



Master Dissertation:

Mechanical characterization of a silicone aortic phantom through inflation

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June 2023

Porto

*"Whether you win or lose... you can always come out ahead by learning from the
experience."*

My Hero Academia

Abstract

Ascending aorta aneurysms (AsAA) present a significant health risk, potentially leading to life-threatening complications. These conditions can greatly impact the mechanical properties of the aorta. Bulge inflation and inflation-extension tests are commonly employed to assess these properties. However, existing tests use liquid fluids to simulate blood flow, which hinders implementation in a hospital environment.

This dissertation aims to explore an unconventional approach, performing inflation-extension tests by using air through the usage of intra-aortic balloons. The objective is to validate this experimental procedure and quantify displacements exhibited by a silicone phantom, assessing its similarity to the behavior of the natural aorta.

Adjustments were made to ensure proper operation of the testing machine, including iterative development of the balloons and evaluation of materials for phantom manufacturing. Displacement measurements were obtained using Digital Image Correlation (DIC). Additionally, computational simulations were conducted using properties representative of aortic tissue. Furthermore, a bulge inflation test was performed using the experimental setup with minimal modifications.

Analysis of the inflation-extension tests revealed challenges in comparing results due to difficulty in replicating the experimental conditions in the computational simulation. However, the results of the bulge inflation test demonstrated a strong similarity between the displacements of the silicone phantom and the computational simulation of aortic tissue.

The experimental setup was successfully implemented, yielding measurable displacements in the silicone phantom, thereby confirming the feasibility of using air as the inflating medium. Additionally, the bulge inflation tests showed that the computational simulation of aortic tissue closely replicated the results obtained from the experimental part using the silicone phantom.

These findings contribute to the development of a reliable testing setup for assessing the mechanical behavior of the aorta. The use of air-based inflation-extension tests and DIC analysis enables valuable insights into the behavior of the silicone phantom, potentially paving the way for future clinical applications and further investigations.

Keywords: Aorta, AsAA, Digital Image Correlation, Inflation, Computational simulation, Phantom.

Resumo

Os aneurismas da aorta ascendente (AsAA) representam um risco significativo para a saúde, podendo levar a complicações potencialmente fatais. Estas condições podem ter um grande impacto nas propriedades mecânicas da aorta. Os testes de insuflação *bulge* e de insuflação-extensão são normalmente utilizados para avaliar estas propriedades. No entanto, os testes existentes usam fluidos líquidos para simular o fluxo sanguíneo, o que dificulta a implementação num ambiente hospitalar.

Esta dissertação tem como objetivo explorar uma abordagem não convencional, realizando testes de insuflação-extensão com ar através da utilização de balões intra-aórticos. O objetivo é validar este método experimental e quantificar os deslocamentos exibidos por um *phantom* de silicone, avaliando a sua semelhança com o comportamento da aorta.

Foram efetuados ajustes para garantir o bom funcionamento da máquina de ensaios, incluindo o desenvolvimento iterativo dos balões e a avaliação dos materiais para o fabrico do *phantom*. As medições de deslocamento foram obtidas usando Correlação Digital de Imagem (DIC). Para além disso, foram efectuadas simulações computacionais utilizando propriedades representativas de tecido aórtico. Adicionalmente, foi efectuado um teste de insuflação *bulge* utilizando a configuração experimental com modificações mínimas.

A análise dos testes de insuflação-extensão revelou desafios na comparação dos resultados devido à dificuldade de replicar as condições experimentais na simulação computacional. No entanto, os resultados do teste de insuflação *bulge* demonstraram uma forte semelhança entre os deslocamentos do *phantom* de silicone e a simulação computacional do tecido aórtico.

A configuração experimental foi implementada com sucesso, produzindo deslocamentos mensuráveis no *phantom* de silicone, confirmando assim a viabilidade da utilização de ar como meio de insuflação. Além disso, os testes de insuflação *bulge* mostraram que a simulação computacional, utilizando propriedades de tecido aórtico, replicou aproximadamente os resultados obtidos na parte experimental utilizando o *phantom* de silicone.

Estes resultados contribuem para o desenvolvimento de uma configuração de teste fidedigna para avaliar o comportamento mecânico da aorta. A utilização de testes de insuflação-extensão utilizando ar e a análise de DIC permitem obter informações importantes sobre o comportamento do *phantom* de silicone, abrindo potencialmente o caminho para futuras aplicações clínicas e investigações adicionais.

Palavras-Chave: Aorta, AsAA, Correlação Digital de Imagem, Insuflação, Simulação Computacional, *Phantom*.

Acknowledgements

Throughout the last few months, this dissertation demanded a great deal of effort and resilience for its completion. However, this work counted on the collaboration of many people who were able to guide and support me unconditionally and, for all of this, I give a sincere thank you.

First of all, a special thanks to my advisers at FEUP, Pedro Sousa and Paulo Tavares, as well as INEGI, Tiago Domingues and Hugo Mesquita, for having assisted me in the realization of this dissertation. For all the time made available for clarifying doubts and suggestions.

To the UMAI team, a special thanks for helping me in different steps along the way, for the availability shown in helping and guiding me.

To all my friends who started the academic life with me, the BOIS, for all the moments and memories that I will carry for life with you, for all the support and mocking given throughout these 5 years that made me feel that all the effort and dedication were worthwhile.

To my parents and brother, who have always been a fundamental presence in my life and for the support they give me in everything I do.

Finally, my boyfriend, Francisco Lázaro, for all his unconditional support throughout this process and for creating a code to send me emails with empowering quotes to make it through the day.

This dissertation was developed in the scope of the project AneurysmTool - Ref. PTDC/EMD-EMD/1230/2021, funded by *FCT Projetos I&D* of the *Fundação para a Ciência e a Tecnologia*.

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Abbreviations

2D - Bidimensional
3D - Tridimensional
AAA - Abdominal Aortic Aneurysms
ABS - Acrylonitrile Butadiene Styrene
AsAA - Ascending aorta aneurysms
AOI - Area Of Interest
AM - Addictive Manufacturing
ATAA - Ascending Thoracic Aortic Aneurysm
BVOH - Butenediol Vinyl Alcohol co-polymer
CAD - Computer-Aided Design
CT - Computed tomography
DIC - Digital Image Correlation
FDM - Fused Deposition Modeling
FEM - Finite Element Method
FOV - Field of View
FSI - Fluid-Structure Interaction
MRI - Magnetic Resonance Imaging
PDMS - Polydimethylsiloxane
PETG - Polyethylene Terephthalate Glycol-modified
PLA - Polylactic Acid
PMMA - Polymethyl Methacrylate
PVA - Polyvinyl Alcohol
PVAc - Polyvinyl Alcohol Cryogel
SLA - Stereolithography Apparatus
TAA - Thoracic Aortic Aneurysm
TTE - Transthoracic Echocardiography
US - Ultrasound

Introduction

1.1 Background and motivation

The cardiovascular system entails an intricate network of organs, vessels, and cells that cooperate to deliver oxygen and nutrients across all parts of the body, preserving, this way, its homeostasis, which is a state of balance among all the body systems required for survival and proper function [1]. At the center of this complex system resides the heart, a muscular organ responsible for pumping blood throughout the body. This organ contracts cyclically, producing pressure that propels blood through a network of vessels comprising arteries, veins, and capillaries, forming a closed circuit that delivers blood from the heart to the tissues of the body and back [1, 2]

The circulatory system can be separated into two main branches: the systemic circulation and the pulmonary circulation. The systemic circulation delivers oxygenated blood, originated from the left ventricle of the heart, to the tissues of the body through the aorta, whereas the pulmonary circulation transports deoxygenated blood, from the right ventricle of the heart, to the lungs, where it is oxygenated before returning to the heart [1].

The aorta is the largest artery on the human body and it can be mainly divided in two sections: thoracic and abdominal. As depicted in Figure 1.1, the ascending aorta is located between the aortic valve and the aortic arch and it is the first blood vessel of the systemic circulation which receives the oxygenated blood from the left ventricle and directs it to the remaining blood vessels [2, 3, 4].

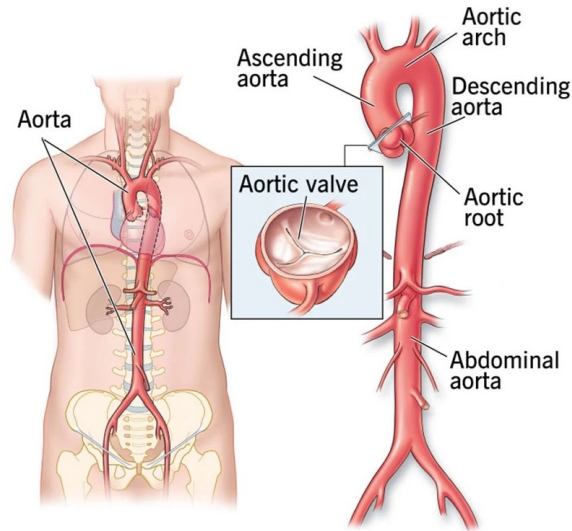


Figure 1.1: Location and constitution of the aorta. Adapted from [5].

The aorta is considered a muscular and elastic artery due to being primarily composed of vascular smooth muscle and elastin [4]. The arterial wall, which is responsible for the vessel's integrity and function, is comprised of three distinct layers: intima, media and adventitia [3, 4, 6].

During systole, the arterial walls expand and blood is stored within the elastic walls, leading to a temporary increase in blood pressure. The arterial walls recoil as the heart relaxes during diastole, and the accumulated blood is propelled forward, preserving blood flow through the arteries and preventing a sharp drop in blood pressure. This procedure is called Windkessel effect and it is derived from the elastic properties of the arterial wall of the thoracic aorta, which have up to 40% elastic fibers in its constitution. These elastic fibers play a crucial role in maintaining the steady and continuous flow of oxygen and nutrients to the body's tissues, thereby safeguarding the integrity and vitality of the cardiovascular system [2, 7, 8].

Aortic aneurysms, characterized by a pathological swelling of the aorta, are a common type of aortic disease. While they can occur in any segment of the aorta, abdominal aortic aneurysms are more common than thoracic aortic aneurysms. Ascending aortic aneurysms are particularly common among the latter [2]. The phenomena of ascending aorta aneurysms will receive special emphasis in this discussion.

Ascending aorta aneurysms (AsAA), although frequently asymptomatic, can result in life-threatening complications such as aortic rupture or dissection, ultimately affecting the mechanical properties of the aorta [9, 10]. At present, the prevailing approaches to detect these ailments comprise of continual observation, through ultrasound (US), computed tomographic imaging (CT), and magnetic resonance imaging (MRI), or transthoracic echocardiography (TTE) in some cases [9]. Furthermore, elective surgery is the standard treatment [11].

Current guidelines state that individuals with an enlarged ascending aorta measuring 55 mm or more are advised to undergo replacement surgery [9, 12, 13]. However, using diameter as a predictor of AsAA risk is not a completely secure method, as ruptures have been observed in people with diameters smaller than 55 mm [9, 12, 13, 14].

Given the significance of AsAA pathogenesis, there has been a trend toward using biomechanical analysis to acquire a better understanding of the underlying mechanisms at work [15].

The present dissertation aims to conduct inflation experiments on an existing prototype of a testing machine. The fundamental goal of this research is twofold: first, to perform inflation tests to check that the system is functioning properly, and second, to examine the similarities between the phantom materials manufactured and the results from computational simulation using aortic tissue. By attaining these goals, the study seeks to validate this experimental approach and quantify the displacements exhibited by a silicone phantom using Digital Image Correlation (DIC).

1.2 Objectives

This dissertation was developed in the scope of the project AneurysmTool - Ref. PTDC/EMD-EMD/1230/2021, funded by *FCT Projetos I&D* of the *Fundação para a Ciência e a Tecnologia*, and has as main objective to corroborate the efficacy of the of the inflation testing system, previously developed in the scope of this project by UMAI (*Unidade de Monitorização e Integridade Estrutural*) at INEGI (*Instituto de Ciência e Inovação em Engenharia Mecânica e Engenharia Industrial*), and quantify the displacements exhibited by a silicone phantom using Digital Image Correlation, comparing them to those obtained through computational simulation. In order to attain these overarching objectives, a series of intermediate goals have been meticulously established from the outset:

- Conduct a review of the healthy tissue of the aorta, as well as the mechanical behavior of aneurysms, phantom materials which are able to mimic the mechanical properties of the aorta, inflation trials in the aorta, and, finally, DIC;
- Manufacture of the phantoms and balloons in order to conduct experiments where they will be inflated with air to simulate diastolic and systolic pressure for each trial in the chosen materials;
- Perform inflation tests with phantom and selected balloon material;
- Analysis of the obtained geometry using DIC;
- Conduct a Finite Element Method (FEM) simulation mimicking the performed experimental test;
- Comparison of the experimental results with the results obtained from computer simulations.

1.3 Structure

This dissertation is divided into eight chapters:

- Chapter 1: presents a context for the addressed topic, as well as the proposed aims during the period of the current study;
- Chapter 2: provides a synthesis of the current literature related to the main topics of this dissertation, namely phantoms that replicate the mechanical properties of aortic tissue and inflation, as well as DIC;
- Chapter 3: contains a description and contextualization of the experimental setup and the preparatory steps undertaken to execute the experimental procedure. This chapter provides a comprehensive account of the methodology employed in the study, including the materials used for the balloon and the phantoms, as well as the manufacturing process of these components. This chapter provides a detailed account of each of these steps, leading to the execution of the experiment;
- Chapter 4: contains the description of all the methodology used in the experimental part to achieve the DIC procedure, as well as the experimental results obtained;
- Chapter 5: presents the methodology employed for conducting the computational simulation, alongside an examination of the geometry fitting process, the boundary conditions adopted as well as the post-processing necessary to obtain the intended results and, finally, the results of this analysis;
- Chapter 6: presents another type of test performed by the same experimental setup with minimal modifications, to further have more results to compare the validity of the silicone used. This chapter presents the methodology for the new experimental procedure, as well as the steps to perform the computational simulation, presenting the results from both the methodologies;
- Chapter 7: compares the results obtained and presented in previous chapters, as well as a thorough critical analysis of them. The experimental results from each test are compared with their counterparts from the computational simulation results. Finally, it also highlights the strengths and limitations of the study and discusses their implications;
- Chapter 8: presents relevant conclusions drawn from the study and future work that could continue and enrich the work done during the dissertation. This chapter provides a succinct summary of the main findings of the study and their implications. It also highlights the limitations of the study and the opportunities for future research that could advance the field.

Literature Review

2.1 The aorta

The aorta, being the largest artery in the human body, serves as the main channel through which blood flows from the heart to the rest of the body. It is anatomically divided into several sections, as can be depicted by Figure 2.1 [2, 3, 4]. This artery can be classified based on the anatomical compartment or according to its route and blood flow direction [3]. According to the anatomical compartment, it is categorized as the thoracic aorta from the heart to the diaphragm, and the abdominal aorta from the diaphragm to the bifurcation of the aorta, respectively. According to its route and blood flow direction, the aorta first extends from the heart as the ascending aorta before turning in a hairpin shape in the aortic arch. The aorta then continues as the descending aorta, which is further classified into two sections: the thoracic aorta and the abdominal aorta. Finally, the aorta ends at a point where it divides into two main blood vessels: the common iliac arteries and the median sacral artery, a smaller midline vessel [2, 3, 4].

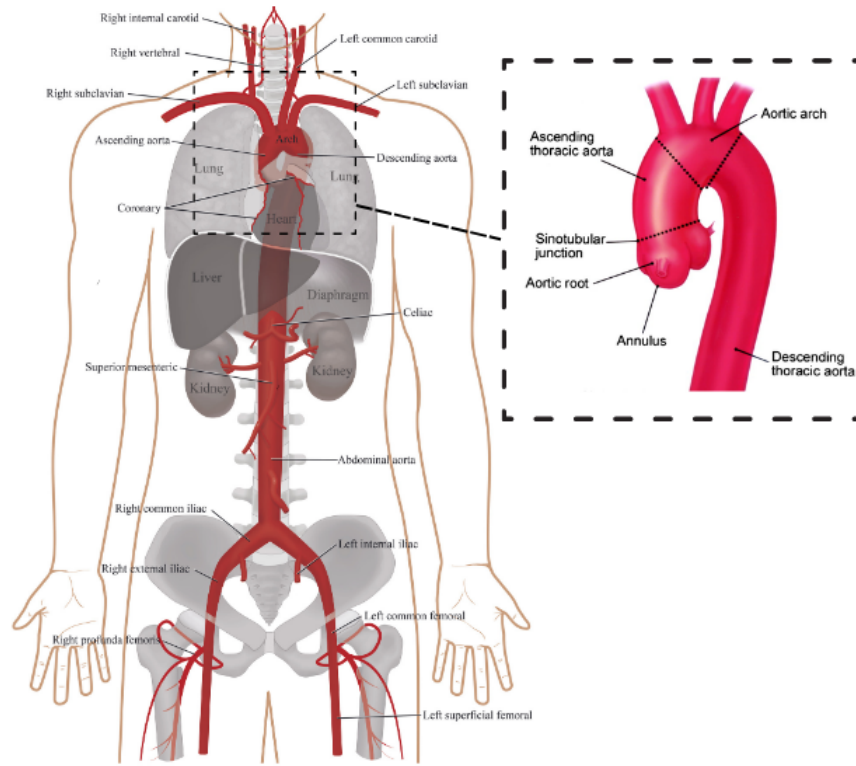


Figure 2.1: Schematic of the aorta and its segments, with focus on the thoracic aorta. Adapted from [16, 17].

2.1.1 Anatomy and Physiology of the Healthy Aorta

The aorta presents the characteristics of a hyperelastic and viscoelastic material [6]. It is considered a muscular and elastic artery due to being constituted of vascular smooth muscle and elastin [4]. Furthermore, its primary passive structure is composed primarily of collagen and elastin fibers [6].

The arterial wall, which is responsible for the vessel's integrity and function, is comprised of three distinct layers: the intima, media and adventitia, as is depicted in Figure 2.2 [3, 4, 6]. The intima, which is the innermost layer, is thin and easily damaged [4, 6, 18]. The media, which is made up of laminated but interweaving sheets of elastic tissue, is responsible for providing strength to the artery [6, 10]. The outermost layer is adventitia, which is mostly made of collagen and helps to keep the artery from over inflating and stretching [6]. Together, these three layers work in tandem to provide the aorta with the necessary physical characteristics to perform its critical function of transporting blood throughout the body [3, 18].

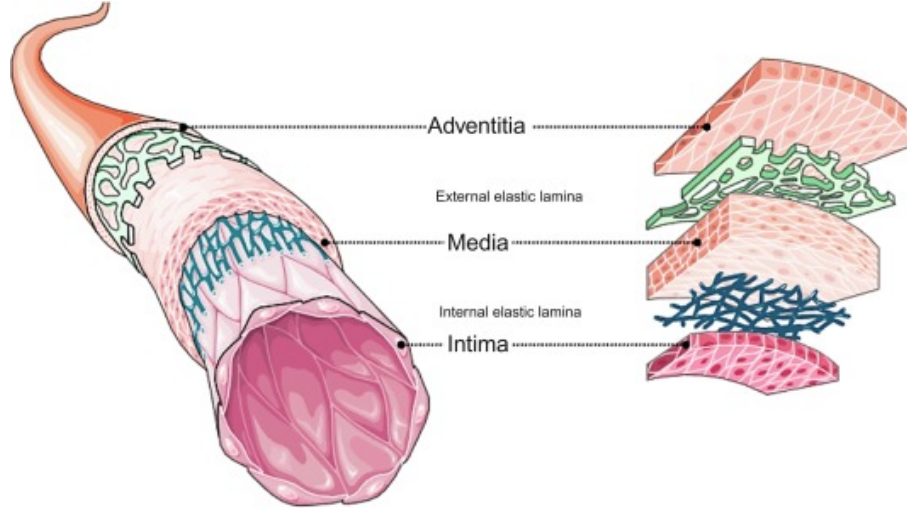


Figure 2.2: Structure of the arterial wall [19].

Eliathamby et al. [20] conducted an extensive investigation of ascending aortic dimensions in both healthy individuals as well as patients with acute aortic dissections and thoracic aortic aneurysms (TAAs). The healthy aorta exhibited a median diameter and length of 33.4 mm and 83.2 mm, respectively, in accordance with the study.

2.1.2 Aortic aneurysms

Aneurysms, by definition, are a persistent and localized expansion of an artery, with a diameter increase of at least 50% above the expected normal diameter of the affected artery. Although all three arterial layers (intima, media, and adventitia) may be visible, the intima and media within big aneurysms may become so thin that they are indistinguishable in certain parts of the arterial wall [9].

Aortic aneurysms are one of the most commonly encountered aortic ailments. In 2010, their global death rate was of 2.49 per 100,000 population was 2.49 [21]. While aneurysms can manifest in any segment of the aorta, the abdominal aorta is significantly more susceptible than the thoracic region [2, 10]. Nonetheless, among TAAs, the ascending aorta is the most frequently affected segment [2]. As a result, this dissertation focuses on geometries of thoracic aortic pathologies, with particular emphasis on the ascending aorta. The focus on this specific location is attributed to the AneurysmTool project's primary goal, which centers on studying aortic diseases in this particular region. Eliathamby et al. [20] showed that AsAA exhibited a median diameter and length of 50.3 mm and 107 mm, respectively, in accordance with the study.

2.1.2.1 Pathophysiology of AsAA

Aneurysmal aortas exhibit structural differences in comparison to their healthy counterparts, which may give rise to life-threatening complications [9, 10, 22]. One of the most critical complications include the degradation of elastin and collagen fibers [10, 22]. Aortic disorders are frequently associated with microstructural alterations in elastin or collagen composition or architecture. Irregularities in the number and/or architecture of these fibers inside the aorta wall has a direct impact on elasticity and strength, resulting in the mechanical and functional abnormalities associated with these diseases [22].

The aorta is normally very elastic and has a microstructure that aids in the delivery of blood from the heart to the systemic circulation [6, 22]. The elastic characteristics and strength of the aorta are imparted by elastin and collagen connective fibers within this microstructure, respectively. However, in aneurysmal aortas, these fibers become damaged, resulting in weakened vessel walls that are more prone to rupture [6, 10, 17, 22].

Additionally, alongside the structural differences between healthy and aneurysmal aortas, biomechanical changes occur. Compliance and stiffness, two critical properties of the aorta, are significantly altered in an aneurysmal aorta [23]. Compliance, which refers to the aorta's ability to respond to changes in pressure by expanding and contracting, is reduced in an aneurysmal aorta [12, 24, 25, 26]. In contrast, stiffness, which refers to the aorta's resistance to these changes in pressure, is increased [25]. These changes in compliance and stiffness may result in altered hemodynamics, leading to an increased risk of rupture [25, 27].

Furthermore, another important biomechanical alteration is increased wall stress and strain. Wall stress is the force exerted on the vessel wall, while strain is the resulting deformation. There is a significant increase in wall stress values as aneurysm size increases, demonstrating a clear link between aneurysm size and elevated wall stress levels. This increased wall stress, caused by the aneurysm's raised blood pressure and changed artery geometry, increases the chance of rupture or dissection [2, 9, 28]. In addition to the changes in this referenced parameters, other biomechanical alterations in aneurysmal aorta have also been identified, such as altered flow patterns and turbulence. These alterations are important because they can impact the hemodynamics of the vessel, leading to further changes in wall stress and strain [12].

2.1.2.2 Diagnosis and Monitoring

The ascending aorta dilatation is a slow and stealthy process that requires years to develop and is initially asymptomatic. Patients with systemic illnesses that cause ascending aortic aneurysms might display symptoms of the primary disease, such as those observed in Marfan syndrome, alerting clinicians to the risk of dilatation. However, in most cases, this illness goes unnoticed until catastrophic complications occur or it is detected incidentally on cardiovascular imaging for other reasons. In order to receive proper treatment and avert life-threatening consequences, aneurysm detection is vital [9, 29]. Currently, methods such

as ultrasound (US), computed tomographic imaging (CT), and magnetic resonance imaging (MRI), as well as transthoracic echocardiography (TTE) in some cases, are the most commonly utilized as tools for the detection of these ailments and to determine the risk for future complications [9].

While non-invasive, ultrasound can be time-consuming and has limitations in visualizing particular regions of the aorta [29]. Both CT and MRI provide accurate images of the aorta and its surrounding tissue, however CT entails risks associated with exposure to radiation and contrast injection, whereas MRI may not be suitable for individuals who are claustrophobic or have metallic implants, adding the fact that it is also an expensive procedure [2, 4, 9, 29]. Meanwhile, TTE is a simple and suggested screening and follow-up technique, but it may not provide adequate imaging of all aortic segments [29]. These diagnostic methods, when used together, form a potent toolkit for the early detection and effective therapy of aortic aneurysms, lowering the risk of complications and improving patient outcomes [9, 29].

Furthermore, while these approaches have been helpful in detecting aortic aneurysms, they also have limitations. Some imaging methods' lack of sensitivity and specificity could culminate in mistaken or false positive diagnoses, while the cost and availability of such procedures may limit their application in particular populations. Furthermore, existing imaging techniques may not provide sufficient information on the underlying mechanisms of aortic aneurysm formation and progression, making a reliable prediction of rupture risk challenging to accomplish [9, 29].

2.1.2.3 Treatment

Even with swift detection along with effective treatment, complications from AsAA can have devastating effects. This emphasizes the necessity of detecting a pathologically dilated ascending aorta in a timely manner and providing enough follow-up to commence medical therapy and, if necessary, prescribe preventative surgical repair [29].

Elective surgery is the primary treatment for ascending aortic aneurysms, which has been shown in multiple researches to drastically lower mortality rates. The choice to conduct corrective surgery is usually based on the size of the aneurysm and the likelihood of complications outweighing the dangers of operation [9, 29].

Current medical standards recommend surgical intervention for asymptomatic persons in particular circumstances. Patients with an ascending aorta diameter of 55 mm or more should be considered for surgical repair [13]. Individuals with Marfan syndrome should undergo elective surgery at smaller diameters (40 to 50 mm, depending on the severity of the problem) to prevent acute dissection or rupture. Those with an aorta that is less than 55 mm in diameter and possesses a growth rate of more than 5 mm per year should also be evaluated for surgery. Furthermore, patients who necessitate aortic valve repair or replacement and have an ascending aorta that exceeds 45 mm should be examined for ascending aortic replacement. These recommendations are the result of significant research and analysis of the risks and

benefits of surgical intervention in each of those settings [9]. Finally, timely detection and surgical intervention are critical for improving outcomes and lowering the risk of complications associated with AsAA [9, 29].

Although using diameter as a predictor of AsAA risk is a common practice, its reliability is not absolute, as several cases of ruptures have been reported in individuals with aortic diameters below 55 mm, indicating that other factors may also influence the risk of rupture [9, 12, 13, 14].

As a result, the AneurysmTool project intends to provide a comprehensive collection of tools to facilitate the biomechanical assessment of each individual case, allowing for the derivation of more objective conclusions about the risk of aneurysm rupture.

2.2 Aortic Phantom materials

Phantoms, also denominated medical models, are vital tools in medical research and device testing, given the difficulty of procuring and working with real human tissue samples. They are physical representations or replicas of biological structures that are used for various purposes in medical and healthcare fields. These models are designed to simulate the characteristics, properties, and behavior of real anatomical structures or physiological systems [12, 15, 30, 31].

Significant progress has been made in the quality of *in vitro* models in the last decade, with the ability to manufacture models and replicas of many artery illnesses. Due to this, they are being increasingly used for surgical planning, verification of diagnostic methods, medical computational models, algorithm verification and validation, and medical devices development [12, 15, 30, 31]. However, although a realistic representation of the geometry can be easily obtained, the modeling of biomechanics and fluid dynamics is still a challenging issue, making urgent the need for realistic *in vitro* models in order to evaluate endovascular procedures [12, 30, 32, 33].

The anatomical and mechanical behavior of the human and diseased aortas requires access to materials that can simulate cardiovascular tissue for clinicians to practice, plan surgical procedures, and verify devices or techniques [12]. Currently, efforts are being undertaken towards creating such materials, known as cardiac phantom materials, which have the potential to accurately simulate the human aortas of specific patients [12, 34]. Developing suitable aortic replicas for *in vitro* studies involves carefully choosing both material and geometry. Matching the grade of geometry details with the mechanical response of the materials is an ongoing research topic [12].

2.2.1 Phantom types

Phantom materials are important tools in medical research and device testing, and the development of aortic phantoms is a crucial aspect of creating realistic *in vitro* models [12]. Different types of medical phantoms are presented in literature, including tissue-mimicking, radiologically relevant, and physiological medical phantoms [31]. The specific type of phantom chosen depends on the imaging modality or device being tested and the intended application [31, 35, 36]. Regarding this three types of medical phantoms [31]:

- Tissue-mimicking medical phantoms: These phantoms are used to evaluate the efficacy of imaging devices and techniques. By mimicking the geometric and material features of biological tissue, these phantoms provide a clinically realistic imaging environment;
- Radiologically relevant medical phantoms: These phantoms are intended to replicate the absorption and scattering properties of human tissue. They are frequently used in radiation therapy studies to assess the dose distribution of different treatment schemes;
- Physiological medical phantoms: These phantoms are designed to mimic the physiological properties of human organs and tissues, such as their mechanical and fluid dynamic properties. They are commonly used in medical device testing to evaluate the performance of devices.

In the context of aortic phantoms, the physiological medical phantoms are the most relevant, as they are designed to mimic the mechanical and fluid dynamic properties of the aorta and its surrounding tissues [31]. Furthermore, with respect to aortic phantoms, these can be classified into two main categories: rigid and compliant [37].

Some research has employed rigid phantoms, which have showed potential in experimental modeling investigations [37, 38, 39, 40, 41]. However, studies that used fluid-structure interaction (FSI) concluded that rigid wall assumptions overestimated the value of wall shear stress when compared to a compliant phantom [35, 37, 41, 42, 43, 44]. Also, it was observed that the pulse waveform differed between stiff and compliant arteries [35, 37]. This is significant proof that employing rigid wall assumptions may not appropriately predict flow phenomena in places critical to the evolution of cardiovascular disease [37, 41]. Thus, mimicking vascular compliance during experimentation is critical for obtaining meaningful hemodynamics, as well as a precise model [35, 37, 45]. Consequently, compliant phantoms have become a crucial tool in medical research and device testing, notably in the creation of innovative cardiovascular technologies [37]. As mentioned in the previous section, compliance refers to the aorta's ability to respond to changes in pressure by expanding and contracting [24, 25]. Given that wall thickness is inversely related to compliance, it is an important consideration when creating effective compliant phantoms. However, establishing uniform wall thickness is difficult, and significant variance might result in unphysiological compliance distribution and unrealistic flow fields, decreasing the clinical value of experimental data [35].

Furthermore, current manufacturing processes are still limited in producing thin-walled compliant phantoms of various shapes [35, 45]. This can limit the efficiency of these phantoms as models for *in vivo* arterial function while also making it difficult to accurately duplicate physiological conditions [35]. Advances in manufacturing processes, on the other hand, provide promise for overcoming these barriers and improving the accuracy and dependability of compliant artery phantoms [35, 37, 36].

2.2.2 Manufacturing Processes

To create aortic phantoms, a variety of manufacturing processes are available, ranging from traditional molding techniques to advanced 3D printing technologies.

One of the most commonly used manufacturing techniques for rigid phantoms is lost-core casting. This process involves using a male mold placed in a casting box. The casting box can be made of various materials, including polymethyl methacrylate (PMMA), glass, and various resins or polymers. Polydimethylsiloxane (PDMS) silicone is then poured into the casting box and forms around the male mold. Once the curing process is complete, the casting box is removed, and the male mold is dissolved or mechanically removed [37].

Sherman et al. [46] utilized lost wax casting in order to produce aortic phantoms. The research team used stereolithography (SLA) to create a mold using a photo-reactive liquid resin and an ultraviolet laser. This mold was then used to make a separate silicone mold, capable of producing multiple casts of the aneurysm vessel, created by injecting low-melting-point wax into the mold and allowing it to cure. The wax mold is then delicately removed, resulting in a patient-specific wax aneurysm vessel. These vessels were then submerged in a PDMS material within a custom-designed water-tight box to obtain the phantom.

Biglino et al. [39] produced a rigid phantom through SLA using transparent resin. Aycock et al. [47] showed that rigid phantoms can be directly printed with an inkjet 3D printer using transparent resin.

Different manufacturing methods have been used for the construction of compliant phantoms. As with rigid phantoms, molding is a common technique used to create compliant phantoms that mimic the compliance of artery walls [35, 37, 48, 45]. It involves using a female mold to create thin-walled silicone models based on a 3D model. The male mold is positioned within the female mold, and after the silicone is poured and hardened, both molds are removed. The process for removing the male mold is similar to rigid phantoms, while the female mold can be unbolted without damage. However, surface smoothness and distortion are important factors that can affect the quality of the phantom. Achieving a suitable surface polish can be challenging due to the thin wall thickness and potential geometric distortion in compliant phantoms [37]. Micco et al. [48] employed fused deposition modeling (FDM) to construct a bespoke mold divided into six separate modules. Through this process, they produced a male counterpart using Acrylonitrile Butadiene Styrene (ABS) material. Subsequently, the ABS male underwent a delicate process of dissolution through acetone after the

solidification of the silicone material. Yazdi et al. [35] presents a new approach for the fabrication of compliant phantoms suitable for optical measurement. They employed 3D printing technology, such as FDM and SLA, to create customized male and female molds. The molds were designed with features like supports and funnels for precise alignment, even silicone injection, and stress reduction. The use of high-resolution 3D scanners and Computer-Aided Design (CAD) software enabled accurate reproduction of anatomical geometries. Surface smoothing techniques, including acetone treatment and sandpaper polishing, were applied to enhance the surface finish. These advancements resulted in more accurate and refined phantoms, facilitating better simulations and experiments in medical and research applications.

Dip coating and dip-spin coating are two further approaches for producing compliant phantoms [37, 36]. Dip coating, also called mold dipping, is repeatedly dipping a rapid prototyped mold in silicone or latex liquid to get a complete coating. However, controlling wall thickness is difficult [35, 37]. Dip-spin coating reduces wall thickness fluctuation by spinning the mold after dipping to obtain a more consistent wall thickness. These methods are relatively expensive since they require a considerable amount of silicone that is discarded 80-90 % of the time [35, 36, 37].

Using 3D printing, more specifically by PolyJet technology, to create the phantom directly has also shown great promise in fabricating realistic phantoms and structures for medical applications, in vitro studies and device testing [31]. Biglino et al. [49] printed a phantom of the arterial geometry of a rubber-like material named TangoPlus using with PolyJetTM technology by Stratasys Ltd., an Addictive Manufacturing (AM) technique that deposits a liquid photopolymer layer by layer through orifice jetting and then solidifies it by UV exposure. Kwon et al. [30] employed PolyJet technology, as well as FDM, presenting an integration of pattern embedding within the 3D-printed structure. By embedding patterns that mimic the orientation of collagen fibers found in real aortic tissue, they aimed to replicate the complex mechanical behavior of the aorta more accurately. To achieve this, they designed and printed a scaffold-like structure, experimenting with different materials that exhibit similar mechanical properties to aortic tissue. The scaffold was specifically designed to incorporate patterned regions that simulated the collagen fiber orientation. The patterns were carefully embedded within the scaffold during the 3D printing process. This involved controlling the deposition of the material in a way that aligned with the desired pattern. The researchers used precise printing parameters and optimized the pattern design to ensure accurate replication of the collagen fiber orientation.

2.2.3 Materials

Numerous studies have been conducted with regard to the development of compliant phantoms, revealing a diverse range of materials that have been extensively tested and evaluated. These studies have contributed to a broad and thorough understanding of the qualities and characteristics of numerous materials, allowing researchers to make informed decisions

depending on their individual needs. However, replicating the mechanical properties of biological tissues remains challenging. Materials having specific mechanical properties such as elasticity, as well as precise control over shape and thickness, are required. Standardized fabrication and characterization methodologies are also required for consistency among research groups [32, 37, 48].

Silicone-based materials are widely utilized in phantom fabrication, with PDMS being the most common choice [35, 37, 45, 46]. PDMS offers optical transparency, low shrinkage rate, ease of fabrication, and desirable elastic properties [37]. Sylgard 184, a commonly used type of PDMS, exhibits good elastic properties but is sensitive to temperature and cross-linking agent variations [35, 37, 46, 45]. Other PDMS products like RTV 615 and RTV 3040 have also been employed. However, working with elastomers like PDMS can present challenges due to their tendency to stick to molds, necessitating the use of release agents [37]. Additionally, Prochima Cristal Rubber is another silicone-based material frequently used in blood vessel prototyping [32, 48]. Silicone-based materials are commonly utilized in the production of commercial phantoms, including those offered by Elastrat. These materials have demonstrated their suitability and effectiveness in creating phantoms for various applications [50].

Apart from silicone-based materials, other elastomers have been explored, albeit less commonly than PDMS. TangoPlus, a rubber-like material, has found applications in 3D printing processes, through PolyJet technology [30, 31, 34, 35, 37, 45, 49]. Polyvinyl alcohol (PVA) has been used to create anisotropic materials, which are desirable for mimicking the non-linear anisotropic mechanical properties of vessels. However, PVA is not transparent, and PVA hydrogels are prone to swelling and degradation over time, limiting their lifespan [37, 51]. A variant of PVA, polyvinyl alcohol cryogel (PVAc), that results from its mixture with cryogel, exhibits remarkable properties, including nonlinear elastic behavior, allowing them to emulate the mechanical response of biological soft tissues, such as arterial tissue, to a certain extent. These cryogels can be easily molded and offer flexibility in phantom fabrication by adjusting specific parameters to tailor their strength [51]. Moreover, successive cure cycles can enhance the strength and elasticity of PVAc cryogels, enabling the creation of intricate phantoms using various molding techniques. However, it is important to note that PVAc, being a water-based gel, exhibits relatively poor mechanical strength and temporal stability. Therefore, careful storage and maintenance practices are necessary to ensure optimal performance over time [52].

Overall, silicone-based materials such as PDMS and Sylgard 184 are the most extensively used materials for aortic phantom fabrication due to their good elastic properties, optical transparency, and ease of fabrication. The choice of material depends on the specific needs of the experiment and the desired mechanical properties of the phantom [37].

2.3 Inflation

Given the unique structure of biological tissues, accurately assessing their mechanical properties is challenging. Aortic tissue properties can vary significantly on an individual and regional basis, and they can change over time in response to changes in the surrounding environment or as a result of pathological conditions. Furthermore, the distribution, arrangement, and proportion of elastin and collagen fibers within the tissue can result in highly nonlinear and anisotropic properties that necessitate complex mathematical models to accurately describe tissue behavior [15].

These mathematical models can be developed and validated with the help of experimental methods such as inflation testing. By varying the inflation pressure or strain rate and measuring the corresponding forces and deformation, it is possible to assess the mechanical response of the tissue and obtain data to calibrate the mathematical models. Nevertheless, careful control of the experimental conditions, such as wall thickness and boundary conditions, is required to ensure that the resulting data accurately reflects the mechanical behavior of the tissue [15].

2.3.1 Inflation testing methods

Understanding the underlying physical properties of the aortic wall is critical for understanding the mechanisms enabling the formation of cardiovascular diseases such as aneurysms. However, modeling complex, anisotropic tubes like the aorta remains a difficult endeavor. A variety of testing methods is used to evaluate the mechanical properties of arteries *in vitro*, including uniaxial and biaxial tensile tests, as well as inflating experiments. These inflation tests are classified as either bulge inflation tests or inflation-extension tests in literature, as depicted in Figure 2.3 [14, 15].

In order to preserve the samples used in the analysis, the project has deliberately excluded the utilization of uniaxial and biaxial tests. Consequently, the current review will primarily concentrate on inflation-extension tests, as they hold central importance in the context of this dissertation.

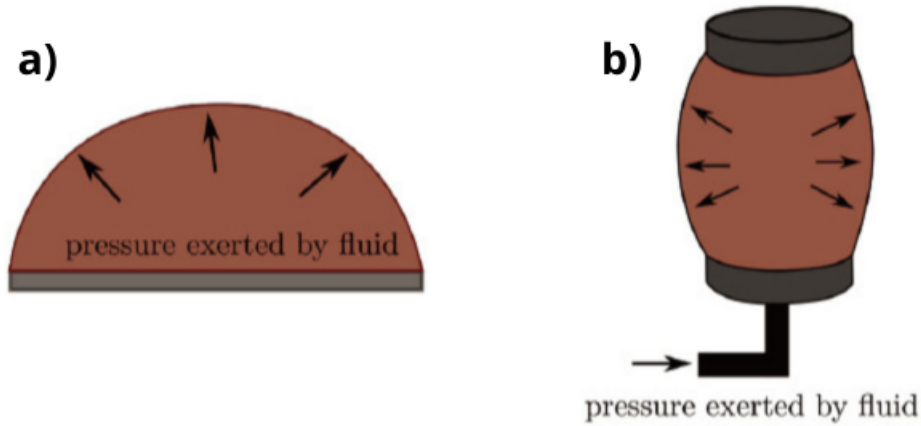


Figure 2.3: Types of inflation tests for arterial mechanics: a) Bulge inflation; b) Inflation-extension testing [14].

The bulge inflation test is a commonly used method for investigating the biaxial mechanical behavior of materials. This procedure, also referred to as the membrane bulging test, involves clamping a sample on an inflation device, forming a hermetically sealed cavity, followed by the controlled release of fluid, typically water. To capture the expansion of the specimen and collect its related strain field, optical tracking methods leveraging sophisticated digital imaging techniques are used [14, 15]. Figure 2.4 visually illustrates the methodology employed to acquire the square-shaped specimen, encapsulating the meticulous approach taken by the researchers. In this particular investigation, the focus of their inquiry rested upon the distinct media and adventitia layers, which were conscientiously separated and subjected to comprehensive analysis and scrutiny [53, 54].

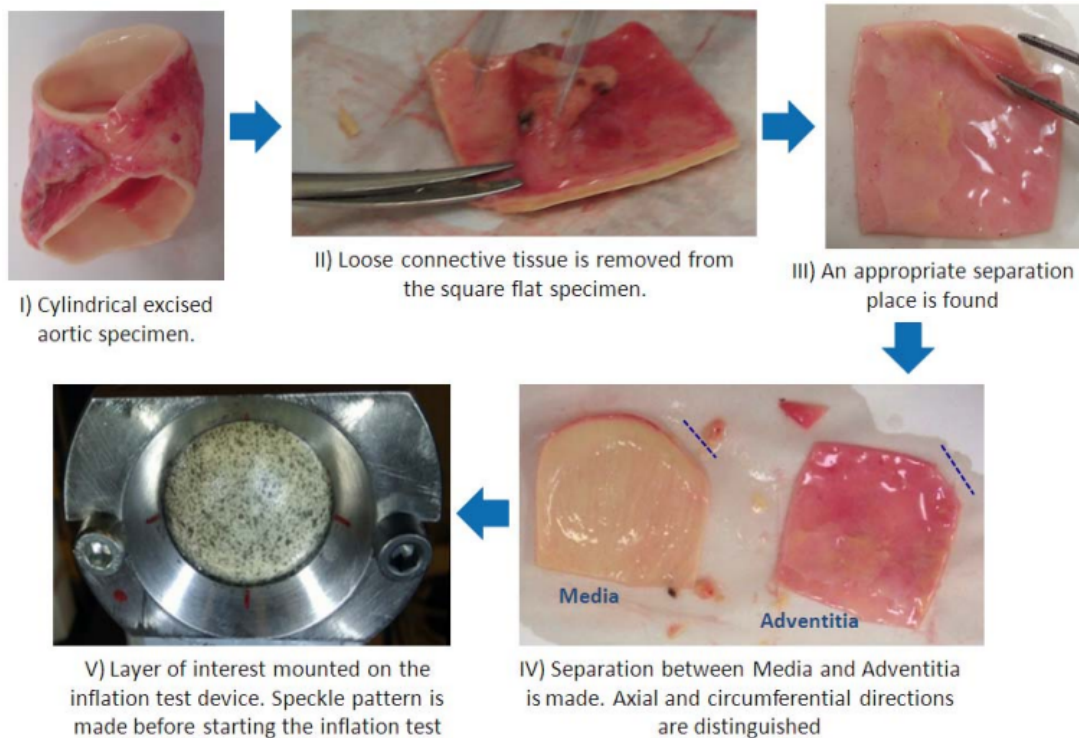


Figure 2.4: Methodology employed to obtain the squared shaped specimen and its utilization in the inflation machine [54].

The bulge inflation test has been used in past studies on healthy human and pig aortas, providing valuable information on their biaxial mechanical behavior [55, 56, 57]. These investigations found that inflation testing is a useful approach for mechanically analyzing vascular tissue. The results showed that the bulge inflation testing approach, in particular, can provide crucial information on the anisotropy and nonlinear mechanical behavior of arterial tissue under physiological loading circumstances. However, these studies have also underscored the importance of refining and standardizing inflation testing techniques to enhance the consistency and reliability of results. While the bulge inflation test provides valuable material information in response to biaxial loading, it falls short of fully replicating proper cylindrical loading conditions. Therefore, further advancements are needed to achieve more precise and consistent inflation testing, ensuring the robustness and accuracy of the obtained data [15, 55, 56, 57].

This technique has also been employed to investigate the mechanical behavior TAA and the underlying mechanisms of rupture [53, 58, 59, 60, 61, 62]. Duprey et al. [58] conducted bulge inflation tests on human ascending thoracic aortic aneurysm (ATAA) samples, using a fluid (water). They found a correlation between patient age and ATAA rupture properties, with younger patients exhibiting greater strength. The study also revealed the importance of collagen and elastin network quality in preventing rupture, with elastin contributing to

extensibility and collagen providing strength. Romo et al. [53] performed bulge inflation tests on human ATAA tissues, using water to impose the pressure, and observed that rupture did not consistently occur at the point of maximum stress. Instead, it happened in weakened zones characterized by strain concentration and localized thinning of the tissue wall. Kim et al. [60] used DIC to analyze the failure of human aortic tissue in a bulge inflation test. They identified localized strain concentrations and tracked the progression of failure, providing insights into microstructural changes and damage mechanisms. The study also investigated the influence of sample orientation and loading conditions on tissue failure behavior. Ohashi et al. [62] investigated the rupture properties of porcine descending thoracic aortas using a pressure-imposed test by inflating a rubber balloon. In this study, the authors used air instead of water, which is more commonly used in inflation tests, to apply pressure to the blood vessel walls. The study reported a strong correlation between strength and a characteristic stiffness modulus. Figure 2.5 depicts the setup of the bulge inflation test used in this study.

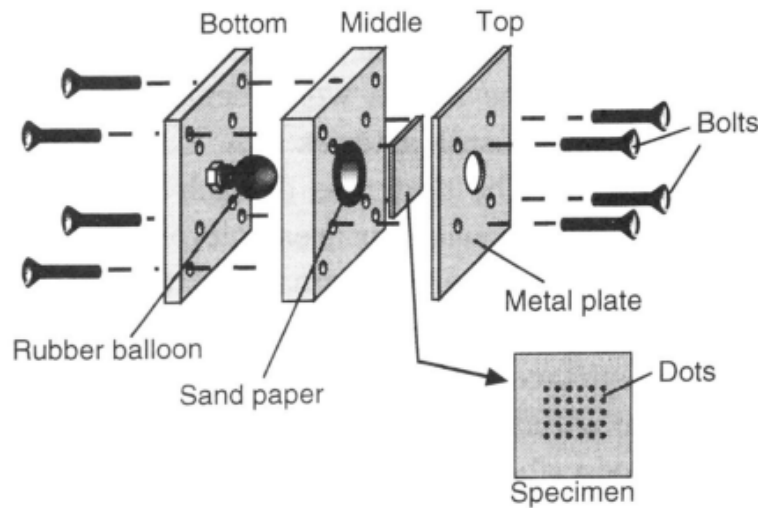


Figure 2.5: Schematic diagram of the experimental setup for the bulge inflation test [62].

Furthermore, another method for testing inflation is recognized as the inflation-extension test in the scientific community. This method stands apart from the previous one since it attempts to simulate the cylindrical loading scenario of an artery *in vivo*. A portion of an undamaged artery is expanded to simulate axial loading, and fluid is pumped through it to simulate circumferential loading. Complex optical technologies are used to monitor the displacement and force of predefined markers on blood vessels [14, 15].

This way, combining extension and inflation provides a comprehensive approach to studying arterial mechanics, accounting for both sample curvature and residual stress. This way, this test can provide a more accurate representation of blood vessel biomechanical function under physiological loading conditions [14, 15, 63].

The utilization of the inflation-extension test for evaluating AsAA is currently infrequent in the field of research. This scarcity can be related to the inherent constraints imposed by the limited size and curvature of readily accessible surgical specimens, which prevents further examination [15].

2.3.2 Inflation-Extension Apparatus

In order to accurately replicate the physiological conditions of the aorta in phantoms, the use of inflation machines is necessary. These machines enable controlled pressurization of the phantom, enabling the simulation of the behavior of blood vessels during cardiac cycles. Among the various tests available, the inflation-extension test has gained prominence as a preferred method for estimating *in vivo* stress. This is because it closely mimics the motion of the aortic wall during the cardiac cycle, making it highly suitable for the objectives of the project at hand [14, 15].

However, it is worth noting that there is limited literature available on the application of the inflation-extension test specifically to TAAs. Existing studies often employ non-standardized and complex protocols, where the sample is pressurized at different fixed lengths or under varying axial loads. These factors can introduce challenges and complexities to the inflation-extension technique. Despite these challenges, the potential benefits and relevance of the inflation-extension test make it a worthwhile endeavor [14, 15].

This section aims at providing an overview of different inflation machines that have been developed and utilized in the literature. These machines offer distinct features and capabilities, which contribute to the advancement of research in this field. By understanding the range of inflation machines available, researchers can make informed decisions regarding the selection of the most suitable equipment for their specific experimental needs.

Mascarenhas et al. [64] developed an experimental setup to simulate the inflation and deflation of the aorta, shown in Figure 2.6. The setup consisted of an axial piston pump (A), a circular water compartment (B), a water tank (C), two mechanical valves (D) and a vertical air column (G). The axial piston pump controlled the flow of water, mimicking the systole and diastole phases of the cardiac cycle. During systole, water was pumped from the circular compartment to the aortic segment, simulating aortic inflation. Mechanical valves allowed one-way flow, replicating the function of the aortic valve. During diastole, water flowed back from the water tank to the circular compartment, causing aortic deflation. The setup also included a silicon tube connected to a vertical air column with adjustable resistances, regulating water flow.



Figure 2.6: Experimental setup for the extension-inflation test [64].

Kim et al. [63] developed a computer-controlled system consisting of motion, data acquisition, and vision systems to measure the biaxial deformation of a porcine thoracic aorta during the inflation test. The setup included a polycarbonate test chamber, moving crossheads driven by a stepper motor, and a syringe pump for cyclic inflation and deflation of the specimen. Pressure transducers and a load cell were used to measure transmural pressure and axial load, respectively. The system also incorporated a stereo vision system for 3D marker tracking. Figure 2.7 depicts the experimental setup used for the inflation-extension test.

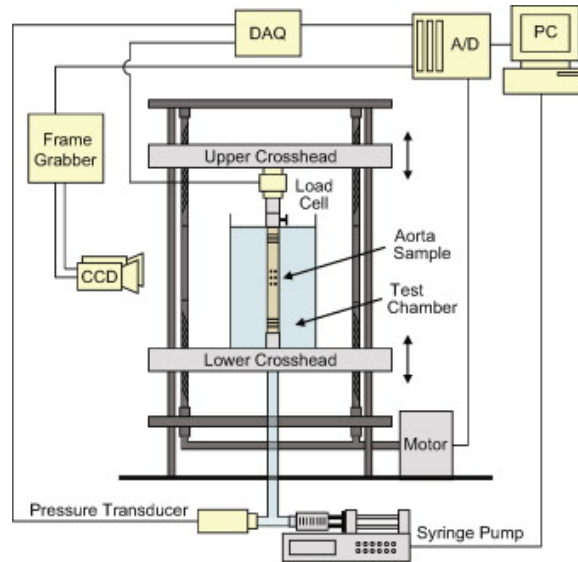


Figure 2.7: Schematic diagram of the experimental setup for the extension-inflation test [63].

Micco et al. [48] developed an experimental setup involving a parametrized aortic phantom housed in a pulse duplicator designed to replicate systemic circulation. The setup included components such as the left ventricle, aortic valve, ascending aorta, systemic resistances, left atrium, and mitral valve. A chamber connected to an electromagnetic motor mimicked the left ventricle's contractility, while a Windkessel tank simulated vessel compliance. The circuit was controlled by a program which regulated the flow and pressure by adjusting the bellows' amplitude, air pressure in the Windkessel tank, and peripheral valve opening. The setup allowed for the generation of pulsatile flow through the system, replicating physiological ranges for stroke volume and mean aortic pressure.

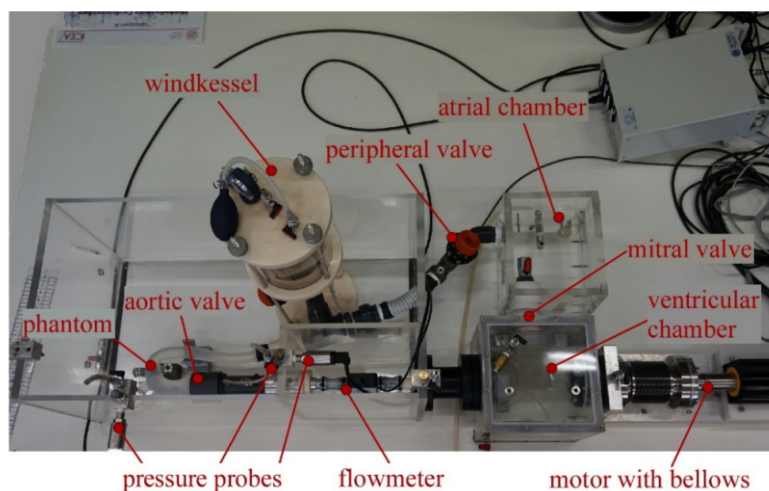


Figure 2.8: Experimental setup for the extension-inflation test [48].

The prevailing approach in inflation tests, as evidenced by the literature review, involves the utilization of fluid mediums to mimic the dynamic properties of blood flow. However, the project in which this dissertation is inserted has as its ultimate goal to be used in a hospital environment, in order to measure the deformation of aortic tissue, which consequently makes the use of liquid fluids more challenging. The current investigation introduces an experimental setup employing air inflation through the use of an intra-aortic balloon, thereby facilitating the measurement of aortic displacements using DIC, which will be developed subsequently.

2.4 Digital Image Correlation

The measurement of aortic displacement is critical for the diagnosis and treatment of aortic diseases such as aneurysms and dissections [65, 66]. Digital Image Correlation is a promising technique for measuring aortic displacement due to its ability to provide high-resolution and real-time displacement measurements [67, 68]. Additionally, it is non-invasive and can be used *in vivo*, which is a significant advantage for biomechanical applications [65].

DIC is a full-field optical technique that uses digital image processing algorithms to track the displacement of surface features of an object over time [69]. By accurately tracking speckle patterns, which are random, DIC provides a comprehensive understanding of a material's behavior under loading conditions, offering a full-field measurement of the visible surface of the test object, in both small and large deformations for any size and material of specimens, making it a versatile tool [65, 68, 70, 71, 72]. Moreover, DIC requires optical access to the specimen and surface preparation, which can be challenging in some applications. By fully grasping the potentialities and limitations of DIC, those who implement this technique can employ it with precision and efficacy in diverse fields of study [65].

In this dissertation, a comprehensive review of DIC is accomplished, including its operating principle and its categories, as well as its implementation procedures, such as speckle pattern fabrication, image acquisition techniques and deformation measurement. Finally, a brief context of this technology in biomechanics will be presented.

2.4.1 Operating Principle

To perform DIC, sets of images (in digital form) of the surface of the specimen in the undeformed (reference) and deformed states are captured, and the images are divided into smaller sub-images (subsets) for analysis [65]. A matching algorithm is then used to compare the subsets between the reference and deformed states (performing a correlation search), and the displacement field is finally computed based on the subset matching for all subsets. Subsequently, the strain field can be obtained by differentiation [65, 66, 68].

For the surface deformation information to be detected, the surface should be covered with areas of high visual contrast, usually accomplished by applying a random speckle

pattern.[65, 66, 68, 69, 71, 73, 74, 75, 76, 77, 78, 79]. The basic operation of DIC is tracking this speckle pattern, in a sequence of images. The basics of the process can be divided into three steps [69]:

1. Obtain a pattern on the sample for tracking;
2. Capture images of the sample during motion/deformation;
3. Analyze the images to compute the sample surface's displacements.

In the case of 3D DIC, a calibration of the stereo system is also necessary [69].

2.4.2 DIC categories

Within the last few decades, significant advancements have been achieved in DIC algorithms aimed at enhancing the resolution and robustness of the technique [68, 74]. The DIC technique can currently be broken down into three primary groups based on the set of analyses and/or measurements performed, as can be observed in Figure 2.9, namely:

- In plane or Bidimensional Digital Image Correlation (2D-DIC): This version of the system captures images with a single camera and enables displacement measurements in only two dimensions [68, 69]. As a result, this system can only measure deformations that occur within this plane and on surfaces with at most a small amount of curvature. Outside of this plane deformations can cause measurement errors, while image distortions can introduce inaccuracies into DIC measurements, limiting their effectiveness if uncorrected [69];
- Tridimensional Digital Image Correlation (3D-DIC): employs two synchronized cameras for recording images, although it is also a possibility to use only one camera with specially designed light-splitting system for pseudo stereo image formation of the tested surface [73]. It can measure displacements in all three dimensions with high accuracy on both flat and curved surfaces, without introducing error in 3D measurements due to out-of-plane deformations, as 2D DIC can [68]. Furthermore, lens distortions are corrected in 3D DIC via a calibration process that calculates intrinsic and extrinsic parameters for posterior image correction, effectively correcting lens distortions and aberrations as well as position related parameters [69];
- Digital volume correlation (DVC): It is a powerful tool that broadens the capabilities of 2D-DIC into the three-dimensional realm. It relies on the comparison of volumetric images obtained using a 3D imaging device, such as an X-ray CT system, making it an effective means of measuring the internal deformation of solid opaque objects as well as semi-transparent biological tissue [73]. Unlike 3D DIC, which only measures

surface displacements, DVC may employ a material's full structural information and be combined with finite element solvers to calculate internal strain fields using a continuum assumption. The lengthy acquisition time required to obtain high-quality 3D images, on the other hand, remains a significant limitation of DVC, as does the requirement of existing features in the interior of the object that the technique can follow upon deformation [68].

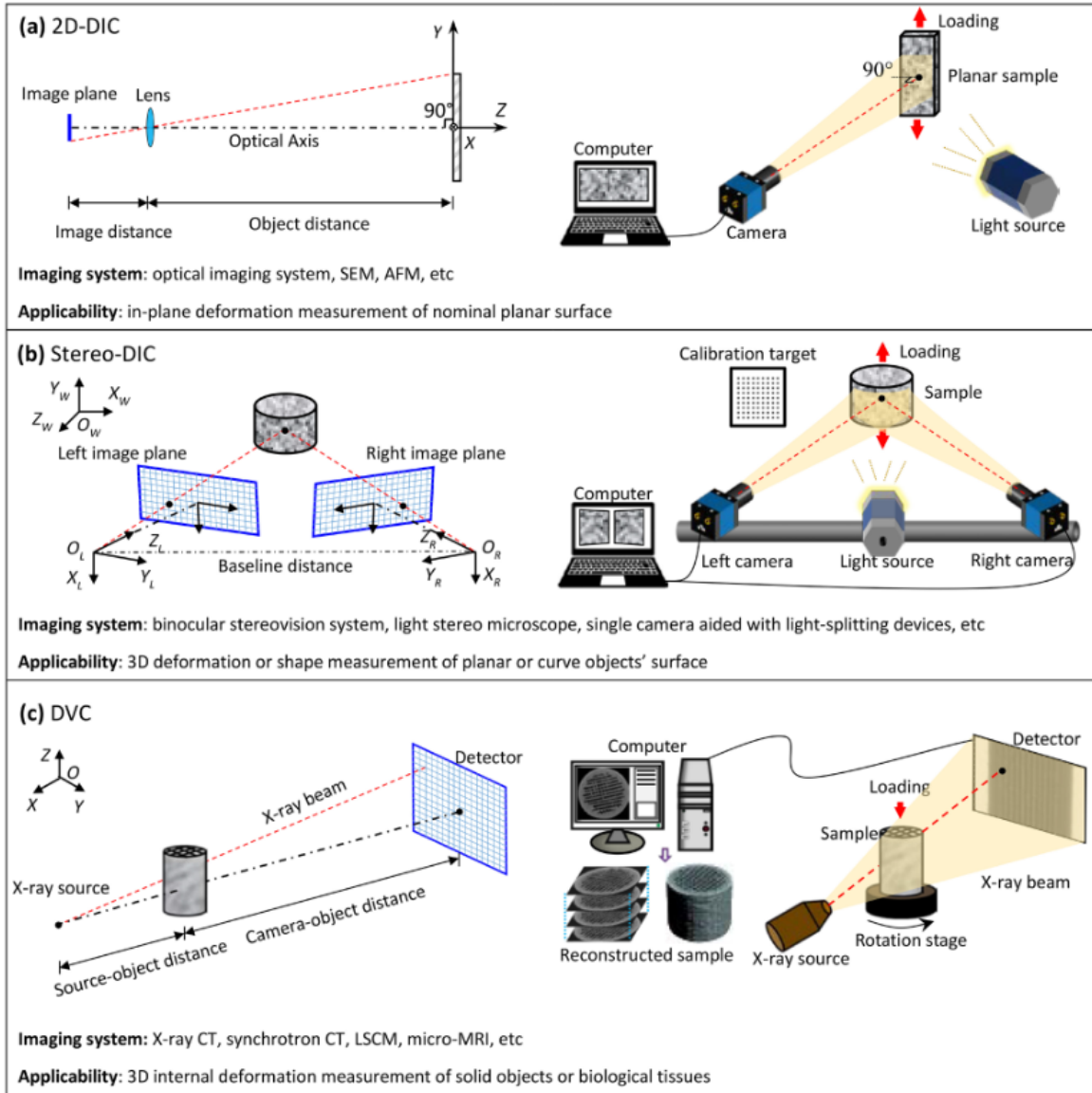


Figure 2.9: DIC techniques and their imaging models, imaging systems and applicability [73].

2.4.3 Implementation procedures of 3D DIC techniques

The process of ascertaining surface deformation utilizing DIC methodologies typically encompasses a sequence of three consecutive phases [69]:

1. Speckle pattern application ;
2. Image acquisition;
3. Displacement and deformation calculation.

2.4.4 Speckle Pattern Application

To enable optimal DIC application, the surface of interest must possess a high-contrast random speckle pattern that moves and deforms in conjunction with the specimen surface, allowing the software to unequivocally identify and track each point on the specimen's surface across subsequent images acquired while the specimen is displaced and/or strained [65, 68, 73]. Owing to this, it is essential to take into account the texture or inhomogeneity of the surface being studied. If the specimen already possesses a natural and non-uniform texture, the system may employ it directly. If the surface lacks this type of pattern, a random pattern must be generated and applied to the surface in order for the system to capture and analyze deformation information effectively [65, 66, 68, 69, 73]. The speckle pattern should ideally meet the following requirements to ensure the accuracy and precision of the computed displacements and strains [65]:

- It covers entirely the surface of interest for the analysis [69];
- High contrast, in the interest of minimizing the amount of noise in the acquired images [65, 69, 70, 80];
- A black/white ratio of 50:50, in order to avert regions which may not be recognized properly [65, 80];
- Random distribution (non-periodic and non-repetitive pattern) to facilitate full field displacement mapping [65, 66, 68, 69, 71, 73, 74, 75, 76, 77, 78, 79];
- Roughness should be minimized as much as possible to avoid modifying the surface geometry [65, 74];
- The speckle size should be consistent and have a minimum of 3-5 pixels to over-sample the features in each speckle pattern [68, 69, 70, 80, 81]. Speckles smaller than this size are not properly identified, culminating in a grey image [70];
- It should be isotropic (it should not exhibit any bias in any particular direction) [65, 80].

Figure 2.10 and Figure 2.11 represent examples of bad speckle patterns and good speckle patterns, respectively.

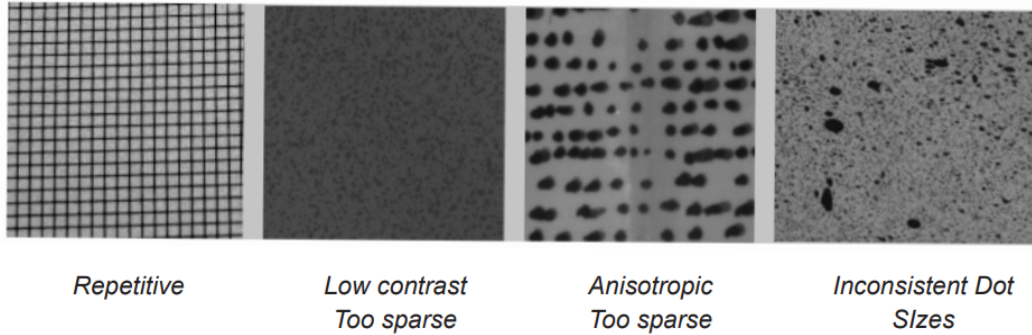


Figure 2.10: Examples of poor speckle patterns [80].

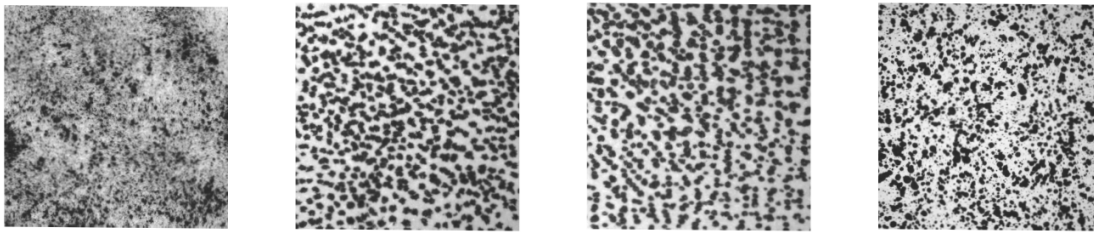


Figure 2.11: Examples of good speckle patterns [80].

2.4.4.1 Speckle patterning methods

In the context of DIC, the utilization of an optimal speckle pattern is a critical factor in minimizing the measured noise and enhancing the accuracy of the overall results [80]. However, creating this ideal artificial speckle pattern can be challenging given that it is heavily reliant upon the particular application being used, including factors such as the material composition of the specimen, the extent of the observed area, as well as other variables [69, 70].

Speckle patterns can be categorized based on their scale, with two main categories being macro-scale and micro/nano-scale patterns [74]. The present dissertation focuses particularly on macro-scale patterns due to their broad applicability in various production processes and with the process presented in this dissertation. The analysis will emphasize the unique features of such patterns, which are characterized by grains of macroscopic dimensions.

The most commonly used techniques to prepare a speckle pattern are:

- **Spraying:** This method entails the use of spray paint bottles to create a random pattern on the sample being tested, making it an extremely accessible and simple process

for creating speckle patterns [74, 80]. It is recommended to use spray paint with a matte finish to reduce the effect of lighting reflection and with sufficient elasticity and adhesion for any type of surface [82]. Since paint colors other than black and white will inherently have less contrast, these colors are the most recommended for usage at room temperature [74]. The optimal approach for creating speckles is to use white paint as the background and black paint for the speckles, as this arrangement provides greater contrast than the reverse order [68, 69, 74, 75, 80]. Typically, this technique involves applying several thin coats of white paint to the specimen's surface, followed by the speckles after the base coat has partially dried. It should be noted that if the base coat is still moist, the paints may mix and lose their distinctiveness [80]. Despite being a widely used technique that can produce a suitable speckle pattern, spray paint is subject to substantial variation in its quality, due to the difficulty in achieving consistent speckle size, given the unpredictable nature of the paint's projection. Additionally, the fine mist that accompanies the paint spray can contribute to the speckles appearing blurred and, thus, causing a reduction in contrast [80, 82];

- **Airbrushing:** This highly versatile technique bears similarities to traditional spray painting methods. As the latter, the process typically commences by carefully applying a series of thin, even layers of white paint to the surface of the object in question. Once these layers have dried completely, the next step involves depositing carefully curated speckles of black paint on top of the white base [65, 68, 69]. However, airbrushing offers an added degree of precision and control that sets it apart from its predecessor. By utilizing the airbrush, a higher level of refinement and nuance in the speckle pattern application can be achieved [74];
- **Stamping:** This technique involves the application of ink to the specimen by rolling dots onto its surface using stamps. The stamps come in various sizes of dots to produce the appropriate speckle pattern which is dependent on the camera resolution and field of view (FOV). It is typically essential to apply a base coat, usually white paint, to the surface before the stamping process commences [80];
- **Sharpie/permanent marker:** Creating a speckle pattern using a sharpie marker is a highly effective technique with minimal impact on the surface of the specimen, allowing for precise measurements of high strain. It is also versatile, being able to be applied to specimens with complex geometries and textures, as well as providing control over the speckle size. A white base coat with black markers can provide excellent contrast, but it is also possible to use a silver sharpie on a black base coat or surface, also providing excellent contrast against the black surface [80].

2.4.5 Digital Image Acquisition

As previously noted, various classifications of Digital Image Correlation (DIC) exist. The present dissertation, however, shall concentrate its attention specifically on the category of 3D DIC.

Calibration is an essential step in initializing the spatial correlation processes of DIC in 3D applications [65]. It involves obtaining intrinsic and extrinsic parameters that enable accurate triangulation. The intrinsic parameters include the center point, lens distortion coefficients, and focal length of each camera and the extrinsic parameters involve determining the translation and rotation between the two cameras [73].

This process involves capturing images of precise calibration targets in various poses within a calibration volume. The calibration software computes the intrinsic system parameters by minimizing a cost function [68]. By performing these calibration procedures, DIC systems establish the necessary parameters and optimize their accuracy for subsequent displacement measurements [68].

Figure 2.12 presents an example of four commonly used planar calibration targets.

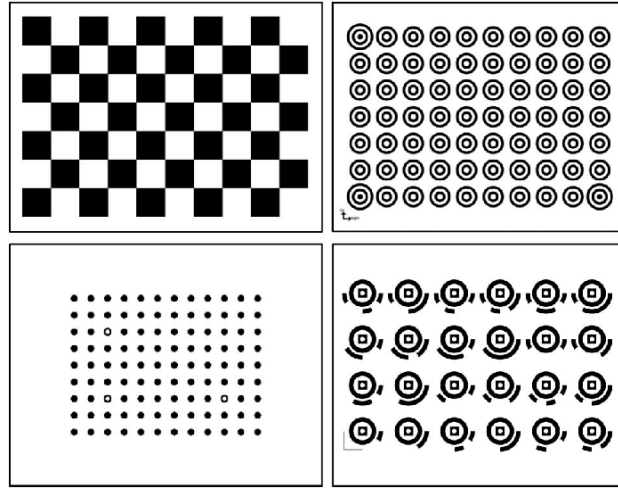


Figure 2.12: Examples of four commonly used camera calibration targets [83].

2.4.5.1 3D DIC Setup

Below is an illustration of a 3-D DIC system as a prime example. This setup comprises two cameras equipped with lenses, stereo lights accompanying the cameras and an additional fiber optic light for back-lighting the glass calibration grid [69].

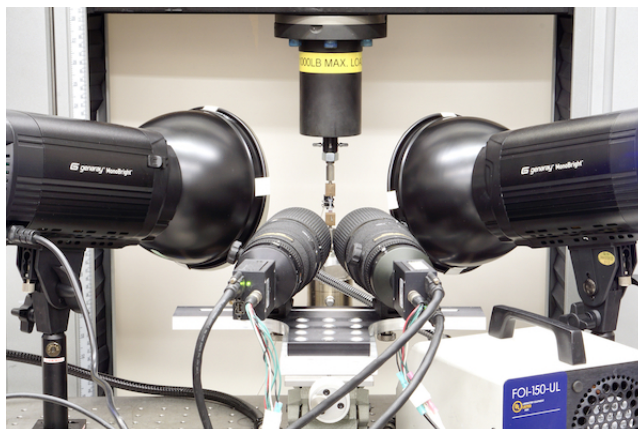


Figure 2.13: Usual setup of 3D DIC [69].

2.4.6 Displacement and deformation calculation

By using the calibrated parameters and disparity maps obtained through DIC algorithms, it becomes possible to reconstruct the 3D coordinates of measurement points. This process enables the calculation of surface displacement fields by subtracting the 3D cloud points obtained before and after deformation. These displacement fields provide valuable information regarding the magnitude and direction of displacements [73].

2.4.7 Applications in Biomechanics

The measurement of displacement and strain is critical in the field of experimental biomechanics for assessing biological tissues, organs, and their interactions with biomedical equipment [65]. Traditional mechanical or optical transducers can only detect one-dimensional elongation between two levels, yet DIC overcomes this limitation by measuring displacement fields on a specimen's visible surface. This is especially useful for non-uniform and irregularly shaped biological soft tissue samples undergoing non-uniform elongation [67]. This way, DIC is a versatile technique with significant potential for investigating various types of materials, including both soft and hard biological tissues, regardless of their mechanical behavior and for capturing small or large deformations [65, 84].

Many studies investigating bulge inflation and inflation-extension tests have successfully employed DIC as a valuable tool for their analyses [53, 58, 60, 63, 72, 78, 85]. The versatility and accuracy of DIC in measuring displacement and strain make it a promising technique for advancing our understanding of the mechanical behavior of biological tissues and enhancing biomechanical research in diverse applications.

Materials and Methods

To initiate the experimental procedure, a series of preparatory steps were diligently undertaken to ensure the successful execution of the test. This involved procuring the materials necessary and fabricating the balloon and phantom needed for the experiment. The manufacturing process of these components constituted a significant portion of the pre-experimental phase. Furthermore, it was important to understand the test requirements and make any necessary adjustments to ensure proper test operation. This chapter provides a detailed account of each of these steps, leading up to the execution of the experiment.

3.1 Materials

This study involved the use of various materials for manufacturing both the balloon and the phantom of the aorta. Specifically, ecoflex 00-30 was selected as the material for the balloon, while cristal rubber, sylgard 184, and HB FLEX 5540 A+B were evaluated to determine their suitability as phantom materials for the experimental procedure.

In this section, a comprehensive investigation is conducted on all the employed materials. The analysis begins with an explanation regarding the rationale behind their selection. Each material's properties and characteristics are thoroughly explored in relation to their role and significance within the experimental framework.

3.1.1 Ecoflex 00-30

The fabrication of the balloons involved the use of pre-existing materials in the laboratory, specifically ecoflex 00-30 and 00-50 rubbers, which were readily available due to a previous research regarding materials for the intended use. Ecoflex rubbers are characterized by their remarkable flexibility, capable of stretching significantly without tearing and returning to their original shape without distortion [86, 87]. Furthermore, their water white translucency is advantageous as it ensures minimal interference with the phantom.

To minimize any potential interference during the analysis and to fulfill their primary role of facilitating phantom inflation, ecoflex 00-30 was selected due to its lower shore hardness [86]. By employing this material and constructing balloons with thin walls, the aim was

to achieve the desired outcome while minimizing any disruptive effects on the analysis. Furthermore, its design ensures minimal interference with the results by facilitating quick and immediate inflation of the phantom. Its process of development is described in Subsection 3.2.1. Table 3.1 presents the properties of the material chosen for the balloon.

Table 3.1: Properties of the Ecoflex 00-30 silicone from Smooth-On [86]

Mixture ratio	1:1
Color	Translucent
Pot life	45 minutes
Cure time	4 hours
Hardness (Shore A)	30
Viscosity	3 Pa*s
Elongation at Break	900 %
Tensile Strength	1.38 MPa
Tear Strength	6.7 N/mm
Specific gravity	1.07

3.1.2 Prochima Cristal Rubber

Prochima Cristal Rubber is a PDMS, which is a type of silicone elastomer that exhibits rubber-like properties. One of the main factors for this choice of material was its common use in blood vessel prototyping [32]. Its elasticity and flexibility, closely mimicking the overall mechanical response of arterial tissue, enables the fabrication of compliant phantoms that can simulate the deformation and behavior of real arteries under different physiological conditions [48].

Another important parameter for choosing this material was its ease of use, given that it offers a convenient mixing ratio of 1:1, simplifying the fabrication process. The equal parts mixing ratio eliminates the need for complex calculations and provides consistent results. This ease of use contributes to efficient phantom production and minimizes the potential for error. Moreover, the presence of distributors in Europe added significant value, given the relatively short timeframe.

The curing time of this material from Prochima was a significant factor in the selection process. With a curing time of 24 hours at 23°C, it offered a relatively short waiting period compared to other rubber materials. This characteristic allows for faster fabrication of arterial phantoms, facilitating a more efficient workflow. Finally, this material was less expensive than other materials usually utilized for phantom construction. Table 3.2 shows the present material properties.

Table 3.2: Properties of the Cristal rubber silicone from Prochima [88]

Mixture ratio	1:1
Color	Transparent
Pot life	90 minutes
Cure time	24 hours
Hardness (Shore A)	20
Viscosity	9 Pa*s
Elongation at Break	350 %
Tensile Strength	1.7 MPa
Tear Strength	3.2 N/mm
Specific gravity	1.05

3.1.3 Sylgard 184

Sylgard 184, a specific formulation of PDMS, offers several advantages when used as a phantom material to simulate the aorta. This elastomeric silicone has been gaining popularity in the field of medical modeling and flow simulation [35, 37, 46, 45]. Firstly, it has remarkable elasticity, closely mimicking the compliance and mechanical qualities of the artery wall [37]. This characteristic makes it possible to create lifelike and anatomically accurate phantoms. Furthermore, the material has a relatively quick curing period, of 48h at 25°C, ensuring rapid solidification and enabling fast and simplified phantom fabrication [89]. Moreover, this material had a distributor in Portugal, which also contributed for choosing the present choice. Table 3.3 depicts the properties of this material.

Table 3.3: Properties of the Sylgard 184 silicone from Dow Corning [89]

Mixture ratio	10:1
Color	Transparent
Pot life	90 minutes
Cure time	48 hours
Hardness (Shore A)	43
Viscosity	3.5 Pa*s
Tensile Strength	6.7 MPa
Specific gravity	1.03

3.1.4 HB FLEX 5540 A+B

To accommodate the growing demand for vascular phantoms, specialized companies have emerged. These businesses specialize in the creation of high-quality vascular phantoms to meet the needs of researchers and medical practitioners. Elastrat, based in Geneva,

Switzerland, is one significant example [90]. This company provides a selection of silicone phantoms that accurately reproduce anatomical vasculature, including diseases such as aneurysms and stenoses. Soft and rigid phantoms are available in their product line. Elastrat also specializes in developing customized models from STL files supplied by clients. These phantoms provide a realistic simulation environment for endovascular treatments, pre-surgery training, research studies, and instructional reasons for interventionists and other medical professionals [50]. Due to the expected need to intentionally alter and damage phantoms for experimental purposes, the associated cost was prohibitively high for the present study. As a result of our study objectives not aligning with the goals of their project, it was decided the optimal course of action involved engaging a reputable portuguese company capable of formulating a silicone composition exhibiting similar characteristics. Table 3.4 and Table 3.5 present the properties of the flexible silicone from Elastrat, as well as the material obtained from Hb Química, respectively.

Table 3.4: Properties of the flexible silicone models from Elastrat [50]

Color	Translucent
Hardness (Shore A)	40
Tensile Strength	6.7 MPa
Elongation at break	400 %
Tear Strength	27 N/mm
Specific Gravity	1.09

Table 3.5: Properties of the silicone HB FLEX 5540 A+B obtained from Hb Química

Mixture ratio	1:1
Color	Translucent
Pot life	30 minutes
Cure time	3 hours
Hardness (Shore A)	40
Viscosity	5 Pa*s
Elongation at Break	300 %
Tear Strength	17.5 N/mm
Specific gravity	1.20

One noticeable feature of this material is its exceptionally quick cure time, which takes only 3 hours to complete. This shortened duration contrasts sharply with alternative materials frequently used in phantom fabrication. As a result, the accelerated curing process allows for the manufacture of a greater number of phantoms in a substantially shorter time frame.

3.2 Methodology Analysis

After the selection and analysis of the involved materials, this section discusses the techniques used throughout the study. Methodologies for the fabrication of the balloons and phantoms are addressed, as well as the functioning of the inflation system and the faced challenges.

3.2.1 Preparation of the balloons

For the experimental procedure at hand, the development of an optimally sized mold for the balloon was a crucial step, aimed at minimizing interference during the inflation process. This mold underwent an iterative refinement process in order to determine the most effective option. Additional insights into the specific iterations will be provided subsequently. The flowchart in Figure 3.1 displays the whole procedure of obtaining the balloons.

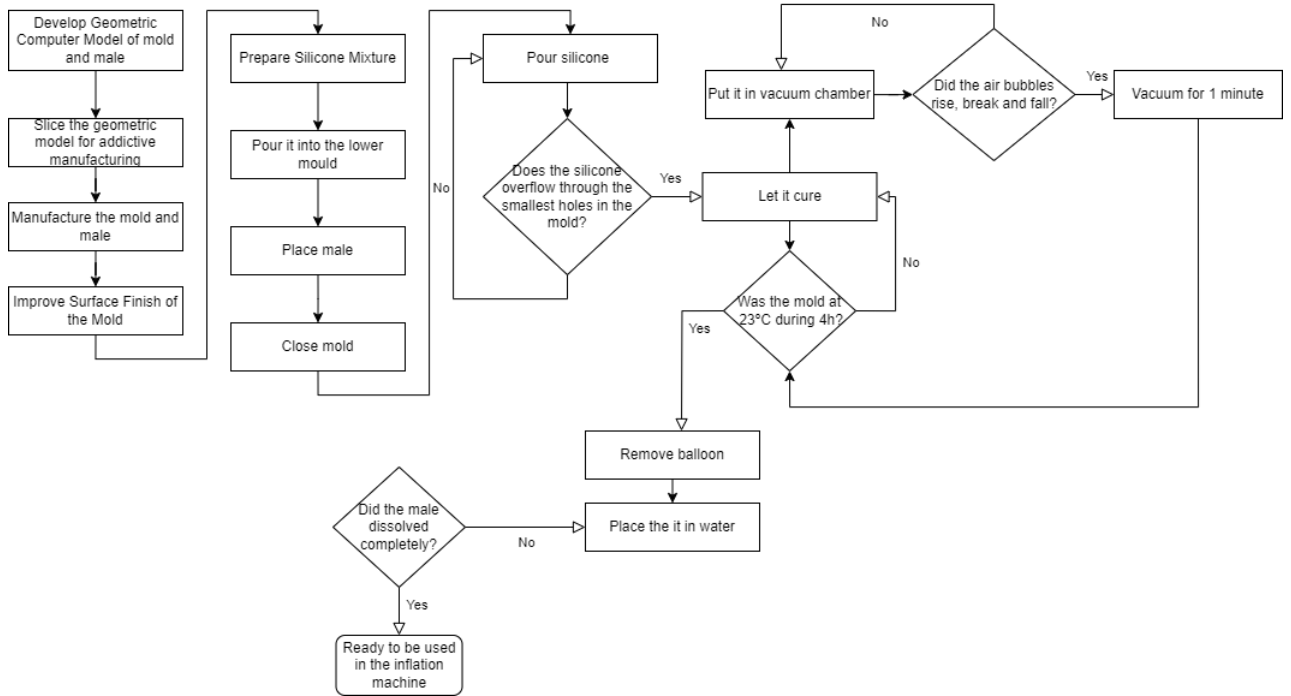


Figure 3.1: Process diagram for the manufacture of the balloons.

The first stage of balloon manufacture entailed developing a computer-aided design (CAD) model. A pre-existing balloon model had already been made using the mold indicated in Figure 3.2. This balloon was created with the use of a Polylactic Acid (PLA) male, which was then retrieved after the silicone had cured. The removal operation, however, entailed outward pulling, which undermined the structural integrity of the balloon tips. Consequently,

this technique resulted in areas of decreased strength, requiring the research of alternative methods to achieve greater uniformity during the inflation process. The presence of balloon heterogeneity would thus prevent an equitable expansion, thus influencing the results in a detrimental way. Therefore, it was imperative to devise a solution that could mitigate such biases and promote optimal and consistent inflation throughout the structure.

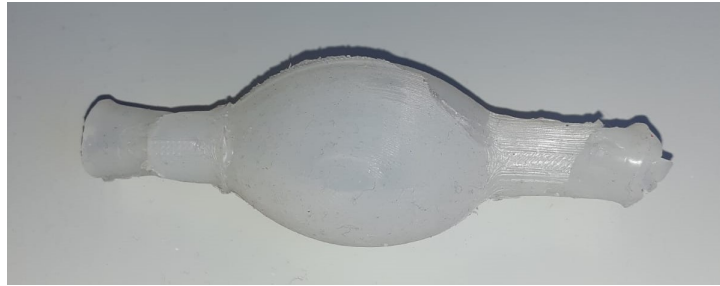


Figure 3.2: First balloon made using a removable PLA male.

The removal of the existing mold presented an additional challenge. The mold, as shown in Figure 3.3, had a notch for connecting purposes. As a result, after the silicone had solidified, the presence of silicone residue within this notch complicated the extraction process and caused it to gradually degrade with each subsequent use. Figure 3.3 shows that the mold had already deteriorated at the corners as a result of repeated attempts to forcefully remove it.



Figure 3.3: Mold for the first balloon.

Furthermore, considering the mold was produced by FDM, which has lower resolution along the Z axis compared to other additive manufacturing techniques, layering effects and

undesirable surface roughness may have occurred, necessitating post-print surface refinement [37]. As a result, simply extracting the balloon after silicone curing resulted in unevenness within the silicone structure. As a result, a better solution was required to address the shortcomings of the existing approach.

To solve the first problem, water-soluble cores were used, which were created using FDM. To that end, Verbatim Butenediol vinyl alcohol co-polymer (BVOH) was used. As a result, after curing the silicone, immersion of the male component in water resulted in the dissolution of the water-soluble core, allowing for the creation of a more uniform and homogeneous balloon structure, as shown in Figure 3.4.



Figure 3.4: First iteration of the balloon after using the BVOH filament.

Nevertheless, the utilization of this material posed certain challenges that necessitated the adjustment of various parameters to achieve optimal performance. Specifically, the material was printed utilizing vase mode, enabling the production of objects with a single-wall thickness, thereby minimizing print time and filament consumption. However, to effectively employ this mode, it was imperative to significantly reduce the standard print speed to a mere 7 mm/s while incorporating a brim during the printing process. Also, to ensure proper printing quality, the corners of the object were transformed into a conical shape. Figure 3.5 shows the printing process with the parameters talked about and Figure 3.6 shows the final product of the impression.

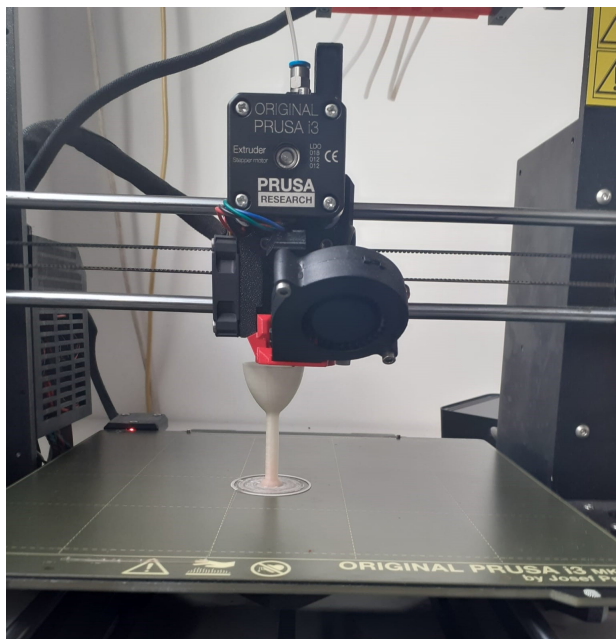


Figure 3.5: 3D printing of the male with verbatim BVOH with the required printing specifications.



Figure 3.6: Final product of the 3D printing process.

When the balloon underwent testing in the inflation setup, a significant vulnerability in its structural integrity, particularly in its wall, was revealed, resulting in a susceptibility to increased pressure, as shown in Figure 3.7. This occurrence could be attributed to the previously mentioned absence of a facilitating mechanism for easy removal from the mold. To address this, a release agent, specifically petroleum jelly, was used to speed up and streamline the detachment process, resulting in improved uniformity.

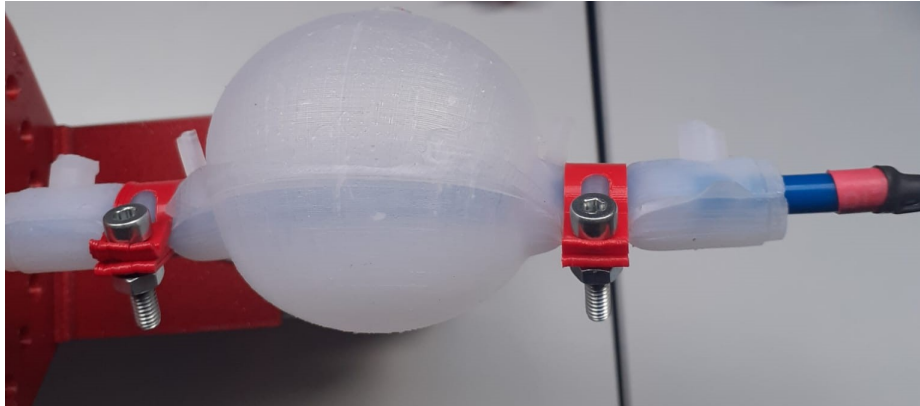


Figure 3.7: Inflated balloon.

Furthermore, an alternative mold fixation method involving the implementation of screw-based mechanisms was adopted, replacing the previous notch connection. Figure 3.8 shows the newly designed CAD mold, along with the resulting refined product, depicted in 3.8, exemplifying the successful implementation of these modifications.

In conjunction with these modifications, the dimensional specifications of the balloon's oval section were revised, increasing its diameter to closely match the larger inner diameter of the phantom. This adaptation allowed for the prompt and simultaneous inflation of both the balloon and the phantom upon pressurization. By increasing the diameter, the balloon begins to inflate immediately and simultaneously fills the phantom, avoiding the scenario in which the balloon would only expand until it reached the phantom's diameter before beginning the filling process. As a result, this change in diameter was deemed necessary to optimize the inflation dynamics.

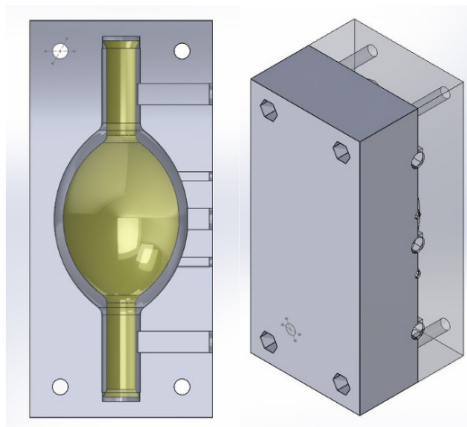


Figure 3.8: CAD model of the final balloon mold.

The final mold was then precisely dimensioned and fabricated using FDM, Figure 3.9. The required threaded components, in the form of nuts, were integrated concurrently with mold production. Following that, the male component was printed with the previously mentioned water-soluble filament material, allowing for its easy dissolution in subsequent steps. Finally, the surface treatment was improved by using a release agent.

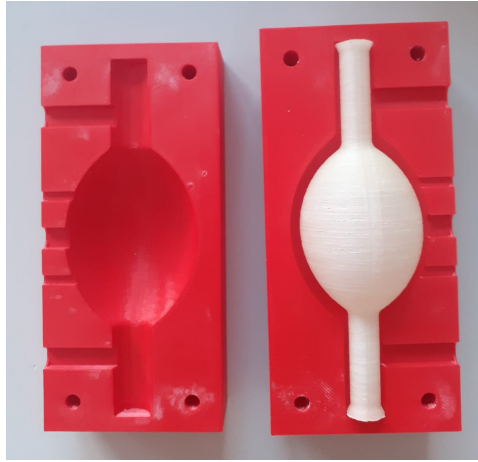


Figure 3.9: Final version of the mold and male manufactured.

Following the completion of the component fabrication, the following phase was the preparation of the silicone mixture. The guidelines outlined in the material's information sheet were followed at this stage, ensuring precise execution. These steps are depicted in Figure 3.10 and Figure 3.11.

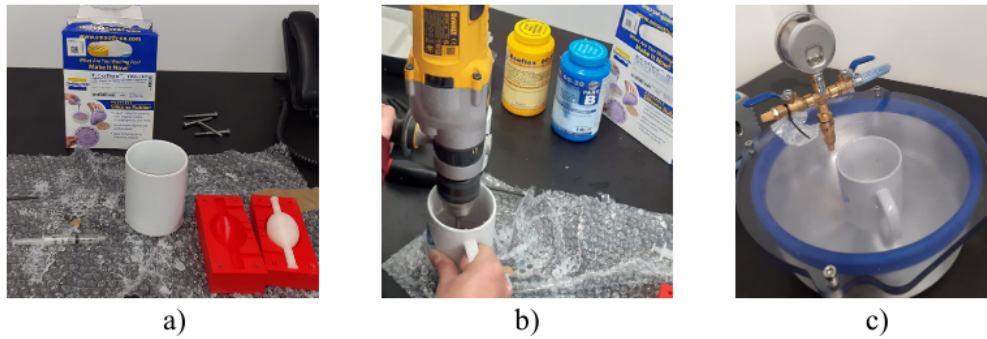


Figure 3.10: Preparation of the silicone mixture: a) setup; b) mixing of 1:1 part A and B; c) mixture in the vacuum chamber to eliminate air bubbles.

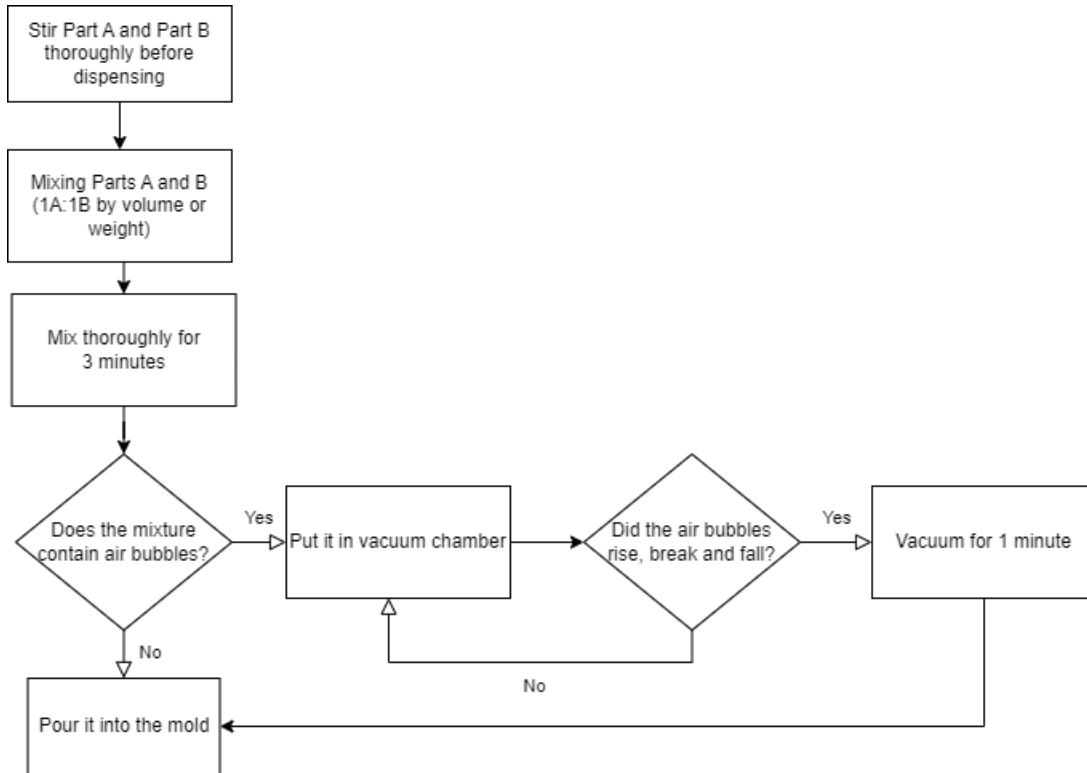


Figure 3.11: Process diagram for the preparation of the silicone mixture for the balloon.

When the silicone mixture was finished, a small amount was carefully poured into the lower mold, ensuring a partial filling of the designated area. Following that, the male component was positioned within the lower mold, followed by mold closure and secure screw fastening to ensure structural integrity. Simultaneously, the remaining silicone was slowly poured through the mold's larger diameter orifices. As the silicone gradually emerges from the smaller holes, indicating full mold occupancy, the pouring process is concluded. For the ecoflex 00-30 silicone, a curing period of 4 hours at 23°C was required. Once the prescribed duration had elapsed, the mold was unsealed, and the product was carefully extracted and subsequently immersed in water to facilitate the dissolution of the male component. This process is depicted in Figure 3.12.

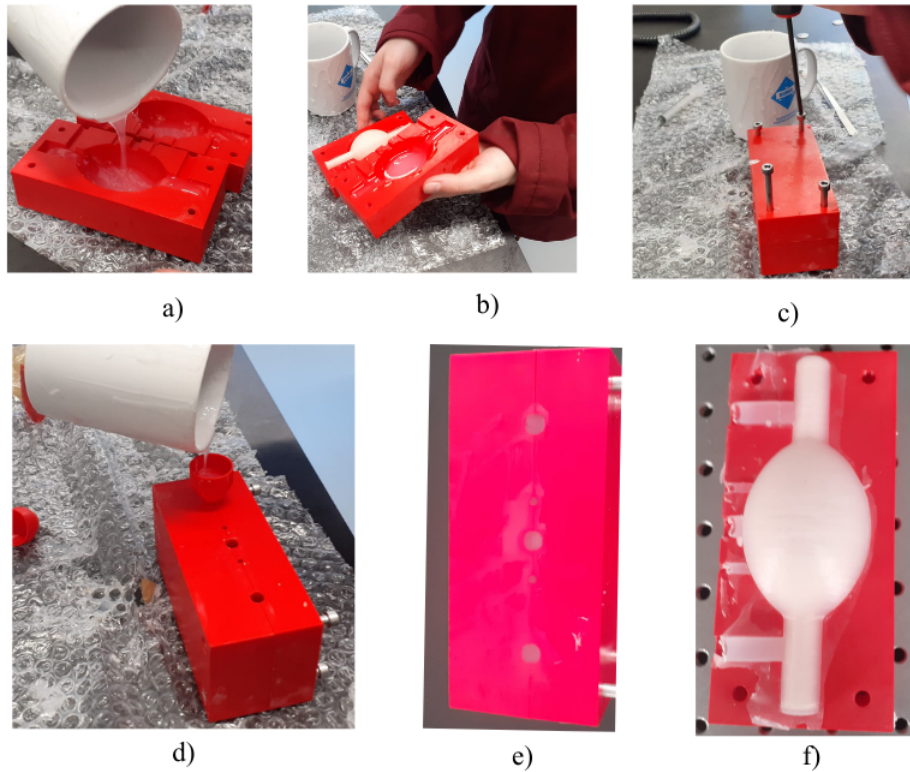


Figure 3.12: Process of obtaining the balloon: a) Pouring the mixture of silicone into the lower mold; b) inserting the male; c) closing the mold; d) pouring the mixture slowly into the orifice; e) stop pouring when the mixture comes out through the lower diameter orifices; f) product after curing of the silicone.

3.2.2 Phantom Fabrication

The fabrication of the phantom can be divided into three major steps [37]:

1. Definition of a 3D model;
2. Mold manufacture and casting;
3. Extraction and commissioning.

The 3D model was dimensioned based on the obtained real aorta model, with considerations for achieving a 2 mm thickness. This obtained model of the aorta had a length of about 80 mm and diameters of approximately 29 mm and 36 mm.

Both the mold and male components were 3D printed using PETG material. To ensure durability and optimize resource utilization, a reusable male component was employed instead of using a disposable male that would require frequent replacement. This approach minimizes material consumption and reduces production time.

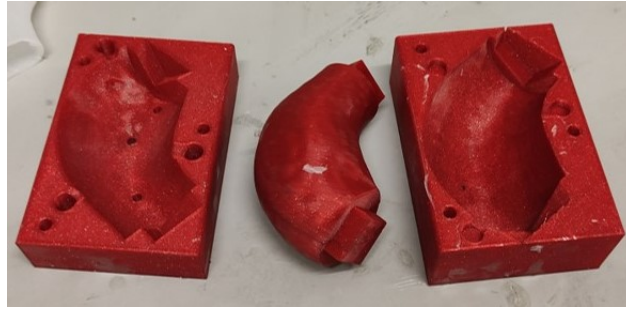


Figure 3.13: Mold and male used for the phantom production.

Given the preexistence of the mold, the first stage was skipped, allowing for a direct progression into the execution of the following two processes, encompassing the utilization of different materials. Figure 3.14 depicts the whole process necessary in order to produce the phantoms.

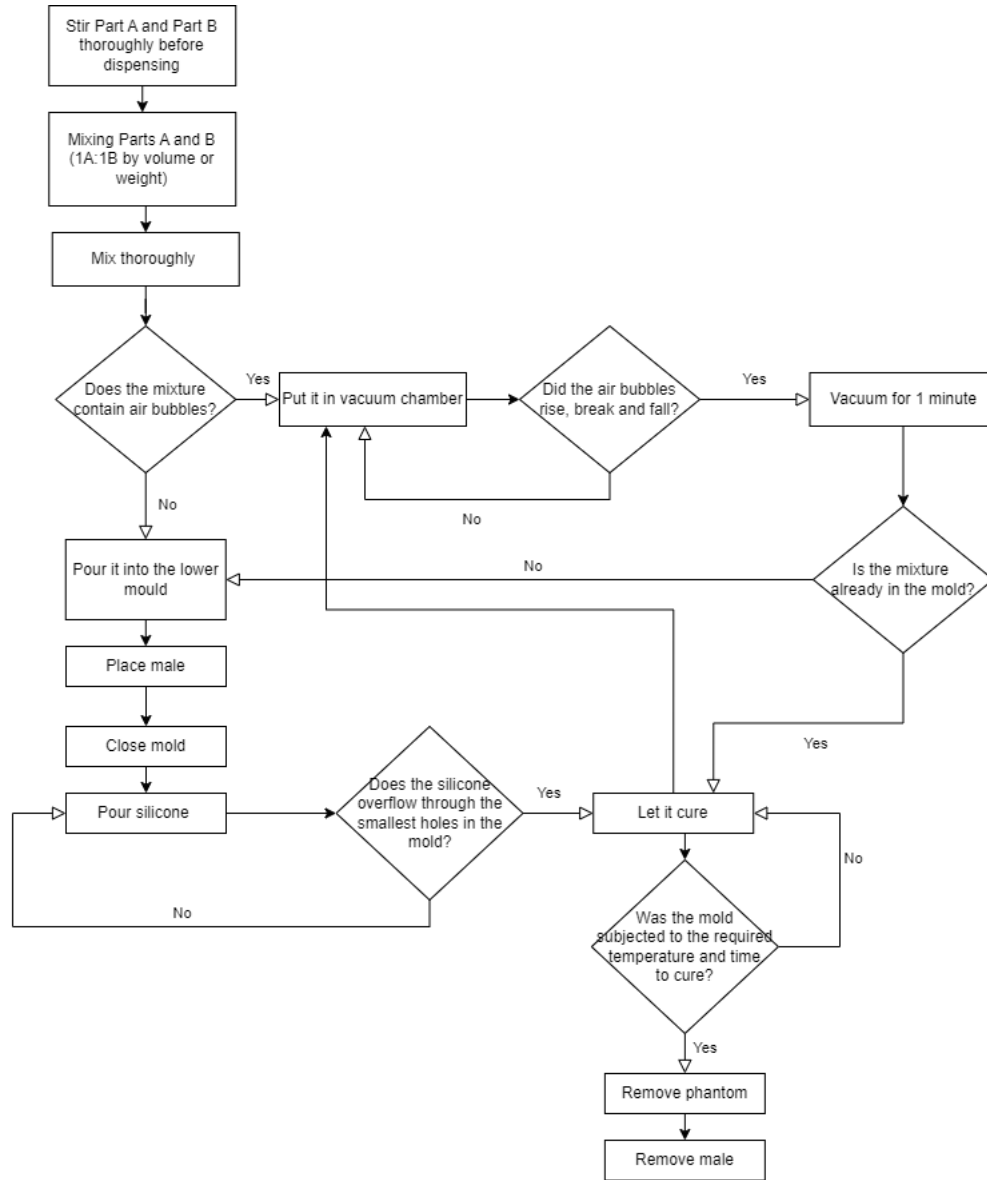


Figure 3.14: Fluxogram of the whole process of manufacturing the phantom.

3.2.2.1 Prochima Cristal Rubber

The cristal rubber material was manufactured in accordance with the instructions provided in the product brochure from Prochima and the steps outlined in Figure 3.14. However, after the 24-hour curing time at 23°C, the phantom was not completely filled and it still presented bubbles despite the vacuum treatment process depicted in the flowchart. Due to its frailty, the phantom tore when attempting to extract the male.

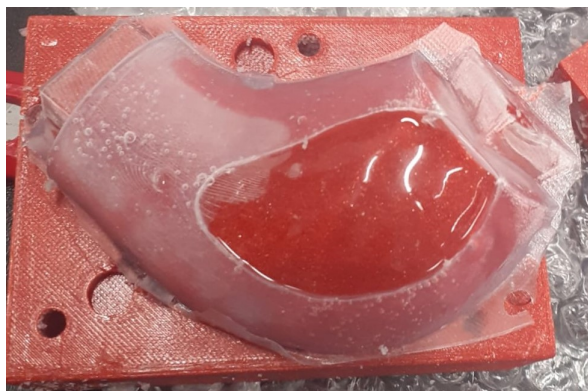


Figure 3.15: Resulting phantom using the cristal rubber silicone from Prochima.

Several factors could have contributed to the phantom's collapse. One of the main factors is that, in comparison to the other silicone materials, it has a higher viscosity. This way, when inserting the mixture it took longer to enter given the few inlet holes. Subsequently, the mold did not fill sufficiently until the pot life was over. Therefore, additional openings for silicone insertion should have been included to ensure thorough dispersion and eliminate weak spots. Furthermore, even after the process of vacuuming the air bubbles for about 40 minutes, air bubbles still remained, indicating that longer vacuuming times are required for better results for this material. Finally, the approach used to create the phantom may not have been optimal, and using a destructible male, while taking more time, may have resulted in more favorable results.

Due to time constraints and the need to size a new mold and manufacture another male, this material was not pursued.

3.2.2.2 Sylgard 184

The procedure outlined in the flowchart in Figure 3.14 was followed for Sylgard 184, with a mixing ratio of 10:1. However, when compared to the other silicone components employed, the final mixture was found to be more liquid, as seen in Figure 3.16.



Figure 3.16: Liquidity of the Sylgard 184 mixture in the mold.

As a result of the mixture's liquid state, when it was poured into the mold, it started leaking. To address this issue, a new mold design would be required to prevent the material from spilling. Given the limited time available, the Sylgard 184 material was not used in the tests.

3.2.2.3 Silicone from Hb Química

For the present silicone, the steps outlined in the fluxogram in Figure 3.14 were followed, with the product having a 1:1 mixing ratio. After a curing time of 3 hours, it was confirmed that the fabrication process yielded satisfactory results. The resulting silicone material demonstrated sufficient strength and showed compatibility with the method employed for removing and reusing the phantom, as depicted in Figure 3.17.

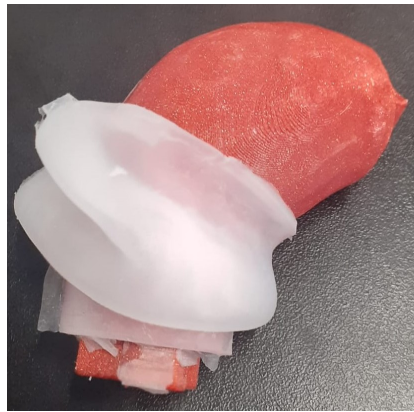


Figure 3.17: Removal of the male from the phantom.

Consequently, the fabrication procedure successfully produced ready-to-use phantoms, as displayed in Figure 3.18. Subsequently, the upcoming chapter will outline the procedural steps involved in utilizing these phantoms to obtain the final results.



Figure 3.18: Final phantom.

3.2.3 Experimental Apparatus

The system is composed of interconnected components that contribute to making the system simpler to operate, as can be observed in Figure 3.19. It commences with the actuator rod, which moves linearly, forcing the rod of the pneumatic cylinder to move in response. The movement of the cylinder rod compresses the air within the cylinder. This compressed air is then sent through a pneumatic tube with a predetermined opening, where it exerts pressure on the balloon.

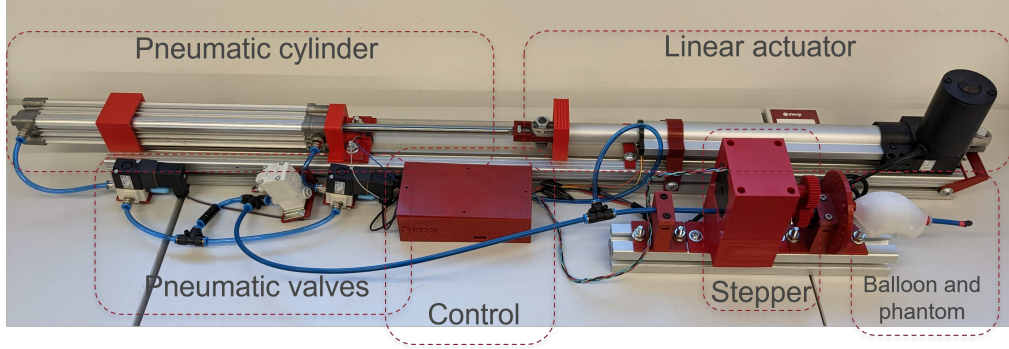


Figure 3.19: Setup of the inflation system.

The primary aim of the experimental procedure is to apply pressure equivalent to the diastolic arterial pressure (80 mmHg) and systolic arterial pressure (120 mmHg) utilizing the inflation machine, thereby simulating the pressures experienced during a cardiac cycle. The subsequent objective is to quantify the displacement occurring between these distinct pressure states.

This way, the process begins by activating the machine and adjusting it to the desired pressure, specifically the diastolic blood pressure. Once the setup reaches this pressure, the experimental procedure commences, involving the initiation of the DIC process. Subsequently, the system is configured to reach the systolic blood pressure, and upon reaching this point, the process is halted. Ultimately, the resulting outcomes are obtained by analyzing the displacements that occur between these two distinct pressures, which correspond to different phases of the cardiac cycle. A comprehensive elucidation of the methodology employed to achieve and measure this displacement is presented in Chapter 4.

For the achievement of this inflation process, several setup developments had to be performed in order to perform this experimental procedure, as will be further developed in the next subsection.

3.2.3.1 Setup Development

Figure 3.20 illustrates an initial challenge encountered with the setup, which involved the repeated breakage and degradation of the balloons during the fixing process. To address

this issue, plastic clamps were implemented as an alternative solution. The use of these clamps significantly reduced material degradation and enhanced the overall durability of the balloons.

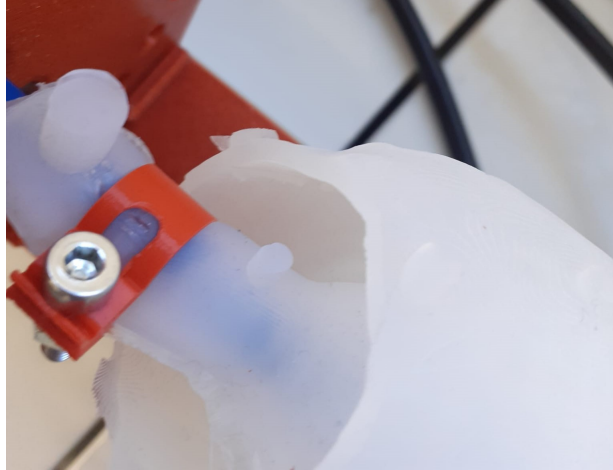


Figure 3.20: Degradation of the balloon in the setup.

An additional challenge emerged when inflating the balloon to 120 mmHg, the maximum pressure required for the experimental test. Figure 3.21 illustrates that the balloon would dislodge from the phantom and fail to reach 120 mmHg. To address this issue, appropriate measures were implemented to rectify the situation and ensure the reliable and accurate assessment of balloon displacement.



Figure 3.21: Dislodgement of the balloon from the phantom at a pressure below 120 mmHg.

Consequently, supplementary lateral supports were constructed to prevent the balloon from protruding beyond the phantom. Determining the optimal angle of the phantom with

the inserted balloon posed a considerable challenge. However, after multiple attempts, the final supports were successfully developed, as depicted in Figure 3.22.



Figure 3.22: Setup of of the balloon and the phantom with the supports.

Following the integration of these supports, the assembly was completed, enabling the inflation test to be performed. The entire testing process, including the use of DIC, will be laid out in Chapter 4, with the results reported in Chapter 7.

Experimental Procedure

The process of ascertaining surface deformation utilizing DIC methodologies typically encompasses a sequence of three consecutive phases [69]:

1. Speckle Pattern fabrication;
2. Image Acquisition;
3. Displacement and Strain field calculation.

Figure 4.1 depicts the experimental process of using a DIC methodology to obtain the displacement results.

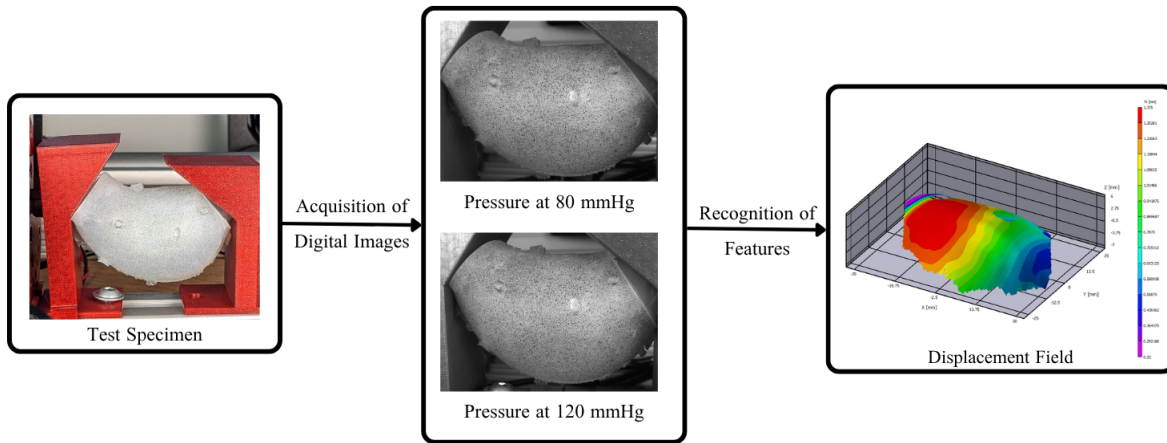


Figure 4.1: Workflow of DIC displacement measurement.

4.1 Speckle Pattern fabrication

In the context of Digital Image Correlation, the choice of an appropriate speckle pattern plays a vital role in reducing measurement noise and improving the accuracy of the obtained results [80].

DIC relies on a high-contrast pattern applied to the surface of the object or specimen being tested. A number of requirements and guidelines are suggested to ensure an appropriate speckle pattern, which are fully covered in Section 2.4 of the literature review.

Hence, the first step in the experimental procedure was to prepare the phantom. Given the advantages of an airbrush for producing a detailed and refined application of the speckle pattern, and the availability of an airbrush within the laboratory, this technique was chosen for speckle application (Figure 4.2) [74]. For this, the largest diameter needle was used, 0.5 mm, in order to apply this speckle.



Figure 4.2: Airbrush used for the speckle application.

One of the requirements for the project related to this dissertation was the utilization of water-based paint for the speckle pattern. This requirement emerged from the need to apply speckle patterns to the live fabric without causing any disruption to its pathological histology, allowing for the fabric to be easily washed and the speckle patterns to be subsequently removed. Due to this, water-based paints were regarded vital. As a result, initial attempts were made to add speckle using Titanlux water-based paints.

Knowing that the experimental procedure entailed the phantom's enlargement, it was decided to start by only using black paint for the speckle application. This decision was inspired by the concern that the balloon's expansion may result in a slight paint flaking, which would impair the speckle pattern. Furthermore, the phantom's manufacturing approach (FDM) resulted in a textured surface, and the addition of white paint could enhance this texture and result in a bad speckle pattern. Given that the phantom was white and minimally opaque, it was decided to just use black paint for the speckle application.

Given that the water-based paint used in the project was not expressly developed for airbrushing, its viability for achieving the desired speckle pattern was questioned. Hence, issues emerged while painting with this product. Since the paint was quite dense, it was difficult to get it out of the airbrush, which generally has a much more liquid paint. Water had to be added continuously in an attempt to resolve this problem. However, even after adding the

water, when tested on a sheet of paper, it did not produce any apparent speckles, producing instead a dense pattern, which is clearly not desirable. Despite these difficulties, an attempt was made to paint the phantom with this water-based paint, as observed in Figure 4.3.



Figure 4.3: Attempt to create a speckle pattern with Titanlux paint.

However, as can be noticed, using this paint no speckle could be clearly noticed, which would not happen with airbrush paint. Figure 4.4 shows the final painted phantom after attempting to add speckle patterns with this paint.



Figure 4.4: Result obtained from painting with the water-based paint.

Nevertheless, an attempt was made to use the DIC's "Focus" tool, which allows the detection of speckle patterns. Thus, the assumption was confirmed, no speckle being detected in the phantom, as can be seen from Figure 4.5.

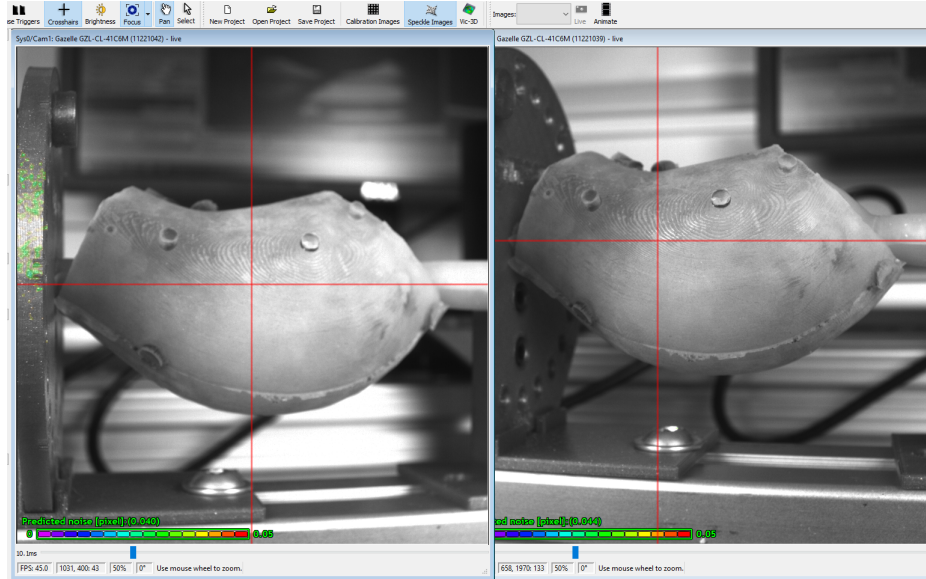


Figure 4.5: Speckle detection using a Correlated Solutions tool [81].

Following this unsuccessful attempt, another was made to create a speckle pattern, carried out by using paint intended for airbrushing, from Pro-Color. An iterative technique was used to create the speckle pattern to be employed for the phantom. On a sheet of paper, different kinds of speckle were tested. Afterwards, the cameras that would be used for analysis, Gazelle GZL-CL-41C6M-C from Point-Grey with 16 mm lenses were used to collect photos of the various patterns. The photos were then magnified to the point that individual pixels could be seen. A speckle dot was chosen from each pattern region and evaluated to determine if it came within the appropriate size range of 3-5 pixels. This procedure enabled the selection of the appropriate speckle pattern. This pattern was then recreated on the phantom surface by rapidly spraying the paint in sweeping motions around the object. Figure 4.6 depicts the resulting phantom with the new speckle pattern.

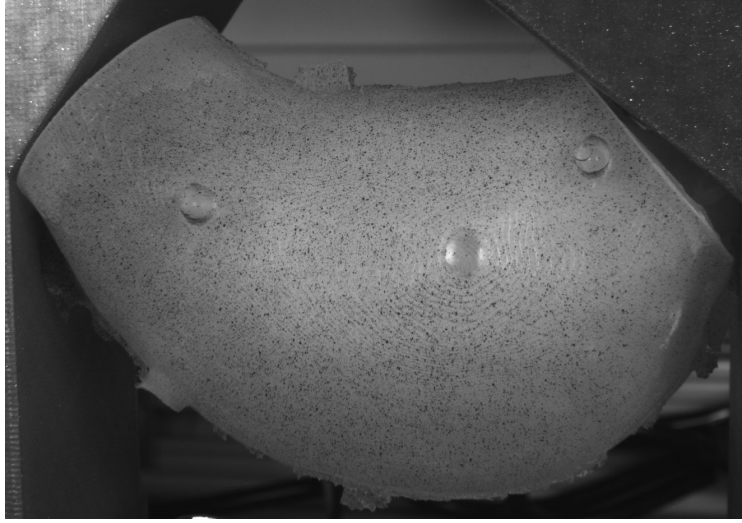


Figure 4.6: Speckle produced.

The created speckle pattern was assessed to determine if it was suitable for analysis via a Correlated Solutions tool. After verification, it became apparent that, whilst the speckle was not bad, it was not optimal. As a result of the pattern's visible wide spacing between points, it was determined that improving the speckle pattern would necessitate increasing the intensity of points, maintaining the same size.

Furthermore, the approach used for this objective was trial-and-error strategy, using the same speckle pattern that was tested on a paper sheet before being placed on the phantom. This way, an effectively optimized pattern was achieved.

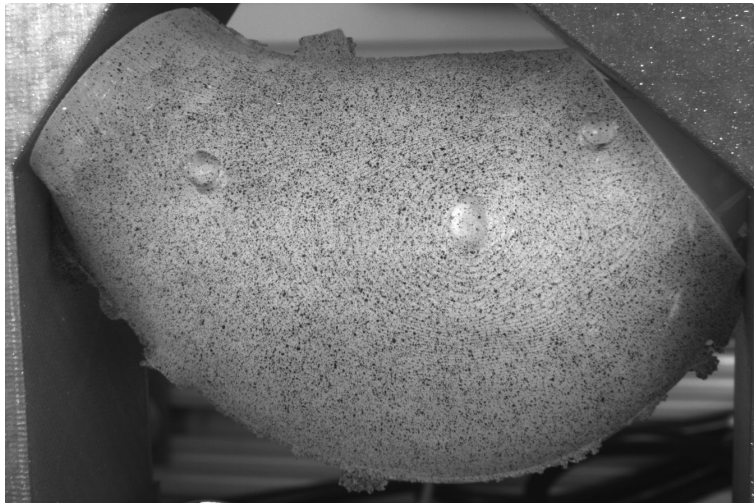


Figure 4.7: Final speckle pattern produced.

4.2 Image Acquisition

Images were recorded using cameras Gazelle GZL-CL-41C6M-C from Point-Grey, with 16 mm lenses (resolution: 2048 x 2048 pixels).

A calibration was necessary to initialize the spatial correlation processes of DIC [65]. For the present application, a stereo calibration was used, where where intrinsic and extrinsic parameters are calculated together. This way, the calibration target is imaged simultaneously in both cameras, and the synchronized target images are used to fully calibrate the system in one step [81].

Given the larger size of the phantom, a calibration target with the approximate size was chosen, witch was a target of 5.5 mm with a grid of 12x9, as can be seen in Figure 4.8.

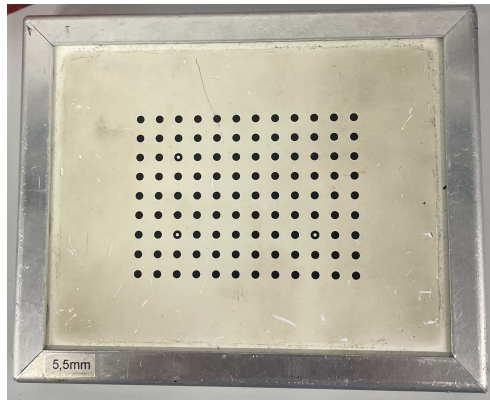


Figure 4.8: Target used for the calibration with 5.5 mm with a grid of 12x9.

For the setup, besides the lighting in the laboratory, additional luminaries were used. Figure 4.9 depicts the setup of the experimental procedure.

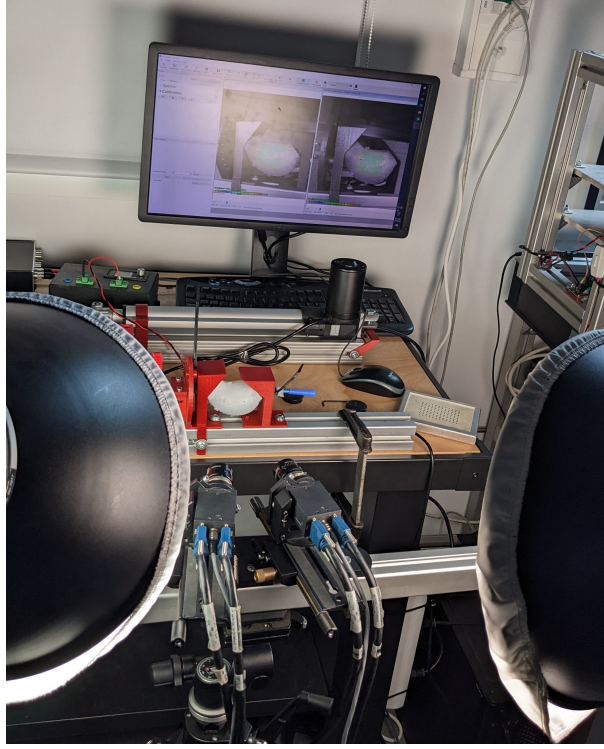


Figure 4.9: Setup for the experimental procedure.

After the calibration was performed, speckle images were subsequently taken. For these images, the setup described in Subsection 3.2.3 is put into operation, performing the test starting with a pressure of 80 mmHg and ending when 120 mmHg is reached. For this, pictures were taken using the DIC's time capture feature, thus obtaining the speckle images of the process, which will be used for the correlation.

The calibration parameters for the experimental procedure are depicted in Table 4.1.

Table 4.1: Calibration parameters

Parameter	Camera 1	Camera 2	Camera 1 $\rightarrow$ 2 transformation	
Center x (pixels)	1025.49	1006.91	Apha (degrees)	-0.574401
Center y (pixels)	1048.11	1015.92	Beta (degrees)	-21.7087
$f S_x$ (pixels)	3072.59	3066.78	Gamma (degrees)	1.16285
$f S_y$ (Pixels)	3072.59	3066.78	T_x (mm)	86.8931
Skem	0	0	T_y (mm)	0.905628
κ_1 normalized	-0.195294	-0.187099	T_z (mm)	19.0205
κ_1	0.200936	0.191915	Baseline (mm)	88.9551

4.3 Displacement calculation

Once the experimental procedure was completed, image processing began. As can be depicted by Figure 4.10, the Area Of Interest (AOI), was selected through the “polygon” tool, removing the zones that could give more errors through the “circle” tool, which were the mold orifice zones. Based on the recommendation from the software, a facet size of 53 pixel, which is calculated to give an optimal match confidence of 0.01 pixel for a given assumed noise level, and a facet step of 8 pixel were chosen based on the recommendation from the software.

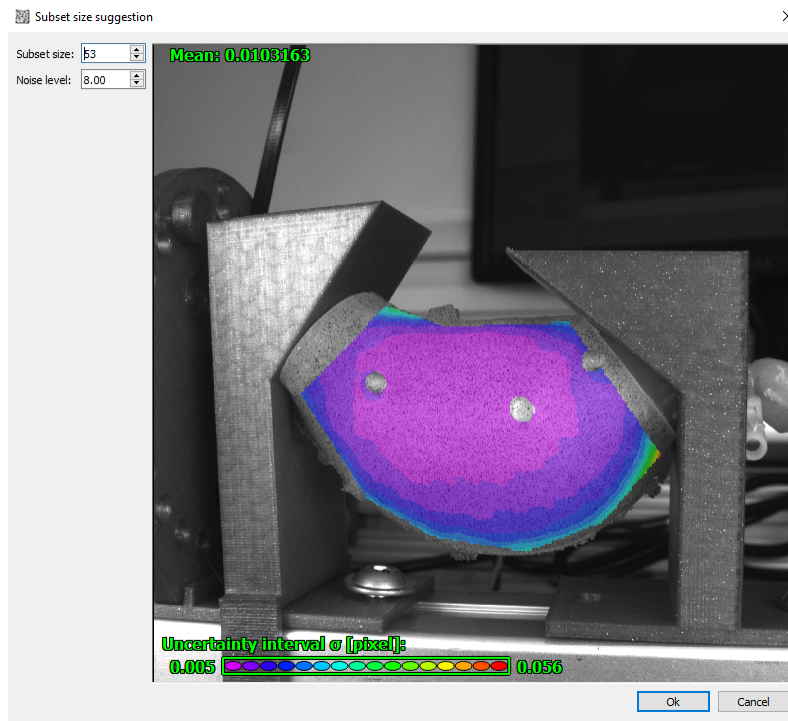


Figure 4.10: AOI with facet size and step recommendation.

After having this parameters defined, the correlation is run and the plot is obtained, which will be further talked about in subsequent chapters.

4.4 Experimental results

This section will discuss the results obtained through Digital Image Correlation employing the inflation setup. By configuring the inflation system to attain a pressure of 80 mmHg, which aligns with the diastolic arterial pressure, the experiment was initiated and

progressed until reaching the systolic arterial pressure of 120 mmHg. As a consequence, the ensuing results are derived from the assessment of the displacements occurring between these two distinct pressures that constitute the cardiac cycle.

4.4.1 Displacement in the X direction

Figure 4.11 and Figure 4.12 depict the displacements results in the X direction represented in both 3D and 2D plots.

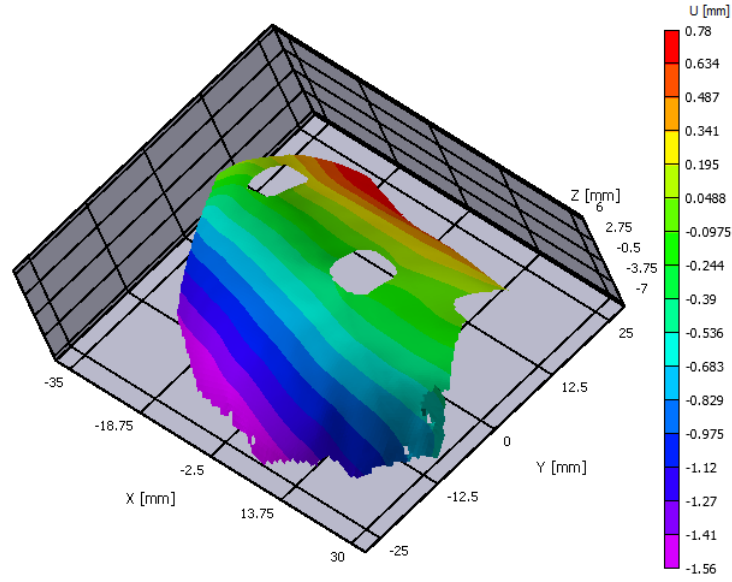


Figure 4.11: 3D plot for the displacement in the X direction.

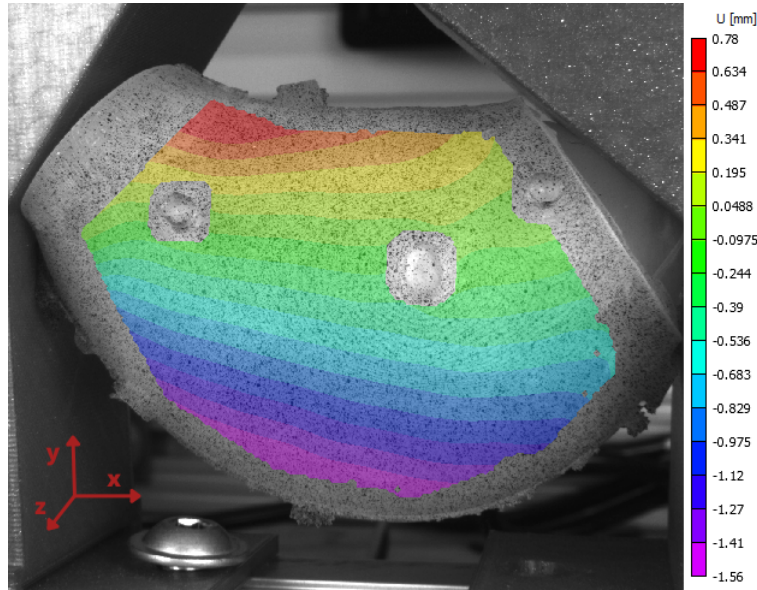


Figure 4.12: 2D plot for the displacement in the X direction.

When evaluating displacement in the X direction, a distinct clockwise movement was observed for the whole phantom. Notably, the lower region of the phantom had a larger negative displacement, reaching a maximum of 1.56 mm, indicating movement in that direction. The upper left side, on the other hand, exhibited a positive displacement reaching a maximum of 0.78 mm, indicating a displacement for the higher right side. Furthermore, the center area exhibited more neutral movement, indicating a relatively stable region.

The underlying mechanism of this rotational movement can be attributed to a combination of factors inherent to the experimental setup. The fixation mechanisms employed to restrain the balloon within the phantom played a crucial role. The left support effectively prevented any protrusion of the balloon, resulting in minimal displacement. However, the right support allowed some degree of movement from the balloon in the outside region, although doing its main objective which was keeping the balloon from completely exiting the phantom. This way, when the inflation process starts, and subsequently the balloon starts to expand, it exerts pressure on the phantom's tips, striving to expand beyond its boundaries. The stronger left-side restriction immediately limits the phantom's left tip, effectively locking its position. On the other hand, it causes the right tip of the phantom to be pushed downwards, resulting in a downward displacement.

4.4.2 Displacement in the Y direction

Regarding the Y direction, Figure 4.13 and Figure 4.14 depict the displacements for this direction in both 3D and 2D plots, respectively.

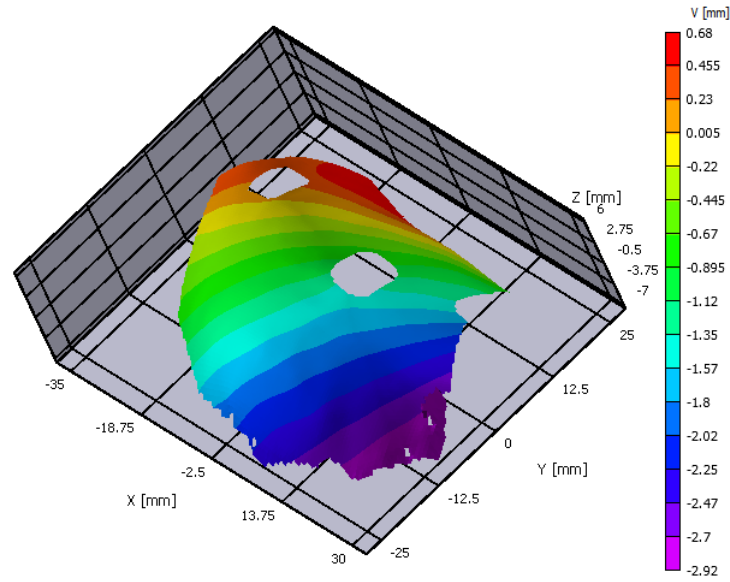


Figure 4.13: 3D plot for the displacement in the Y direction.

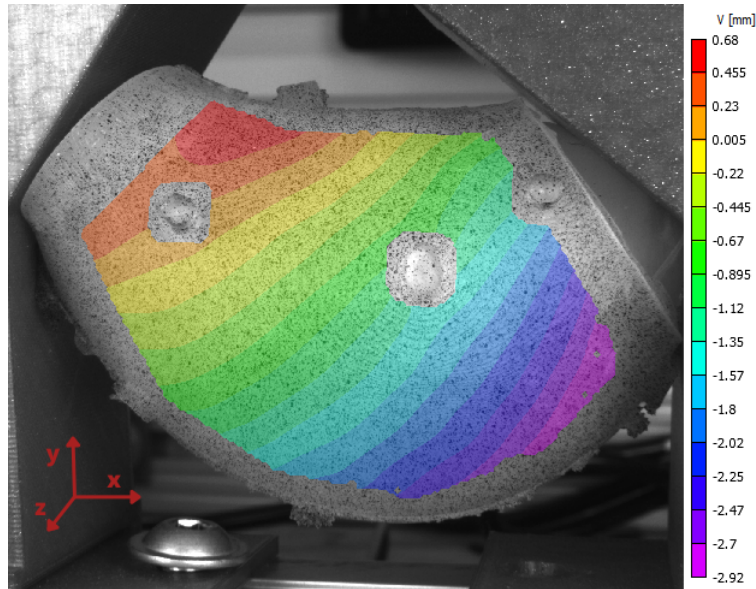


Figure 4.14: 2D plot for the displacement in the Y direction.

The displacement pattern in the Y direction shows similarities to that of the X direction, but with a noticeable difference in the position of the neutral zone, which is the zone where the displacement has the values closer to zero. In this case, the neutral zone appears to be shifted more towards the upper part of the phantom, in concordance with the upper tip of the right side of the phantom.

The similarity between the displacement patterns in the X and Y directions can be attributed to the consistent nature of the movement. The motion pattern exhibited a significant downward displacement in the lower right region of the phantom, with a maximum displacement magnitude of 2.92 mm. This indicates a prominent negative displacement within this specific zone, wherein the movement is directed downward and attains the aforementioned maximum value. The upper left side, on the other hand, shows a positive displacement, reaching a maximum of 0.68 mm, meaning that there is an upwards movement in this zone. This way, it can then be observed an upward movement for parts above the middle zone and a downward movement for regions below this zone, having a middle region with a more neutral displacement pattern. This is in agreement with the movement described in the X direction concerning the existence of the lateral supports.

Consequently, through the analysis of the displacement patterns acquired for both the X and Y axis, it becomes feasible to derive an estimation of the axis of rotation. This estimation is achieved by identifying the intersection of the two lines illustrated in Figure 4.15, indicative of the regions where displacement approaches null values. The resulting axis of rotation is in the perpendicular plane in this intersecting point, which actually intersects the tube in which the balloon and phantom are centrally placed. However, it is imperative to acknowledge that due to the substantial deformations experienced along these axis, particularly the Y axis, the derived axis can only be considered an approximation rather than an exact localization.

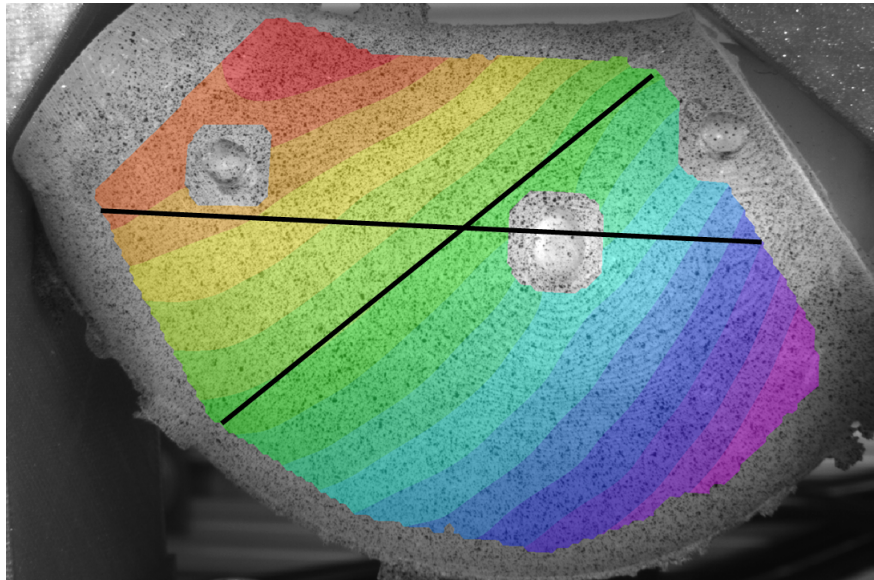


Figure 4.15: Estimation of the axis of rotation.

4.4.3 Displacement in the Z direction

Regarding the Z direction, Figure 4.16 and Figure 4.17 depict the displacements for this direction in both 3D and 2D plots, respectively.

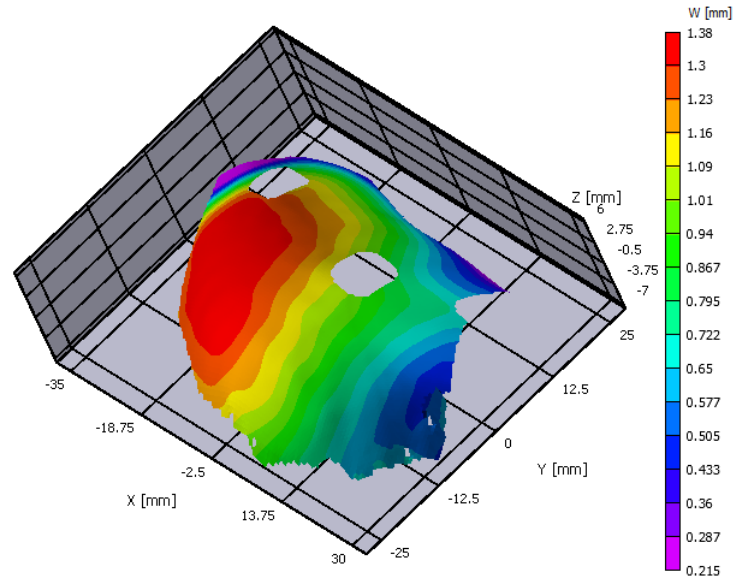


Figure 4.16: 3D plot for the displacement in the Z direction.

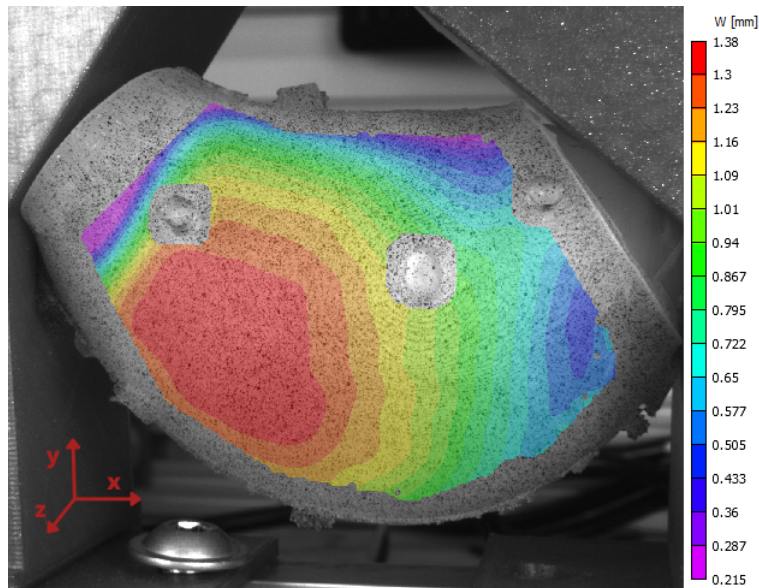


Figure 4.17: 2D plot for the displacement in the Z direction.

Discernible displacement patterns were identified in the Z direction, representative of

the radial orientation. Notably, the region showcasing the most pronounced displacement resides predominantly in the lower left quadrant, exhibiting a maximum displacement of 1.36 mm. Following this region, a discernible gradient of displacement emerges, gradually diminishing as moving farther away from that zone. This gradient manifests in a circular manner, tracing a trajectory that conforms to the shape of the phantom.

Seeing that in this experimental procedure a balloon is inflating the silicone phantom, the expected scenario in the displacement analysis in the radial direction would be an outward expansion of the phantom. As the balloon inflates, it exerts pressure on the silicone phantom, causing it to expand radially. This expansion will manifest as positive displacements in the Z direction, indicating an outward movement or bulging of the phantom. As can be observed by the analysis, only a positive displacement is shown, ranging from 0.215 mm to 1.38 mm.

As previously mentioned, the analysis of the experimental results revealed significant differences in displacement magnitudes along the Z axis, particularly in the left zone. This observation suggests the presence of a weaker surface in that specific area, potentially due to variations in the manufacturing process. When subjected to the pressure from the balloon, this weaker surface undergoes more substantial deformation, resulting in greater displacements. These findings highlight the non-uniform wall strength across the phantom, which contributes to the observed fluctuations in displacement along the Z axis. Further investigation into the manufacturing process and material properties will provide deeper insights into these observations.

Considering the inherent limitations in the fabrication process of the phantoms, it is reasonable to anticipate the occurrence of deviations and inconsistencies. The manufacturing of phantoms involves various steps and variables that may introduce uncertainties and imperfections. These can include factors such as material properties, molding techniques, and dimensional variations. As a result, it is expected that the phantoms may not possess a perfect level of precision or uniformity. Consequently, it is understandable to acknowledge the potential for deviations and variations in the characteristics and behavior of the phantoms, including their response to external forces like inflation.

4.4.4 Total displacement

Regarding the total displacement, Figure 4.18 and Figure 4.19 depict the magnitude of the displacements in both 3D and 2D plots, respectively.

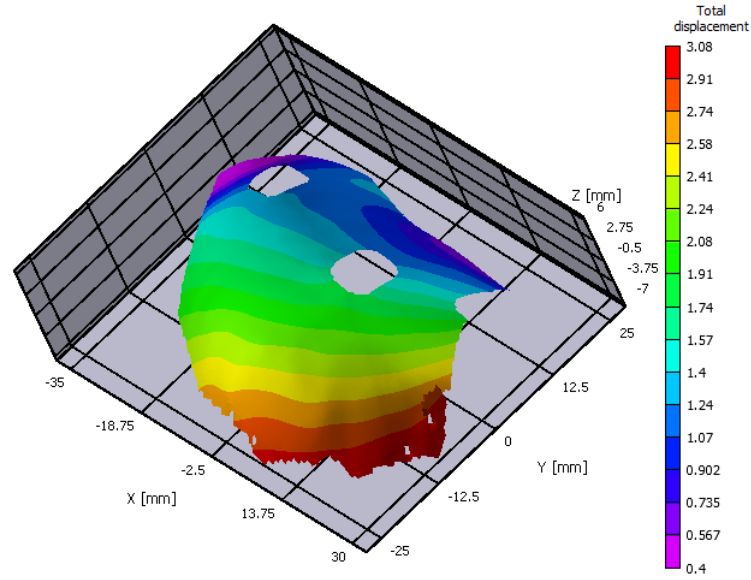


Figure 4.18: 3D plot for the total displacement.

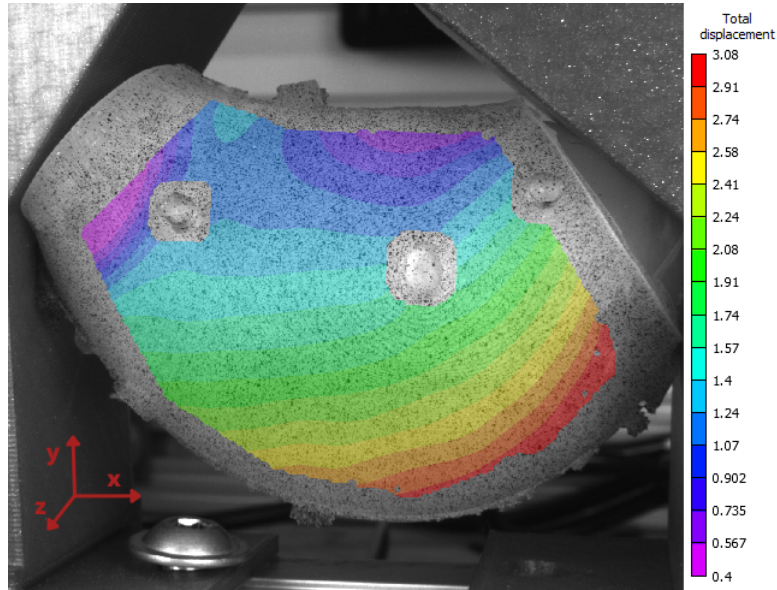


Figure 4.19: 2D plot for the total displacement.

The total displacement refers to the overall movement or change in position of the object of analysis. The largest displacement in this scenario is found in the lower right corner, correlating with higher displacements in both the X and Y directions. Given the influence of gradients on both the X and Y axis, a gradient pattern is evident. The smallest

displacement values are observed primarily along the boundaries, influenced by the lower values obtained in this zone in the Z direction.

Computational Simulation

Given the scarcity of comparable analyses in the current body of literature, computational simulation was employed to compare the obtained results.

The present chapter delineates the methodology employed for conducting the finite element method (FEM) simulation, alongside an in-depth examination of the geometry fitting process. Furthermore, the boundary conditions adopted in the study will be thoroughly explored, shedding light on their significance and implications. Afterwards, the post-processing necessary in order to obtain the intended results will be explained and, finally, the results will be presented and analysed.

5.1 Geometric model generation

This study's model is based on CT imaging of a real aorta. The model was originally obtained from a project partner who had used it in a different software, named SimVascular [91]. As a result, the model file could only be obtained in mesh format, which was compatible with this software.

This way, further refinement of the geometry was necessary to ensure optimal compatibility with the Abaqus software, enabling the acquisition of meaningful and accurate results. The details of this geometry treatment process will be discussed in subsequent sections.

Finally, it is important to acknowledge that due to the inherent limitations of computed tomography (CT), the specific cardiac cycle phase corresponding to the captured images could not be determined. Therefore, for the purposes of this study, the geometry was considered to be unloaded, since the objective was to test this geometry in the inflation setup by applying arterial pressures at diastole and systole and measure the consequent displacement.

The available CAD files consisted solely of mesh data without any defined thickness. Additional efforts were undertaken to associate geometric properties with the mesh and establish a uniform thickness of 2 mm [92]. However, due to the inherent complexity of the CT-derived aortic geometry, characterized by numerous irregularities, the process of obtaining a satisfactory representation proved challenging. Figure 5.1 illustrates the comparison between the provided initial geometry and the modified geometry incorporating the appropriate thickness.

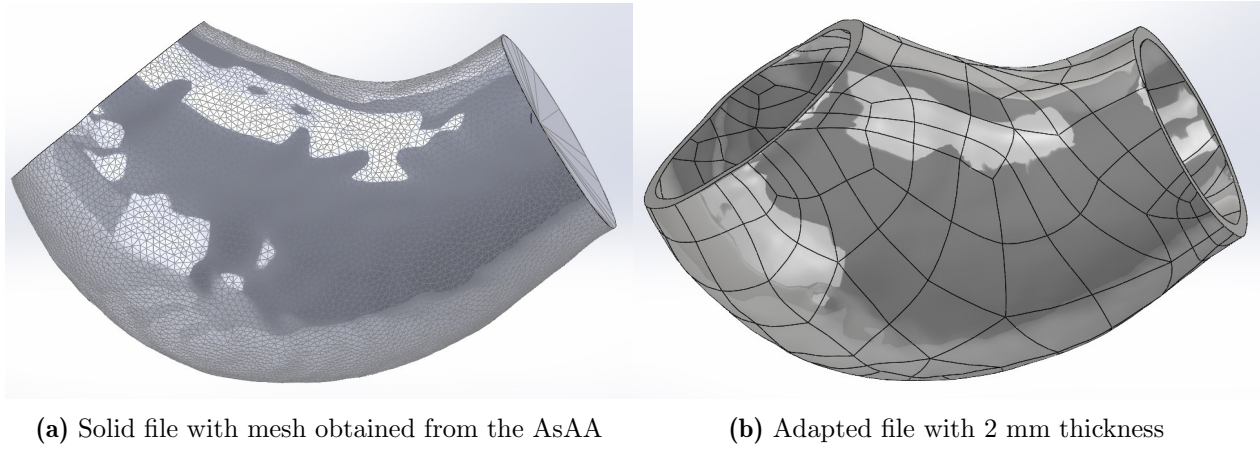


Figure 5.1: Comparison between the initial geometry provided and the modified geometry incorporating the appropriate thickness.

To simplify the handling and treatment of the mesh, the solution was to eliminate the existing mesh and generate a new one from scratch in Abaqus, importing only the geometry.

The imported file was integrated into the model as a part, revealing a zoning pattern inherited from the original mesh. To enhance the geometry's quality and ensure optimal meshing, a smoothing process was applied. This step aimed to refine the geometry and prepare it for subsequent mesh optimization, thereby facilitating the generation of accurate and reliable results. For this process, the Virtual Topology toolset from the mesh module was used. By utilizing the option "Combine faces" the intended result was obtained. Figure 5.2 illustrates the initial appearance of the part following its importation to Abaqus, as well as the ultimate outcome after the comprehensive smoothing process applied to the geometry. The depicted images provide a visual representation of the significant improvements achieved in refining the overall geometry.

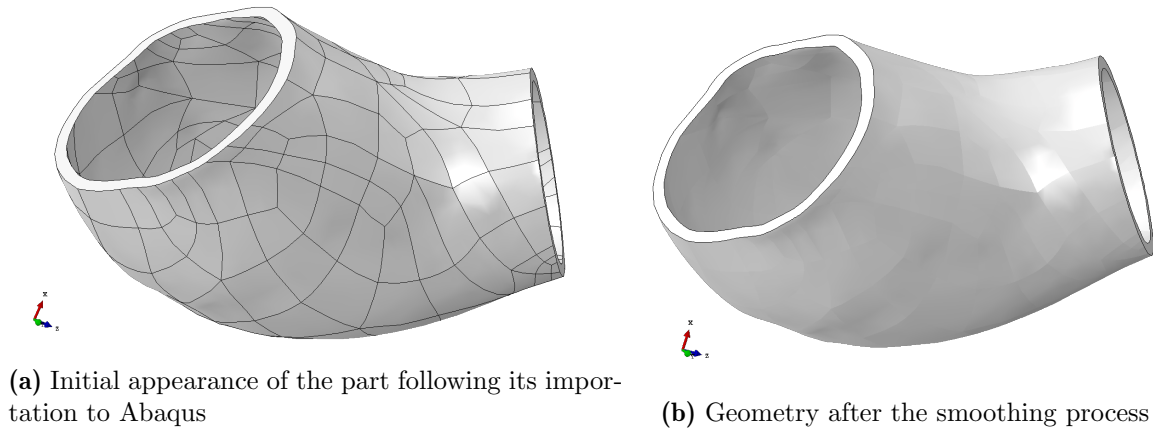


Figure 5.2: Refinement of the overall geometry.

5.2 Material Properties

The aortic structure, as previously discussed, is characterized as an incompressible, anisotropic material exhibiting hyperelastic properties. In light of the limitations of Hooke's law in this context, it becomes imperative to employ a suitable constitutive model that effectively captures the complex strain energy behavior of the material. Due to its ability to approximate the elastic behavior of the aorta, this model presents itself as a suitable option for characterizing the elastic behavior of the aortic structure in the current numerical investigation, given that the present simulation only considered one homogeneous wall to simulate the experimental procedure [91, 93, 64]. The Neo-Hookean model is a hyperelastic material model which can be used for predicting stress-strain behavior of materials undergoing large deformations [64]. This model also assumes linear elastic, isotropic behavior and compressibility, having the strain energy density equation given by Equation 5.1,

$$U = \frac{\mu_s}{2} (I_1 - 3) + \frac{k}{2} (J-1)^2 \quad (5.1)$$

where μ_s and k represent the shear and bulk modulus, respectively, I_1 represents the first invariant of the right Cauchy-Green strain tensor and J represents the Jacobian determinant of the deformation gradient.

The material properties used in this study were derived from a previous investigation that conducted fluid-structure interaction (FSI) simulations to examine the deformation of the cardiac muscle and the behavior of vessel walls under blood pressure. Table 5.1 presents the properties obtained from this study, which were the ones selected to be used in the present simulation [93].

Table 5.1: Mechanical parameters of the aorta [93]

Parameter	Unit	Value
Density (ρ)	kg/m ³	960
Shear Modulus (μ_s)	N/m ²	62.04×10 ⁵
Bulk Modulus (κ)	N/m ²	124×10 ⁻⁵

Regarding the Neo-Hooke model in Abaqus, the strain energy function is represented by the Equation 5.2 [94]:

$$U = C_{10} (I_1 - 3) + \frac{1}{D_1} (J_{el}-1)^2 \quad (5.2)$$

In order to utilize the shear and bulk modulus values in the context of the Abaqus software, a conversion process was undertaken to obtain the corresponding C_{10} and D_1 constants. This values can be depicted in Table 5.2, which specific shows the values employed in Abaqus density and hyperelasticity parameters.

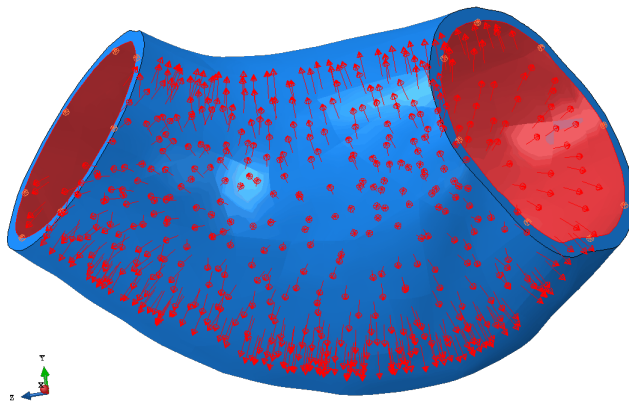
Table 5.2: Values used in the simulation for the Neo-Hooke model [93]

Material Behaviour	Parameters	Values	Unit
Density	ρ	960×10^{-12}	t/mm ³
Hyperelastic	C_{10}	3.102	N/mm ²
	D_1	0.16129	mm ² /N

While the natural structure of the aorta consists of three distinct layers, as discussed in Chapter 2, the focus of this study revolves around comparing a single-layer material phantom. Hence, in the conducted simulation, only one layer is considered to closely align with the experimental setup, aiming to maintain fidelity with the experimental counterpart. Thereby, the previously presented properties were employed in this study, taking into account a single-layer configuration.

5.3 Loads and Boundary Conditions

The accuracy of the results of the numerical study is dependent on the boundary conditions. The objective of the simulation was to replicate, to the greatest extent possible, the conditions observed in the experimental procedure using a static simulation approach. In the experimental setup, the balloon pressure was incrementally raised until it reached the diastolic pressure, allowing for the analysis of displacement from the diastolic pressure (80 mmHg) to the systolic pressure (120 mmHg). Consequently, the load conditions involved the application of a uniform pressure on the inner surface of the aorta, specifically referred to as the luminal surface, as can be observed in Figure 5.3. This condition aimed to mimic the physiological environment and provide insights into the mechanical response of the aorta under varying pressure conditions.

**Figure 5.3:** Application of the pressure in the aorta.

The experimental procedure involved the application of two distinct loads in separate stages. In the initial step (Step-Diastole), a uniform pressure of 80 mmHg, equivalent to 0.0107 MPa in accordance with the units used in Abaqus, was applied. This pressure was then propagated to the subsequent step (Step-Systole), where a new pressure of 120 mmHg, corresponding to 0.016 MPa, was applied. By applying these sequential loads, the study aimed to assess the mechanical response of the system under different pressure conditions and gain insights into its behavior and performance. Table 5.3 showcases the applied loads and their distribution across the simulation.

Table 5.3: Applied loads and their distribution

	Load [MPa]	Load [mmHg]	Step-Diastole	Step-Systole
Diastolic Pressure	0.0107	80	Created	Propagated
Systolic Pressure	0.016	120		Created

Furthermore, three distinct boundary conditions were implemented to emulate the experimental conditions. Firstly, transversal motion at the corners of the aorta was blocked, similar to the experimental process in which support systems were employed on the sides of the phantom to confine the balloon within it and restrict any outward displacement. For each of these boundary conditions, a new coordinate system was created, in order to better define the orientation of the specific boundary condition, given that the geometry was very irregular. In Figure 5.4 the two coordinate systems created can be observed, CSYS-BC1 and CSYS-2, which are used for each of the boundary conditions, BC-1 and BC-2, respectively.

However, it was not feasible to fully replicate the central attachment provided by the presence of the balloon in the aorta. As an alternative solution, a partition was introduced at the top to impede movement in the Z direction, assuming then that the phantom was radially fixed, as depicted in the Figure 5.4 as the BC-3. Considering the inherent challenges associated with handling this specific geometry, this configuration constituted the most accurate attempt at recreating the conditions observed in the experimental setting. For this boundary condition, the same coordinate system as in the boundary condition 2 (BC-2) is used (CSYS-2), as can be seen in Figure 5.4.

However, it should be noted that in the presence of overlapping regions characterized by different coordinate systems, a unique scenario arises. This is particularly evident in the upper region where a partition coincides with the side faces. Consequently, the application of boundary conditions 1 and 2 had to be modified accordingly, given that the job did not submit this way. These boundary conditions were restricted to the section corresponding to the inner diameter, as visually depicted in the accompanying Figure 5.4.

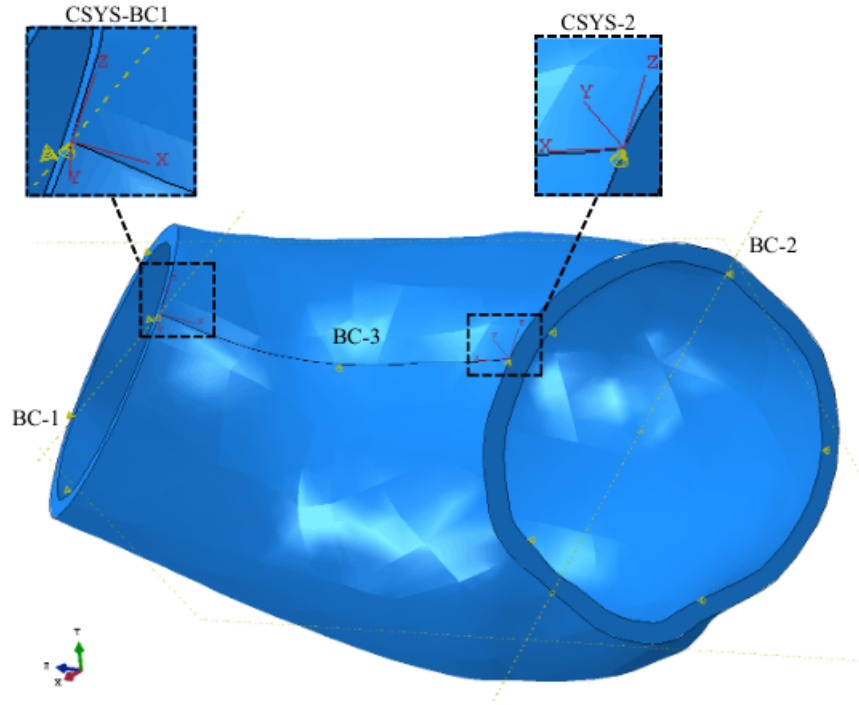


Figure 5.4: Boundary conditions applied and coordinate systems created.

Table 5.4 showcases the applied boundary condition and their distribution across the simulation.

Table 5.4: Applied boundary conditions and their distribution

	Condition	Initial	Step-Diastole	Step-Systole
BC-1	$U_1=0$	Created	Propagated	Propagated
BC-2	$U_1=0$	Created	Propagated	Propagated
BC-3	$U_3=0$	Created	Propagated	Propagated

5.4 Mesh generation

In cases where the geometry is irregular, it can be challenging to generate a hexahedral mesh that conforms accurately to the geometry. Tetrahedral elements, such as C3D4 or C3D10, on the other hand, are more versatile and can better handle complex and irregular geometries [95].

The utilization of C3D10 elements for the meshing of the ascending aorta model was a deliberate choice driven by the need for accurate representation of its geometric complexity and mechanical behavior. These elements, also known as quadratic tetrahedral elements,

offer a higher-order interpolation of displacement and strain fields compared to their linear counterparts. This enables a more precise description of the intricate shape and deformation patterns within the aorta. By employing C3D10 elements, the mesh can capture finer details and localized effects, resulting in improved spatial resolution and accuracy of the simulation. However, it is important to note that C3D10 elements are computationally more demanding compared to other element types. The increased complexity and computational requirements arise from the higher number of nodes and degrees of freedom associated with these elements. As a result, simulations involving C3D10 elements may require additional computational resources, such as processing power and memory, to ensure efficient and accurate calculations. However, given the computational resources provided their usage did not pose any significant challenges.

5.5 Post-Processing

To align the computational analysis with the experimental part, post-processing steps were carried out. After the successful completion of the simulation job, a new coordinate system was introduced, named CSYS-1. The reference coordinate system was established by identifying a point situated at the midpoint between the line formed by the midpoints of each phantom tip. Subsequently, meticulous efforts were undertaken to align the directions as closely as possible to the experimental procedure. Figure 5.5 illustrates the position and orientation of each axis in the modified reference coordinate system.

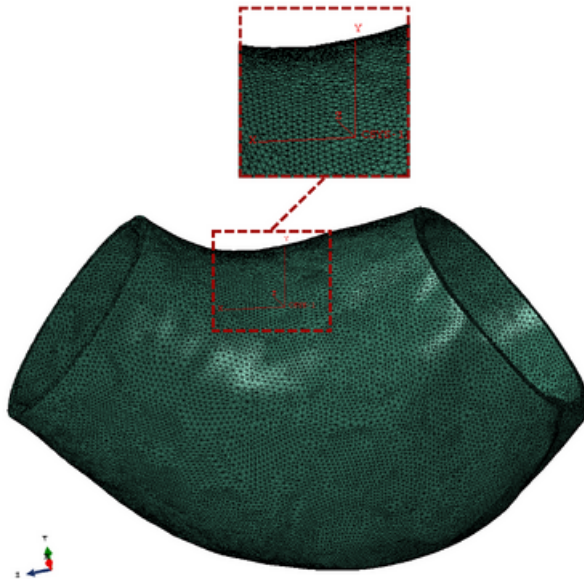


Figure 5.5: Reference coordinate system used in the post-processing step.

In addition to the aforementioned simulation process, a field output was introduced to measure the difference in displacement between the two pressures applied: diastolic blood pressure (80 mmHg), applied in the Step-Diastolic and systolic blood pressure (120 mmHg), applied in the Step-Systolic.

To generate field output data for further analysis and post-processing, the "Create from Frames" command in Abaqus was used. This command, accessible via the "Tools" menu, enables the generation of field output data based on existing frame data within the model.

By selecting the "Create from Frames" function, as well as Step-Diastolic and Step-Systolic in their concluding increments, and assigning a negative value of "-1" for the scale factor for Step-Diastole, the computation of displacement differences was enabled. Through this process, the desired result was successfully attained. Figure 5.6 depicts the described process. In this interface, the Step-Diastolic is designated Step 1 and the Step-Systolic by Step 2.

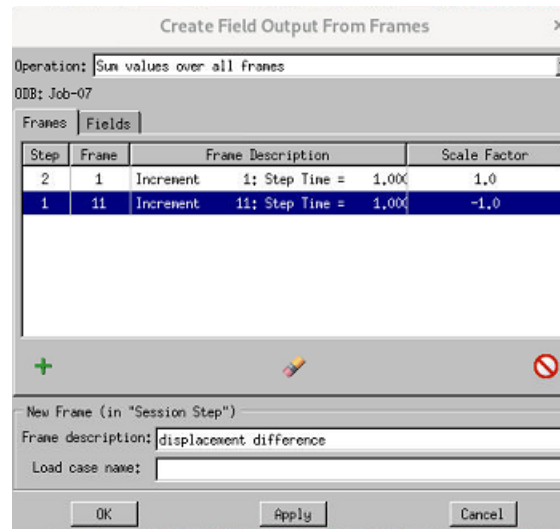


Figure 5.6: Process of creating the Field Output from the desired frames in order to observe the difference in displacement between steps.

By accessing the "Results" tab, selecting "Step/Frame" and choosing the "Session Step" option, it was possible to access the corresponding results (Figure 5.7).

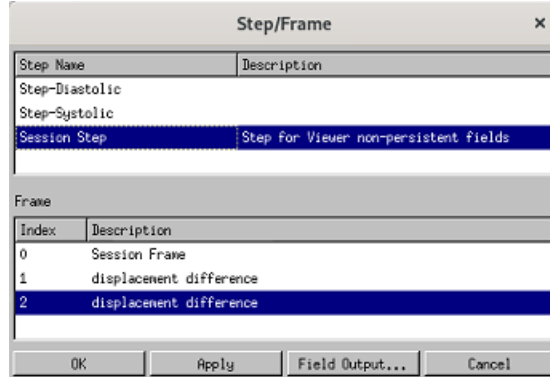


Figure 5.7: Process of choosing the desired step in order to obtain the final results.

The difference in displacements was then calculated and the obtained results will be presented subsequently.

5.6 Simulation Results

The primary focus of this study revolves around the analysis of displacements though the comparison of geometric variations between the systolic and diastolic states. Consequently, the displacements between these two conditions were computed. To ensure compatibility with the experimental investigation, the post-processing techniques outlined in Section 5.5 were employed, enabling the generation of comparable results.

5.6.1 Displacement in the X direction

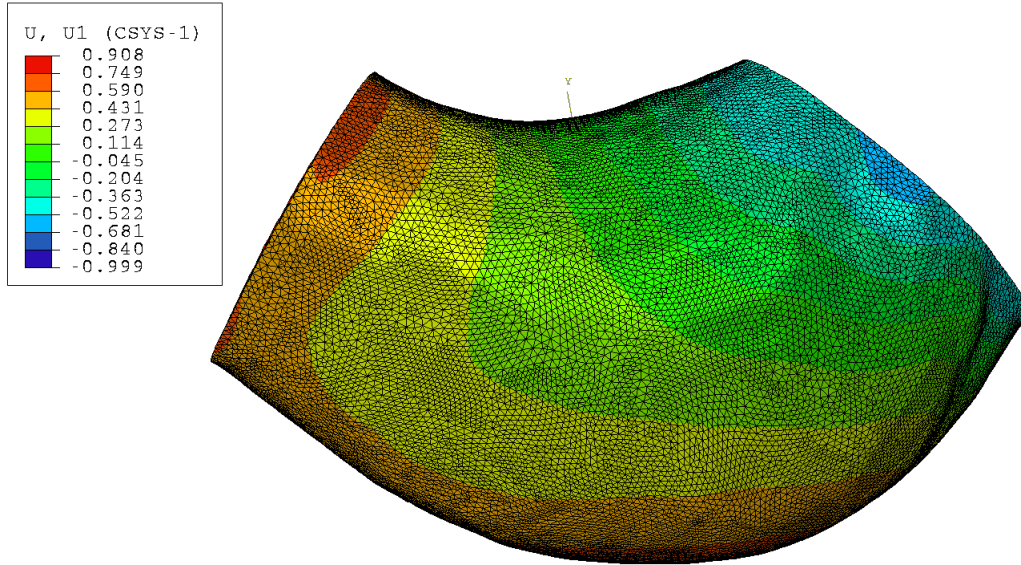


Figure 5.8: Displacement in the X direction.

Upon analyzing the displacement pattern shown in Figure 5.8, several observations can be made. Notably, areas characterized by the most significant displacements are concentrated at the tip of the phantom in the upper left corner and in the corresponding lower left corner, as well as at the base of the phantom, exhibiting the largest magnitude of 0.908 mm. Beyond these regions, a discernible gradient of displacement emerges, progressively diminishing until it reaches a specific zone situated in the upper left quadrant of the phantom, where displacement effectively reaches zero. Subsequent to this zone, displacement undergoes a shift, recommencing its increase in the negative direction. Notably, the displacement attains a magnitude of approximately 0.681 mm at the central region of the right corner.

The observed displacement patterns in the phantom can be attributed to a combination of distinct boundary conditions and geometric considerations. The boundary condition of retention applied in the Z direction within the upper region of the phantom initiates a downward movement, influencing the larger displacements observed in the lower zone. This substantial displacement in the lower region subsequently results in a negative displacement of the right tip in the X direction, directed towards the right side. Similarly, the left tip exhibits a leftward movement, towards the positive side. These movements likely arise from the boundary condition imposed on the inner diameter, which restricts movement in the X direction during displacement. This restriction effectively fixes the tips of the phantom and induces a counteracting movement to oppose the displacement. As a result, the tips exhibit a greater displacement due to the opposing forces. Furthermore, the presence of specific zones

with larger displacements at the tips suggests the existence of thickness variations in those regions, which may arise from the geometric treatment applied to the phantom.

In conclusion, the observed displacement patterns in the phantom demonstrate the significant influence of the applied boundary conditions. The imposed constraints on the upper region induce specific downward movements, which subsequently affect the displacements in the lower and tip regions. Furthermore, the fixed boundary condition applied to the inner diameter results in compensatory movements to counteract the displacements.

5.6.2 Displacement in the Y direction

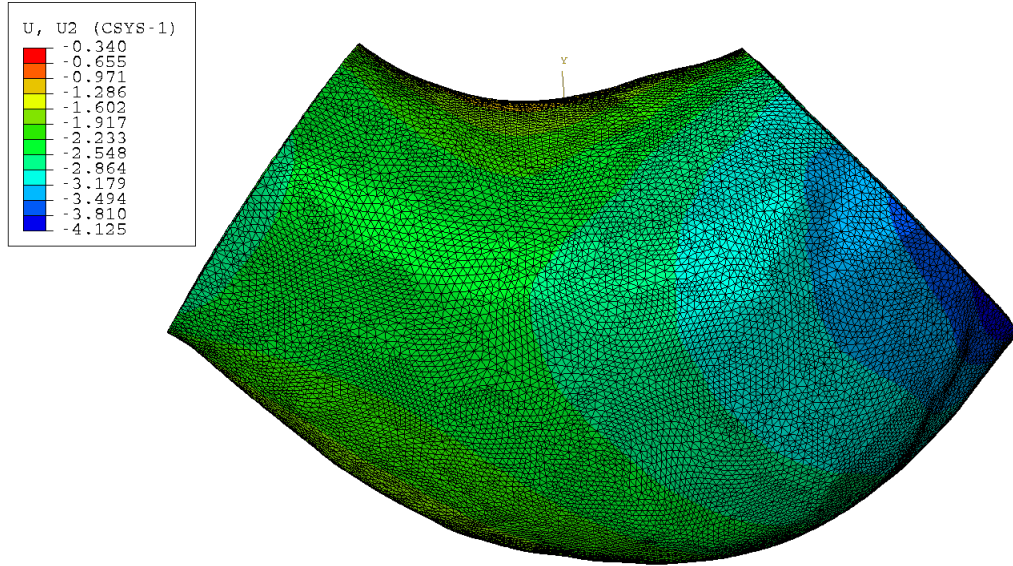


Figure 5.9: Displacement in the Y direction.

Upon analyzing the results, presented in Figure 5.9, a distinct pattern emerges along the lower right side of the phantom tip, displaying a gradient of displacements. The maximum displacement values are observed in this region, reaching up to 4.125 mm. As one moves away from this zone towards the lower left quadrant, a gradual decrease in displacements is observed, characterized by displacements within the yellow range, with a minimum value of 1.602 mm, in both the upper and lower zones of the aorta. However, it is important to note that there is a notable exception to this gradient in the center of the tip on the left side. In this specific area, displacements are significantly higher, reaching up to 3.494 mm. Consequently, this particular zone exhibits greater displacements compared to the surrounding regions on the left side.

Furthermore, a noteworthy observation pertains to the displacement values in the Y direction, which consistently exhibits negative values. This implies that the phantom exclusively undergoes downward movement, aligning with the boundary condition enforced in the upper zone to restrict motion in a specific direction.

Initially, given the whole process of adapting the geometry of the aorta (Section 5.1), when inserting it into Abaqus, it was alerted that it presented geometric problems at the tips. Thus, in the Y direction, slightly larger displacement is observed in the middle portion of the left tip, which wasn't expected initially. This may have occurred due to the existence of these geometric errors that, likely coupled with an in-homogeneous geometry, as can be seen in the previous analysis in turn caused stress concentration effects, thus presenting a larger displacement in this area of the tips. Notably, Figure 5.10 demonstrates that the zones with greater displacement align with the areas with the biggest stress distribution.

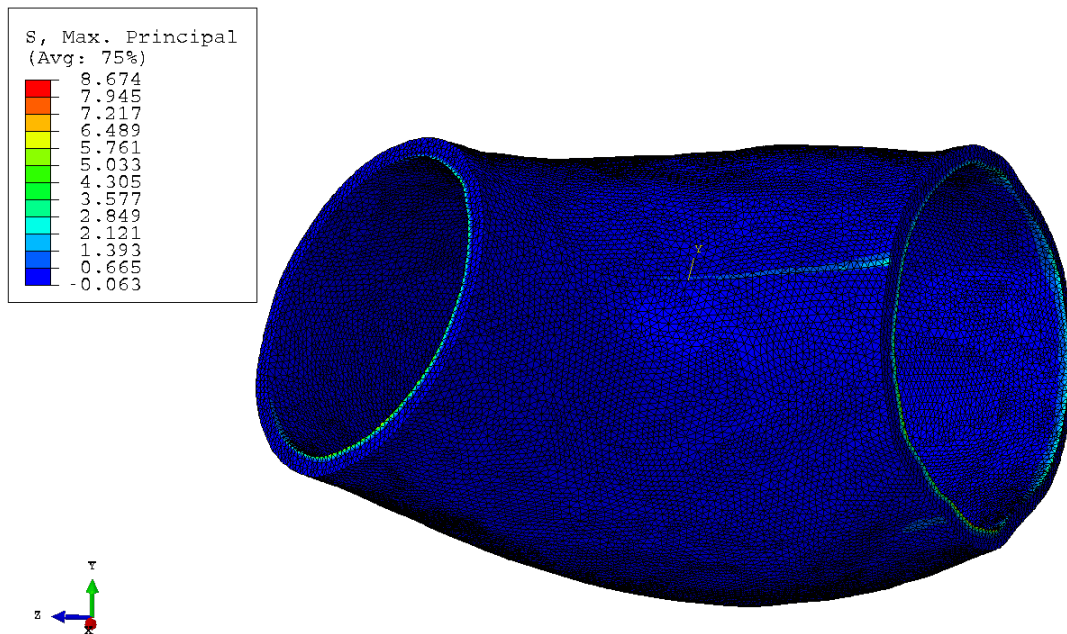


Figure 5.10: Stress distribution in the simulation due to the geometry.

5.6.3 Displacement in the Z direction

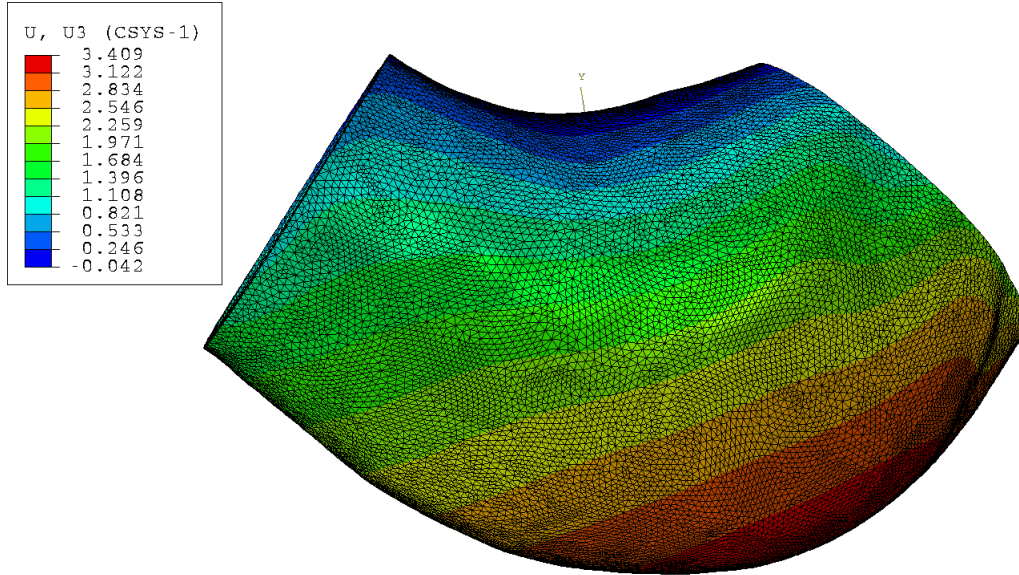


Figure 5.11: Displacement in the Z direction.

Upon examining the distribution of displacements, represented in Figure 5.11, a distinct gradient becomes evident, characterized by minimal displacements in the upper portion of the ascending aorta gradually increasing towards the right corner. Eventually, the maximum displacements are observed in this specific location, reaching an impressive value of 3.409 mm.

The displacement being unambiguously negligible in the upper part of the phantom was already expected due to the imposition of a zero displacement boundary condition in this direction. This condition was included to approximate with the experimental requirements and allow the simulation to run successfully. More information on this boundary condition can be found in the previously stated Section 5.3.

As you move towards the inferior part of the model, where the diameter is larger, no such constraint is applied. This allows for greater freedom of movement in the Z direction. As a result, the displacement can increase gradually as you move towards the region with the larger diameter in the radial direction. The absence of the Z direction constraint in this region allows the material to deform and displace more freely, resulting in higher displacement values.

5.6.4 Displacement Magnitude

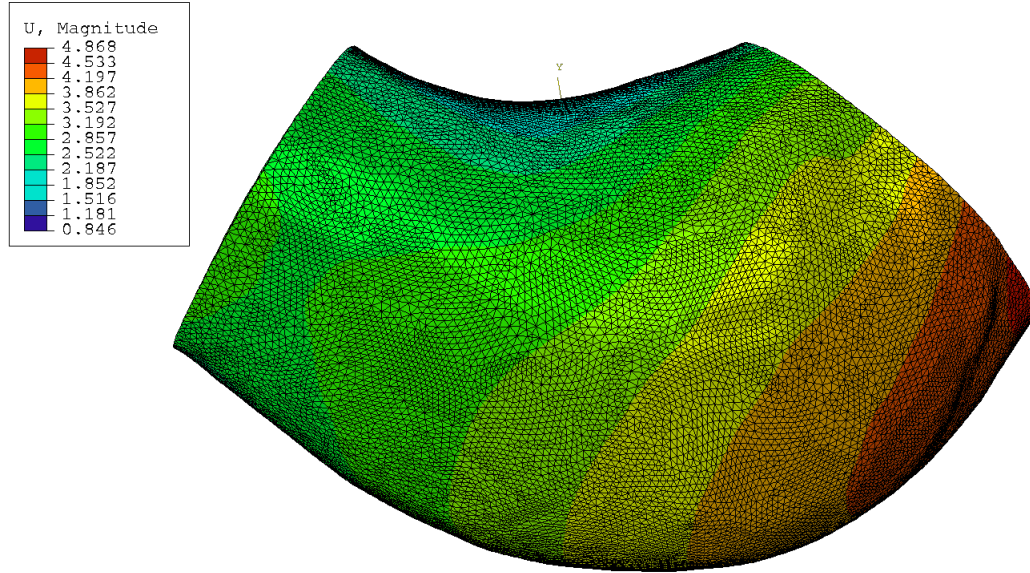


Figure 5.12: Magnitude of the displacement.

The magnitude of displacement, also known as total displacement, represents the object's overall movement or positional change.

On the right side of the ascending aorta, visible in Figure 5.12, visible a noticeable trend of greater magnitude of displacement is evident, with a gradient that gradually reduces towards the tip on the left side. This pattern is predominantly driven by the Y and Z displacements, which are higher in this zone, measuring 4.125 mm and 3.409 mm, respectively.

The upper portion of the phantom, on the other hand, has a considerably reduced displacement magnitude, owing mostly to the enforced boundary constraint in the Z direction to restrict displacement. Without the presence of this discrete zone of reduced displacement, it is likely that merely a gradient of displacement would be detected.

Bulge Inflation test

Given the absence of literature on comparable experimental conditions and considering the disparities observed between the experimental and computer simulation results, a subsequent test was conducted. The objective was to enable a more comprehensive comparison between these two methodologies and ascertain whether they yield convergent outcomes under simplified conditions.

The test conducted in the present chapter is known as the bulge inflation test, as previously mentioned in Section 2.3. This test entails securing a square-shaped specimen onto an inflation device, which forms a hermetically sealed cavity. Subsequently, a controlled release of fluid, typically water, is initiated at a predetermined rate [14, 15]. However, in the current investigation, a departure from the conventional approach was made by employing air instead. This adjustment was made to maintain consistency with the previous test and to ensure the utilization of the same experimental setup with minimal modifications. Consequently, the following description will outline the setup employed for this test, along with the specific conditions implemented to ensure accurate and reliable results setup.

6.1 Experimental apparatus

Figure 6.1 depicts the mechanism in which the silicone sample will be inserted for the inflation test. By the analysis of this mechanism, a perforation is visible, through which a pneumatic tube is meant to be connected to aid the inflation. A circular groove is also present, deliberately placed to ensure consistent dispersion of the incoming pressure across the entire phantom. A rectangular slit is also visible, which serves as the designated entry point for the material with the specified thickness. Following insertion, the visible screws illustrated in the following image secure the slot. These structural features help to provide the precise and controlled testing environment required for reliable measurement and analysis.

The material employed in this test was the same used for the previous inflation test, HB FLEX 5540 A+B. The used sample was taken from a phantom, ensuring a standardized thickness of 2 mm throughout the specimen.



Figure 6.1: Mechanism used to secure the sample for the experimental procedure.

Figure 6.2 exhibits the apparatus employed for the bulge inflation test, with the silicone sample securely fixed within the mechanism. At the rear portion of the mechanism, the connection area where the pneumatic tube is inserted can be observed. To ensure a tight seal and prevent any air leaks, silicone material was placed around the pneumatic tube, guaranteeing a controlled and precise inflation process. It is worth noting that the pneumatic tube, which was originally utilized for the previous inflation setup, was modified and repurposed to suit the requirements of the present test.

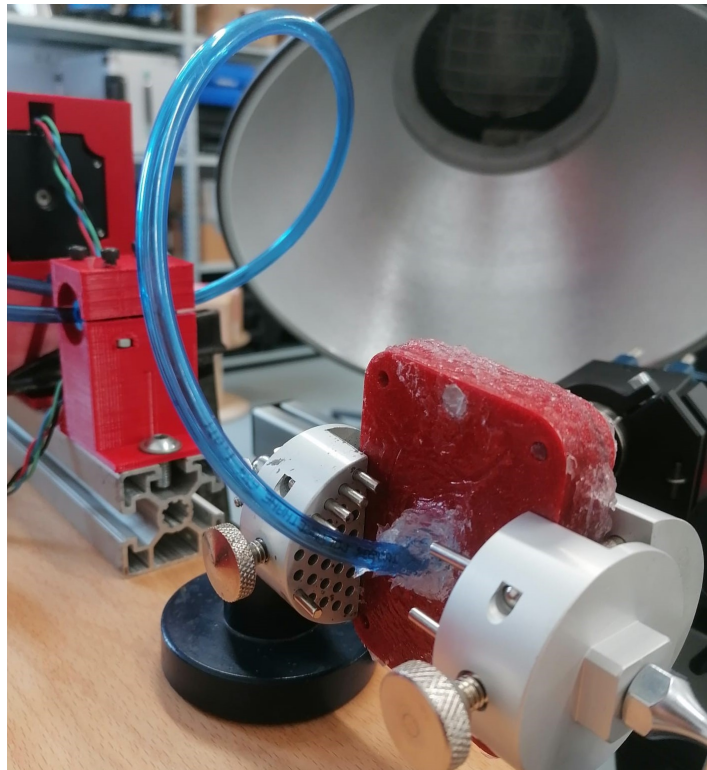


Figure 6.2: Apparatus for the experimental procedure.

6.2 Experimental Procedure

In order to achieve the displacement results for the experimental analysis, a series of steps had to be ensured, as depicted in Figure 6.3.

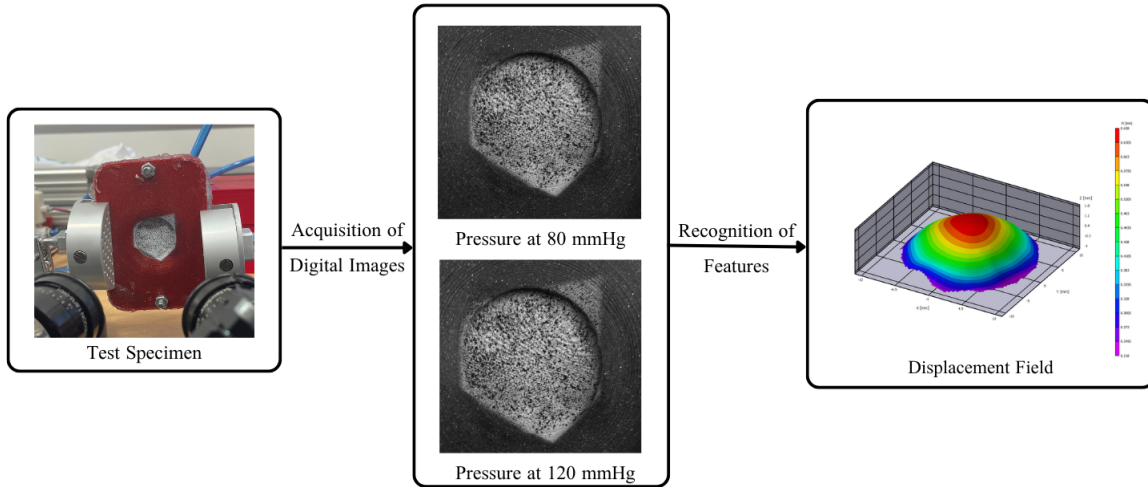


Figure 6.3: Workflow of the DIC procedure for the experimental test.

The speckle pattern was created by employing the same conditions as described in Section 4.1. The speckle pattern was applied employing an airbrush with a 0.5 mm needle size, alongside airbrush paint. However, the main difference in this case was the usage of a white background, as the observation area was smaller and the estimated displacements were smaller than in the previous analysis. This modification improved visibility and contrast for reliable displacement measurements. In Figure 6.4, the final speckle pattern obtained can be observed.

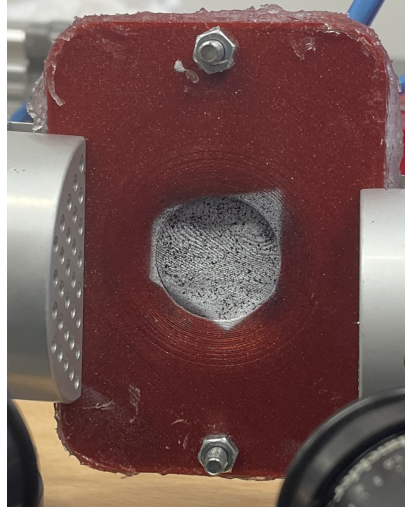


Figure 6.4: Speckle pattern of the silicone sample.

Regarding the image acquisition, as in the previous experimental test (Section 4.2), images were recorded using Gazelle GZL-CL-41C6M-C cameras from Point-Grey with 16 mm lenses (resolution: 2048 x 2048 pixels). The camera parameters can be observed in Table 6.1.

Table 6.1: Calibration parameters

Parameter	Camera 1	Camera 2	Camera 1 $\rightarrow$ 2 transformation	
Center x (pixels)	1024.29	1002.87	Apha (degrees)	-0.405956
Center y (pixels)	1033.58	1021.14	Beta (degrees)	-27.5547
$f S_x$ (pixels)	3153.04	3141.52	Gamma (degrees)	1.09855
$f S_y$ (Pixels)	3153.04	3141.52	T_x (mm)	79.2695
Skem	0	0	T_y (mm)	0.828355
κ_1 normalized	-0.204804	-0.216532	T_z (mm)	15.7135
κ_1	0.253511	0.245802	Baseline (mm)	80.8161

Given the smaller size of the specimen used for the test, a calibration target of 4 mm with a 12x9 grid was used, as can be seen in Figure 6.6.

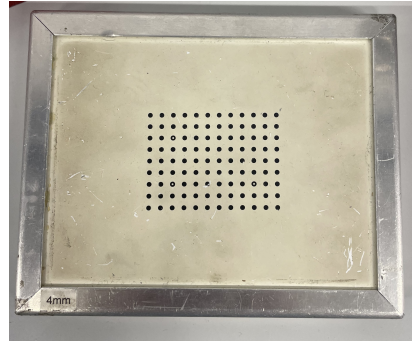


Figure 6.5: Target of 4 mm with a 12x9 grid used for the calibration.

After successfully completing the calibration process, the actual test was initiated. The pressure within the system was gradually increased from a diastolic blood pressure of 80 mmHg, and the accompanying displacements were recorded until reaching a systolic pressure of 120 mmHg, the moment the procedure was terminated. Afterwards, the Area of Interest was carefully chosen to encompass the entirety of the region of observation, choosing the recommended facet size of 43 pixel and a facet step of 8 pixel, as can be observed in Figure 6.6.

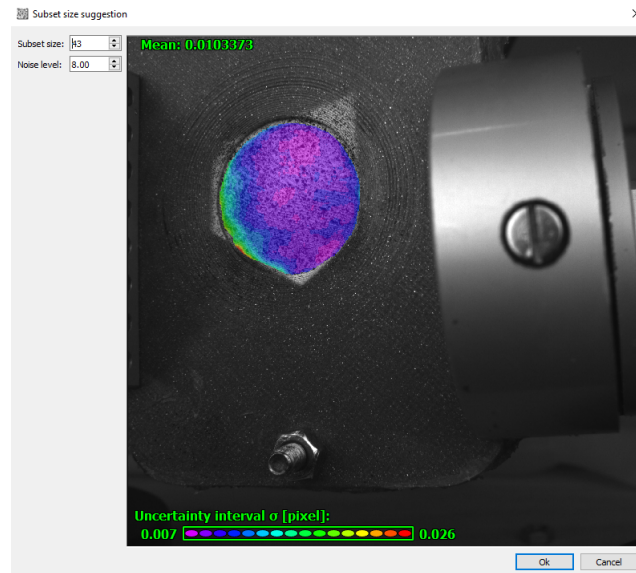


Figure 6.6: AOI with the facet and facet step recommended.

Subsequent to AOI selection, the DIC analysis was performed. The results obtained will be discussed in the following section.

6.2.1 Results

6.2.1.1 Displacement in the X direction

Regarding the X direction, Figure 6.7 and Figure 6.8 depict the displacements for this direction in both 3D and 2D plots, respectively.

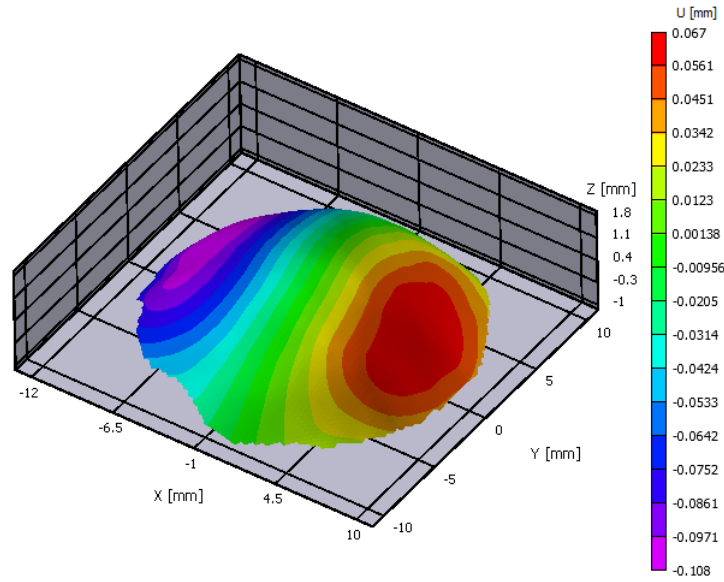


Figure 6.7: 3D plot for the displacement in the X direction.

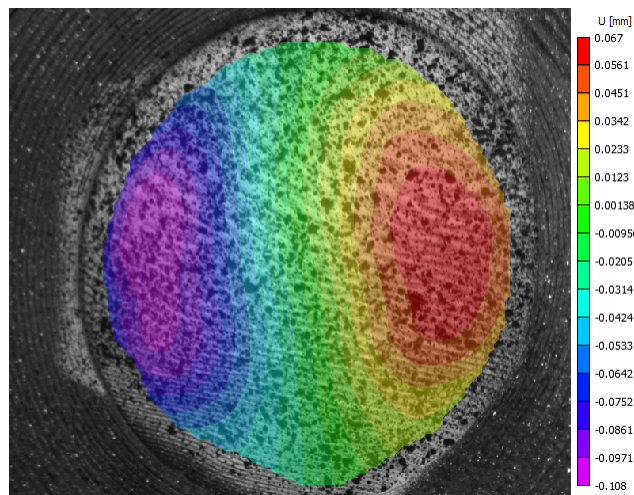


Figure 6.8: 2D plot for the displacement in the X direction.

Upon careful observation, it becomes apparent that there is near-symmetrical distribution of displacement values on both sides. On the positive side of the displacement, located

on the right side, the range of values is approximately 0.067 mm, while on the negative side of the displacement, situated on the left side, the range is approximately 0.108 mm.

Outside those corner zones, a noticeable trend of stability may be noticed, with successively declining values approaching zero. This pattern continues until the central zone, as there is displacement in the Z axis but no movement in the X axis.

It is noteworthy that in this circumstance, a symmetric distribution of displacement values would be predicted. However, one side appears to have slightly bigger displacement than the other. This difference can be explained by the pneumatic tube projecting slightly further and being slightly tilted to the left. Nonetheless, it is crucial to point out that the difference in displacement values is minimal, indicating only a slight variance.

6.2.1.2 Displacement in the Y direction

Regarding the Y direction, Figure 6.9 and Figure 6.10 depict the displacements for this direction in both 3D and 2D plots, respectively.

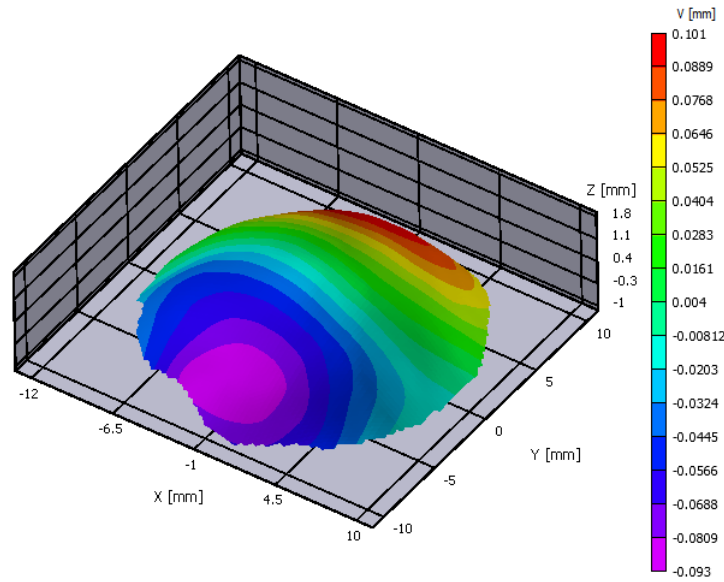


Figure 6.9: 3D plot for the displacement in the Y direction.

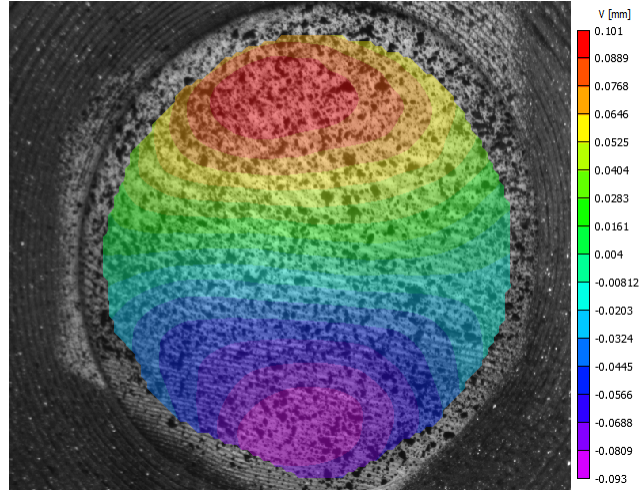


Figure 6.10: 2D plot for the displacement in the Y direction.

In terms of Y axis variation, a similar pattern of near-symmetrical distribution of displacement values is apparent. Positive displacements with a maximum value of 0.101 mm are observed in the upper zone, whereas negative displacements with a maximum value of 0.093 mm are found in the opposite direction. The difference between these two maximum values can be depicted as insignificant. Beyond these specific zones, in the middle area, a stability zone with displacement values near or exceeding zero is seen.

6.2.1.3 Displacement in the Z direction

Regarding the Z direction, Figure 6.11 and Figure 6.12 depict the displacements for this direction in both 3D and 2D plots, respectively.

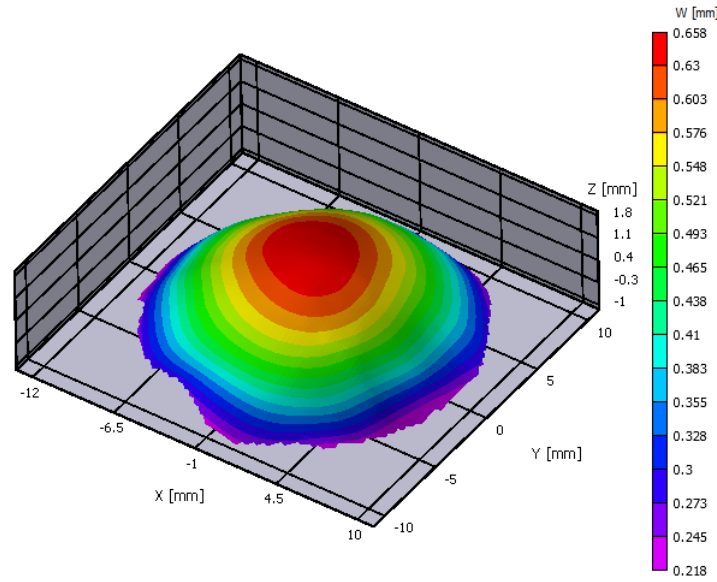


Figure 6.11: 3D plot for the displacement in the Z direction.

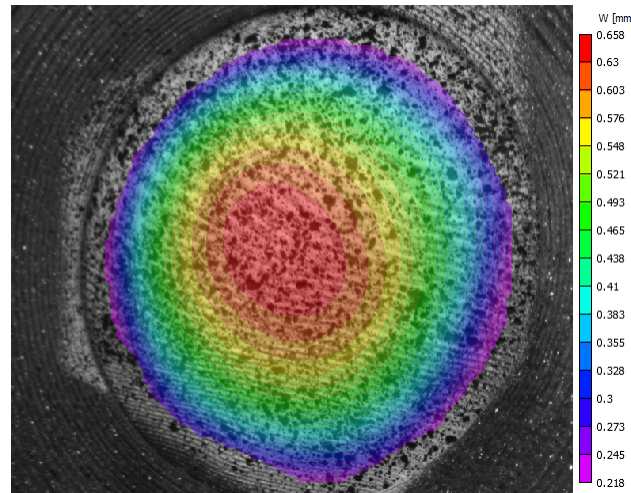


Figure 6.12: 2D plot for the displacement in the Z direction.

The analysis of displacement in the Z direction reveals the presence of a distinct central zone characterized by the most significant displacement magnitude, attaining a maximum value of 0.658 mm. As we move away from this central zone towards the tips, a gradual reduction in displacement is observed, ultimately reaching minimum values of 0.218 mm.

This way, the response obtained was already expected, given that it was anticipated that the experimental results in the radial direction would exhibit a radial expansion or bulging behavior of the silicone sample, due to the controlled inflation process applied through the

pneumatic tube connected to the mechanism. As the pressure is gradually increased, the silicone material is expected to deform and expand radially from its initial state, following the shape of the circular groove and the geometry of the specimen.

6.2.1.4 Total Displacement

Regarding the total displacement, Figure 6.13 and Figure 6.14 depict the 3D and 2D plots, respectively.

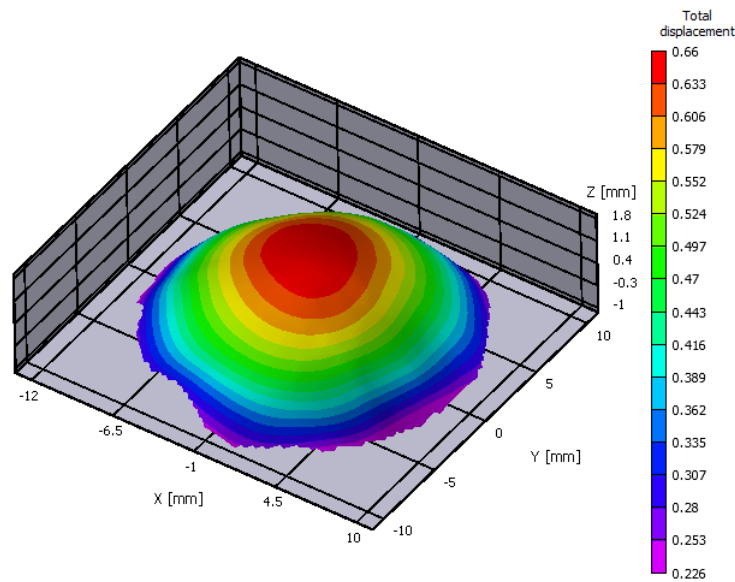


Figure 6.13: 3D plot for the total displacement.

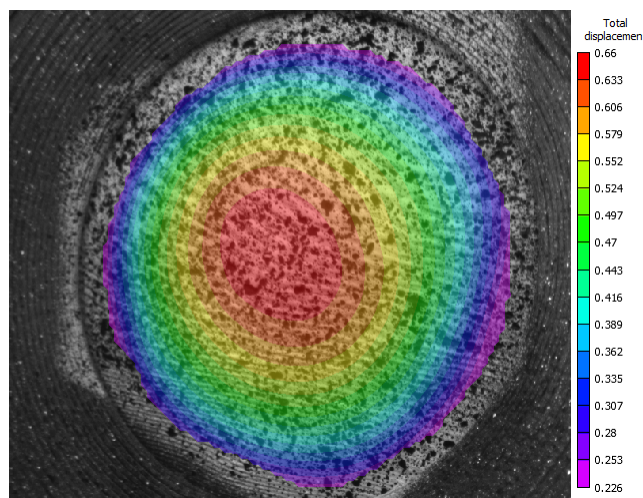


Figure 6.14: 2D plot for the total displacement.

In terms of the total displacement analysis, the experimental test reveals a maximum magnitude of displacement of 0.66 mm in the middle zone, gradually decreasing to values as low as 0.226 mm.

6.3 Computational Simulation

The geometry used in the simulation was based on the external area of the silicone specimen, resulting in a circular shape with a diameter of 25 mm and a thickness of 2 mm to ensure consistency in thickness values used throughout the study.

In order to simulate the behavior of the aortic material, the same material model used in Section 5.2 was employed, in order to maintain uniformity and coherence throughout the experimental procedures. This way, the Neo-Hooke parameters used are shown in Table 5.2. The two steps created in the simulation are also similar to the described in Section 5.3: Step-Diastole and Step-Systole. These steps represent the distinct pressure conditions applied during the experimental procedure.

In contrast to the previous test, the boundary conditions were altered to better simulate the experimental setup. Initially, a “fixed” boundary condition was implemented across the thickness of the sample region to restrict any movement of the silicone specimen within the mechanism. This boundary condition accurately reflects the fixation of the specimen inside the mechanism, where no movement is permitted in any direction.

Additionally, a load was defined for each step to replicate the corresponding pressure levels. In the Step-Diastole, a pressure of 0.107 MPa (80 mmHg) was applied to simulate the diastolic blood pressure, while in the Step-Systole, a pressure of 0.16 MPa (120 mmHg) was implemented to emulate the systolic blood pressure. The pressure load was specifically applied to one side of the specimen to represent the region within the mechanism where the pneumatic tube exerts pressure. This pressure is distributed across the specimen’s circular groove. The applied boundary condition (highlighted in orange) and the pressure load (highlighted in purple) can be observed in Figure 6.15.

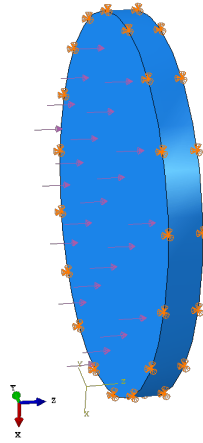


Figure 6.15: Load and boundary condition applied in the model.

The meshing process in the simulation utilized C3D10 elements due to the availability of substantial computational resources. These elements were chosen to ensure accurate and efficient representation of the specimen's geometry and behavior during the analysis.

Following the completion of the simulation, post-processing procedures, as described in Section 5.5, were carried out to generate a field output. This output was specifically created to calculate and assess the differences between the displacements obtained from the simulation.

6.3.1 Results

6.3.1.1 Displacement in the X direction

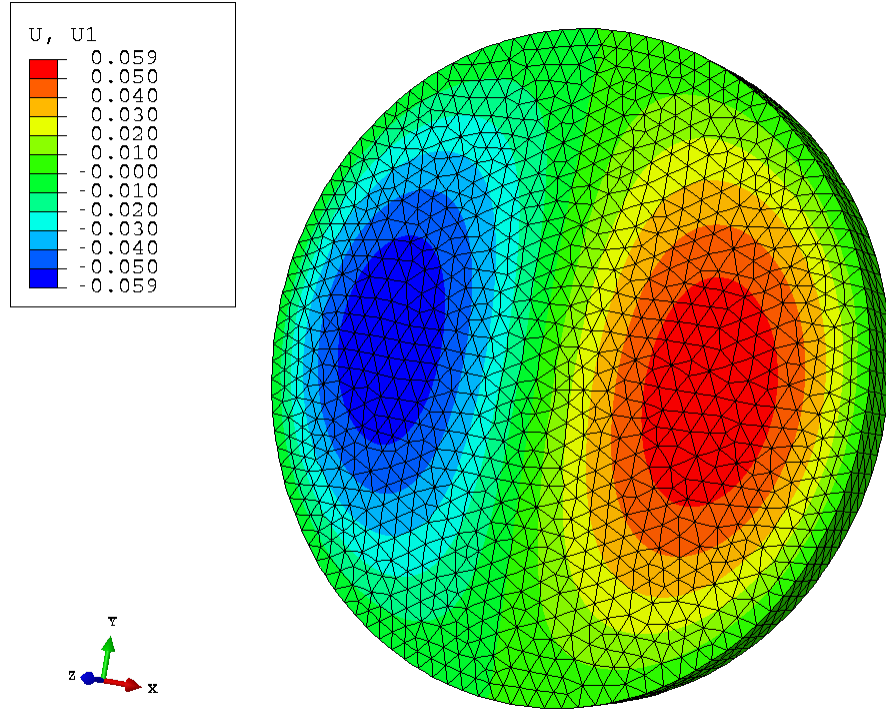


Figure 6.16: Displacements in the X direction.

As can be observed from Figure 6.16, the displacement in the X direction exhibits a symmetrical pattern at the tips, with positive displacement values on the right side and negative displacement values on the left. The largest displacement magnitude observed is 0.059 mm, with the magnitude gradually decreasing towards the surrounding regions. The central zone, on the other hand, exhibits negligible displacement, with near-zero displacement in its center and relatively insignificant values close to this zone.

6.3.1.2 Displacement in the Y direction

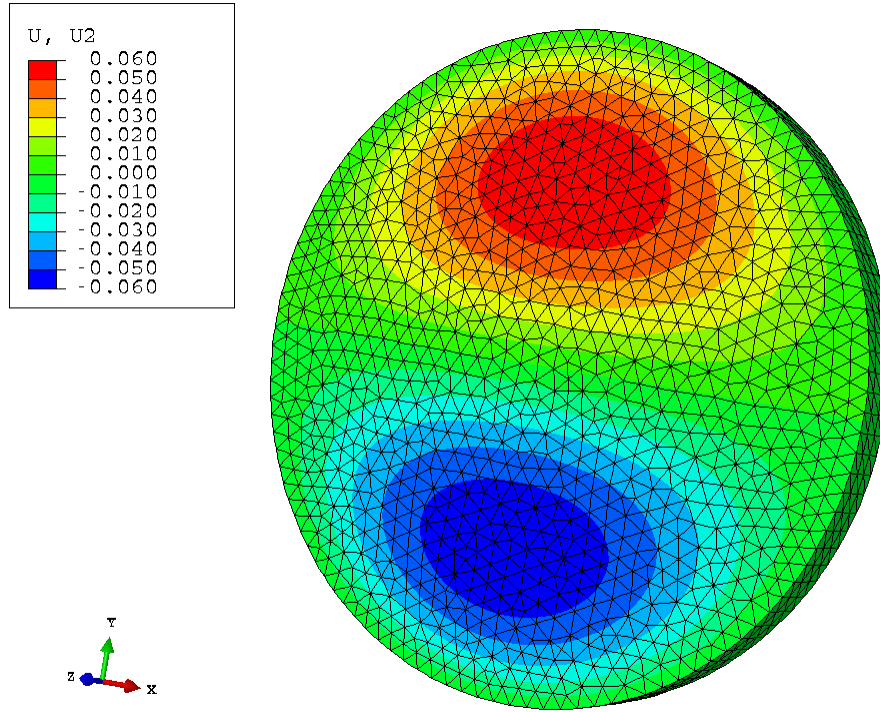
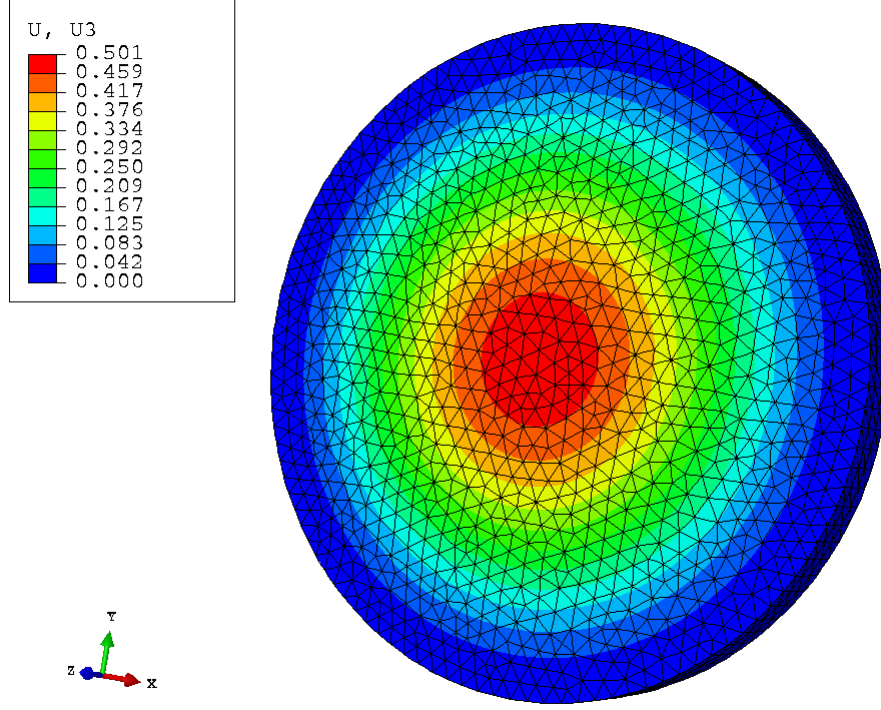


Figure 6.17: Displacements in the Y direction.

As can be observed from Figure 6.17, the displacement in the Y direction exhibits a symmetrical pattern, such as in the X direction, with positive displacement values on the upper side and negative displacement values on the lower side. The largest displacement magnitude observed is 0.060 mm, with the magnitude gradually decreasing towards the surrounding regions. The central zone, on the other hand, exhibits negligible displacement, with null displacement in its center and relatively insignificant values close to this zone.

6.3.1.3 Displacement in the Z direction

**Figure 6.18:** Displacements in the Z direction.

In terms of displacement in the Z direction, Figure 6.18 shows a distinct pattern with higher displacements in the central area. The displacement reduces gradually from the center and towards the the corners. The largest displacement magnitude measured is 0.501 mm, showing the considerable deformation detected in the central area. However, when the measurement locations get closer to the tips, the displacement decreases and eventually reaches zero. This behavior can be linked to the boundary condition of encastre, which restricts mobility and promotes stability at the specimen's outermost regions. The observed displacement pattern verifies the anticipated mechanical reaction of the silicone sample during the bulge inflation test.

6.3.1.4 Displacement Magnitude

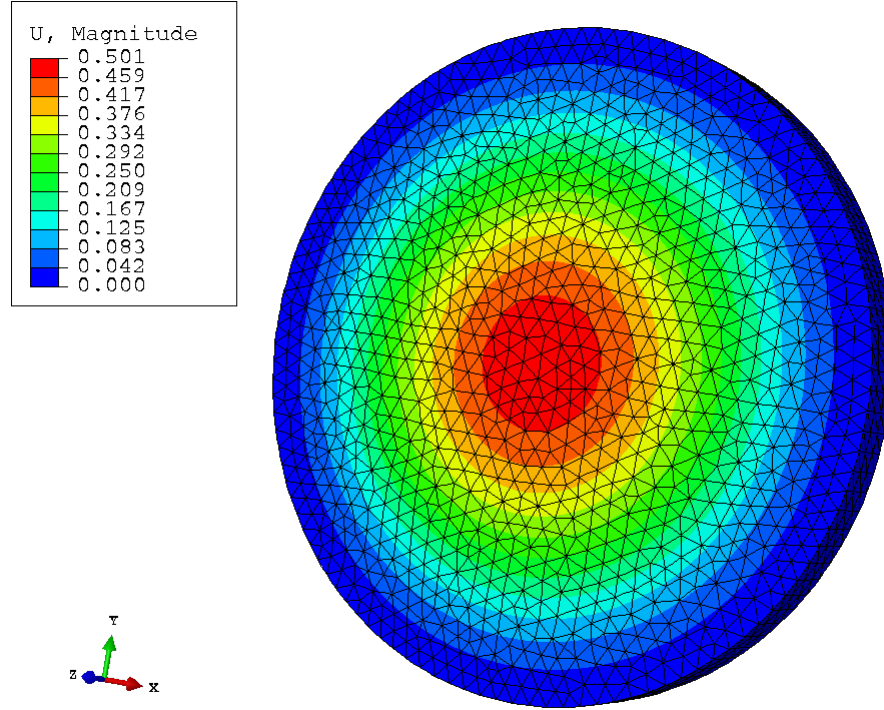


Figure 6.19: Displacement magnitude.

From the magnitude the overall displacement can be observed, showing a maximum magnitude of displacement of 0.501 mm in the middle zone, decreasing subsequently until it reaches a null displacement at the corners.

Results and Discussion

In the preceding sections, the experimental findings of the inflation-extension test conducted on a silicone phantom (Chapter 4) and the corresponding computer simulation results employing aortic tissue properties (Chapter 5) were presented. In addition to these outcomes, comprehensive analyses were performed in each direction for both the experimental and simulation datasets. In view of the observed disparities in the results, which made direct comparisons challenging due to inherent differences in the experimental and simulation conditions, an extra test was performed, as previously explained (Chapter 6). This test was designed to provide a more robust assessment of the comparative analysis between the experimental and simulation approaches, under conditions that were relatively similar in nature. Consequently, the results of this test will be thoroughly examined and analyzed in this chapter.

The present analysis will be performed based on the observation and comparison of the displacement maps obtained from each method.

7.1 Inflation-extension test

7.1.1 Displacement in the X direction

Figure 7.1 presents the displacement results in the X direction obtained from the experimental part and from the computational simulation, respectively.

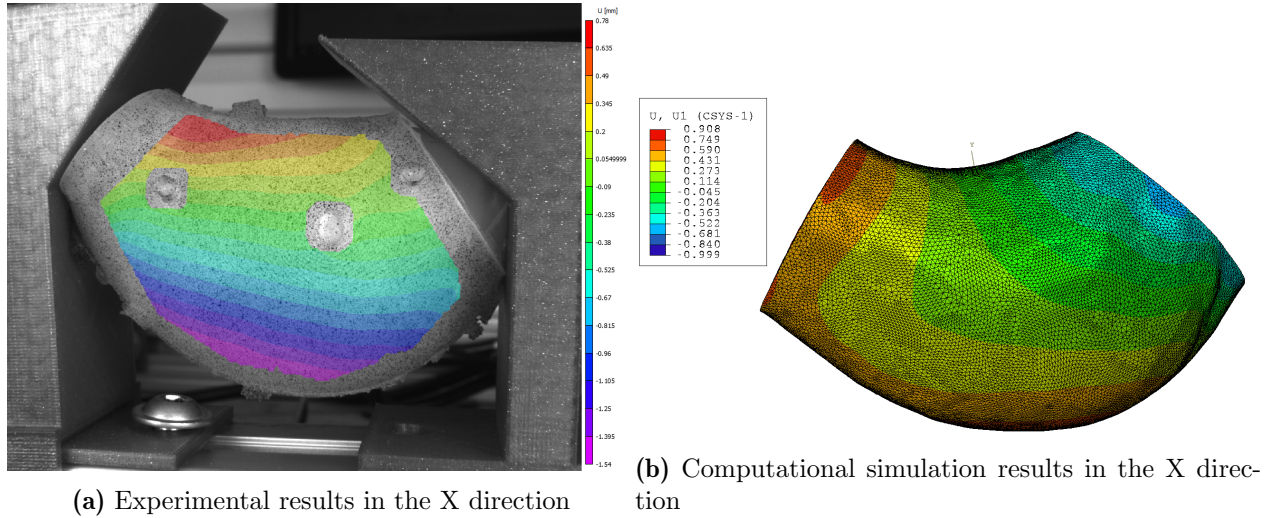


Figure 7.1: Comparison of the displacement in the X direction between experimental and computational simulation results.

Regarding the results in the X direction, given the existence of a clockwise rotation in the experimental part, referenced earlier in Section 4.4.1, it is challenging to establish a clear comparison between these results and the computational simulation ones.

The agreement between the two approaches is visible at the zone of highest displacement, which corresponds to the phantom's lower portion. However, there are evident differences at the figure's corners, where the simulation exhibits significant displacement compared to the different values exhibited in the experimental results.

These discrepancies in the results can be attributable to a variety of reasons, including slight discrepancies in the reference coordinate system and the inherent difficulty of simulating the exact experimental conditions, ending up imposing boundary conditions that influenced the results.

The displacement range acquired by experimental analysis is between -1.56 mm and 0.78 mm, while in the simulation, the range falls between -0.999 mm and 0.908 mm. Considering the differences in materials used, the proximity of the displacement value ranges in both methods is advantageous, showing a reasonable agreement between the experimental and simulated results.

7.1.2 Displacement in the Y direction

Figure 7.2 presents the displacement results in the Y direction obtained from the experimental part and from the computational simulation, respectively.

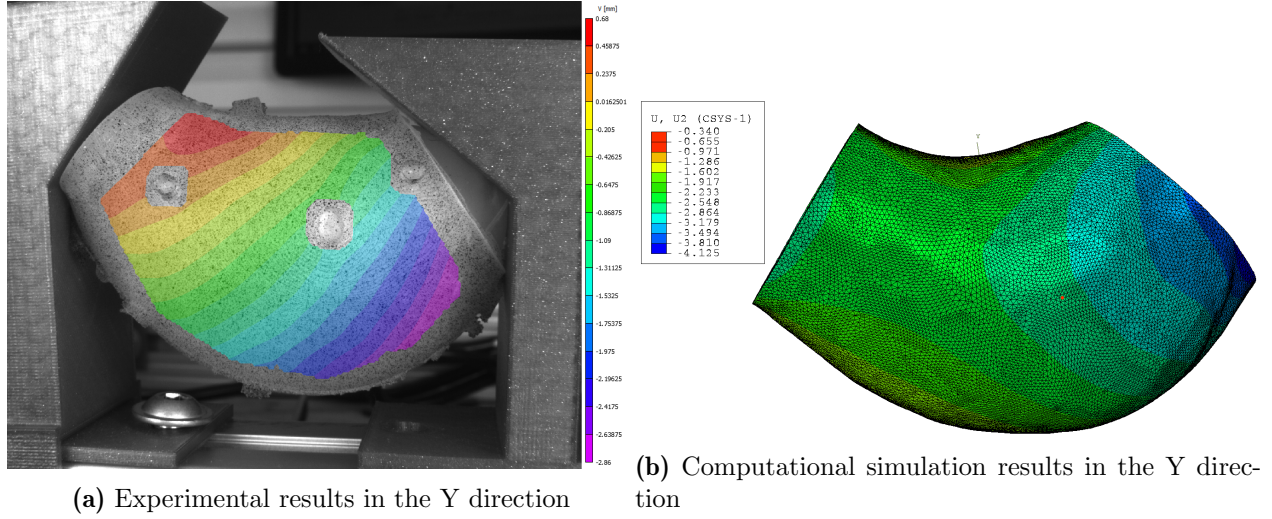


Figure 7.2: Comparison of the displacement in the Y direction between experimental and computational simulation results.

In this situation, there is a substantial mismatch in value ranges, as the DIC findings show a range of values from 0.68 mm to -2.92 mm, while the simulation shows a range of -0.34 mm to -4.125 mm.

The disparity in results can primarily be attributed to differences in boundary conditions and the reference coordinate system. In the experimental setup, the presence of the supports induces clockwise movement, leading to both upward and downward displacements in different regions. However, in the computer simulation, only the inner part of the aortic tips in the X direction and the upper zone of the aorta in the Z direction are fixed, restricting movement in specific directions. Consequently, greater motion is observed primarily in the lower zone where these boundary conditions are absent. Additionally, the misalignment of the reference coordinate system with the experimental setup further contributes to the deviation of results.

7.1.3 Displacement in the Z direction

Figure 7.3 presents the displacement results in the Z direction obtained from the experimental part and from the computational simulation, respectively.

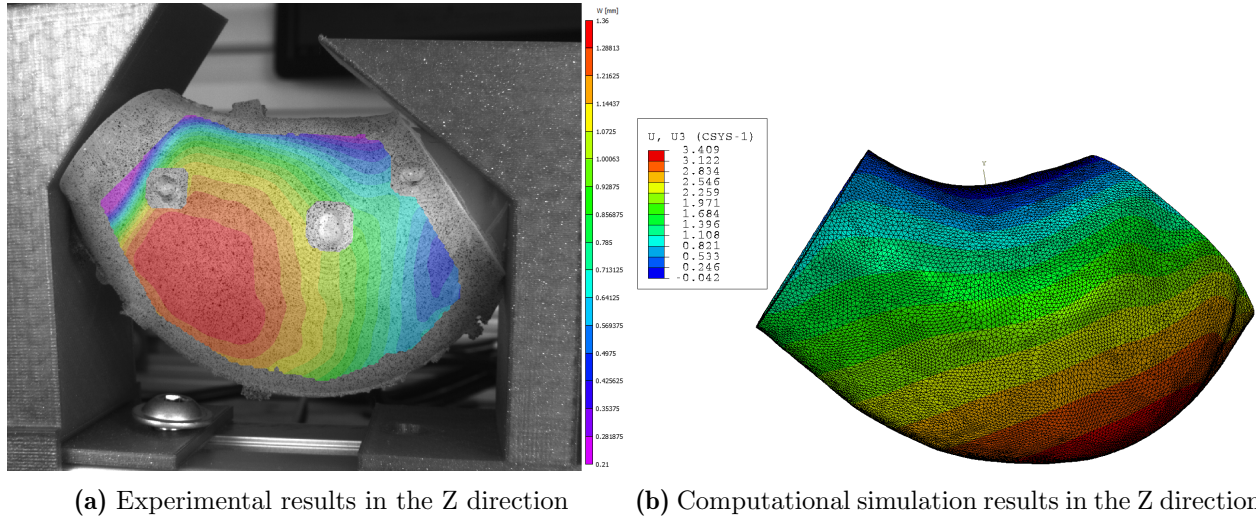


Figure 7.3: Comparison of the displacement in the Z direction between experimental and computational simulation results.

When juxtaposing the findings in the Z direction, there is a significant variance in the range of displacement values. The DIC analysis provides results ranging from 0.215 mm to 1.38 mm, however the simulation produces values ranging from -0.042 mm to 3.409 mm.

The biggest displacement zones differed between the two approaches as well. As previously indicated in Section 4.4.3, in the experimental analysis, the biggest displacement is detected near the lower left corner. The computational simulation, on the other hand, reveals the highest displacement in the lower right corner. However, both approaches demonstrate a displacement gradient beyond these zones. As previously explained, these results may be derived from non-homogeneity along the area of biggest displacement in the experimental part, and due to the imposed boundary conditions in the computational simulation.

An additional discrepancy is observed in the upper region of the phantom. The simulation exhibits negligible displacement in this area, in contrast to the experimental results which demonstrate noticeable displacement values. However, this near-zero displacement in the simulation's upper region can be attributed to the imposed boundary conditions.

Despite the observed differences, a notable similarity can be observed between the experimental and computer simulation results in the middle zone. The displacement values in this region exhibit relatively close agreement, suggesting some degree of similarity between these results.

7.1.4 Total displacement

Figure 7.4 presents the total displacement results obtained from the experimental part and from the computational simulation, respectively.

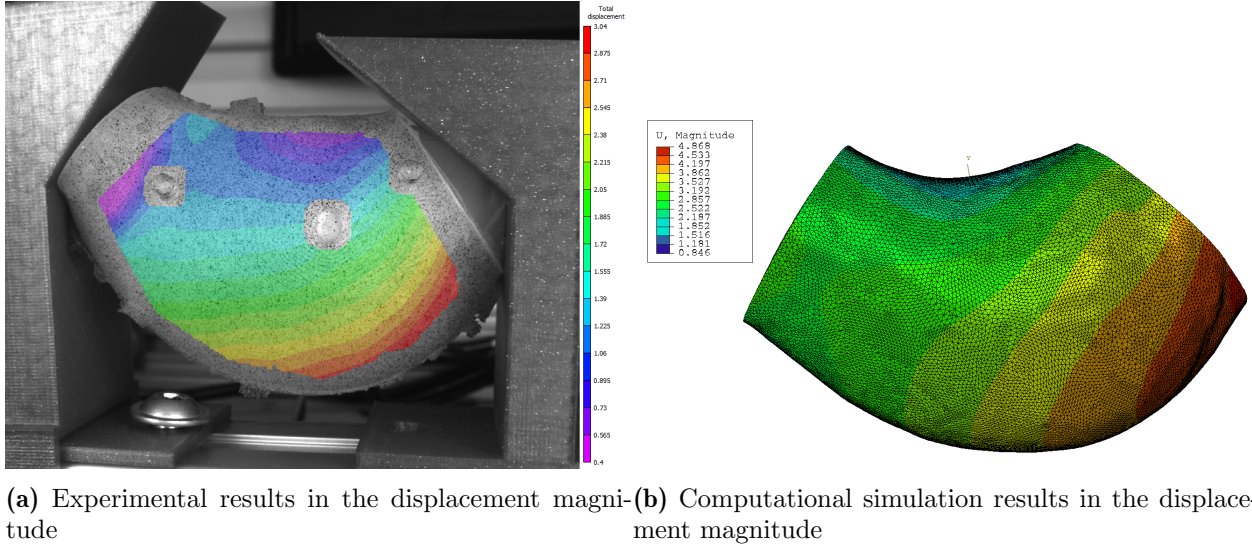


Figure 7.4: Comparison of the displacement magnitude between experimental and computational simulation results.

In terms of magnitude of the displacement, the experimental and computer simulation results have broadly comparable ranges. The displacement range in the experimental analysis is 0.4 mm to 3.08 mm, while in the simulation it is 0.846 mm to 4.868 mm. Additionally, the magnitude pattern for both approaches is comparable, with the zone of maximum displacement being in the same region for both.

Following the region of highest displacement, the intermediate zones exhibit some variations. In the experimental results, these zones display values ranging from 2.545 to 1.225 mm, while in the simulation, the corresponding central zone shows values ranging from 3.863 to 2.857 mm. Despite being different, the values are relatively proximate to each other.

The most visible variations occur in the upper zone. This location has displacements ranging from 0.4 mm to 1.39 mm in the experimental analysis, yet values ranging from 1.181 mm to 2.187 mm in the simulation. If the attributes of the two were identical, the displacement in the simulation would be smaller due to the enforced boundary constraint that restricts displacement in the Z direction. This boundary condition has an effect on the magnitude values, lowering them.

Additionally, smaller displacements are noticed in the corners of the phantom in the experimental analysis, which is not the case in the simulation. These lesser displacements could be attributable to limited Z displacements in this specific zone, which varies from the simulation.

7.2 Bulge inflation

7.2.1 Displacement in the X direction

Figure 7.5 presents the displacement results in the X direction obtained from the experimental part and from the computational simulation, respectively.

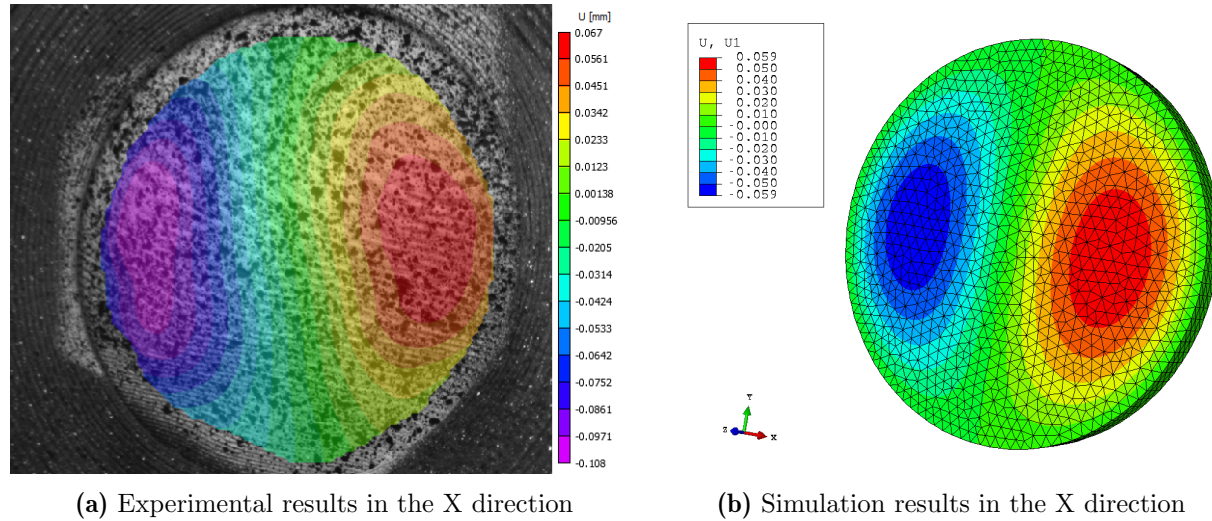


Figure 7.5: Comparison of the displacement in the X direction between experimental and computational simulation results.

Here already a greater consistency of results can be observed, with both methods exhibiting similar displacement patterns. Although the experimental result gives a near-symmetrical distribution of displacement values on both sides, for reasons already explained in Subsection 6.2.1.1, this difference is quite minimal, not being very significant.

Regarding the experimental part, as already mentioned in Subsection 4.4.1, on the left side the maximum value obtained is 0.067 mm, while on the negative side of the displacement, situated on the left side, this value is 0.108 mm. For the computer simulation, the largest displacement, symmetric on both sides, reaches a maximum value of 0.059 mm, which is not far from the experimental values.

The main distinction observed is that in the computer simulation, there is a region with zero or nearly zero displacement not only in the central area but also at the tips. On the other hand, in the experimental part, the only region with zero or nearly zero displacement is in the central area, and at the tips, the displacement still increases. This disparity could be attributed to the material not being completely embedded, allowing it to move more freely in the corners. Additionally, upon observing the image, it becomes apparent that the area analyzed by DIC does not extend to the edges. This characteristic is likely to contribute

to the slightly divergent results observed. However, despite these observed differences, it is important to note that both methods exhibit a largely congruent behavior when considered as a whole. While there may be variations in specific regions, the overall trends and patterns of displacement are comparable between the experimental and computer simulation approaches. This suggests that despite the nuanced distinctions, the fundamental behavior and response of the system are captured reasonably well by both methods.

7.2.2 Displacement in the Y direction

Figure 7.6 presents the displacement results in the X direction obtained from the experimental part and from the computational simulation, respectively.

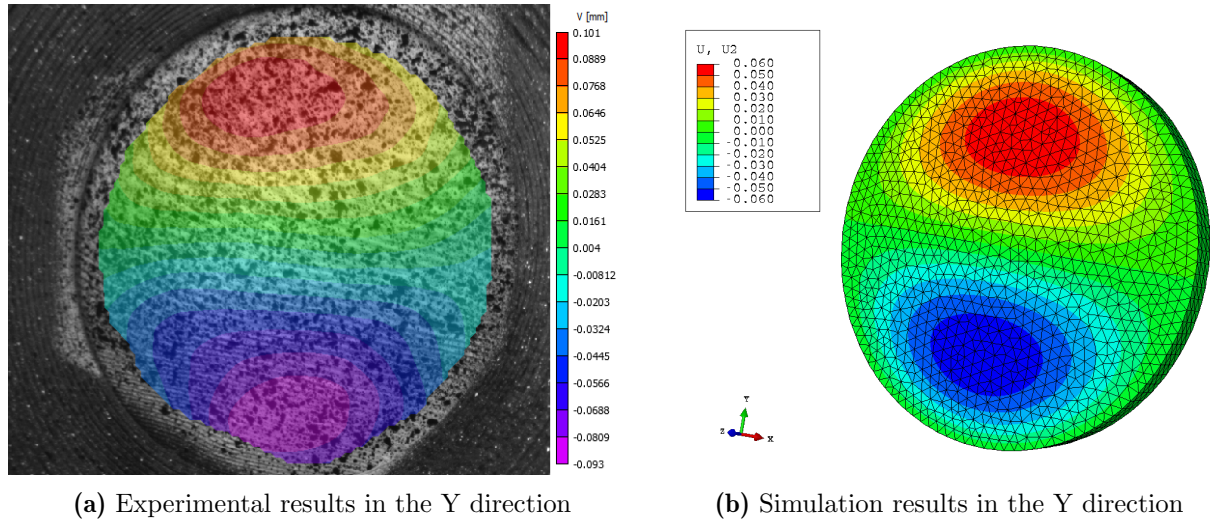


Figure 7.6: Comparison of the displacement in the Y direction between experimental and computational simulation results.

As can be observed, both methods exhibiting similar displacement patterns. For the experimental part, positive displacements with a maximum value of 0.101 mm are observed in the upper zone, whereas negative displacements with a maximum value of 0.093 mm are found in the opposite direction. For the computational simulation, the largest displacement magnitude observed is 0.060 mm. Although the values obtained from the two methods may show some disparities, they are relatively close in proximity and do not deviate significantly from each other. The observed variations, though present, can be considered relatively minor and may be attributed to factors such as experimental uncertainties or inherent limitations of the simulation model. Overall, the differences between the values are not substantial enough to suggest a substantial discrepancy between the two methods.

The main distinction observed is the same for the X direction, where for the computer simulation, there is a region with zero or nearly zero displacement not only in the central

area but also at the tips and for the experimental part the only region with zero or nearly zero displacement is in the central area. As mentioned earlier in Subsection 7.2.1 regarding the X direction, the observed difference in displacement could be attributed to the material not being entirely fixed, allowing for relatively freer movement in the corners, or the limited coverage of the analyzed area by DIC, which does not extend to the edges, being likely to contribute to the slightly divergent results.

However, by the analysis of the overall behaviour of the methods, a largely similar behavior is observed.

7.2.3 Displacement in the Z direction

Figure 7.7 presents the displacement results in the X direction obtained from the experimental part and from the computational simulation, respectively.

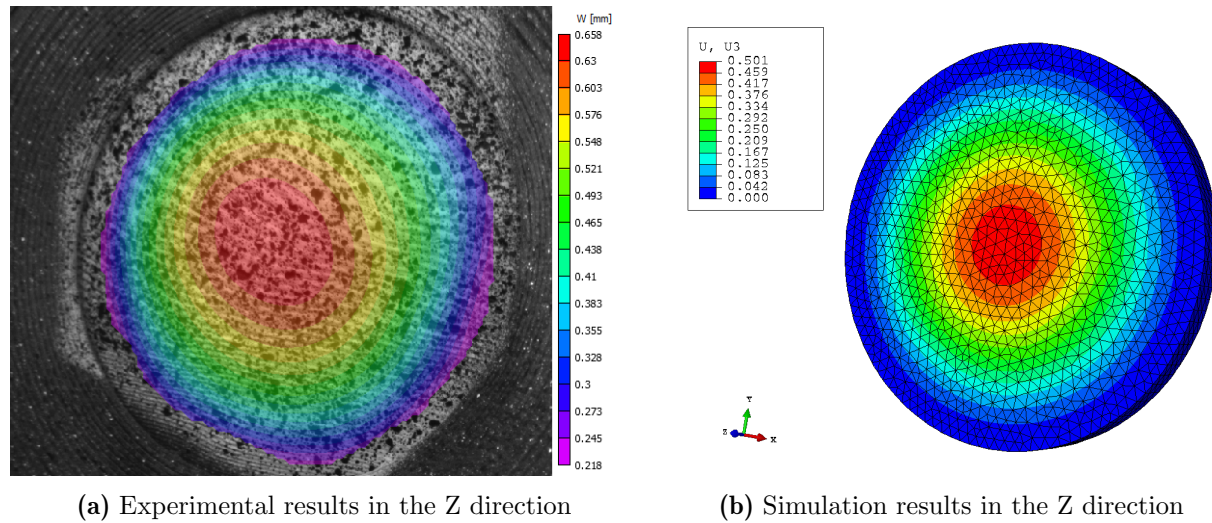


Figure 7.7: Comparison of the displacement in the Z direction between experimental and computational simulation results.

Both the experimental and simulation analyses demonstrate the presence of a distinct central zone with higher displacements. In the experimental analysis, the maximum displacement magnitude recorded is 0.658 mm, while in the simulation, it reaches 0.501 mm. This indicates significant deformation in the central area in both cases.

However, there are some disparities between the two methods. In the experimental analysis, the displacement gradually decreases as we move away from the central zone towards the tips, with a minimum value of 0.218 mm. In contrast, the simulation results show a reduction in displacement towards the corners, eventually reaching zero. The displacement behavior in the corners may be influenced by the factors mentioned previously, such as the

material's relative freedom of movement in the experimental part, which can cause variations and the limited coverage of the analyzed area by DIC, excluding the edges, introducing uncertainties.

Overall, while the magnitudes of displacement differ slightly between the experimental and simulation results, the overall trends align with the expected mechanical response of the silicone sample during the bulge inflation test. These findings suggest that both methods capture the fundamental behavior of the sample during the inflation process, albeit with some nuanced variations. In conclusion, the experimental testing conducted with the silicone sample can be considered to closely approximate the simulation results obtained using the properties of aortic tissue.

7.2.4 Total displacement

Figure 7.8 presents the displacement magnitude from the experimental part and from the computational simulation, respectively.

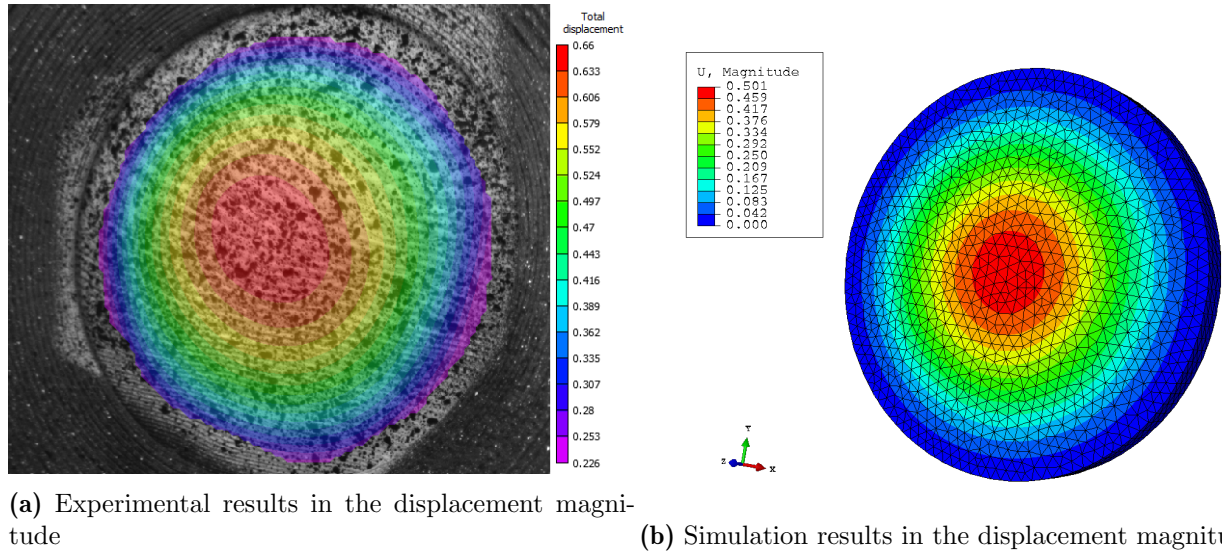


Figure 7.8: Comparison of the displacement magnitude between experimental and computational simulation results.

In terms of the total displacement analysis, the experimental test reveals a maximum magnitude of displacement of 0.66 mm in the middle zone, gradually decreasing to values as low as 0.226 mm. Conversely, in the simulation, the overall displacement pattern demonstrates a maximum magnitude of 0.501 mm in the middle zone, followed by a progressive reduction in displacement towards null values at the corners.

Despite some variations in the specific magnitude of displacements, both the experimental and simulation results consistently depict a central zone of higher displacements, with

decreasing values towards the corners. This similarity in the displacement trends indicates that the silicone sample used in the experimental setup closely approximates the mechanical response of the aortic tissue simulated in the study. The findings from the experimental test contribute valuable insights into the overall displacement behavior, supporting the validity of the simulation approach and enhancing our understanding of the biomechanical characteristics involved in the bulge inflation test.

Conclusions and Future works

8.1 Conclusions

The present study aimed to investigate the feasibility of an unconventional inflation setup, specifically using air in the inflation-extension process. The main objectives were to validate this experimental approach and quantify the displacements exhibited by a silicone phantom using Digital Image Correlation (DIC). Based on the results provided by the experimental part, it can be concluded that the experimental setup was successfully implemented and yielded measurable displacements in the silicone phantom. This confirms the feasibility of using air as the inflating medium and demonstrates the potential of this approach for further investigations.

Furthermore, it is important to underscore that the experimental apparatus owes its feasibility to the creation of the balloon-like structure enabling inflation without imposing unnecessary and unrealistic conditions, such as constraining the aorta within the corners of the apparatus with overly restrictive fixations to prevent any air leakage. Given the practical application of this setup on a real aorta, such rigid attachments would be unsuitable. As a result, multiple iterations were carried out to enhance the balloon's design, with the goal of optimizing the process and making it as suitable as possible for its intended purpose. These iterations sought to achieve a compromise between practicability and preserving the integrity of the experimental conditions, thereby ensuring the accuracy and reliability of the experimental test.

In the process of testing the setup, three different types of silicone materials were considered. However, during the experimentation phase, it became evident that the chosen manufacturing method was not suitable for some of these materials, leading to the utilization of only one silicone for the present analysis. Although the sample size was restricted, the obtained findings still provide valuable preliminary insights into the behavior and displacement measurements of the tested silicone phantom. While this study clearly illustrates the potential of the chosen inflation setup and its applicability in capturing displacement measurements, it also raises questions regarding the optimization of the manufacturing process and the selection of suitable silicone materials. The limitations encountered during the experimentation phase have created further curiosity about alternative manufacturing techniques and their influence on the performance and behavior of the phantoms.

Furthermore, due to the scarcity of available comparative data in the existing literature and the inherent difficulties associated with acquiring human or animal aorta specimens, computer simulation emerged as a viable alternative for comparison purposes. For the computational simulation, a balance was struck between replicating the experimental conditions and introducing necessary modifications to render the simulation feasible. Furthermore, time constraints necessitated certain simplifications, which inevitably influenced the results. Due to this, an exact and compelling comparison between experimental and computer simulation results was challenging to establish in some cases. Despite these challenges, a combined examination of experimental and simulation provided useful insights and contribute to an in-depth understanding of the phenomenon under study.

Finally, given the disparities between the two approaches, in order to further compare the used silicone to see whether its properties could assimilate those of aortic tissue, a supplementary inflation test, namely the bulge inflation test, was employed. This test involved minor modifications to the experimental setup and facilitated further comparisons between the experimental and simulation outcomes. Notably, these results revealed improved agreement, which can be attributed to a greater match between the experimental and simulation conditions. The convergence of results between these two approaches strengthens the simulation model's validity and applicability in capturing the essential aspects of the inflation dynamics.

8.2 Future Works

Considering the work done in this thesis and the results obtained, some developments can be performed to improve them. Some of those suggestions, as future steps remarks, are the following:

- Obtain the displacement difference map between the experimental results and the results obtained through computer simulation;
- Dimension and recreate lateral fixtures with the ability to rotate in order to observe the total phantom behavior through the DIC;
- Improve the molds used for the phantoms to facilitate the utilization of the existing silicone materials;
- Improve the speckle pattern painting procedure;
- Recreate the interaction of the balloon with the aorta through computer simulation;
- Performed the inflation-extension experiment using a human or porcine aorta to effectively validate the obtained results.

These proposed measures aim to strengthen the validity and applicability of the research findings, while also addressing potential areas for improvement in the experimental setup.

Bibliography

- [1] National Cancer Institute. Introduction to the cardiovascular system. <https://training.seer.cancer.gov/anatomy/cardiovascular/>. Accessed on 2023-03-10.
- [2] Clive Rosendorff. *Essential cardiology: principles and practice*. Springer, 2005.
- [3] Up Huh, Chung-Won Lee, Ji-Hun You, Chan-Hee Song, Chi-Seung Lee, and Dong-Man Ryu. Determination of the material parameters in the holzapfel-gasser-ogden constitutive model for simulation of age-dependent material nonlinear behavior for aortic wall tissue under uniaxial tension. *Applied Sciences*, 9(14):2851, 2019.
- [4] Hunter J White, Stephen Bordes, and Judith Borger. Anatomy, abdomen and pelvis: Aorta. 2019.
- [5] Cleveland Clinic. Aorta. <https://my.clevelandclinic.org/health/articles/17058-aorta-anatomy>. Accessed on 2023-10-2.
- [6] Lei Wang. *Modelling dissection of the arterial wall*. PhD thesis, University of Glasgow, 2016.
- [7] Ryan Gouveia e Melo, Goncalo Silva Duarte, Alice Lopes, Mariana Alves, Daniel Caldeira, Ruy Fernandes e Fernandes, and Luis Mendes Pedro. Incidence and prevalence of thoracic aortic aneurysms: a systematic review and meta-analysis of population-based studies. In *Seminars in thoracic and cardiovascular surgery*, volume 34, pages 1–16. Elsevier, 2022.
- [8] Gustav G Belz. Elastic properties and windkessel function of the human aorta. *Cardiovascular drugs and therapy*, 9:73–83, 1995.
- [9] Loren F Hiratzka, George L Bakris, Joshua A Beckman, Robert M Bersin, Vincent F Carr, Donald E Casey Jr, Kim A Eagle, Luke K Hermann, Eric M Isselbacher, et al. 2010 accf/aha/aats/acr/asa/sca/scai/sir/sts/svm guidelines for the diagnosis and management of patients with thoracic aortic disease: a report of the american college of cardiology foundation/american heart association task force on practice guidelines, american association for thoracic surgery, american college of radiology, american stroke association, society of cardiovascular anesthesiologists, society for cardiovascular angiography and interventions, society of interventional radiology, society of thoracic surgeons, and society for vascular medicine. *Circulation*, 121(13):e266–e369, 2010.

- [10] Amélie Pinard, Gregory T Jones, and Dianna M Milewicz. Genetics of thoracic and abdominal aortic diseases: aneurysms, dissections, and ruptures. *Circulation research*, 124(4):588–606, 2019.
- [11] Samar A Tarraf, Benjamin Kramer, Emily Vianna, Callan Gillespie, Emídio Germano, Kelly B Emerton, Rouzbeh Amini, Robb Colbrunn, Jennifer Hargrave, Eric E Roselli, et al. Lengthwise regional mechanics of the human aneurysmal ascending thoracic aorta. *Acta Biomaterialia*, 162:266–277, 2023.
- [12] André Mourato, Rodrigo Valente, José Xavier, Moisés Brito, Stéphane Avril, José César de Sá, António Tomás, and José Fragata. Computational modelling and simulation of fluid structure interaction in aortic aneurysms: A systematic review and discussion of the clinical potential. *Applied Sciences*, 12(16):8049, 2022.
- [13] Caitlin Martin, Wei Sun, and John Elefteriades. Patient-specific finite element analysis of ascending aorta aneurysms. *American Journal of Physiology-Heart and Circulatory Physiology*, 308(10):H1306–H1316, 2015.
- [14] Sonja Pejcić, Syed M Ali Hassan, David E Rival, and Gianluigi Bisleri. Characterizing the mechanical properties of the aortic wall. *Vessel Plus*, 3:32, 2019.
- [15] Andrea Avanzini, Davide Battini, Lorenzo Bagozzi, and Gianluigi Bisleri. Biomechanical evaluation of ascending aortic aneurysms. *BioMed Research International*, 2014, 2014.
- [16] UF Health Aortic Disease Center. Anatomy of the aorta. <https://ufhealth.org/uf-health-aortic-disease-center/aorta-anatomy>. Accessed on 2023-03-15.
- [17] Eric M Isselbacher. Thoracic and abdominal aortic aneurysms. *Circulation*, 111(6):816–828, 2005.
- [18] T Christian Gasser, Ray W Ogden, and Gerhard A Holzapfel. Hyperelastic modelling of arterial layers with distributed collagen fibre orientations. *Journal of the royal society interface*, 3(6):15–35, 2006.
- [19] Claudie Petit, S Jamaledin Mousavi, and Stéphane Avril. Review of the essential roles of smcs in ataa biomechanics. *Advances in biomechanics and tissue regeneration*, pages 95–114, 2019.
- [20] Daniella Eliathamby, Mariana Gutierrez, Aileen Liu, Maral Ouzounian, Thomas L Forbes, Kong Teng Tan, and Jennifer Chung. Ascending aortic length and its association with type a aortic dissection. *Journal of the American Heart Association*, 10(13):e020140, 2021.

-
- [21] Uchechukwu KA Sampson, Paul E Norman, F Gerald R Fowkes, Victor Aboyans, Yanna Song, Frank E Harrell Jr, Mohammad H Forouzanfar, Mohsen Naghavi, Julie O Denenberg, Mary M McDermott, et al. Global and regional burden of aortic dissection and aneurysms: mortality trends in 21 world regions, 1990 to 2010. *Global heart*, 9(1):171–180, 2014.
- [22] Alkiviadis Tsamis, Jeffrey T Krawiec, and David A Vorp. Elastin and collagen fibre microstructure of the human aorta in ageing and disease: a review. *Journal of the royal society interface*, 10(83):20121004, 2013.
- [23] Ghassan S Kassab. Biomechanics of the cardiovascular system: the aorta as an illustrative example. *Journal of the Royal Society Interface*, 3(11):719–740, 2006.
- [24] Helmut F Kuecherer, Armin Just, and Hartmut Kirchheim. Evaluation of aortic compliance in humans, 2000.
- [25] João L Cavalcante, João AC Lima, Alban Redheuil, and Mouaz H Al-Mallah. Aortic stiffness: current understanding and future directions. *Journal of the American College of Cardiology*, 57(14):1511–1522, 2011.
- [26] David J Tobey, Tyler S Reynolds, George E Kopchok, Carlos E Donayre, Ali Khoyneshad, and Rodney A White. In vivo assessment of ascending and arch aortic compliance. *Annals of vascular surgery*, 57:22–28, 2019.
- [27] Kevin E Boczar, Munir Boodhwani, Luc Beauchesne, Carole Dennie, Kwan Leung Chan, George A Wells, and Thais Coutinho. Aortic stiffness, central blood pressure, and pulsatile arterial load predict future thoracic aortic aneurysm expansion. *Hypertension*, 77(1):126–134, 2021.
- [28] George Koullias, Raj Modak, Maryann Tranquilli, Dimitris P Korkolis, Paul Barash, and John A Elefteriades. Mechanical deterioration underlies malignant behavior of aneurysmal human ascending aorta. *The Journal of thoracic and cardiovascular surgery*, 130(3):677–e1, 2005.
- [29] Emile Saliba, Ying Sia, Annie Dore, and Ismael El Hamamsy. The ascending aortic aneurysm: When to intervene? *IJC Heart & Vasculature*, 6:91–100, 2015.
- [30] Jaeyoung Kwon, Junhyeok Ock, and Namkug Kim. Mimicking the mechanical properties of aortic tissue with pattern-embedded 3d printing for a realistic phantom. *Materials*, 13(21):5042, 2020.
- [31] Sina G Yazdi, PH Geoghegan, PD Docherty, Mark Jermy, and Adib Khanafer. A review of arterial phantom fabrication methods for flow measurement using piv techniques. *Annals of Biomedical Engineering*, 46:1697–1721, 2018.

- [32] Giulia Comunale, Luigi Di Micco, Daniela Paola Boso, Francesca Maria Susin, and Paolo Peruzzo. Numerical models can assist choice of an aortic phantom for in vitro testing. *Bioengineering*, 8(8):101, 2021.
- [33] Abdulrazzaq Sulaiman, Loïc Boussel, Frédéric Taconnet, Jean Michel Serfaty, Hasan Alsaid, Cherif Attia, Laurent Huet, and Philippe Douek. In vitro non-rigid life-size model of aortic arch aneurysm for endovascular prosthesis assessment. *European journal of cardio-thoracic surgery*, 33(1):53–57, 2008.
- [34] Aidan J Cloonan, Danial Shahmirzadi, Ronny X Li, Barry J Doyle, Elisa E Konofagou, and Tim M McGloughlin. 3d-printed tissue-mimicking phantoms for medical imaging and computational validation applications. *3D printing and additive manufacturing*, 1(1):14–23, 2014.
- [35] Sina G Yazdi, Larissa Huetter, Paul D Docherty, Petra N Williamson, Don Clucas, Mark Jermy, and Patrick H Geoghegan. A novel fabrication method for compliant silicone phantoms of arterial geometry for use in particle image velocimetry of haemodynamics. *Applied Sciences*, 9(18):3811, 2019.
- [36] Karina Arcaute and Ryan B Wicker. Patient-specific compliant vessel manufacturing using dip-spin coating of rapid prototyped molds. 2008.
- [37] Sina G Yazdi, PH Geoghegan, PD Docherty, Mark Jermy, and Adib Khanafer. A review of arterial phantom fabrication methods for flow measurement using piv techniques. *Annals of Biomedical Engineering*, 46:1697–1721, 2018.
- [38] Diego Gallo, Utku Gülan, Antonietta Di Stefano, Raffaele Ponzini, Beat Lüthi, Markus Holzner, and Umberto Morbiducci. Analysis of thoracic aorta hemodynamics using 3d particle tracking velocimetry and computational fluid dynamics. *Journal of Biomechanics*, 47(12):3149–3155, 2014.
- [39] Giovanni Biglino, Alessandro Giardini, Catriona Baker, Richard S Figliola, Tain-Yen Hsia, Andrew M Taylor, Silvia Schievano, MOCHA Collaborative Group, et al. In vitro study of the norwood palliation: a patient-specific mock circulatory system. *ASAIO journal*, 58(1):25–31, 2012.
- [40] Rong Fung Huang, Ten-Fang Yang, and Y-K Lan. Pulsatile flows and wall-shear stresses in models simulating normal and stenosed aortic arches. *Experiments in fluids*, 48:497–508, 2010.
- [41] Sina G Yazdi, Paul D Docherty, Petra N Williamson, Mark Jermy, Natalia Kabaliuk, Adib Khanafer, and Patrick H Geoghegan. In vitro pulsatile flow study in compliant and rigid ascending aorta phantoms by stereo particle image velocimetry. *Medical Engineering & Physics*, 96:81–90, 2021.

-
- [42] AS Anayiotos, SA Jones, DP Giddens, S Glagov, and CK Zarins. Shear stress at a compliant model of the human carotid bifurcation. 1994.
- [43] Philippe Reymond, Paolo Crosetto, Simone Deparis, Alfio Quarteroni, and Nikos Stergiopulos. Physiological simulation of blood flow in the aorta: comparison of hemodynamic indices as predicted by 3-d fsi, 3-d rigid wall and 1-d models. *Medical engineering & physics*, 35(6):784–791, 2013.
- [44] Patrick H Geoghegan, Mark C Jermy, and David S Nobes. A piv comparison of the flow field and wall shear stress in rigid and compliant models of healthy carotid arteries. *Journal of Mechanics in Medicine and Biology*, 17(03):1750041, 2017.
- [45] Larissa Hütter, Patrick H Geoghegan, Paul D Docherty, Milad S Lazarjan, Donald Clucas, and Mark Jermy. Fabrication of a compliant phantom of the human aortic arch for use in particle image velocimetry (piv) experimentation. *Current Directions in Biomedical Engineering*, 2(1):493–497, 2016.
- [46] JR Sherman, Hussain S Rangwala, Ciprian N Ionita, AC Dohatcu, JW Lee, Daniel R Bednarek, Kenneth R Hoffmann, and Stephen Rudin. Investigation of new flow modifying endovascular image-guided interventional (eigi) techniques in patient-specific aneurysm phantoms (psaps) using optical imaging. In *Medical Imaging 2008: Visualization, Image-Guided Procedures, and Modeling*, volume 6918, pages 631–641. SPIE, 2008.
- [47] Kenneth I Aycock, Prasanna Hariharan, and Brent A Craven. Particle image velocimetry measurements in an anatomical vascular model fabricated using inkjet 3d printing. *Experiments in Fluids*, 58:1–8, 2017.
- [48] Luigi Di Micco, Giulia Comunale, Stefano Bonvini, Paolo Peruzzo, and Francesca Maria Susin. Distensibility of deformable aortic replicas assessed by an integrated in-vitro and in-silico approach. *Bioengineering*, 9(3):94, 2022.
- [49] Giovanni Biglino, Peter Verschueren, Raf Zegels, Andrew M Taylor, and Silvia Schievano. Rapid prototyping compliant arterial phantoms for in-vitro studies and device testing. *Journal of Cardiovascular Magnetic Resonance*, 15:1–7, 2013.
- [50] Elastrat. Anatomical vascular models. <https://www.elastrat.com/>. Accessed on 2023-02-20.
- [51] V Pazos, R Mongrain, and JC Tardif. Deformable mock stenotic artery with a lipid pool. 2010.

- [52] Natasha Alves, Angela Kim, Jeremy Tan, Germain Hwang, Talha Javed, Bogdan Neagu, and Brian K Courtney. Cardiac tissue-mimicking ballistic gel phantom for ultrasound imaging in clinical and research applications. *Ultrasound in Medicine & Biology*, 46(8):2057–2069, 2020.
- [53] Aaron Romo, Pierre Badel, Ambroise Duprey, Jean-Pierre Favre, and Stéphane Avril. In vitro analysis of localized aneurysm rupture. *Journal of Biomechanics*, 47(3):607–616, 2014.
- [54] Aaron Romo, Stéphane Avril, Pierre Badel, Jérôme Molimard, Ambroise Duprey, and Jean-Pierre Favre. Mechanical characterization of the thoracic ascending aortae. In *IRCOBI (International Research Council on Biomechanics of Injury)*, pages IRC–12, 2012.
- [55] Renate W Boekhoven, Mathijs FJ Peters, Marcel CM Rutten, Marc R van Sambeek, Frans N van de Vosse, and Richard GP Lopata. Inflation and bi-axial tensile testing of healthy porcine carotid arteries. *Ultrasound in Medicine & Biology*, 42(2):574–585, 2016.
- [56] Steven P Marra, Francis E Kennedy, Jeffrey N Kinkaid, and Mark F Fillinger. Elastic and rupture properties of porcine aortic tissue measured using inflation testing. *Cardiovascular Engineering*, 6:123–131, 2006.
- [57] Dinesh Mohan and John W Melvin. Failure properties of passive human aortic tissue. ii—biaxial tension tests. *Journal of biomechanics*, 16(1):31–44, 1983.
- [58] Ambroise Duprey, Olfa Trabelsi, Marco Vola, Jean-Pierre Favre, and Stéphane Avril. Biaxial rupture properties of ascending thoracic aortic aneurysms. *Acta biomaterialia*, 42:273–285, 2016.
- [59] Olfa Trabelsi, Frances M Davis, Jose F Rodriguez-Matas, Ambroise Duprey, and Stéphane Avril. Patient specific stress and rupture analysis of ascending thoracic aneurysms. *Journal of biomechanics*, 48(10):1836–1843, 2015.
- [60] Jin Kim, Pierre Badel, Ambroise Duprey, Jean-Pierre Favre, and Stéphane Avril. Characterisation of failure in human aortic tissue using digital image correlation. *Computer Methods in Biomechanics and Biomedical Engineering*, 14(Supplément 1):73–74, 2011.
- [61] David M Pierce, Franz Maier, Hannah Weisbecker, Christian Viertler, Peter Verbrugghe, Nele Famaey, Inge Fourneau, Paul Herijgers, and Gerhard A Holzapfel. Human thoracic and abdominal aortic aneurysmal tissues: Damage experiments, statistical analysis and constitutive modeling. *Journal of the mechanical behavior of biomedical materials*, 41:92–107, 2015.

-
- [62] Toshiro Ohashi, Syukei Sugita, Takeo Matsumoto, Kiichiro Kumagai, Hiroji Akimoto, Koichi Tabayashi, and Masaaki Sato. Rupture properties of blood vessel walls measured by pressure-imposed test. *JSME International Journal Series C Mechanical Systems, Machine Elements and Manufacturing*, 46(4):1290–1296, 2003.
- [63] Jungsil Kim and Seungik Baek. Circumferential variations of mechanical behavior of the porcine thoracic aorta during the inflation test. *Journal of biomechanics*, 44(10):1941–1947, 2011.
- [64] Edgar José Sanches Mascarenhas. *Mechanical characterization of aortas using 2D ultrasound elastography*. PhD thesis, 2014.
- [65] Marco Palanca, Gianluca Tozzi, and Luca Cristofolini. The use of digital image correlation in the biomechanical area: a review. *International biomechanics*, 3(1):1–21, 2016.
- [66] Francisco Moutinho de Magalhães. *Digital Image Correlation Applied in the Evaluation of the Mechanical Behavior of Arterial Tissue*. PhD thesis, Universidade do Porto, 2014.
- [67] Giacomo Lionello, Camille Sirieix, and Massimiliano Baleani. An effective procedure to create a speckle pattern on biological soft tissue for digital image correlation measurements. *Journal of the mechanical behavior of biomedical materials*, 39:1–8, 2014.
- [68] Alessandro Freddi, Giorgio Olmi, and Luca Cristofolini. *Overview of Digital Image Correlation*. Springer Cham, 2015.
- [69] The Advanced Materials Design Group. A practical guide to dic. <https://digitalimagecorrelation.org/>. Accessed on 2023-02-21.
- [70] Giacomo Lionello and Luca Cristofolini. A practical approach to optimizing the preparation of speckle patterns for digital-image correlation. *Measurement Science and Technology*, 25(10):107001, 2014.
- [71] Ming-Tzer Lin, Cosme Furlong, and Chi-Hung Hwang. *Advancement of Optical Methods & Digital Image Correlation in Experimental Mechanics*, volume 2. Springer, 2021.
- [72] Jinfeng Ning, Shaowen Xu, Ying Wang, Susan M Lessner, Michael A Sutton, Kevin Anderson, and Jeffrey E Bischoff. Deformation measurements and material property estimation of mouse carotid artery using a microstructure-based constitutive model. *Journal of biomechanical engineering*, 132(12), 2010.
- [73] Bing Pan. Digital image correlation for surface deformation measurement: historical developments, recent advances and future goals. *Measurement Science and Technology*, 29(8):082001, 2018.

- [74] YL Dong and Bing Pan. A review of speckle pattern fabrication and assessment for digital image correlation. *Experimental Mechanics*, 57:1161–1181, 2017.
- [75] WS LePage, JA Shaw, and SH Daly. Optimum paint sequence for speckle patterns in digital image correlation. *Experimental Techniques*, 41:557–563, 2017.
- [76] Yaroslav Blikharsky, Nadiia Kopiika, Roman Khmil, Jacek Selejdak, and Zinoviy Blikharsky. Review of development and application of digital image correlation method for study of stress–strain state of rc structures. *Applied Sciences*, 12(19):10157, 2022.
- [77] Szabolcs Szalai and Gábor Dogossy. Speckle pattern optimization for dic technologies. *Acta Technica Jaurinensis*, 14(3):228–243, 2021.
- [78] Dongsheng Zhang, Charles D Eggleton, and Dwayne D Arola. Evaluating the mechanical behavior of arterial tissue using digital image correlation. *Experimental mechanics*, 42:409–416, 2002.
- [79] Dongsheng Zhang and Dwayne D Arola. Applications of digital image correlation to biological tissues. *Journal of Biomedical Optics*, 9(4):691–699, 2004.
- [80] Correlated Solutions Inc. Speckle pattern fundamentals. <https://correlated.kayako.com/article/38-speckle-pattern-fundamentals>. Accessed on 2023-03-01.
- [81] Inc. Correlated Solutions. Vic-3d 9 manual and testing guide. <https://correlated.kayako.com/article/60-vic-3d-9-manual-and-testing-guide>. Accessed on 2023-03-01.
- [82] José Davide Lima Rosas. *Implementação de um sistema para medição das deformações no plano pela técnica de correlação digital de imagem*. PhD thesis, Instituto Politecnico do Porto (Portugal), 2019.
- [83] Hieu Nguyen, Hien Kieu, Zhaoyang Wang, and Hanh ND Le. Three-dimensional facial digitization using advanced digital image correlation. *Applied Optics*, 57(9):2188–2196, 2018.
- [84] Hubert Schreier, Jean-José Orteu, Michael A Sutton, et al. *Image correlation for shape, motion and deformation measurements: Basic concepts, theory and applications*, volume 1. Springer, 2009.
- [85] Yufei Wang, David Joannic, Patrick Delassus, Alain Lalande, Patrick Juillion, and Jean-François Fontaine. Comparison of the strain field of abdominal aortic aneurysm measured by magnetic resonance imaging and stereovision: A feasibility study for prediction of the risk of rupture of aortic abdominal aneurysm. *Journal of Biomechanics*, 48(6):1158–1164, 2015.

-
- [86] Smooth-On. Ecoflex™ 00-30. <https://www.smooth-on.com/products/ecoflex-00-30/>. Accessed on 2023-02-30.
- [87] Smooth-On. Ecoflex™ 00-50. <https://www.smooth-on.com/products/ecoflex-00-50/>. Accessed on 2023-02-30.
- [88] Prochima. Goma siliconica trasparente da colata cristal rubber a+b. <https://www.prochima.it/gomma-siliconica-trasparente-da-colata-cristal-rubber-ab/>. Accessed on 2023-02-20.
- [89] Dow Corning. Technical data sheet. <https://www.dow.com/content/dam/dcc/documents/en-us/productdatasheet/11/11-31/11-3184-sylgard-184-elastomer.pdf?iframe=true>. Accessed on 2023-02-20.
- [90] Philipp Seidel. Smart medical phantoms: State of the art in construction and usage of robotic and sensory equipment. Master's thesis, Karlsruhe Institute of Technology Department of Informatics Institute for Anthropomatics and Robotics (IAR) - Health Robotics and Automation (HERA), Germany, 2021.
- [91] Rodrigo Valente. Estudo numérico da interação fluido-estrutura numa aorta ascendente aneurismática. Master's thesis, Universidade NOVA de Lisboa, Lisboa, Portugal, 2021.
- [92] Nusrat Choudhury, Olivier Bouchot, Leonie Rouleau, Dominique Tremblay, Raymond Cartier, Jagdish Butany, Rosaire Mongrain, and Richard L Leask. Local mechanical and structural properties of healthy and diseased human ascending aorta tissue. *Cardiovascular Pathology*, 18(2):83–91, 2009.
- [93] Marwa Selmi, Hafedh Belmabrouk, and Abdullah Bajahzar. Numerical study of the blood flow in a deformable human aorta. *Applied Sciences*, 9(6):1216, 2019.
- [94] Michael Smith. *ABAQUS/Standard User's Manual, Version 6.9*. Dassault Systèmes Simulia Corp, United States, 2009.
- [95] Huda S Basloom. A comparison study between tetrahedral mesh and hexahedral mesh. In *2018 21st Saudi Computer Society National Computer Conference (NCC)*, pages 1–6. IEEE, 2018.