs, the reduction of the internal space has increased the efficiency of the flip-flop fluidic mechanism. However, the resistance in the exhalation part of the valve is high in the Version 3-A and B designs, which is why the iWOB values are high, except for the Version 3-D where the exhalation part is enlarged. Version 3-D, utilized the minimal internal volume and non-fragile cover, leading to the greatest reduction of breating out off all offered solutions. From the Table 5.2, 5.3 and 5.4 you can see that the total iWOB for the Version 3-D was varying in the values of 0.60 J/L, while for reference CV circuit it was 0.93J/L.

Overall, the results confirm that the use of a infant flow geometry in is appropriate to reduce iWOB, but it must have a low internal volume and minimized resistance in the exhalation portion of the valve. The material structure also has a certain influence on the flow inside the fluidic element, which in the case of 3D printing may not be smooth in all circumstances.

Design of all vaiants of solutions were done in the Fusion 360 software. The PRUSA MK 3 printer was used to complete the printing of design.

7 Conclusions

In the conclusion, I would like to say that, as the part of this diploma thesis, high frequency jet ventilation circuit for the neonate care was analyzed and possible solutions for the LifePort adapter was offered from the point of view of reducing work of breathing of patients. For providing continuous flow of air inside lungs of the patient CPAP device was used. The offered solutions of the design, allowed CPAP device to connect to the LifePort adapter of HFJV, allow continuous flow of air towards the lung and the passive exhalation to the atmosphere. Infant flow geometry was used to redirect the exhaled air with the air from the CPAP machine to the atmosphere. After the designing and 3D printing of the mentioned connectors, iWOB was evaluated on the ASL5000 Lung simulator. The main aim of the thesis to reduce the work of breathing of the patient during the High frequency Jet ventilation, was completed. The greatest work reduction of breathing was detected for the design solution V3-D. Overall results proved that the infant flow geometry if used correctly, with least amount inner volume and robust cover, will reduce the work of breathing of the patient. The subject of the further research could be optimization of infant flow geometry used in this work.

The printing style must be set to very fine and slow printing, or a different technology and material must be used, for example SLA (Stereolithography aparatus) to increase the accuracy.

8 List of literature

[1] Blencowe H, Cousens S, Oestergaard M, Chou D, Moller AB, Narwal R, Adler A, Garcia CV, Rohde S, Say L, Lawn JE. *The Lancet.* DOI: 10.1016/S0140-6736(12)60820-4. 2012, Jun 9.

[2] Liu L, Oza S, Hogan D, Chu Y, Perin J, Zhu J, et al. *Lancet.* [Global, regional, and national causes of under-5 mortality in 2000-15: an updated systematic analysis with implications for the Sustainable Development Goals.] doi: 10.1016/S0140-6736(16)31593-8. 2016, Nov 11.

[3] *Flow measuring during neonatal high frequency jet ventilation using orifice plate.* Petr Kudrna1, Martin Rožánek1, Barbora Hřibalová1. Faculty of biomedical engineering, Czech Technical Univesity in Prague, Czech Republic, : 2014.

[4] J. Bert Bunnell, Sc.D.Chairman and Chief Executive Officer Bunnell Incorporated. *High-Frequency Jet Ventilation of an infant children.* 436 Lawndale Drive.Salt lake city, Utah 84108 :

[5] WHO. World health organisaton . *https://www.who.int/news-room/fact-sheets/detail/newborns-reducing-mortality.* [Online] 2020.

[6] *Premature Infant.* Rice, Sandy Calhoun. https://www.healthline.com/health/pregnancy/premature-infant#prevention. 2016.

[7] *Common respiratory conditions of the newborn.* David J. Gallacher, Kylie Hart, and Sailesh Kotechacorresponding. doi:10.1183/20734735.000716.2016 march 12.

[8] *Surfactant administration in neonates: A review of delivery methods.* Nina Nouraeyan, MD,1 Alicia Lambrinakos-Raymond, MD,1 Marisa Leone, RRT,2 and Guilherme Sant'Anna, MD PhD. PMID: 26078618 2014 :

[9] *Encyclopedia of Medical Devices and Instrumentation* . Webster, John G. ISBN: 978-0-471-26358-6 March 2006, Sv. Second Edition.

[10] *High Frequency Ventilation.* Murthy, Pooja R. a AK, Ajith Kumar. PMID:33085298. July 30, 2021.

[11] JJ, Fredberg. *frequency, Pulmonary mechanics during high frequency ventilation.* Acta Anaesthesiol Scand Suppl 1989. 90:170-1.

[12] *Effects of Orifice Plate as a Flow Sensor in an Endotracheal Tube on the Ability of Elimination CO 2 during HFJV-Animal Study.* Petr, Kudrna. 2021/05/06, Sv. 11. 10.3390/app11094210.

[13] Thoracic Key. *https://thoracickey.com/6-high-frequency-ventilators/.* [Online] 19. Dec 2021.

[14] V. Courtney Broaddus, Robert J. Mason,Joel D. Ernst,Talmadge E. King, Jr.,Stephen C. Lazarus,John F. Murray,Jay A. Nadel,Arthur S. Slutsky,Michael B. Gotway. *Murray and Nadel's Textbook of Respiratory Medicine.* 2016.

[15] Continuous positive airway pressure. *Wikipedia.* [Online] https://www.wikiwand.com/en/Continuous_positive_airway_pressure.

[16] *Current methods of non-invasive ventilatory support for neonates.* MAHMOUD, Ramadan A., Charles Christoph ROEHR and Gerd SCHMALISCH. 2010, 12(3). DOI: 10.1016.

[17] *Nasal continuous positive airway pressure (CPAP) for the respiratory care of the newborn infant.* 1, Robert M Diblasi. PMID: 19712498. 2009.

[18] *Inspiratory work of breathing on flow-by and demand-flow continuous positive airway pressure.* C S Sassoon 1, A E Giron, E A Ely, R W Light. DOI: 10.1097/00003246-198911000-00004. 1989 Nov;17(11).

[19] Inspiration Healthcare. [Online] https://www.inspiration-healthcare.com/products/neonatal-intensive-care/resuscitation/inspire-rpap.

[20] obgynkey. [Online] https://obgynkey.com/nasal-cpap-an-evidence-based-assessment-2/, 2016 (9).

[21] *Revolution for the first breath. 500-119 (2) : Inspiration Healthcare.* rPAP:, Inspire. England : autor neznámý, 2017-10.

[22] *DIBLASI, RM Nasal continuous positive airway pressure (CPAP) for the respiratory care of the newborn infant. PMID:19712498.* 2009 54 (9), 1-27.

[23] IngMar Medical, RespiSim System,. *ASL 5000: User's Manual.* . Pittsburgh: : 2016.

[24] IngMarMedics ∕[Online] https://www.medicalexpo.com/prod/ingmar-medical/product-68861-429175.html.

[25] Yartsev, Alex. Work of breathing and its components. [Online] 23. 09 2021.https://derangedphysiology.com/main/cicm-primary-exam/required-reading/respiratory-system/Chapter%20041/work-breathing-and-its-components.

[26] *Imposed work of breathing during high-frequency oscillatory ventilation: a bench study.* . van Heerde, M., van Genderingen, H.R., Leenhoven, T. 2006 Feb;10(1). DOI: 10.1186/cc398801.

[27] *Imposed work of breathing during high-frequency oscillatory.* Marc van Heerde, Huib R van Genderingen, Tom Leenhoven, Karel Roubik, Frans B Plötz and Dick G Markhorst. 2006. doi:10.1186/cc3988.

[28] Gupta S, Donn SM. Continuous positive airway pressure: Physiology and comparison of devices, Seminars in Fetal & Neonatal Medicine. 2006 Feb;10(1): R23. DOI: 10.1186/cc3988.

[29] Archana Gupta, Phyllis Della-Latta, Betsy Todd, Pablo San Gabriel. Neonatal Intensive care. 2000 DOI:10.1086/502380.

[30] *Flows and function of the Infant Flow neonatal continuous positive airway pressure device investigated with computational fluid dynamics.* Thomas Drevhammar, Niclas Berg,| Kjell Nilsson, Baldvin Jonsson. Östersund & Stockholm, Sweden : 2021 MAR;110(3)811-817. DOI:10.1111

[31] 31. *MEDIJET nCPAP generator optimization. Diploma thesis,CTU FBMI. Thesis supervisor Ing. Petr Kudrna Ph.D.* SUROVCOVÁ, Markéta. Kladno : autor neznámý, 2016.

Annex A: Contents of enclosed CD

MATLAB_Script:

data_Nacteni.m

data_uprava.m

data_vyhodnoceni.m

vypocet_iWOB.m

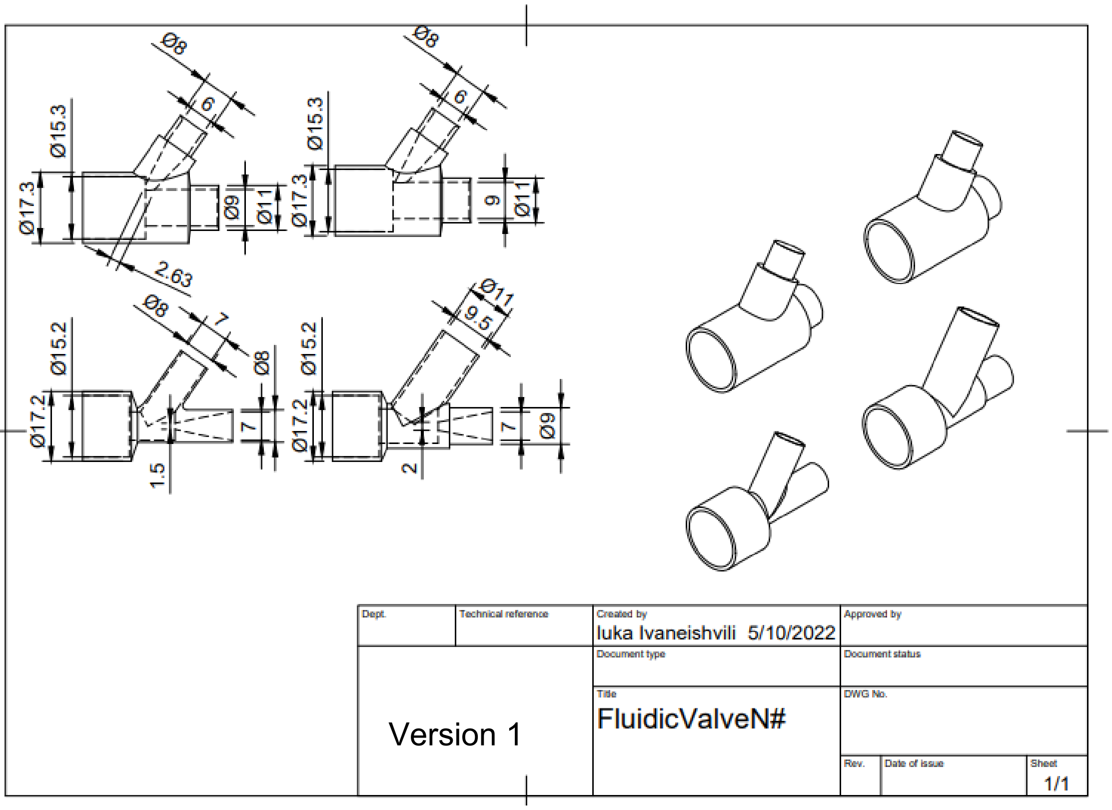
Fusion 360_designs:

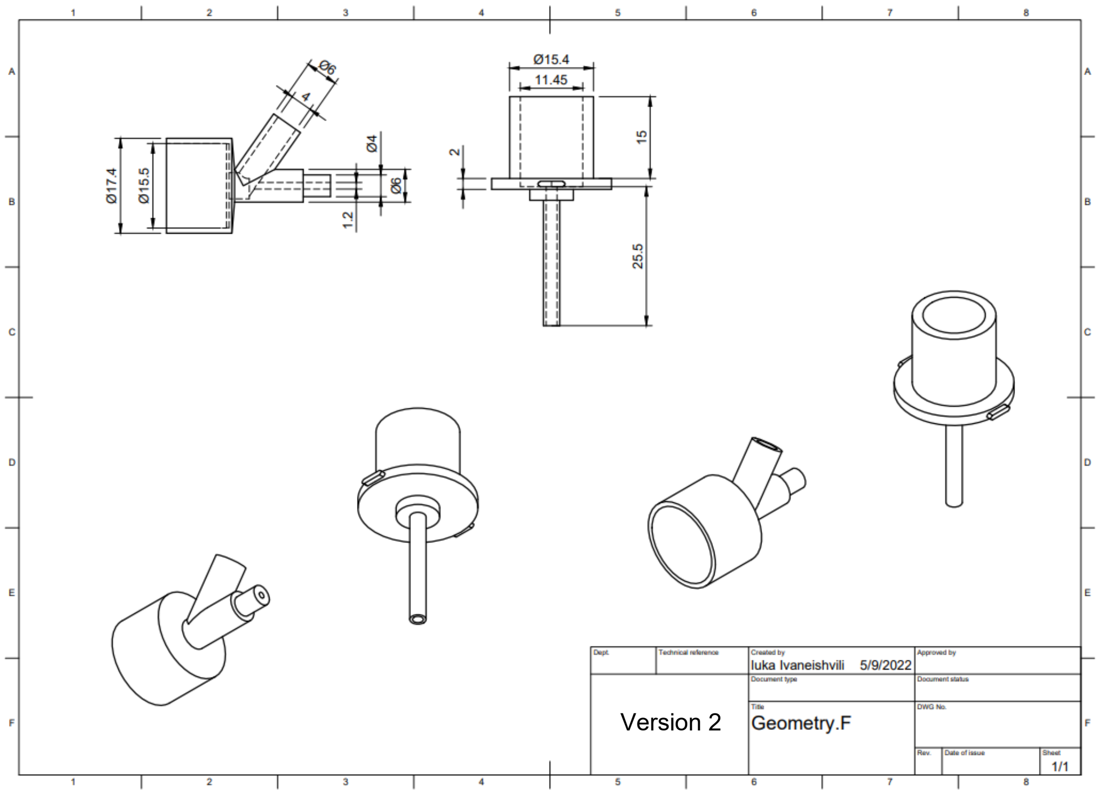
Version 1.stl

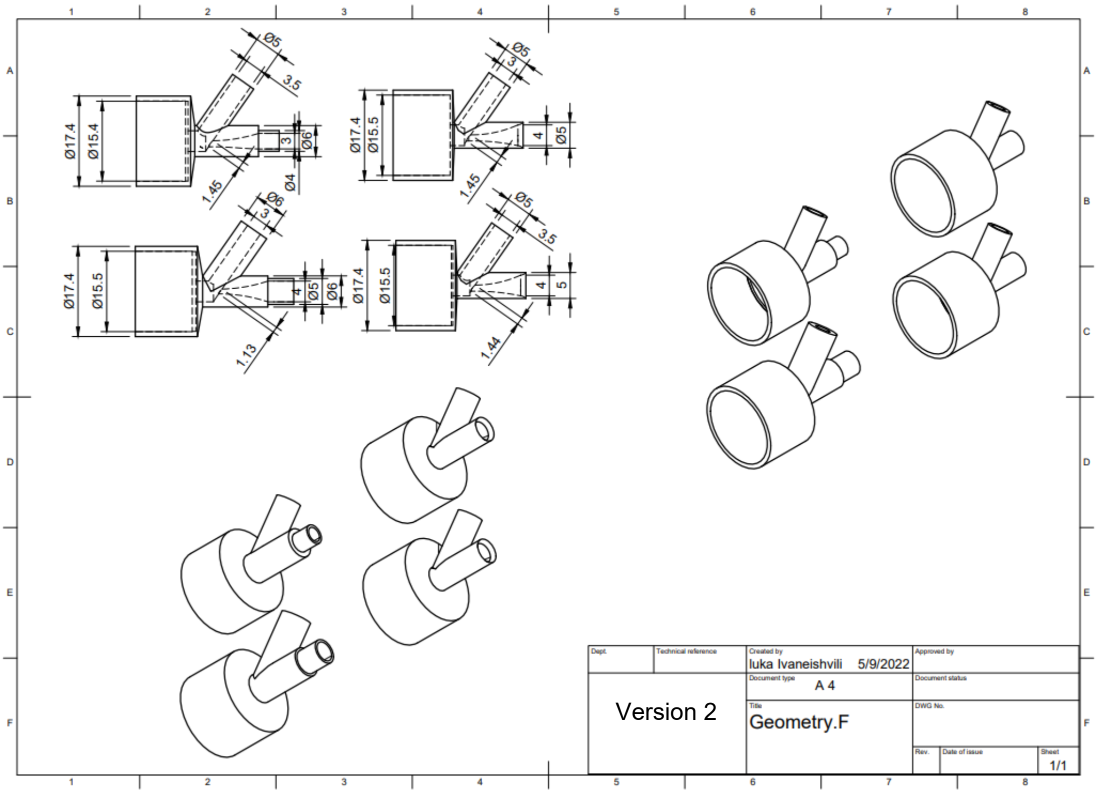
Version 3.stl

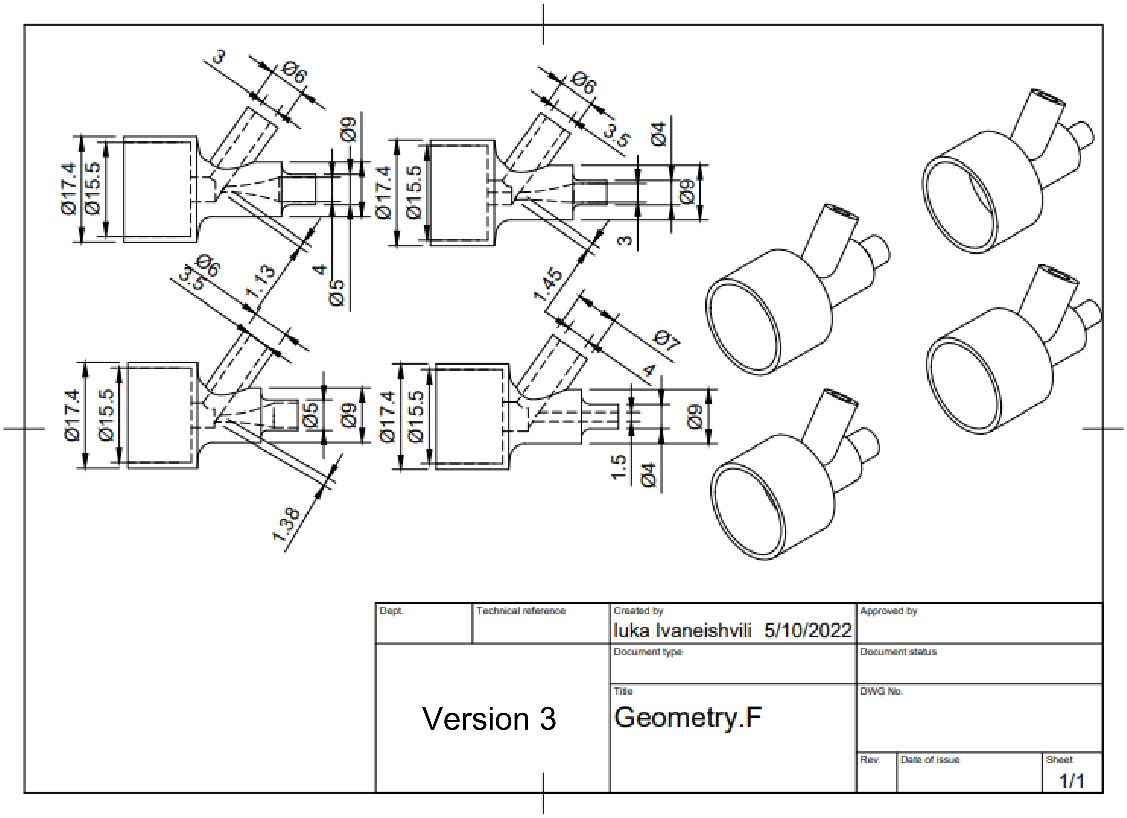
Annex B

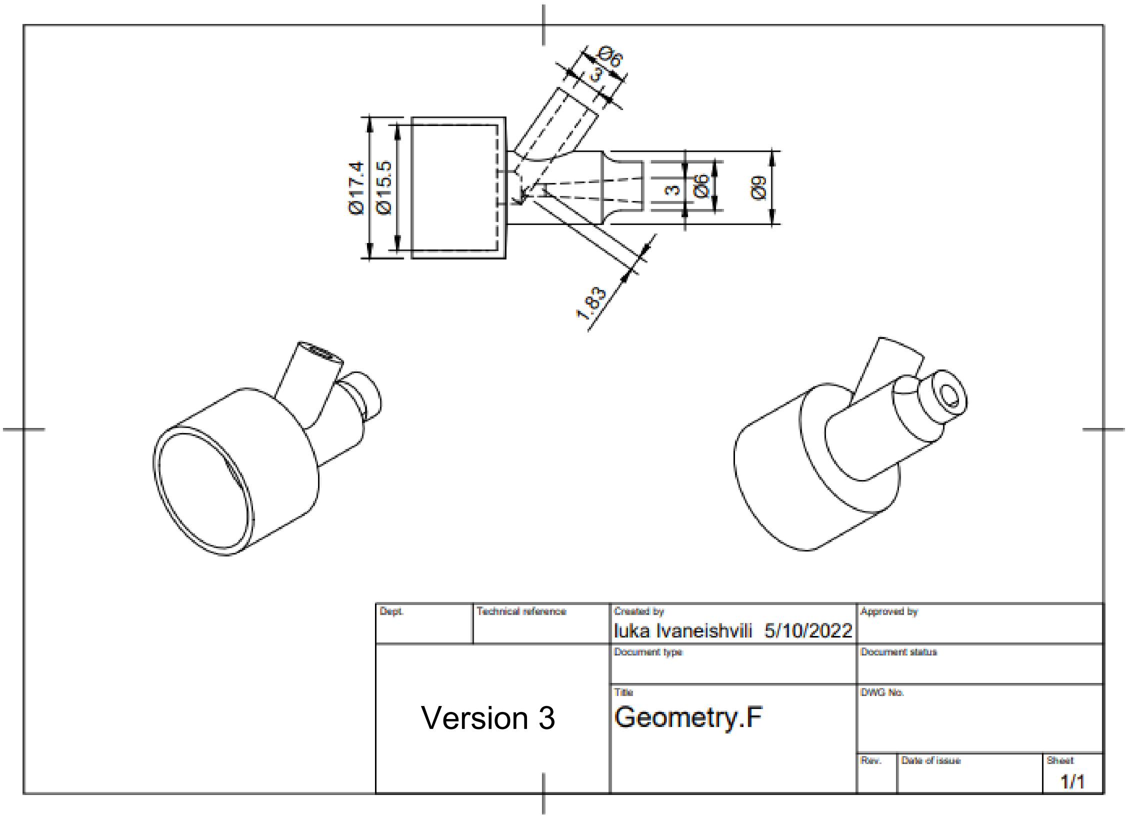
Technical documentation of designs solutions






Version 2
Created by
Iuka Ivaneishvili 5/9/2022
Document type
A 4
Title
Geometry.F
Sheet
1/1


Version 3
Created by
Iuka Ivaneishvili 5/10/2022
Title
Geometry.F
Sheet
1/1



Ø6
3
Ø17.4
Ø15.5
3
Ø6
Ø9
1.83
Dept.
Technical reference
Created by
Iuka Ivaneishvili 5/10/2022
Approved by
Document type
Document status
Version 3
Title
Geometry.F
DWG No.
Rev.
Date of issue
Sheet
1/1