UNIVERSITA' POLITECNICA DELLE MARCHE

Engineering Faculty

Master's degree in biomedical engineering

DESIGN AND TESTING OF AORTIC TEST BENCH

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1. Introduction

Aortic valvular disease occurs when the valve between the aortic arch and the left ventricle fails to function effectively through an open and close mechanism in response to blood flow. A damaged aortic valve can be caused by aortic stenosis or aortic regurgitation. In aortic stenosis, the valve is constricted and does not fully open. This lowers or completely prevents blood flow from the left ventricle to the aorta and the rest of the body. While in aortic regurgitation permits part of the blood pumped out of the left ventricle to return. As the left ventricle works harder to maintain blood flowing through the aortic valve, it may expand and weaken.

Cardiologists are eager to replace the damaged valve with a new one to address the essential issue of aortic valve failure, which can be completely invasive or less invasive, such as transcatheter aortic valve implementation, The valve replacement depends on using prosthetic or bioprosthetic valves taking into consideration the hemodynamic physiological characteristics like pressure and fluid flow.

The aim of this thesis project is to design and testing of an aortic test bench for the evaluation of hemodynamic parameters of prosthetic valve of the left-sided part of the heart which in turn may be used as a guide for surgeons when replacing a damaged valve.

The creation of revolutionary test benches for aortic valves resulted from the advent of new less invasive surgical procedures for heart valve implantation, as opposed to riskier open-heart surgery.

The bench must meet the requirements for the ability to perform fluid dynamical and optical measurements at the aortic valve level. As a result, special materials and components are required for the bench's construction. Flat and clear plexiglass panels are required to give optical access while doing optical tests. The material utilized to recreate these anatomical components is elastic, allowing the ventricle to contract and relax and the aorta to expand and rebound, much like the genuine ones.

The construction of the test bench with anatomical components that are morphologically close to the genuine one might aid in obtaining a more realistic value for hemodynamic parameters, in terms of flow and pressure changes.

The test takes into consideration pulsatile flow which represents blood flow in a normal cardiac cycle that passes through different compartments of the heart and the method to initiate it.

For this purpose, a control system is designed and used as an engine to simulate the pulsatile flow of fluid passing through the left ventricle and aortic arch synthesized materials which are used in the realization of test bench creation. In this study, we employed a feedback linear actuator whose stroke activates the membrane of the piston pump, stimulating fluid flow via the developed system. The actuator feedback is determined by an external electromotive instrument's input signal. Both devices are equipped with a microcontroller device that takes input readings and processes them using its own software to translate analog data into physical actions.

Measurement sensors including flow and pressure sensors are utilized to gather and record data from the left ventricle and aortic arch during the running of the designed bench. These recorded measurements are then captured by a data collecting system, which processes them through built-in software to display the values of studied parameters as a consequence of fluid flow during the operation of the created bench.

Based on the results obtained by the measurement sensors for the pressure and flow readings during the test which are closely the physiological parameters during the cardiac cycle, the heart surgeon should be able to do the simulation immediately on the test bench before the valve implant on the patient to ensure that they satisfy all requirements in terms of the hemodynamic characteristics of the healthy valve.

1.1 Anatomy and Physiology of the heart

The heart is a muscular organ, pyramidal in shape, consisting of two parallel valved pumps, located within the middle mediastinum, two-thirds to the left of the centerline. The base of the heart is oriented superiorly, whereas the apex points leftward, anteriorly, and slightly inferiorly. The cardiac apex is located at the fifth intercostal space near the midclavicular line. The heart is enclosed by the fibrous pericardium, which is bordered by the diaphragm inferiorly; the sternum and ribs anteriorly; the pleurae laterally; and the esophagus, descending aorta, and vertebrae posteriorly [1].

The heart consists of four chambers:

Right Atrium

The right atrium is a low-pressure capacitance chamber that receives blood from the superior vena cava, inferior vena cava, and coronary sinus. The superior vena cava enters the superior aspect of the right atrium and directs its blood flow toward the tricuspid valve. The inferior vena cava returns blood from the lower body, The coronary sinus returns most of the blood from the heart itself [1].

Right Ventricle

The right ventricle is the most anterior chamber of the heart, It is the

smaller of the two ventricular chambers, separated from the left ventricle by the interventricular septum, which bulges into the right ventricle.

The right ventricle contains three papillary muscles. Chordae tendineae (fibrous cords) extend upward from the papillary muscles and attach to the leaflet edges and to the ventricular side of the tricuspid valve. Blood enters the right ventricle via the tricuspid valve, and passes through the pulmonic valve into the main pulmonary artery [1].

Left Atrium

The left atrium is the left upper posterior chamber of the heart, it is cuboidal shaped, smaller than the right atrium (volume, 55 to 65 mL), but with thicker walls (3 mm) and higher pressure. It receives oxygenated blood from the lungs via four pulmonary veins (two from each lung) [1].

Left Ventricle

The left ventricle is a high-pressure, muscular chamber, 2.5 to 3 times thicker than the right ventricle. Two papillary muscles, the larger anterolateral and the smaller posteromedial arise from the free wall and have a variable number of heads.They anchor the chordae tendineae of the mitral valve, which are thicker than

The tricuspid valve chordae. Blood enters the left ventricle via the mitral valve and is ejected at a 90- to 120-degree angle through the aortic valve. The ejection phase is shorter in the left ventricle, but the pressure is greater compared with right ventricular contraction [1].

CARDIAC VALVES

Tricuspid Valve

The tricuspid valve is the largest of the heart valves and maintains forward flow of blood through the right heart. The functional components of the

tricuspid valve include the three leaflets, commissures, annulus, chordae tendineae, papillary muscles, and the right ventricle. The leaflets are named for their anatomic position: anterior, posterior, and septal. The valve leaflets are attached to a discontinuous fibrous annulus that has a “D” shape [2].

Pulmonic Valve

The pulmonic valve is the most anterior valve of the heart, located between theright ventricular outflow tract and the main pulmonary artery. It is the mirror image of the aortic valve, containing right, left, and anterior cusps (or leaflets) that are thinner than those of the aortic valve [2].

Mitral Valve

The mitral valve is named after the miter, a tall ornamental hat worn by bishops and abbots. The valve, located between the left atrium and the left ventricle, maintains the forward flow of blood in the left heart. The mitral valve has six components: leaflets, commissures, annulus, chordae tendineae, papillary muscles, and left ventricle. There are two mitral leaflets, the anterior and the posterior, which have similar surface areas but different shapes. During atrial contraction, the valve forms an ellipsoid orifice. During ventricular contraction, the atrial side of the leaflets coat, preventing the regurgitation of blood into the atrium. Unlike the tricuspid chordae, mitral chordae do not have insertions into the septum [2].

Aortic Valve

The aortic valve, located between the left ventricle and the aorta, is thicker and stronger than the pulmonic valve. It consists of three semilunar (half-moon) cusps located within the sinuses of Valsalva, three commissures, and an annulus. The three semilunar cusps, left, right, and noncoronary (or posterior), are pocket-like structures. The valve has triangular-shaped. central orifice when fully opened during systole. In diastole, blood fills the pocket-like cusps, causing the valve to close by coating on the ventricular surfaces of the cusps [2].

Vena Cava

These large veins return blood from the body to the right atrium. The superior vena cava enters the upper portion of the right atrium, where its blood flow is directed toward the tricuspid valve. The inferior vena cava is larger than the superior vena cava. It receives blood from the lower body and from the abdominal viscera [3].

Pulmonary Arteries

The main pulmonary artery is the most anterior cardiac vessel. It arises from the base of the right ventricle [3].

Pulmonary Veins

Four pulmonary veins, the right, and left, superior and inferior, return oxygenated blood from the lungs to the left atrium. Occasionally, five or six pulmonary veins may be found. The atrial muscle extends for 1 to 3 cm within the pulmonary veins and functions as a sphincter to prevent the reflux of blood during atrial systole [3].

The Aorta

The aorta arises from the aortic fibrous ring and passes superiorly and to the right as the ascending aorta. The proximal aorta (aortic root) is dilated and contains the aortic valve and the sinuses of Valsalva [3].

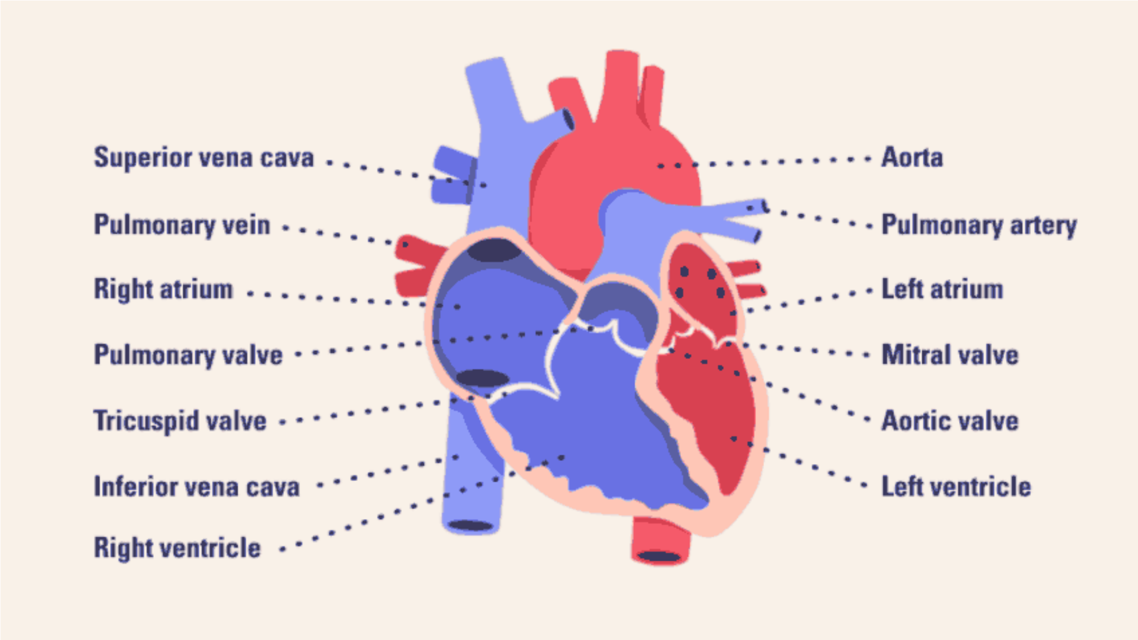

Figure 1. Anatomy of the heart

Heart cycle circulation:

A single cycle of cardiac activity can be divided into two basic phases - **diastole** and **systole**.

Diastole represents the period when the ventricles are relaxed (not contracting). Throughout most of this period, blood is passively flowing from the left atrium (LA) and right atrium (RA) into the left ventricle (LV) and right ventricle (RV), respectively. The blood flows through atrioventricular valves (mitral and tricuspid) that separate the atria from the ventricles. The RA receives venous blood from the body through the superior vena cava (SVC) and inferior vena cava (IVC). The LA receives oxygenated blood from lungs through four pulmonary veins that enter the LA. At the end of diastole, both atria contract, which propels an additional amount of blood into the ventricles.

Systole represents the time during which the left and right ventricles contract and eject blood into the aorta and pulmonary artery, respectively. During systole, the aortic and pulmonic valves open to permit ejection into the aorta and pulmonary artery. The atrioventricular valves are closed during systole, therefore no blood is entering the ventricles; however, blood continues to enter the atria though the vena cavae and pulmonary veins [4].

Aortic pressure is measured by inserting a pressure-measuring catheter into the aorta from a peripheral artery, and the left ventricular pressure is obtained by placing a catheter inside the left ventricle and measuring changes in intraventricular pressure as the heart beats. Left atrial pressure is not usually measured directly, except in investigational procedures; however. Ventricular volume changes can be assessed in real time using echocardiography or radionuclide imaging, or by using a special volume conductance catheter placed within the ventricle.

To analyze systole and diastole in more detail, the cardiac cycle is usually divided into seven phases. The first phase begins with the P wave of the electrocardiogram, which represents atrial depolarization, and is the last phase of diastole. Phases 2-4 represent systole, and phases 5-7 represent early and mid-diastole. The last phase of the cardiac cycle ends with the appearance of the next P wave, which begins a new cycle.

Phases of the heart cycle

<u>Phase 1 atrial contraction (2)</u>

This is the first phase of the cardiac cycle. It is initiated by the P wave of the electrocardiogram (ECG), which represents electrical depolarization of the atria. Atrial depolarization initiates contraction of the atrial musculature. As the atria contract, the pressure within the atrial chambers increases, which forces more blood flow across the open atrioventricular (AV) valves, leading to a rapid flow of blood into the ventricles. Blood does not flow back into the vena cava because of inertial effects of the venous return and because the wave of contraction through the atria moves toward the AV valve thereby having a "milking effect." However, atrial contraction does produce a small increase in venous pressure.

Atrial contraction normally accounts for about 10% of left ventricular filling when a person is at rest because most of ventricular filling occurs prior to atrial contraction as blood passively flows from the pulmonary veins, into the left atrium, then into the left ventricle through the open mitral valve.

After atrial contraction is complete, the atrial pressure begins to fall causing a pressure gradient reversal across the AV valves. This causes the valves to float upward (pre-position) before closure. At this time, the ventricular volumes are maximal, which is termed the **end-diastolic volume** (EDV). The left ventricular EDV (LVEDV), which is typically about 120 ml, represents the ventricular preload and is associated with end-diastolic pressures of 8-12 mmHg and 3-6 mmHg in the left and right ventricles, respectively [4].

Isovolumetric Contraction: (3) Phase 2:

ALL VALVES CLOSED

The AV valves close when intraventricular pressure exceeds atrial pressure. Ventricular contraction also triggers contraction of the papillary muscles with their chordae tendineae that are attached to the valve leaflets. This tension on the the AV valve leaflets prevent them from bulging back into the atria and becoming incompetent (i.e., “leaky”).

During the time period between the closure of the AV valves and the opening of the aortic and pulmonic valves, ventricular pressure rises rapidly without a change in ventricular volume (i.e., no ejection occurs). Ventricular volume does not change because all valves are closed during this phase. Contraction, therefore, is said to be "isovolumic" or "isovolumetric." Individual myocyte contraction, however, is not necessarily isometric because individual myocyte are undergoing length changes. Some individual fibers contract isotonically (i.e., concentric, shortening contraction), whereas others contract isometrically (i.e., no change in length) or eccentrically (i.e., lengthening contraction). Therefore, ventricular chamber geometry changes considerably as the heart becomes more spheroid in shape; circumference increases and atrial base-to-apex length decreases [4].

Phase 3 Rapid ejection (4)

AORTIC AND PULMONIC VALVES OPEN;A-V REMAIN CLOSED

This phase represents initial, rapid ejection of blood into the aorta and pulmonary arteries from the left and right ventricles, respectively. Ejection begins when the intraventricular pressures exceed the pressures within the aorta and pulmonary artery, which causes the aortic and pulmonic valves to open. Blood is ejected because the total energy of the blood within the ventricle exceeds the total energy of blood within the aorta. In other words, there is an energy gradient to propel blood into the aorta and pulmonary artery from their respective ventricles. During this phase, ventricular pressure normally exceeds outflow tract pressure by a few mmHg. This pressure gradient across the valve is ordinarily low because of the relatively large valve opening (i.e., low resistance). Maximal outflow velocity is reached early in the ejection phase, and maximal (systolic) aortic and pulmonary artery pressures are achieved.

Left atrial pressure initially decreases as the atrial base is pulled downward, expanding the atrial chamber. Blood continues to flow into the atria from their respective venous inflow tracts and the atrial pressures begin to rise. This rise in pressure continues until the AV valves open at the end of phase 5 [4].

Phase 4 reduced ejection (5)

AORTIC AND PULMONARY VALVES OPEN; A-V VALVES REMAIN CLOSED

epolarization leads to a decline in ventricular active tension and pressure generation; therefore, the rate of ejection (ventricular emptying) falls. Ventricular pressure falls slightly below outflow tract pressure; however, outward flow still occurs due to kinetic energy of the blood.

Left atrial and right atrial pressures gradually rise due to continued venous return from the lungs and from the systemic circulation, respectively [4].

Phase 5 : Isovolumetric relaxation (6)

ALL VALVES CLOSED

When the intraventricular pressures fall sufficiently at the end of phase 4, the aortic and pulmonic valves abruptly close (aortic precedes pulmonic).After valve closure, the aortic and pulmonary artery pressures rise slightly.The rate of pressure decline in the ventricles is determined by the rate of relaxation of the muscle fibers. Although ventricular pressures decrease during this phase, volumes do not change because all valves are closed [4].

Phase 5 : Rapid filling (7)

A-V VALVES OPEN

As the ventricles continue to relax at the end of phase 5, the intraventricular pressures will at some point fall below their respective atrial pressures. When this occurs, the AV valves rapidly open and passive ventricular filling begins. Despite the inflow of blood from the atria, intraventricular pressure continues to briefly fall because the ventricles are still undergoing relaxation. Once the ventricles are completely relaxed, their pressures will slowly rise as they fill with blood from the atria [4].

Phase 7 : Reduced filling (8)

A-V VALVES OPEN

As the ventricles continue to fill with blood and expand, they become less compliant, and the intraventricular pressures rise. The increase in intraventricular pressure reduces the pressure gradient across the AV valves so that the rate of filling falls late in diastole.Aortic and pulmonary arterial pressures continue to fall during this period[4].

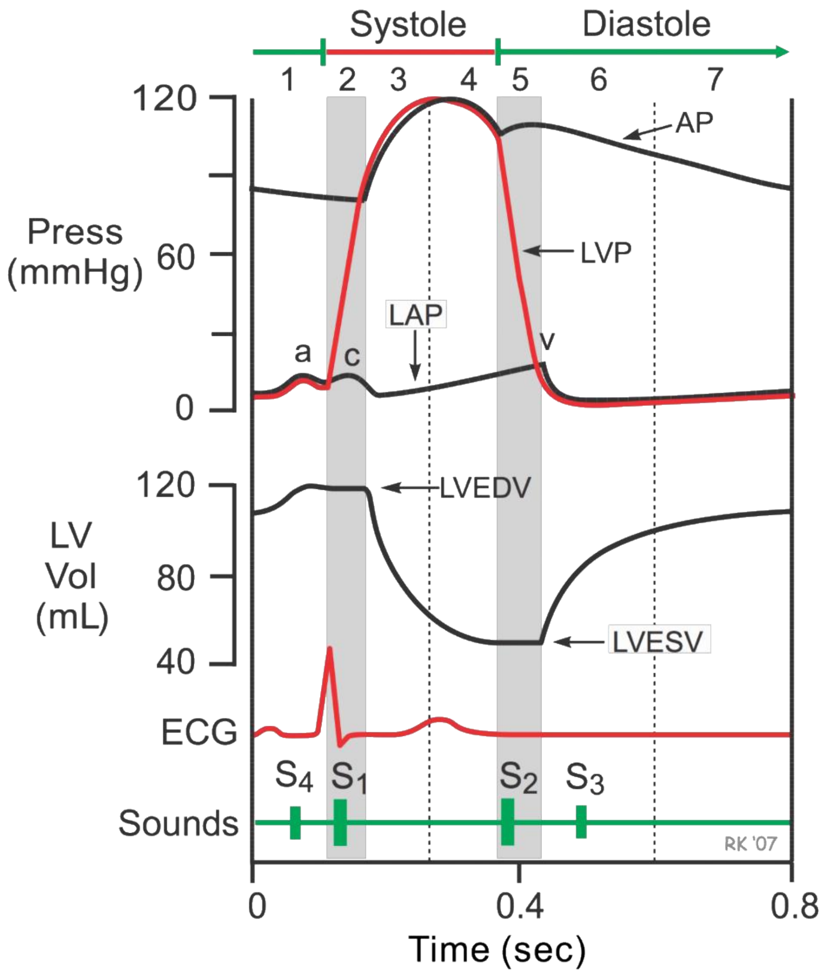

Figure 2. Phases of Cardiac cycle.

1.2 Prosthetic heart valves

The outcomes of individuals with valvular heart disease have significantly improved since the early 1960s when valve replacement surgery was first performed. Currently, 280000 artificial valves are implanted worldwide and 90000 in the United States each year; roughly half of these are mechanical valves and the other half are bioprosthetic valves. Despite significant advancements in surgical techniques and prosthetic valve design over the past few decades, valve replacement does not offer the patient a permanent cure. Instead, "prosthetic valve disease" is substituted for "native valve disease," and the hemodynamics, durability, and thrombogenicity of prosthetic valves have an impact on the outcomes of patients who have their valves replaced[5].

Nevertheless, by carefully choosing the right prosthesis for each patient and managing their medical care and follow-up after implantation, many prosthesis-related issues can be avoided or their effects reduced. This article's goal is to give readers an overview of the present state of knowledge and potential directions in selecting the best prosthesis and managing patients following valve implantation.

Types of Prosthetic Heart Valve Design:

The properties of a typical native valve should be replicated by the ideal valve replacement. It should have excellent implantability, extended durability, high thromboresistance, and excellent hemodynamics in particular. Unfortunately, there is no perfect prosthetic valve replacement, and all of the ones that are currently on the market have drawbacks [6].

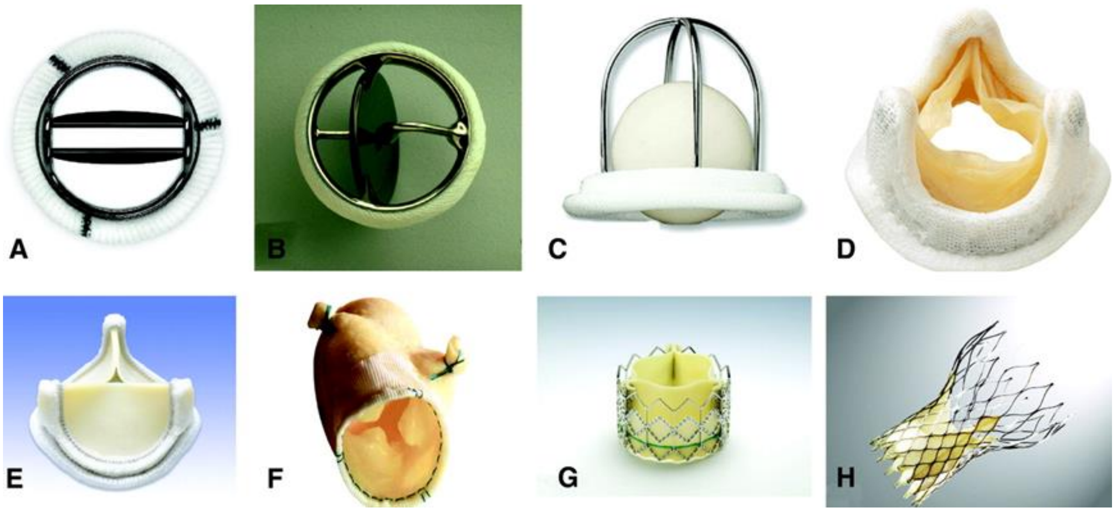

Figure 3 Types of Prosthetic Heart valve Design

Different types of prosthetic valves. A, Bileaflet mechanical valve (St Jude); B, monoleaflet mechanical valve (Medtronic Hall); C, caged ball valve (Starr-Edwards); D, stented porcine bioprosthesis (Medtronic Mosaic); E, stented pericardial bioprosthesis (Carpentier-Edwards Magna); F, stentless porcine bioprosthesis (Medtronic Freestyle); G, percutaneous bioprosthesis expanded over a balloon (Edwards Sapien); H, self-expandable percutaneous bioprosthesis (CoreValve) [7].

Mechanical Valves

Three basic types of mechanical valve design exist: bileaflet, monoleaflet, and caged ball valves (A, B, C)[7].

Caged Ball Valves

Caged ball valves, which consist of a silastic ball with a circular sewing ring and a cage formed by 3 metal arches, are no longer implanted. However, several thousands of patients still have caged ball valves, and these patients require follow-up.

Monoleaflet Valves

Monoleaflet valves are composed of a single disk secured by lateral or central metal struts. The opening angle of the disk relative to valve annulus ranges from 60° to 80°, resulting in 2 distinct orifices of different sizes.

Bileaflet Valves

Bileaflet valves are made of 2 semilunar disks attached to a rigid valve ring by small hinges. The opening angle of the leaflets relative to the annulus plane ranges from 75° to 90°, and the open valve consists of 3 orifices: a small, slit-like central orifice between the 2 open leaflets and 2 larger semicircular orifices laterally.

Bioprosthetic Valves

<u>Stented Bioprostheses</u>

The design of bio prostheses purports to mimic the anatomy of the native aortic valve (D, E)[7]. Porcine bioprosthetic valves consist of 3 porcine aortic valve leaflets cross-linked with glutaraldehyde and mounted on a metallic or polymer supporting stent. Pericardial valves are fabricated from sheets of bovine pericardium mounted inside or outside a supporting stent.

Stentless Bioprostheses

In an effort to improve valve hemodynamics and durability, several types of stentless bioprosthetic valves have been developed (F) [7]. Stentless bio prostheses are manufactured from whole porcine aortic valves or fabricated from bovine pericardium.

Percutaneous Bio prostheses

Percutaneous aortic valve implantation is emerging as an alternative to standard aortic valve replacement (AVR) in patients with symptomatic aortic stenosis considered to be at high or prohibitive operative risk (G, H)[7]. The valves are usually implanted using a percutaneous transfemoral approach. To reduce the problems of vascular access and associated complications, a transapical approach through a small thoracotomy may also be used. At present, the procedure appears promising, but it remains experimental and is currently undergoing further investigation.

Bioprosthetic Vs Mechanical Valve

To maximize the results for patients having valve replacement, it is challenging but crucial to choose the proper valve for the particular patient. Choosing between a mechanical and a bioprosthetic valve is the first stage in this decision-making process [8].The patient's age, life expectancy, preference, indication or contraindication for warfarin medication, and comorbidities are the most crucial considerations that should be taken into account in this initial phase.

The weight allocated to patient age has decreased in the most recent American College of Cardiology/American Heart Association and European guidelines, whilst the patient's preference is now given far greater relevance. The following factors in favor of utilizing a mechanical valve include: The patient is already taking anticoagulant medication (mechanical prosthesis in another position or at high risk for thromboembolism);

- ❖ The patient is aware and wants a mechanical valve;

- ❖ The patient has no contraindications for long-term anticoagulation.
- ❖ The patient is at risk of accelerated bioprosthesis structural deterioration (young age, hyperparathyroidism, renal insufficiency);
- ❖ The patient is under 65 years of age and has a long-life expectancy.

A bioprosthesis, on the other hand, might be preferred in the following circumstances:

- ➢ The informed patient desires a bioprosthesis;
- ➢ Good-quality anticoagulation is not available (contraindication or high risk, compliance issues, lifestyle);
- ➢ The patient is younger than 65 and/or has a limited life expectancy.
- ➢ The patient is a woman of childbearing age.

Young patients and women who are pregnant experience bioprosthetic degeneration more quickly. This means that a lady who has finished having children in her late 30s or early 40s should probably be advised to get a mechanical valve[9].

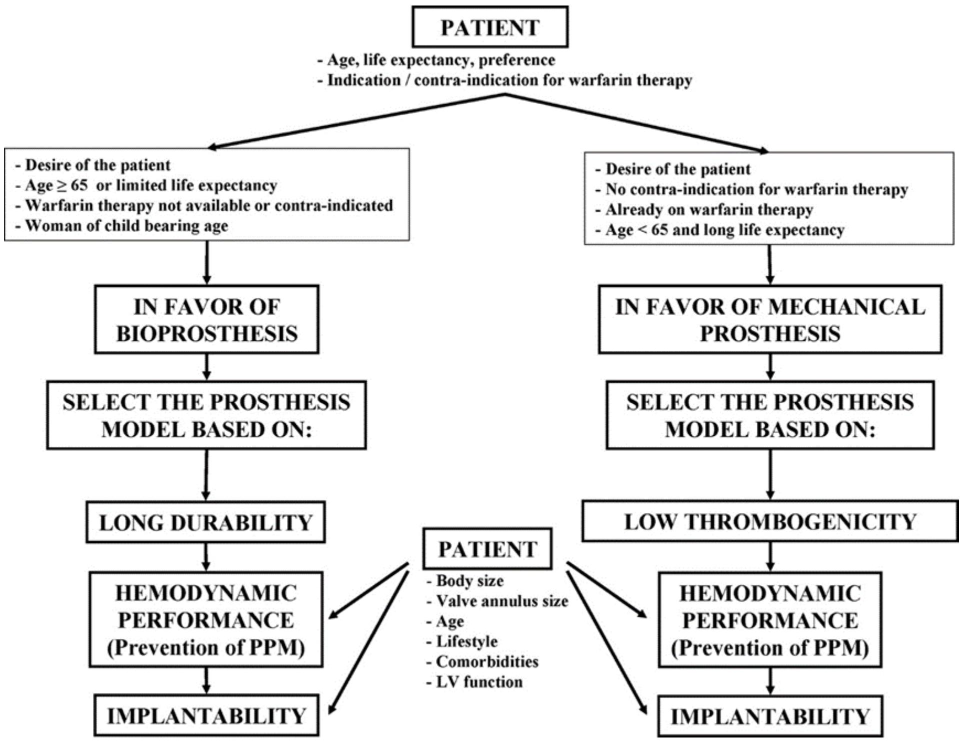

Figure 4 Algorithm for the selection of the optimal prosthesis in the individual patient

To replicate systemic circulation and conduct testing on the functioning of the heart valve, a closed loop hydraulic circuit must be created. It is also useful to evaluate how hemodynamic factors, such as cardiac output or blood pressure, affect the functioning of a cardiac prosthesis.

Closed hydraulic circuits were created to test mechanical valves. This test bench was made up of pulsatile systems controlled by an actuation system, which can recreate an artificial heartbeat, and artificial materials representing some compartments involved in the cardiac cycle, which will aid in evaluating some parameters presented during the cycle and determining whether these measurements match real physiological readings[10].

Cardiologists might utilize it to examine and evaluate the behavior and adaptability of a heart valve before surgery, particularly in the presence of aortic stenosis. Modeling pathological, physiological, and hemodynamic factors might yield enormous advantages[10].

To create the organ as close to human physiology as feasible, additive manufacturing techniques were used in the replication of the left ventricle and aorta. Through the use of appropriate materials, it may be possible to imitate illnesses connected to the organ, the blood artery, or the valve itself. Another critical feature is the replication of blood flow pulsatility, which is dependent on a linear actuator. It should be able to regulate the membrane of a pump, which will determine the volume of water pushed and drawn in the ventricular chamber. To correctly regulate the movement of the membrane, the actuator should be fitted with a potentiometer that can give feedback position when the resistance value changes. Furthermore, to make optical measurements on the cardiac prosthetic valves, clear and flat surfaces around the ventricle are required to ensure optical access. The bench should also enable the testing of various types of valves by putting them between the left ventricle and the aortic arch. Other elements could be installed into the bench design to protect it during operation behavior like Resistance tape and check valve which prevent backflow of fluid.

1.3 State of the art: Aortic Test Bench

There are two types of prosthesis heart valve: biological and mechanical, in the biomedical engineering field, a Technique called polymeric heart valves combined the main advantages of both biological and mechanical valve prosthesis and replicate the native heart, the most important thing that it can be a candidate of a trans catheter aortic valve replacement but due to lack of reliability it is yet not considered in the medical community, therefore more experiments were done on the prosthetic heart valves in the consideration of multiple conditions using an aortic test bench.

Some researches have been made over the few past years on the effect of multiple conditions on the prosthesis heart valve and here's a review of some articles:

Luraghi et al. (2017) shows in their study that the Fluid-Structure interaction analyses (FSI) are better at describing the exact behavior of PHVs than the Finite-Element (FE) analyses. Results have clearly shown that FSI simulations are closer to reality. the differences in leaflet thickness present in the analyzed model is responsible for the non-symmetric valve opening during the low flow rate, considering that the thickness differences are non-physiological, in the physiological scenario the differences between FSI and FE may be irrelevant. Simulations have been run with identical leaflets to exhibit the reliability of this point and the result is also in a non-symmetrical opening of the valve for the FSI simulation. The final conclusion is that the Geometric Orifice Areas (GOA) computed from FSI simulation is more accurate than that derived from a FE simulation. Moreover, the importance of this work rests on using experimental data to validate the in silico models quality and quantity. The material that has been used in this study: The PHV prototypes similar to those presented by De Gaetano et al. (2015a) made of styrenic block copolymer (SBP), Pulsatile tests were conducted on an in-house pulse duplicator (De Gaetano et al., 2015b), also a Structural FE and FSI simulations were performed. Importantly, a linear actuator has been used to provide a linear motion and to give the system a trigger pulse in order to start the experiment [11].

Bazan et al. (2016) describes through their work the experimental validation of a left heart simulator, at Escola Politécnica of the University of São Paulo (EPUSP), the work has been done using three heart rates (HR): 60, 80 and 100 bpm (beats per minute) under physiological conditions and using prosthetic valves with no simulated stenosis nor valve thickening using in vivo data to examine the responses from the cardiac simulator for each HR. the study agreed that the cardiac simulator is appropriate for in vitro evaluation of prosthetic heart valves. Therefore, the cardiac simulator was suitable to these conditions, in conformity with the human physiological parameters. Once the cardiac simulator operating parameters allow valid experimental comparisons of flow through mitral or aortic prostheses, the Hydrodynamic testing of prosthetic heart valves can be started [12].

Moreover, the work of **Azadani (2019)** "Principles of Heart Valve Engineering" Reviewed in one of its chapters an examination of the common in vitro techniques for evaluating the structural and hemodynamic performance of prosthesis heart valves; In vitro experimental testing is required in order to design and verify the implantable medical devices. Prosthetic heart valves are tested for stable forward and backward flow while heart valve substitutes are exposed to pulsatile flow under physiological conditions, this helps to confirm the validity of in vitro pulsatile flow studies. The performance of heart valve substitute's hemodynamics has been examined through steady flow loops across many studies published in the scientific literature. Additionally, pulsatile flow devices may accurately reproduce the features of blood flow within the heart and vasculature system, A pulse duplicator system can be used to simulate ventricular and arterial function for evaluating prosthetic heart valves.. An in vitro closed flow loop system called a pulse duplicator is designed to generate pulsatile flow and mimic the hemodynamic conditions of the left or right side of the heart. Assessment of the hemolytic and thrombogenic capabilities of different prosthetic valve designs largely depends on flow field characterization of prosthetic heart valves. In order to accurately mimic blood flow characteristics, kinematic similarity must be exploited properly in a pulse duplicator system. Kinematic similarity is often attained when both the dynamic and the geometric similarity condition are satisfied. The geometry of a pulse duplicator should be similar to that of the human heart because geometric similarity also pertains to shape similarity. As a result, pulse duplicator devices have been utilized in the past to test aortic heart valves using realistic glass, acrylic, silicone, and cadaveric aortic roots [13].

The study of **Querzoli et al. (2016)** proposed pulse duplicator consists of a passively pulsing ventricle, a proximal aorta that is flexible and coronary arteries linking to the sinuses of Valsalva. The coronary flow, which is based on the contraction and relaxation of the heart muscle during the cardiac cycle, is modulated by a self-regulating device that mimics the physiological mechanism. The aortic root and blood pressure within the vascular resistance are the two fundamental determinants of coronary circulation. The change of the intraventricular pressure, which reaches a maximum value during the systole and lower values during the diastole, predominantly determines the blood pressure. The cyclic compression of coronary arteries by the heart muscle determines the vascular resistance, which it reach its peak during the systolic contraction. The resulting flow rate demonstrates a rapid peak at the start of diastolic relaxation and remains relatively low during the whole systole. The behavior of the aortic pressure causes the flow rate to then progressively drop. Therefore, the diastole is when the majority of coronary blood flow occurs (Rubio and Berne 1975). An aortic root model with a bi-leaflet mechanical prosthetic valve was used to conduct the study. Coronary arteries were left open in a preliminary series of studies. The coronaries were then clamped in order to compare and contrast the impact of coronary flow on the aortic root's fluid dynamics. The flow via the proximal aorta and the left ventricle are both replicated by the pulse duplicator. The water-filled ventricle, which is contained in a closed chamber, changes volume on a periodically and connects with a piston that is powered by a linear motor to create the fluid flow. A personal computer regulates the linear motor velocity, which follows a digitally specified speed law. Two coronary arteries emanating from the Valsalva sinuses are present in the model aorta. Through the use of the chamber, coronary flow can be controlled [14].

2. Materials and Methods

This section represents the used experimental setup and the measurement techniques that are used to implement our approach of the aortic test bench. Then control system section is described and demonstrated; it explains the use of different electronic elements to run the bench which acts as a trigger part of the designed bench. The second part exhibits physiological-based materials which describe the system circulation of our heart. The last section explains the measurement tools used to extract useful information about the fluid flow and pressure changes with physiologically based materials.

2.1 Control System

1. Based on the physiological parameters obtained from numeric physiological parameters and on the existed type of piston pump, there was a limit to choose the right linear actuator to control the inward and outward movement of the pump and to preserve its shape without deforming during its motion, a suitable stroke has been chosen to control the membrane of pump movement.

Sensor and actuators

Since any type of control needs two main parts; sensor and actuator, the sensor in this case was a potentiometer; it is used to measure the position of an actuator system. The potentiometer versions provide a voltage signal that rises as the actuator expands and falls as it retracts.

Linear actuators use potentiometers, which are variable resistors, to provide positional feedback based on how their resistance changes. Like the used Actuator, it that use potentiometers for feedback will also include the three additional wires; wire 1 for the input voltage, wire 2 for the variable resistor, and wire 3 for ground. By measuring the voltage between wire 2, the output, and ground, you can determine the absolute position of the linear actuator from the output of the potentiometers.

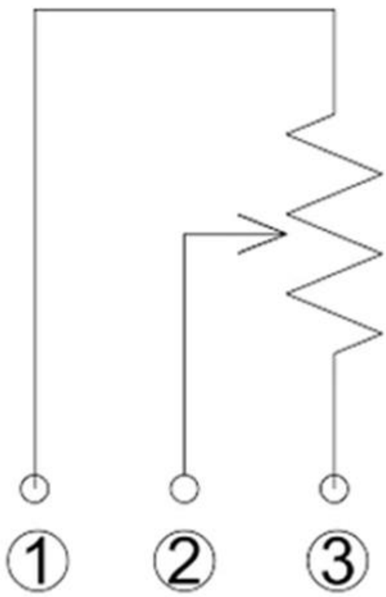

Figure 5 Variable resistor to read voltage when piston moves

We need an analog to digital converter to read the value; this can be done using Arduino, to read this positional value when the actuator moves in order to make use of this feedback. Since piston is an actuator (motor), as all the motors; it needs a driver, the common driver of dc motors is H-Bridge. For this actuator, we will use a specific driver of H-Bridge as will be explained later. The figure below shows our flowchart.

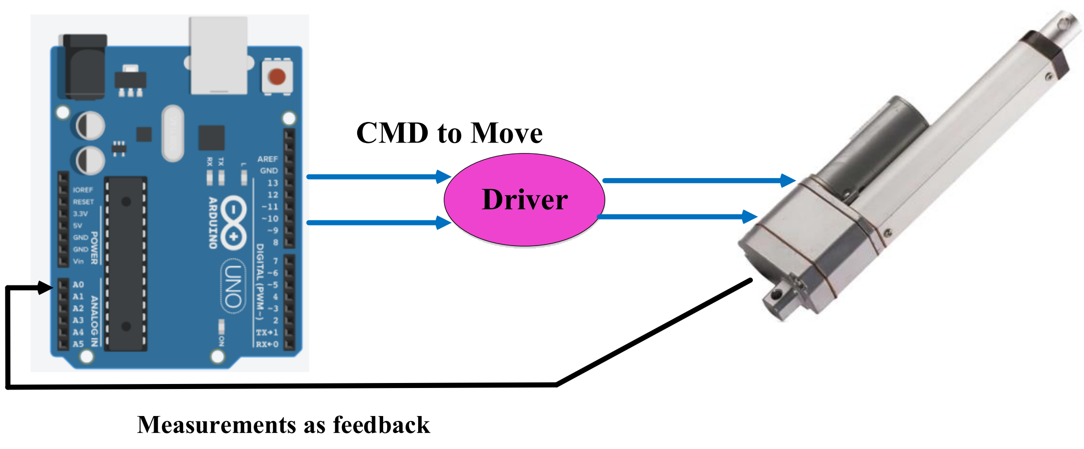

Figure 6 flowchart of the method

To move the piston pump, a Fergelli Feedback linear actuator with a maximum stroke of 2 inches (50 mm) was used. This linear actuator is made up of two wire bundles: 2 for power wires where the red is 12 volts positive polarity, black wire for ground polarity, and the other bundle for internal potentiometer where the white and black represent reference terminals (source of power) and the yellow one describes positional signal. It worth mentions that there are pistons with an input voltage 24 volts; they also can be used in our case, but 12Volt is suitable.

Arduino

Arduino is a free and open-source platform for creating and controlling electronic-based projects. Arduino is made up of a physical programmable circuit board (also known as a microcontroller) and a piece of software called an IDE (Integrated Development Environment) that runs on your computer and is used to create and upload computer code to the physical board. The main components of the Arduino UNO board are as follows:

- The USB connector is used to load an Arduino IDE program onto the Arduino board. This connection can also be used to power the board through Laptop.
- Power port an AC-to-DC converter or a battery can power the Arduino board. Connecting the power supply is as simple as inserting a 2.1mm center-positive connector into the board's power port. In addition, the Arduino UNO board runs on 5 volts but can handle a maximum voltage of 20 volts.
- A voltage regulator (located between the power port and the USB connection) prevents the board from overheating if the board is supplied with a greater voltage. Microcontroller with 28 pins, it is the most visible black rectangular chip. Consider it the brains of your Arduino. Atmel's Atmega328P microprocessor is utilized on the UNO board (a major microcontroller manufacturer).
- The bootloader is pre-programmed on the Atmega328P. This allows you to upload a new Arduino program straight into the device without the need for an

additional hardware programmer, making the Arduino UNO board simple to use.

- Analog input pins: these analog input pins on the Arduino UNO board are designated "Analog 0 to 5." These pins can read an analog sensor signal, such as a temperature sensor, and convert it to a digital value that the system can interpret. Because of their high internal resistance, these pins only monitor voltage and not current. As a result, only a minimal amount of current passes through these pins.
- Digital pins: these pins are designated "Digital 0 to 13." These pins can function as an input or output pins. When used as output, these pins provide electricity to the components attached to them. When used as input pins, they read the signals from the component to which they are attached. When utilized as output pins, digital pins provide 40 milliamps of current at 5 volts, which is more than enough to light an LED.
- Some digital pins include the tilde (~) sign next to the PINs (pin numbers 3, 5, 6, 9, 10, and 11). These pins function normally as digital pins but may also be used for pulse width modulation (PWM) control signals, which replicate analog output such as fading.
- RS232 communication pins: they are located in digital bins (0: Rx, 1: Tx). This is used for communication with other devices.
- Other communication protocol such as Tow Wire I2C and SPI; are common protocols and they depend on the analog pins.

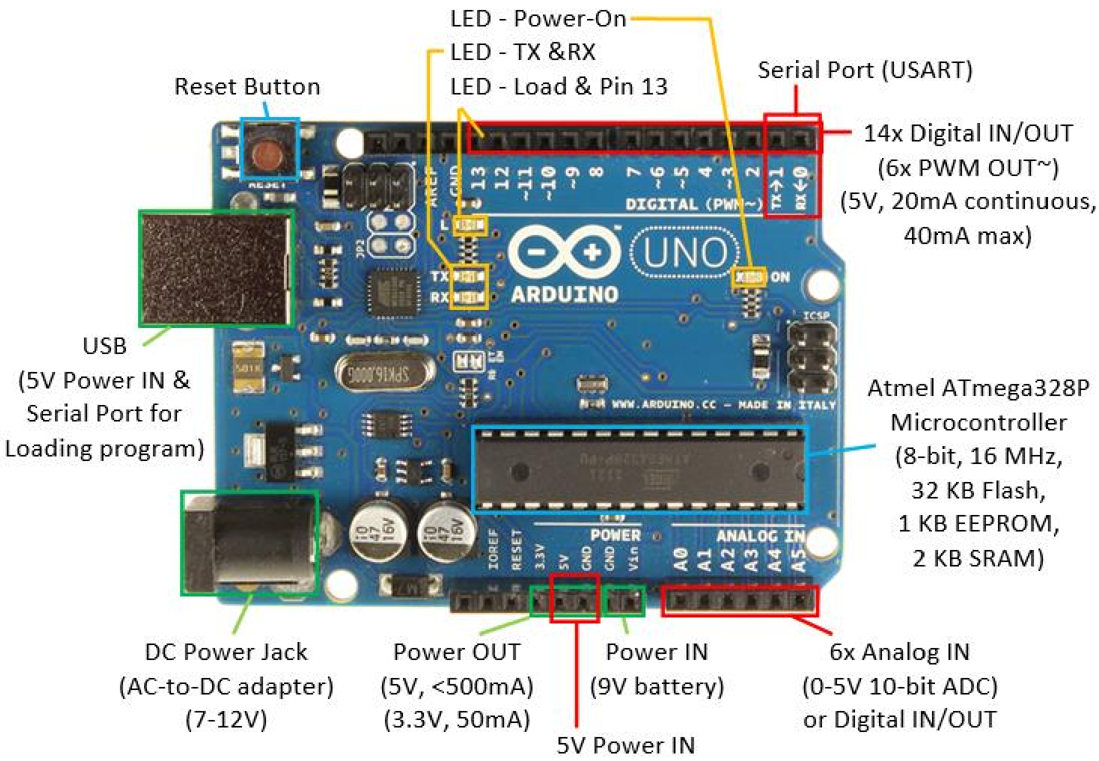

Figure 7 Components of Arduino

They are a family of Arduino products such as Arduino Mega, Leonardo, Nano, Micro, etc. they differ in capabilities, Arduino Uno is suitable for our project. Since the recommended input voltage operating for Arduino is between 7 and 12v so, for more safe work, we intend to use a motor driver as a link between the power wires of the linear actuator and Arduino controller.

Motor Driver

The motor driver is BTS7960; it is a high-current **H** bridge module with complete integration for motor driving applications. Operating voltage of 24V, a maximum continuous current of 43A, PWM capabilities of up to 25 kHz, and active freewheeling. These modules use the PWM (Pulse Width Modulation) technology to drive DC motors or linear actuators. These modules change a constant input voltage into a variable voltage for the motor. The DC motor voltage may be changed to alter the speed. PWMs have a set frequency and can be modified by changing the time the pulse is HIGH (Duty Cycle). This module has 12 pins:

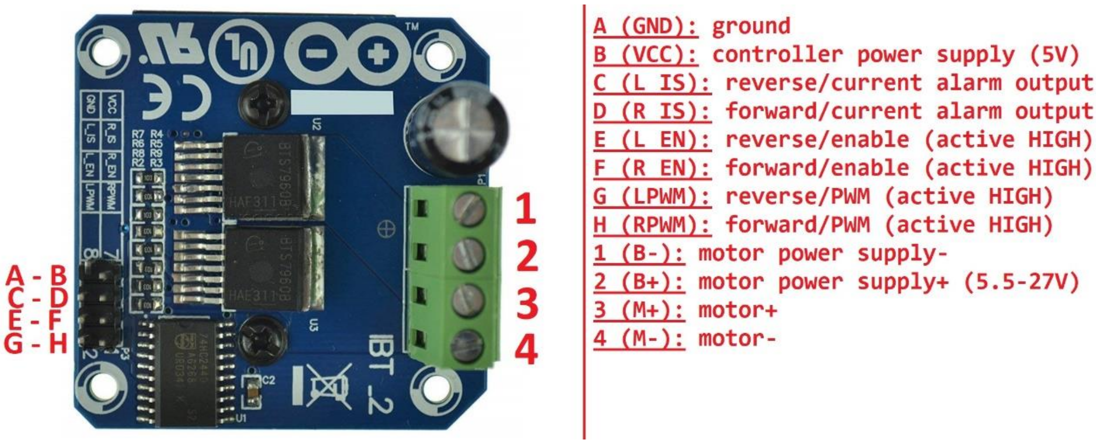

Figure 8 Motor driver pins diagram

Microcontroller pins (Low current):

- **VCC**: Module power supply – 5V
- **GND**: Ground
- **IS-R:** Input signal for detecting high current – Straight rotation
- **IS-L:** Input signal for detecting high current – Inverse rotation
- **EN-R:** Output Signal for controlling motor direction – Straight rotation
- **EN-L:** Output Signal for controlling motor direction – Inverse rotation
- **WM-R:** PWM Signal for controlling motor speed – Straight rotation
- **PWM-L:** PWM Signal for controlling motor speed – Inverse rotation

Motor pins (High current):

- **M+:** Motor Positive
- **M-:** Motor negative
- **B+:** Battery positive
- **B+:** Battery negative

To accomplish the closed loop circuit idea for this actuation system, an input signal should be stimulated into our system such that when the strength of this signal varies (for example, frequency, voltage), the position feedback of the linear actuator would reflect this modification of the input signal.

A potentiometer is a three-wire resistive device that serves as a variable resistor. Turning the knob causes the resistor to increase or decrease, which causes the voltage on the analog pin to which the potentiometer is linked to increase or decrease. The pins are a) Vcc, b) Ground, c) Output signal.

Power supply

The power supply is needed for all parts, a 12-volt power supply is employed as a power source for the specified test bench. Jumper wires are employed to link actuation system components. The use of a breadboard simplifies the interconnection of the components in building up this circuit.

Connections:

Connecting the USB to the PC is necessary not only to power the board up but to program it too. The Uno automatically draws power from either the USB or an external power supply. This is happened to upload our program to the Arduino, then the power supply will be connected.

Then, we connect 12 volts and ground power wires of the actuator with the input power pins of the motor driver, then we power them when the ground wires of the motor driver are connected to the positive and negative outlet of the 12volt power supply, and the 5volt ground pins of the driver are connected to the breadboard.

The right and left sides should be connected to the digital pins in Arduino and the same thing for the right and left PWM to transform the input signal into the output signal, PWM is a way to control analog devices with a digital output.

Voltage and ground wires of the feedback linear actuator are connected to the pins on within the breadboard we take common positive and negative wires connected to 5volt power and ground pins on Arduino respectively.

The potentiometer position signal is connected to the analog A1 pin in Arduino because we need to read its value and convert it to change the displacement of the linear actuator.

The ENABLE input can be driven by a PWM output to control the average motor current.

We implement a code on Arduino-based software to control the stroke of the linear actuator by converting its feedback resistance from 0 to 10 Kohm into the change in displacement, then we command a maximum of 20 mm stroke to trigger pump movement so, at 0 the linear actuator is retracted and 20 mm extracted.

The speed of the linear actuator is determined through the delay command in Arduino software so, if the delay time is increased the actuator will move slowly and the inverse will happen if the delay period is decreased through coding.

The full circuit is shown below:

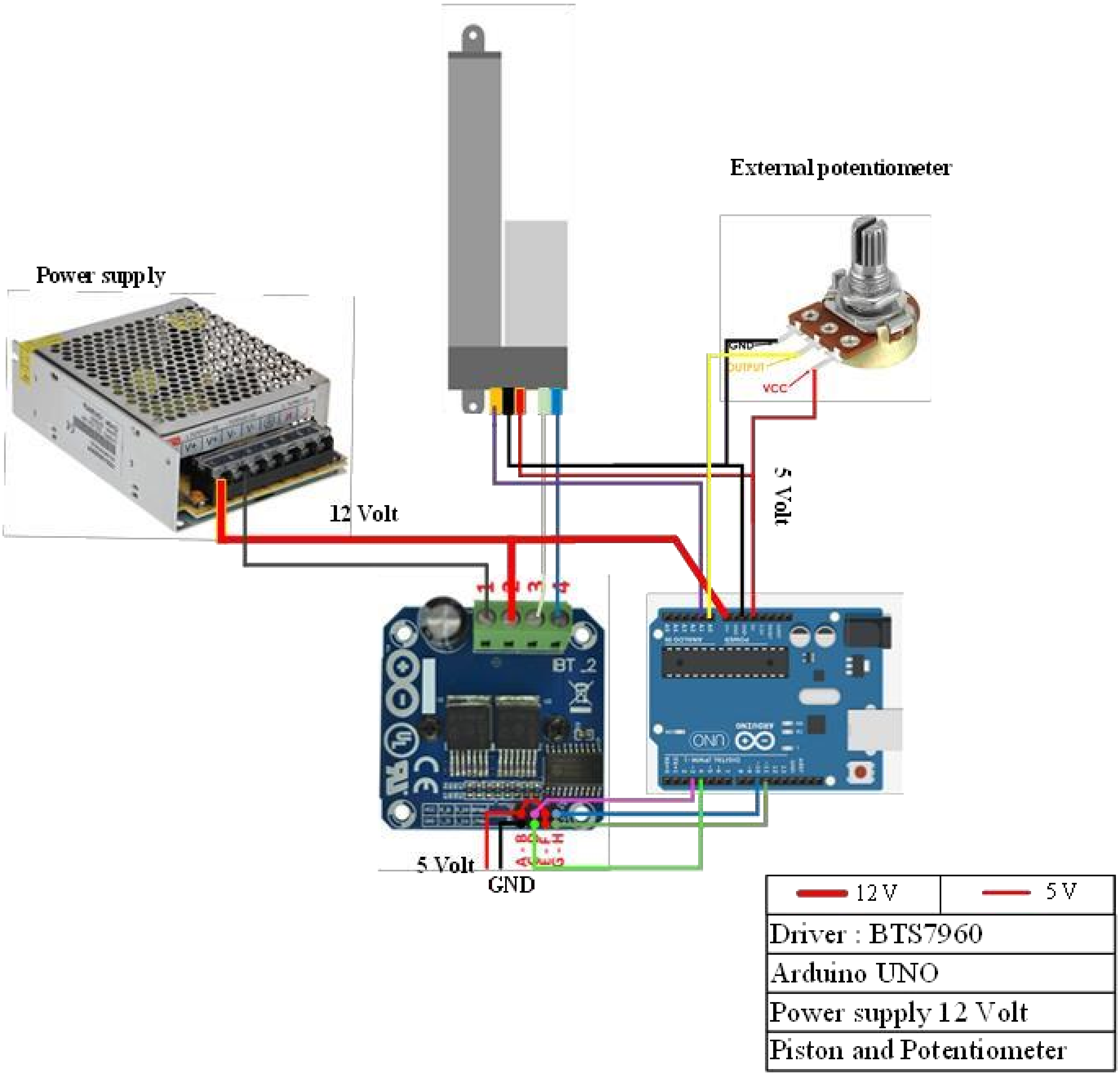

Figure 9 Control system diagram

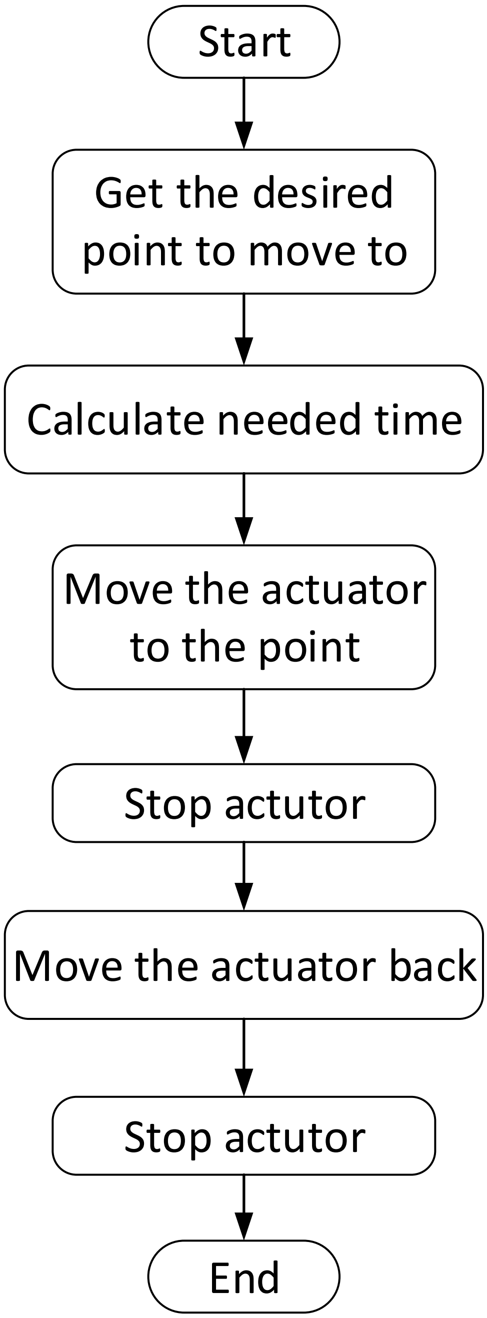

Figure 10 flowchart of controlling code

2.2 Realization of anatomical components

Following the implementation of the closed-loop control system, we will now describe the elements related to the anatomical part that describe the systemic

circulatory system used to construct this bench, which requires a response from the actuation system to induce flow inside the systemic circulation. UNIVPM laboratory team created the prototype of the anatomical components utilized in the test bench implementation, including the left ventricle, aorta, and aortic prosthetic valve.
First, we have a compliance chamber where the left ventricle is located, this chamber is made up of plexiglass material with 2 holes in the upper side to place the left ventricle inside it and the other hole where the piston pump is placed where it is movement triggers the flow inside the chamber during the working of this system
The left ventricle is made up of elastic material to provide flexible compression and relaxation during the systemic circulation cycle. The role of the left ventricle is to pump blood (in normal physiological behavior) into the aorta through the aortic prosthetic valve, which is placed between them. The aortic valve opens, allowing blood to flow from the left ventricle to the aorta. It shuts to stop blood from flowing in the other direction. The closed valve prevents blood from seeping back into your heart from your aorta. For the bench's development, a mechanical valve was employed. as an aortic valve aortic-based mechanical valve is comprised of elastic material that allows for easy opening and closing during test bench operation. It might be a useful method for simulating diseased valves. The aortic arch receives fluid as a consequence of ventricular squeezing or in other words when the ventricle compresses in response to a generated pressure created by the control system via the aortic valve, and subsequently draws fluid out of the compliance chamber directed into the reservoir tank through piping small tubes. We used a compliance chamber synthesized by plexiglass material to include an aortic arch. In addition, the compliance compartment is used to describe how easily the left ventricle chamber expands when it is filled with a volume of blood (water in our case). Physically, compliance (C) is defined as the change in volume (ΔV) divided by the change in pressure (ΔP). In this work, we utilized a mitral valve as a check valve in a pipeline to avoid backflow. A check valve is essentially a one-way valve that allows the fluid to flow freely in one direction but closes if the fluid flows back to safeguard our bench.

Hydraulic resistance tap along pipes between the compliance chamber and reservoir tank is included in designing of the test bench.

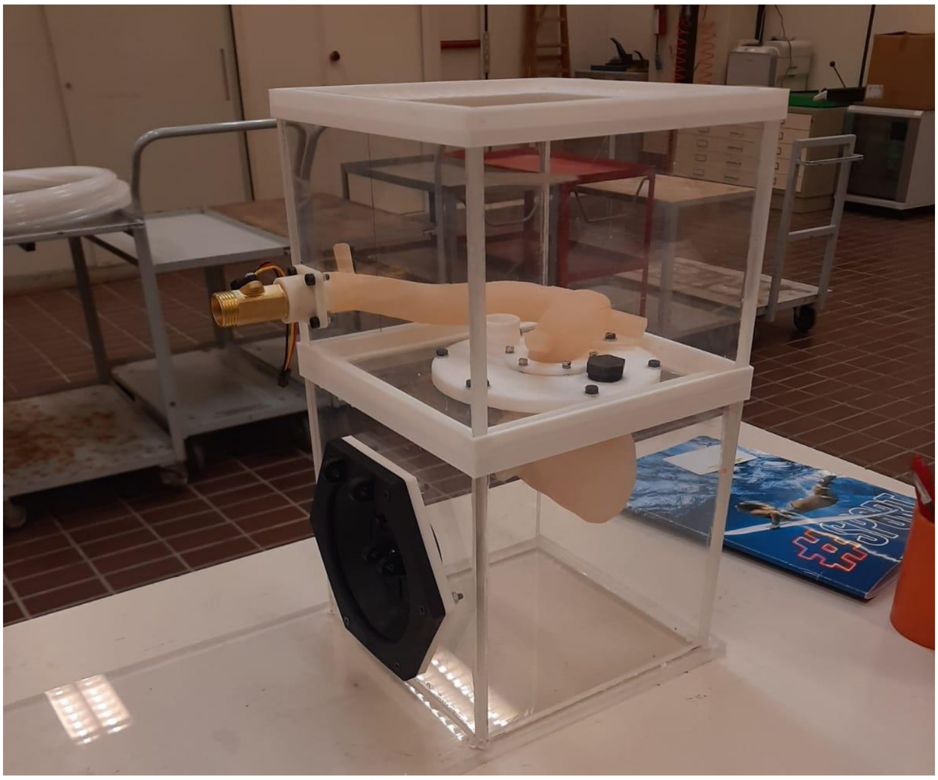

Figure 11 Synthesized Aortic arch and Ventricle

Figure 12 Prosthetic aortic valve

2.3 Measurements procedures

Electromagnetic flowmeters can be employed to detect flow at the aortic valve level. It works on the concept of electromagnetic induction where the voltage produced across the liquid by its passage via a magnetic field supplied to a tube is used to measure fluid flow.

Changes in the pressure of bench elements such as the left ventricle and aortic arch can be measured using a differential pressure sensor. For example, the ventricle is compressed and decompressed by the flow induced by the pump's motion during the mechanism process, so measuring ventricular pressure provides us with useful information about the parameters that characterize our bench.

Fluid flow measurement

2 Hall Effect JS flow sensors are used to measure the fluid flow in our designed test bench, Hall Effect flow sensor is a sensor with 3 connectors:

Figure 13 Hall effect JS flow sensor

Specification of this sensor can be concluded by flow rate range from 0.2 up to 6L/min. It requires 5 volts to 24 volts of working voltage with an output pulse of 4.7 volts. They are based on the principle of Faraday's Law of Electromagnetic Induction, in which liquid generates a voltage at the output level when it flows through a magnetic field. It was chosen because it is flow range matches the normal aortic flow readings during the cardiac cycle.

The first flow sensor is inserted just at the end aorta to measure the value of aortic flow that comes first from the left ventricle through the aortic valve into the Reservoir during the test of our bench. The second flow sensor is integrated close to the Reservoir chamber to better control the flow pattern and to make sure there is no leakage along the hydraulic circuit.

Pressure Measurement

24PCAFA6G differential pressure sensor is used to measure the change of pressure during contraction and relaxation of the left ventricle in the ventricular chamber in response to stimulated movement of the pump's membrane by the actuation system. Change in the pressure in the ventricle is marked with systolic and diastolic phases.

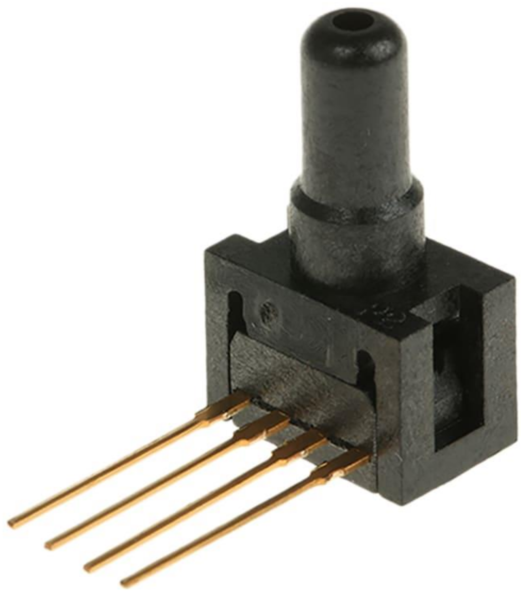

Figure 14 24PCAFA6G differential pressure sensor

This pressure sensor is a proven sensing technology that uses a specialized piezo-resistive micro-machined sensing element to offer high performance, reliability, and

accuracy.
It consists of 4 pins: power voltage, power ground, and the other 2 wires representing the output signal of the sensor. These 2 output signals figure the output difference of pressure inside the ventricular chamber.
To be able to read and visualize information captured by flow and pressure sensors during the testing of designed benches, and to ensure that these measurement readings match the physiological ones, we must connect them to a data acquisition system, which will acquire these measurements and process them using the software.

Monitoring by LabView

LabVIEW is a graphical programming environment in which you drag various building elements around and connect them in a data flow architecture. It's like sketching a block diagram, only you're drawing code. In our situation, we construct a block diagram for each sensor (pressure and flow sensors) and establish the measurement range for each sensor as well as to capture their data; we also drag another block diagram to display the fluctuations in each sensor value as a graph on the screen.
The NI myDAQ is a portable data acquisition (DAQ) device that includes NI LabVIEW-based software instruments for measuring and analyzing collected signals at any time. It is appropriate for sensor measurements. NI myDAQ includes analog input (AI), analog output (AO), digital input and output (DIO), audio, power supply, and digital multi-meter (DMM) features.
Output of flow sensors and pressure are connected to the Ni myDAQ acquisition system to acquire information provided about flow and pressure respectively. The data recorded by the data acquisition system is then transferred to the computer for user interface using the LabVIEW program.

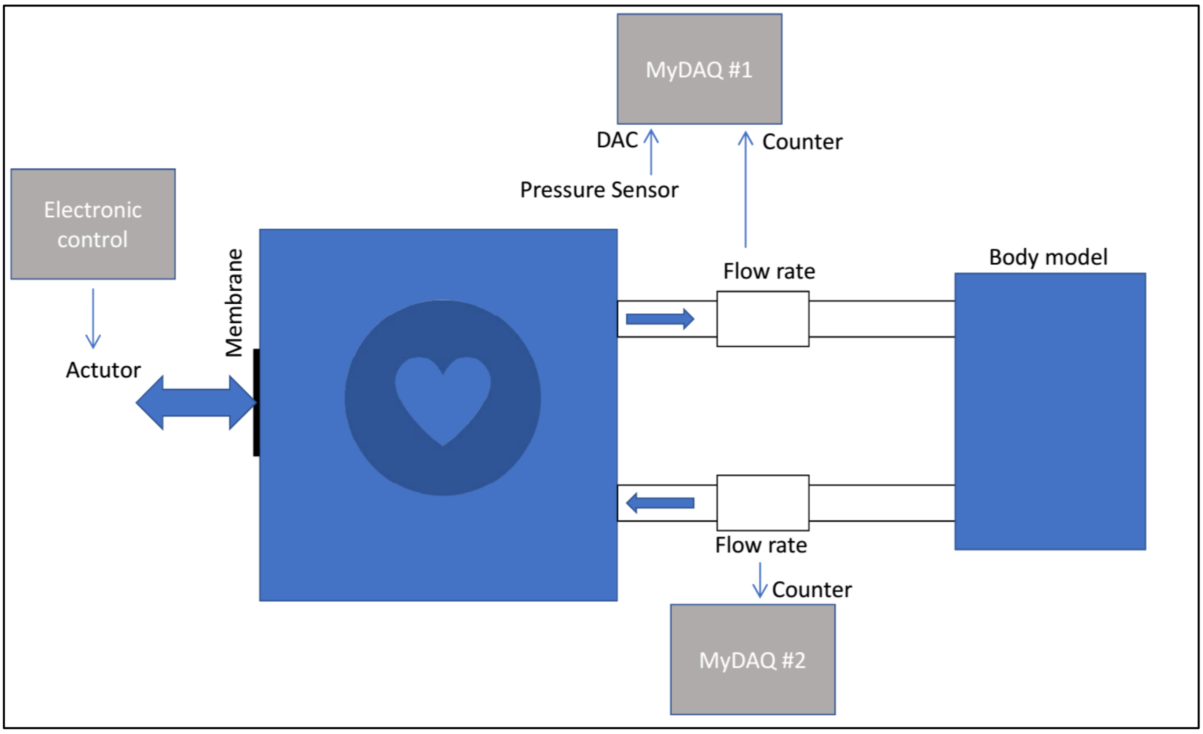

Figure 15 Monitoring of Flow and pressure sensor by MyDAQ instrument

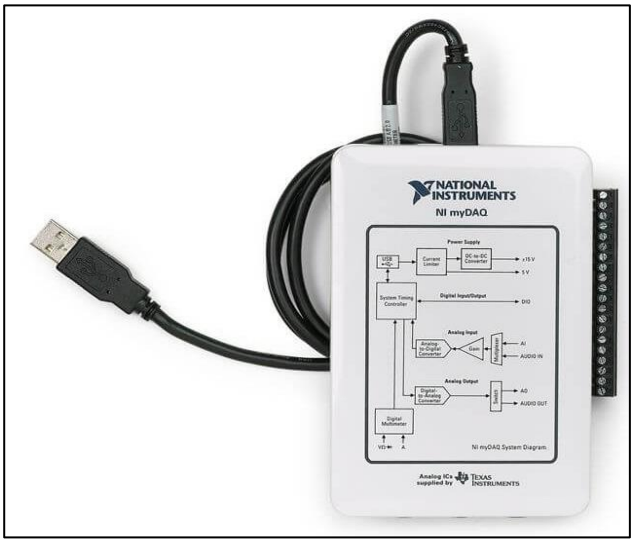

Figure 16 MyDAQ NI device

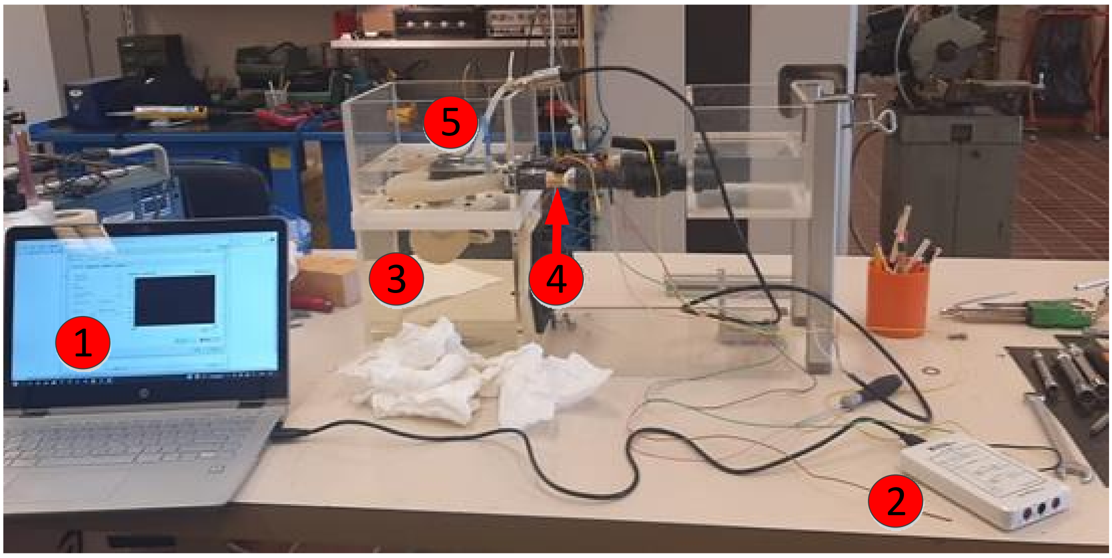

Figure 17 Measurement system

1) Laptop: Labview
2) NI myDAQ
3) The body
4) Hall effect sensor

3. Results

A closed-loop control system able to simulate systemic circulation has been realized starting with the procedures found in the literature to obtain results as close as possible to the physiological ones. The designed bench is composed of all the necessary chambers and components which help to regulate the changes of pressure in the left ventricle and the aortic arc to perform studies on the aortic valve, such as hydraulic resistances and compliance to reproduce the cardiac cycle.

The considered cardiac cycle duration has been 0.8 seconds, corresponding to 75 bpm, as a real heart rate. The compliance chamber which is superior to the ventricular one aims to control the after-load values based on peripheral vascular resistance. During a cardiac cycle, the circuit starts with an actuator able to push and pull the membrane of a pump to move a certain amount of fluid. Furthermore, the left ventricle can be compressed and relaxed due to the change of pressure into the ventricular chamber, controlled by the linear actuator attached to the pump where it is membrane follows inward and outward movement depending on the state of the linear actuator.

It can also compress to mimic the ejection of the left ventricle, pushing 70 ml of fluid thanks to its elastic characteristic. When the pulsatile flow of the fluid arrives through the aortic valve, it allows the aorta to expand and recoil, due to its elastic characteristics. A small tube permits the flow to exit from the compliance chamber and go to the reservoir. The reservoir chamber collects and stores water coming from the aorta during the operation of the aortic test bench.

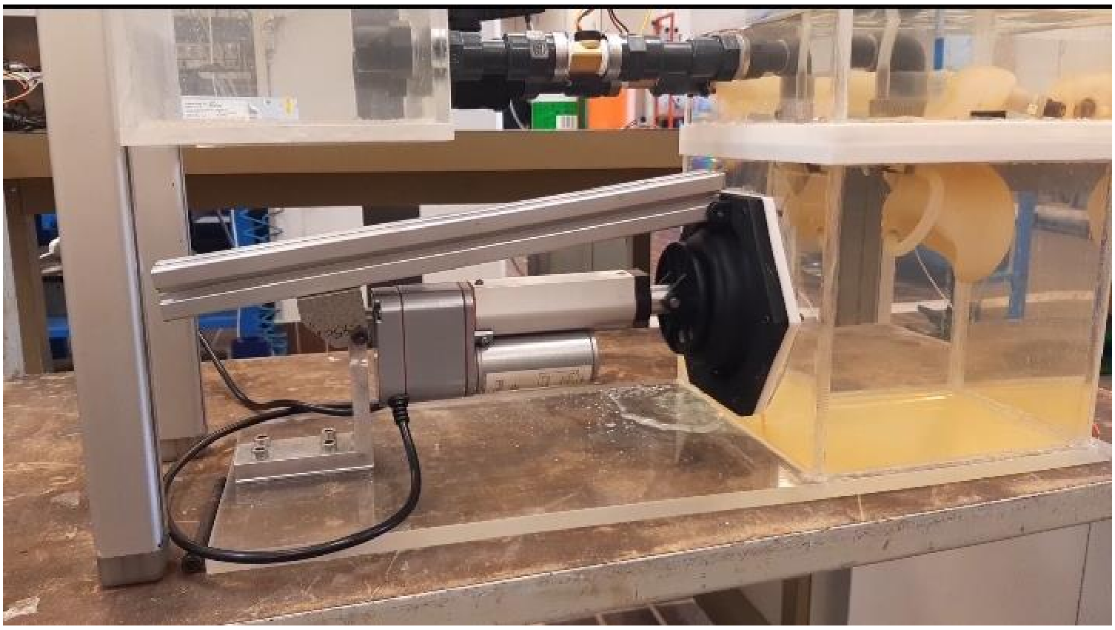

Figure 18 Realization of aortic test bench

The stroke of the pump is 20 mm, and the diameter is 70 mm. From these parameters, the area and the force have been calculated to find the right actuator.
After finding the instantaneous force and the velocity required for the reproduction of the cardiac cycle, the Linear Actuator from the Linak industry has been selected. In particular, the LA12-IC model has been chosen due to its maximum force of 750 N, to its stroke volume which goes from 19 to 130 mm, and to the position feedback (figure 18). This actuator has an integrated controller and a mechanic potentiometer included. So it respects the required characteristic for which, attached to a pump, will be able to move the necessary quantity of fluid.

Approach validation

After choosing the appropriate components and materials to build our bench and choosing the right control system which is used to run our model to initiate the pulsatile flow, we test our bench now and put it into action.

- The first step is to add water to the ventricular chamber, we used water instead of blood due to the difficulties in bringing and storing it.
- Linear actuator motion is initiated by debugging the code written on Arduino

software to control the stroke of the actuator by the feedback potentiometer value which is converted into displacement within the code.

- An external potentiometer is used as an input signal, when we turn it clockwise, the linear actuator will follow this signal voltage so, as we turn the potentiometer to maximum, the linear actuator will move to the maximum position.

In our work, the suitable stroke value for a linear actuator is 20 mm to trigger pump movement without deforming it and within the defined speed of actuator movement. So, within the extension and retraction movement of the linear actuator, the pump will move inward and outward of the ventricular chamber.

During the extended movement of the linear actuator, the pump moves inward which will stimulate flow in the ventricular chamber and due to the change of pressure, the left ventricle will contract and start the systolic phase, and the fluid will pass through the aortic valve into aorta that is located between ventricular and compliance chambers. Finally, the flow will be directed into the reservoir chamber and the process continues until turning the button of the potentiometer back into the initial position (0V). During the retraction of the linear actuator, the pump will go back into its resting state and the left ventricle practices a relaxation phase where fluid will not cause pressure on it or change it, volume and diastolic phase marked this period so, the aortic valve is closed.

The pressure sensor which is inserted just next to the ventricle measures the change in the pressure of the left ventricle during the operation and the testing process of the test bench and these measurements are calibrated by LabView software to present it as shown signal in Figure 19. The graph represents the variation of the ventricular pressure during the systolic and diastolic phases because of contraction and relaxation of the ventricle triggered by pump movement stimulated by linear actuator stroke.

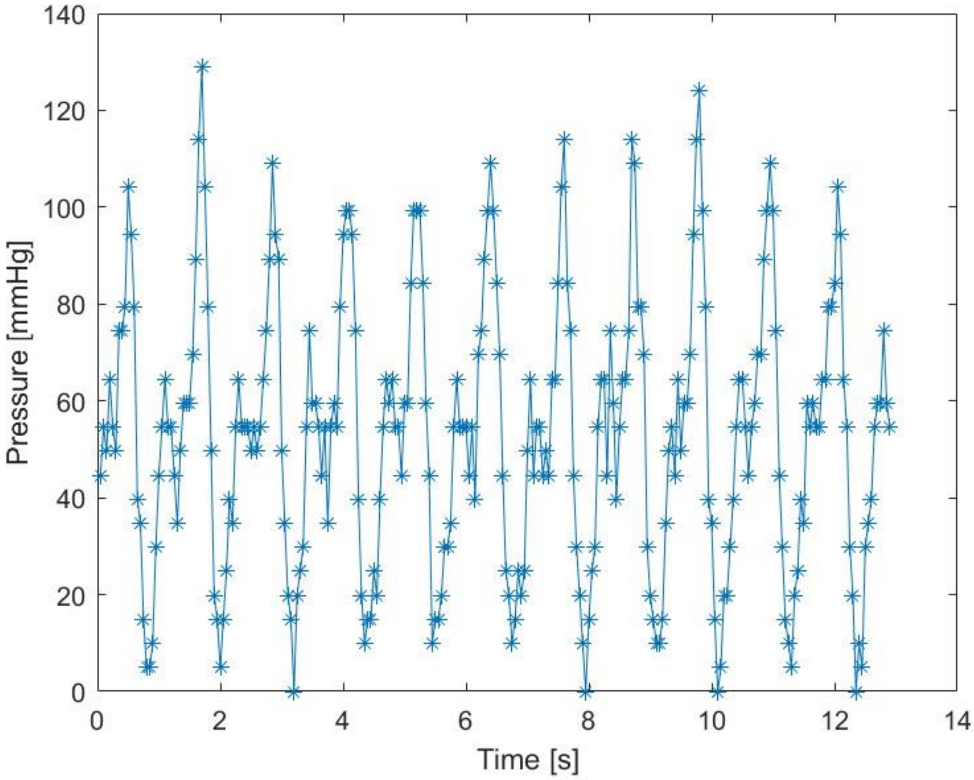

Figure 19. Readings of the pressure sensor within bench testing

It is observed that the shape of the graph follows the trend of the normal physiological pattern of the ventricular chamber, but we used water as fluid to run our test bench, we can't get the ideal shape because the characteristic of the pump is different from the heart and the water characteristics are different from blood.

During the running behavior of the bench, the flow sensor which is mounted at the end of the aorta measures the flow passing through it, and these readings are acquired by a data acquisition system and then calibrated by LabView, which processes this information through a built-in mechanism to visualize the variation of flow measurements as a curve. Another flow sensor is utilized to ensure that there is no leak at the reservoir level and that our system is operating properly.

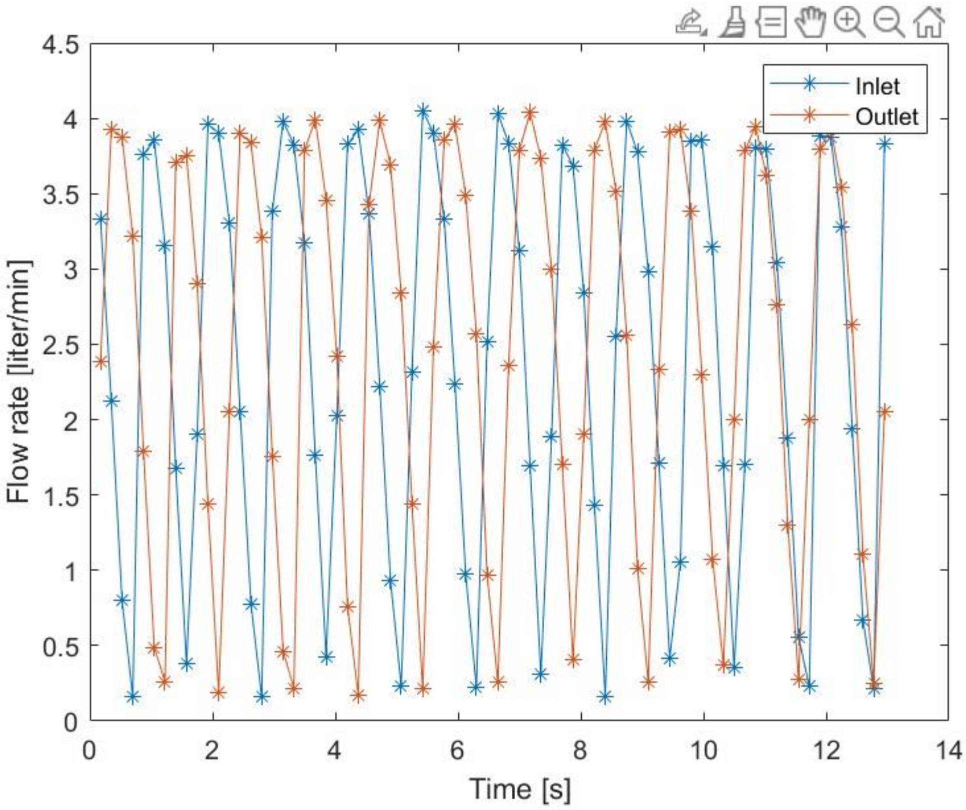

Figure 20. Flow rate variation during bench testing

Figure 20 illustrates the aortic flow curve after crossing into the aorta from the ventricle during the systolic phase and we can consider the aortic flow as a cardiac output since all flow passes through the aorta. From the results shown for the aortic flow values and graph, we can consider that it's close to normal reading values during normal cardiac output with the same curve after systolic and diastolic periods. As we stated before, the water and the pump limit us from reaching the ideal results of the flow due to the previously mentioned reasons in pressure measurements.

4. Discussion and Conclusion

The current test bench shows that the variation of pressure in Figure 21 is in the normal range. According to [3], the pressure changes with respect to time are shown in Figure 1 19

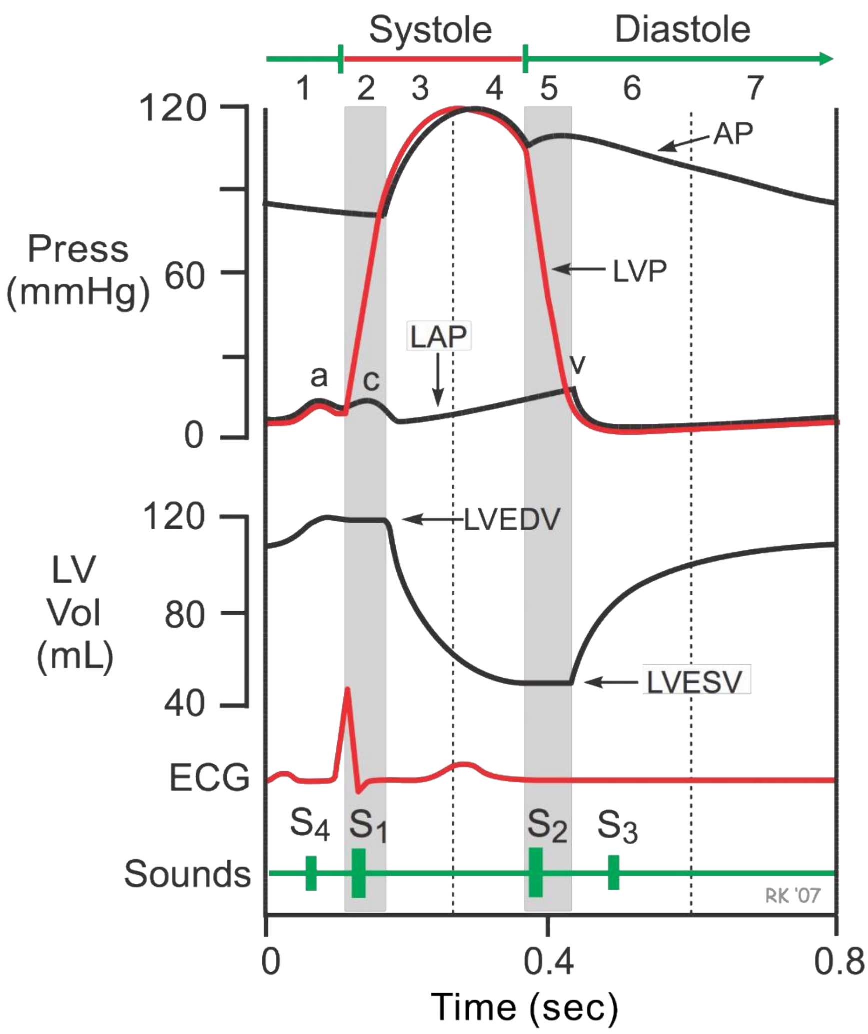

Figure 21 Pressure changes of normal person

So the pressure will varies in range [0-120] mmHg as normal case, in the same second the pressure will increase in systole phase to reach about 120 mmHg then decreases to 0 mmHg in diastole phase. Figure 21 illustrates this discussion, thus the test bench at first run acted as normal cases. compressibility of the air to absorb the pressure wave during the ejection phase and damping the pressure peak and returning the accumulation flow, during the diastole phase, with a more regular trend. The tightness of the plexiglass walls is guaranteed by gluing with a specific bi- component glue Flow rates between end-systolic volume and end-diastolic volume which in turn represent the stroke volume are a little bit inferior to those of physiological parameters but still evident. To ensure our results, let us compare with these results.

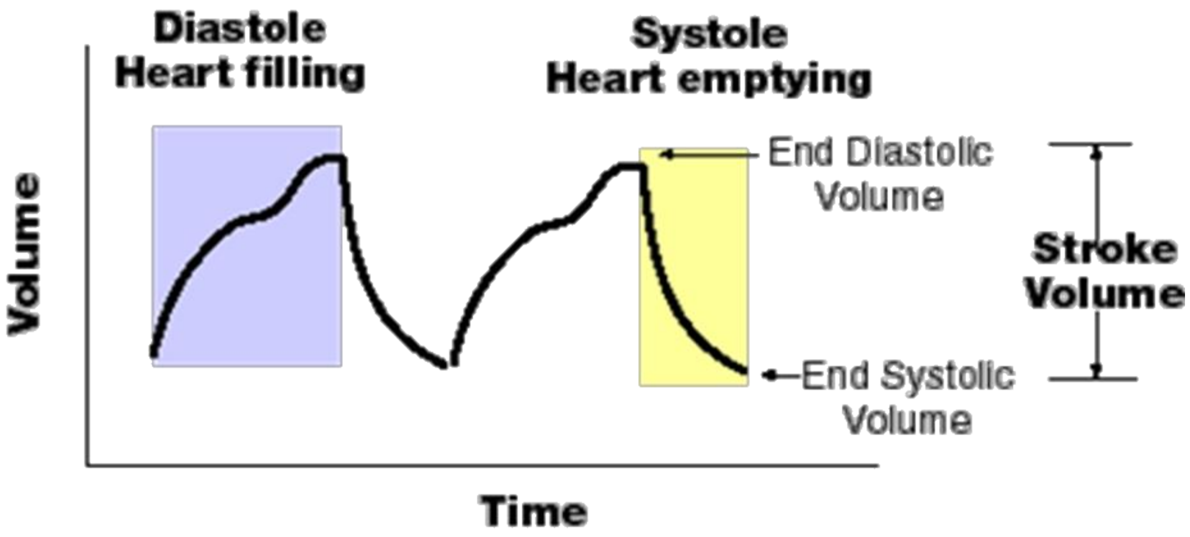

Figure 22. stroke volume during systolic and diastolic phases

.Figure 22 shows the during the diastole phase, there is a filling to reach maximum amount of blood (this divers from one person to another according to age, sex and other conditions, and there is an emptying process during the systole phase.
According to [3] and as it shown in Figure 23 the needed time to fill and empty about 1 second (0.5 for each phase, see Figure 21).
If we compare this with our results in Figure 1 18, we can see the times are correct for normal case, but the in and out flow rate won't be matched till after a period of time then they will be matched or tracked by each other.
The volume of blood differs according to age and sex. We compare only the behavior. Since in our case there is no blood, only air liquid to the initial

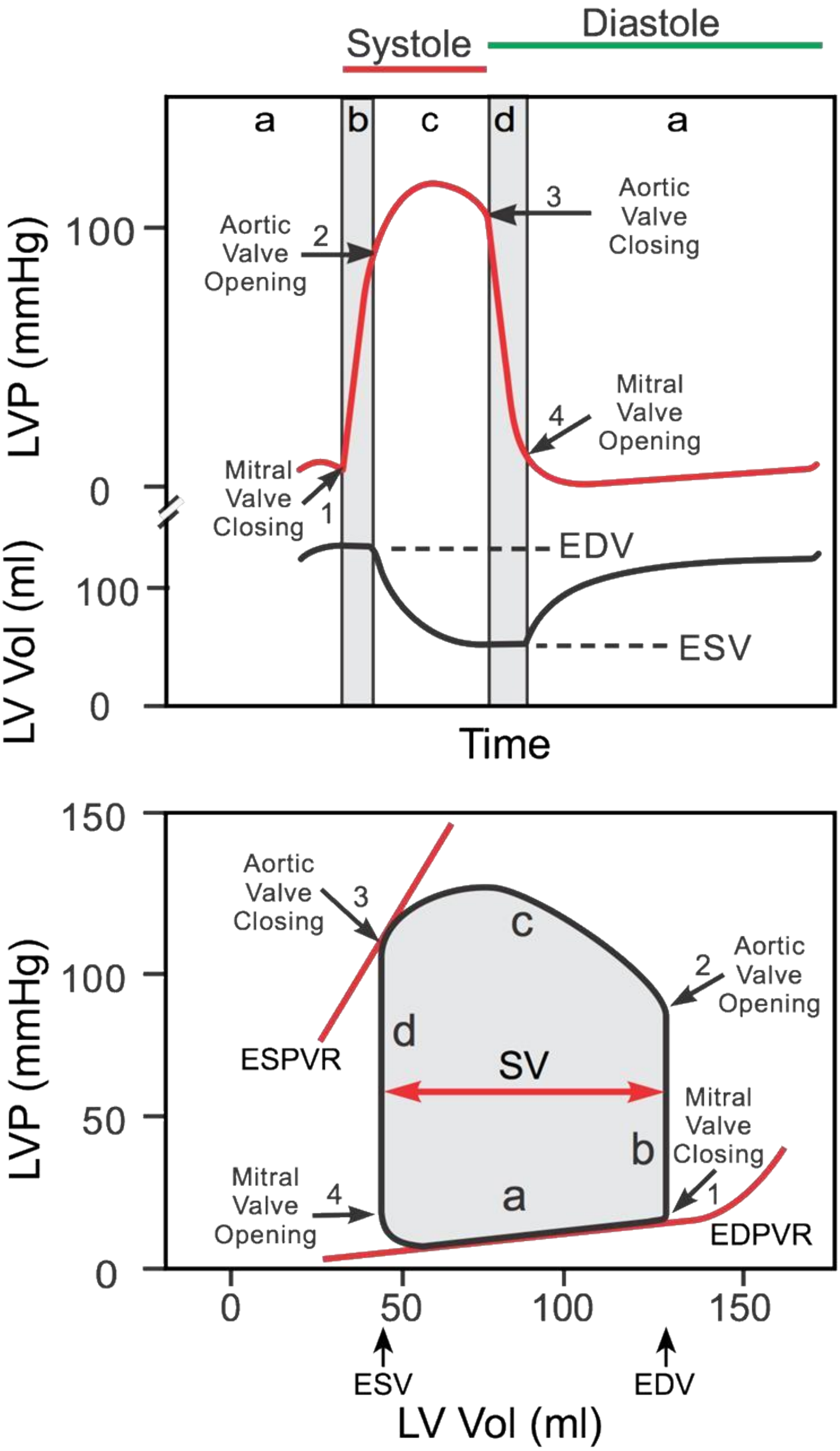

Figure 23. Flow rates of normal person

The left ventricle is one of the most important components of the test bench. It is It worthy mentions that the correct choice of materials and control of a linear actuator to move the pumps plays an important role to match the readings in normal physiological conditions for ventricular pressure and aortic flow during the cardiac cycle. Also, a data acquisition system can be used to capture the information of sensor reading, and with correct calibration by software, we can be able to visualize the variation in values of the sensors in response to the operating test bench.

The current test bench made it possible for cardiologists to test real mechanical prosthetic aortic valve to observe how the hemodynamic changes at the level of pressure and volume response followed it is utilizing and then to perform appropriate aortic valve replacement in case of less invasive procedures like TAVI method or totally surgical replacement method of the valve. The same thing can be done as well in case of using a bioprosthetic valve which is based on biological tissues taken from animals

In this test bench, other types of sensors, such as pressure or temperature sensors, may be used to provide input signals to the control system, which would then map out the stroke of the length of the feedback linear actuator. This test bench might add the left atrium chamber to increase the performance of the existing developed model by measuring atrial pressure as well as the coronary arteries.

Other measurement techniques can be useful to evaluate physiological parameters related to aortic valve implantation like MRI, which is based on using a strong magnetic field and radio waves by using the strong tissue contrast between myocardial and blood to assess the anatomy and function of the heart.

Glycerin in saline solution can be used as the blood analog fluid to test the bench, which has the approximate viscosity of whole blood ~3.5 cP. Surgeons can use blood as well in simulating the test for artificial aortic valve.

Conclusion

The developed aortic bench in this project can be used by surgeons to do aortic valve replacement depending on the results obtained during the testing stage. Moreover, some ideas of additions or replacement (Searching for more accurate tools) some sensors might enhance the performance and the accuracy of the test bench.. The thickness used for the reconstruction of the 3D solid is 3 mm, lower than the real thickness of the ventricle. This choice has been done considering the elastic properties of the material, which changes based on the thickness. A compromise between the physiological parameters and the elasticity of the material has been done to allow the ventricle and the aorta to contract and relax through the change of pressure.

ANNEX A

Arduino code

```
This code is written in order to initiate and control stroke of
linear actuator

#include <Arduino.h>

const int RENA_PIN = 3;  // the Arduino pin connected to the
EN1 pin L298N
const int LENA_PIN = 5;  // the Arduino pin connected to the
EN1 pin L298N
const int RIN1_PIN = 10; // the Arduino pin connected to the
IN1 pin L298N
const int LIN2_PIN = 11; // the Arduino pin connected to the
IN2 pin L298N
#define SET_POINT_POTENTIOMETER A4
// float pos;
#define STROKE_LENGTH 50.8     // PLEASE UPDATE THIS VALUE (in
millimeter)
#define POTENTIOMETER_MAX 1023 // PLEASE UPDATE THIS VALUE
#define POTENTIOMETER_MIN 120  // PLEASE UPDATE THIS VALUE

// the setup function runs once when you press reset or power
the board
void actuatorOff()
{
  digitalWrite(RENA_PIN, LOW);
  digitalWrite(LENA_PIN, LOW);
  digitalWrite(RIN1_PIN, LOW);
  digitalWrite(LIN2_PIN, LOW);
}

// int getActuatorPosition(){
//   int actuatorPos = analogRead(POTENTIOMETER_PIN);
//   actuatorPos = map(actuatorPos, POTENTIOMETER_MIN,
POTENTIOMETER_MAX, 0, STROKE_LENGTH);
//   return actuatorPos;
// }

unsigned long getRunningTime(int distance)
```

```
{
  return (distance / STROKE_LENGTH) * 1000;
}

void moveTheActuator(unsigned long runningTime ,  bool
isExtending = false)
{
  unsigned long timeSnapShot = millis();
  while (millis() - timeSnapShot < runningTime)
  {
      if (isExtending)
  {
    digitalWrite(RENA_PIN, HIGH);
    digitalWrite(LENA_PIN, HIGH);
    digitalWrite(RIN1_PIN, HIGH);
    digitalWrite(LIN2_PIN, LOW);
  }
  else
  {
    digitalWrite(RENA_PIN, HIGH);
    digitalWrite(LENA_PIN, HIGH);
    digitalWrite(RIN1_PIN, LOW);
    digitalWrite(LIN2_PIN, HIGH);
  }
  }
}
int getSetPoint()
{
  int value = analogRead(SET_POINT_POTENTIOMETER);
  return map(value, 0, 1023, 0, 20);
}
void setup()
{
  // initialize digital pins as outputs.

  pinMode(RENA_PIN, OUTPUT);
  pinMode(LENA_PIN, OUTPUT);
  pinMode(RIN1_PIN, OUTPUT);
  pinMode(LIN2_PIN, OUTPUT);
  actuatorOff();
  Serial.begin(9600);
}
// the loop function runs over and over again forever
void loop()
{
  // this is a test code , it will extend the actuator for
about 20mm , hold for 2 seconds
  // then retract and then repeat the process
```

```
  // get Set Point
  int setPoint = getSetPoint();
  unsigned long runningTime = getRunningTime(setPoint) ;
  moveTheActuator(runningTime ,  true );
  actuatorOff();
  delay(250)
;
```

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