Mechanical characterization of the human atria based on patient-specific FEM models

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Summary

The left atrium (LA) of the heart serves vital roles that exerts profound effect on the left ventricle (LV) filling and subsequently the overall performance of the cardiovascular process. There are common diseases such as atrial fibrillation (AF) and atrial stunning (AS) that alters the effective functioning of the atria. In view of this, access to the LA is very vital to perform several minimally invasive cardiac interventions of the left heart to rectify these abnormalities. Transseptal puncture technique (TSP) is the commonest minimally invasive technique that involves the puncturing of the atrial septum by the use of a catheter inserted in the right atrium (RA) through the venous system under a bi-dimensional fluoroscopic guidance. There are associated complications, possibility of multiple puncturing and procedural failures using this technique even though it has been in existence for several years and the success rate basically depends on the experience of the physician.

To resolve these complications, the exact location at which the septum requires to be traversed needs to be simulated. The objective of this project is to develop technologies to help the physicians when performing TSP technique. Using the finite element method (FEM) software (Abaqus) based on FEM, patient-specific analysis of the mechanical behaviour of the atria, specifically the atrial septal wall will be conducted to allow the effective optimization of the TSP technique.
Abstract

Atrial fibrillation (AF) is the most common cardiac arrhythmia in the world affecting more than 9 million of person per year in Europe. AF is described by a dysfunction of the electric conduction system of the cardiac atria. The traditional treatment requires an ablation of the dysfunctional tissues at the atria using a minimally invasive cardiac intervention, called radio-frequency ablation therapy. The success of this procedure is dependent of the experience of the operator, which could be sub-optimal in abnormal anatomy situations.

During this project, we intend to develop patient-specific models of the atria for accurate procedure planning. Starting from a pre-interventional image cardiac acquisition (Computed Tomography or Magnetic Resonance Imaging (MRI)), a manual delineation of the relevant cardiac structures (left atrium, right atrium, superior vena cava, inferior vena cava and atrial septal wall) will be performed. Secondly, it will be developed a strategy for building the meshes for the anatomic models and tissue mechanical properties will be gather from literature. Using a FEM software, like Abaqus or Ansys, based on the finite element method, patient-specific analysis of the mechanical behavior of the atria, particularly of atrial septal wall, will be conducted.

Keywords: Biomedical Engineering, Cardiovascular diseases, Finite Element Method, Biomechanics.
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<td>Bi-dimensional</td>
</tr>
<tr>
<td>3D</td>
<td>Three-dimensional</td>
</tr>
<tr>
<td>AF</td>
<td>Atrial fibrillation</td>
</tr>
<tr>
<td>AFER</td>
<td>Atrial fibrillation induced electrical remodelling</td>
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<tr>
<td>AP</td>
<td>Action potential</td>
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<td>AS</td>
<td>Atrial stunning</td>
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<td>ASM</td>
<td>Active shape models</td>
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<td>AV</td>
<td>Atrioventricular</td>
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<tr>
<td>BEAS</td>
<td>B-spline Explicit Active Surface</td>
</tr>
<tr>
<td>CEL</td>
<td>Coupled Eulerian-Lagrangian</td>
</tr>
<tr>
<td>CHD</td>
<td>Coronary heart disease</td>
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<td>CS</td>
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<td>Computer tomography angiography</td>
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<td>Electrocardiogram</td>
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<td>ECM</td>
<td>Extracellular matrix</td>
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<td>EM</td>
<td>Electromagnetic</td>
</tr>
<tr>
<td>EVF</td>
<td>Eulerian volume fraction</td>
</tr>
<tr>
<td>FEA</td>
<td>Finite element analysis</td>
</tr>
<tr>
<td>FEM</td>
<td>Finite element method</td>
</tr>
<tr>
<td>FO</td>
<td>Fossa ovalis</td>
</tr>
<tr>
<td>FOV</td>
<td>Field-of-view</td>
</tr>
<tr>
<td>GPU</td>
<td>Graphics processing unit</td>
</tr>
<tr>
<td>IAS</td>
<td>Interatrial septal wall</td>
</tr>
<tr>
<td>IVC</td>
<td>Inferior vena cava</td>
</tr>
<tr>
<td>LA</td>
<td>Left atrium</td>
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<tr>
<td>LV</td>
<td>Left ventricle</td>
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<tr>
<td>MNS</td>
<td>Magnetic navigation system</td>
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<tr>
<td>MR</td>
<td>Magnetic resonance</td>
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<tr>
<td>Abbreviation</td>
<td>Full Form</td>
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<td>--------------</td>
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</tr>
<tr>
<td>MRI</td>
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</tr>
<tr>
<td>MV</td>
<td>Mitral valve</td>
</tr>
<tr>
<td>PCI</td>
<td>Percutaneous cardiac interventions</td>
</tr>
<tr>
<td>RA</td>
<td>Right atrium</td>
</tr>
<tr>
<td>RAO</td>
<td>Right anterior oblique</td>
</tr>
<tr>
<td>ROI</td>
<td>Region of interest</td>
</tr>
<tr>
<td>RV</td>
<td>Right ventricle</td>
</tr>
<tr>
<td>SAV</td>
<td>Stenotic aortic valve</td>
</tr>
<tr>
<td>SVC</td>
<td>Superior vena cava</td>
</tr>
<tr>
<td>TA</td>
<td>Transaortic</td>
</tr>
<tr>
<td>TEE</td>
<td>Transesophageal echocardiography</td>
</tr>
<tr>
<td>TSP</td>
<td>Transseptal puncture</td>
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<tr>
<td>VIC</td>
<td>Valve interstitial cells</td>
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</table>
Chapter 1

Introduction

The human heart is an essential organ of the human body parts that is primarily responsible for pumping blood throughout the body. The blood moves through the circulatory system thereby supplying the tissues with oxygen and nutrients as well as removing carbon dioxide and other forms of wastes from the body. The blood is pumped through the arteries which connect to smaller arterioles and then to the capillaries are much smaller. At the capillaries, exchange of nutrients, dissolved gases, waste products and electrolytes are exchanged between the blood and the surrounding tissue. The venules and the veins constitute the Pulmonary Circuit since they send Deoxygenated blood to the lungs to receive oxygen and offload carbon dioxide. The arteries and arterioles form part of the Systemic Circuit since they carry oxygenated blood and nutrients to the body cells as well as removing body wastes[1].

The human heart is a muscular organ which has four chambers situated towards the left of the midline of the thoracic cavity. It is about 14 cm long and 9 cm wide and it is approximately the size of a closed fist of a man. The atria which is the upper two chambers divided by a wall-like structure called interatrial septum whiles the ventricles which consist of the two lower chambers are also divided by the interventricular septum. There are valves between each atrium and ventricle which allows blood flow in one direction.

1.1 - Structure of the heart

The heart which is hollow and cone-shaped is located in the thoracic cavity resting on the diaphragm. The anterior portion of the heart is close to the sternum whiles the posterior border is nearer to the vertebral column.

The wall of the heart has three main layers the outer Pericardium, the middle Myocardium and the Inner Endocardium as shown in figure 1.1.1. The heart is protected against friction by the connective tissues and adipose tissues that constitutes the epicardium. The myocardium is composed predominantly of cardiac muscle tissues and they are organized in planes. It also has a lot of blood capillaries, lymph capillaries and nerve fibers. The endocardium consists of blood vessels and specialized cardiac muscle fibers called Purkinje Fibers, epithelium and connective tissue with elastic and collagen fibers.

Within the heart are four hollow chambers with two on both the left and right. The left atria and the left ventricle is separated from the right atria and right ventricle by the septum (solid wall-like
structure). The septum prevents the blood from each chamber from mixing. The mature heart valves are composed of highly organized extracellular matrix (ECM) as well as valve interstitial cells (VIC) which are surrounded by endothelial cell layer. The distinct biomedical properties of the leaflets and supporting structures are provided by the ECM of the valves which are stratified into Elastin-rich, proteoglycan-rich and collagen-rich layer [2]. The atrioventricular valve (AV Valve) consists of the mitral valve which is located on the left and tricuspid valve which is on the right. These valves ensure the flow of blood in one direction between the atria and the ventricle. The AV valve is so called because they lie between the atria and the ventricle. The superior vena cava and the inferior vena cava as well as a smaller vein (coronary sinus) supplies the right atrium with blood. The tricuspid valve has cusps which are projections and are located between the right atrium and ventricle. The cusps of the tricuspid valve are attached to the chordae tendineae which are strong fibers that originates from the papillary muscles. The tricuspid valve allows blood to flow only from the right atrium into the right ventricle. The muscular wall of the right ventricle which pumps blood to the lungs is thinner compared to the left ventricle which is responsible for pumping blood to all parts of the body. The right ventricle encounters little resistance during its blood flow compared to the left ventricle. At the trunk’s base is a pulmonary valve with three (3) cusps which only allows blood flow through the right ventricle.

Figure 1.1.1: Cross-section of the heart showing the chambers and the direction of flow of blood [1]

The left atrium receives blood from four (4) pulmonary veins. Blood flows from the left atrium into the left ventricle through the mitral valve which also prevents backflow of blood. The mitral valve directs blood into the Aorta by passively closing when the left ventricle contracts. The aortic valve is located at the base of the aorta. The aortic valve opens to allow the flow of blood from the left ventricle during contraction. The valve closes when the left ventricle relaxes, the valve closes to prevent blood from backing up into ventricle. The pulmonary valve and the aortic valve have “Half-Moon” shapes hence they are known as the semilunar valves.
The right atrium receives deoxygenated (low-oxygen) blood from the vena cava and coronary sinus. During the contraction of the right atrium, the blood passes through the tricuspid valve into the right ventricle. The contraction of the right ventricle causes the Tricuspid Valves to close causing blood to move through the Pulmonary Valve into the pulmonary trunk and pulmonary arteries and further entering the Capillaries of the Alveoli of the lungs for gas exchange to take place.

The oxygenated blood enters the heart through the Pulmonary Veins into the left atrium and as the left atrium contracts, blood moves through the Mitral Valve into the left Ventricle. The Mitral Valve closes when the left ventricle contracts causing blood to move through the Aortic Valve into the Aorta and its branches.

The cardiac muscle fibers have a similar function as the skeletal muscle fibers but these cardiac muscles are connected in branched networks. In view of this, when any part of the network is stimulated, impulses are generated and sent throughout the heart causing it to contract as a single unit.

1.2 - Physiological functions of the heart

The heart is responsible for the circulation of blood to all parts of the body through two pathways: The Pulmonary and Systemic Circuits.

One heartbeat or cardiac cycle involves the atria contraction (Atrial Systole) during which the ventricle relaxes (Ventricular Diastole) followed by Contraction of the Ventricles (ventricular Systole) during which the atrial relaxes (Atrial Diastole). After this, there is a brief period of relaxation of both the atria and the ventricles. This process is well coordinated to ensure that the various actions by the chambers of the heart are effective.

The main parameters responsible for the closure and opening of the valves and the flow of blood through the heart chambers during one cardiac cycle are the changes in the pressure and volume in the heart chamber. A single heartbeat makes a double thumping sound which is heard during the use of the stethoscope. This sound is heard due to the vibrations of the heart tissues related to the closure of the Valves. The first sound is heard when the ventricle contracts when the AV valve closes and the second sound is heard when the ventricle relaxes when the Pulmonary and Aortic valves closes as shown in the figure 1.2.1 below. Table 1.2.1 is a summary of the heart valves and their functions.
<table>
<thead>
<tr>
<th>Heart Valve</th>
<th>Location</th>
<th>Action</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tricuspid valve</td>
<td>Between right atrium and right ventricle</td>
<td>During ventricular contraction, it prevents blood from moving from right ventricle into the right atrium</td>
</tr>
<tr>
<td>Pulmonary valve</td>
<td>At entrance to pulmonary trunk</td>
<td>During ventricular relaxation, it prevents blood from moving from pulmonary trunk into right ventricle</td>
</tr>
<tr>
<td>Mitral (bicuspid) valve</td>
<td>Between left atrium and left ventricle</td>
<td>During ventricular contraction, it prevents blood from moving from left ventricle into left atrium</td>
</tr>
<tr>
<td>Aortic valve</td>
<td>At entrance to aorta</td>
<td>During ventricular relaxation, it prevents blood from moving from aorta into the left ventricle</td>
</tr>
</tbody>
</table>

Table 1.2.1: Heart valves and their functions [1]

![Ventricular pressure-volume relationship](image)

Figure 1.2.1: Ventricular pressure-volume relationship [3]

1.3 - Physiological structure of the atrium

Right atrium (RA) has a larger volume coupled with thinner walls (approximately 2mm) compared with the left atrium (LA). The right atrium (RA) is both anterior and to the right of the left
atrium (LA) and extends inferior to it. The walls of the RA forms the right superior surface, the convex right (pulmonary) surface and little of the right side of the anatomical base. The superior RA is composed of the superior vena cava (SVC) and the right atrial appendage which is an extensive muscular pouch which projects anteriorly to overlap the right portion of the aorta (figure 1.3.1). The inferior surface of the RA can be divided into three (3) regions: the smooth-walled venous component, tricuspid valve and the appendage. The SVC receives blood from the superior portions of the body (head, neck and upper limbs), chest walls and the oesophagus. The inferior vena cava (IVC) however drains blood from all structures below (inferior) including the diaphragm into the lowest part of the atrium near the septum. The IVC is larger than the SVC. Blood circulation between the RA and the right ventricle (RV) is controlled by the tricuspid valve. The coronary sinus (CS) opens into the venous atrial component between the orifice of the IVC, the fossa ovalis and the vestibule. CS is a collection of vessels that recieves blood the myocardium draining into the RA[4].

The LA serves three main roles that exerts profound effect on the left ventricle (LV) filling as well as the overall cardiovascular performance. It is a contractile chamber that actively empties immediately before the onset of LV systole and establishes final LV end-diastolic volume[5]. LA is also a reservoir that stores pulmonary venous returns during LV contraction and isovolumic relaxation after the closure and before the opening of the mitral valve[6]. Finally, it is a conduit that empties its contents into the LV down a pressure gradient after the opening of the mitral valve[7] also passively transfer pulmonary venous blood flow during LV diastasis. LA has thicker walls (approximately 3cm) hence smaller with respect to the RA. LA is cuboidal in shape with four (4) pulmonary veins and atrial appendage at the superior as shown in figure 1.3.1. At the inferior, the mitral valve is associated with controlling the blood circulation to the left heart[8]. It must also be noted that the aorta artery and the pulmonary artery cover externally the LA and the thin fibrous pericardium seperates the LA from the oesophagus[9].

The interatrial septal wall (IAS) is a muscular structure found between the LA and the RA. It is composed of the fusion of the septum primum (LA septum) and the septum secundum (RA septum)[10]. The region where they fuse is known as the limbus (the area of fusion of the muscular septum secundum and the thin portion of the septum primum) which is the thickest region which has a depression called fossa ovalis (FO) in its middle. Both septae extend from the roof of the atria toward the endocardial cushions. The septum primum, which is the LA septum is absorbed superiorly leaving the septum secundum or the RA septum to cover this superior defect and separate the atria [11]. FO is oval or circular shaped that can only be identified from the RA. It has an average area of 1.5 to 2.4 cm² located posteriorly at the junction of the mid and lower third of the RA precisely between the IVC and CS[10]. It is also composed of thin fibrous tissues and it is the thinnest region of the IAS[12]. Patients with dilated aortae and those with mitral valve disease and a bulging LA, the FO can be located more superiorly or inferiorly respectively [13].
1.4 - Common diseases associated with the heart

Globally, the most common cause of death is cardiovascular disease (CVD). In Europe, it is responsible for about 45% of all death which amounts to over four (4) million death per year with coronary heart disease (CHD) being the most common single cause of death forming about 19% and 20% of deaths in men and women respectively far more than deaths associated with cancer[14]. It is also noteworthy that CVDs causes about 40% and 49% of all death among men and women respectively in Europe. The death ratio of CVDs increases with age and less common in younger age groups. CHDs is the type of cardiac disease in which a waxy substance referred to as plaque builds up gradually within the coronary arteries (responsible for supplying oxygenated blood to the heart muscles) and with time ruptures or hardens reducing the flow of blood through the arteries [15].

There are several diseases associated with the human heart and since the four (4) chambers of the heart are independent they thus have different associated cardiac diseases with each chamber. Some of the commonest heart diseases are inflammatory heart disease, Ischemic heart disease, hypertensive heart disease just to mention but a few. In this work, I shall dwell on the diseases associated with the atrium in accordance with my motivation, the existing procedures to correct or cure these diseases as well as the deficiencies of the existing corrective procedures. Valve disease (Stenosis or reguritation) is a significant public health problem which results in approximately 20,000 deaths annually[16].

Atrial fibrillation (AF) is the type of arrhythmia where the atria beat irregularly and usually out co-ordination with respect to the ventricles[17]. AF is the commonest acquired cardiac
and the possibility of acquiring this disease increases with age precisely each decade of adult life[20], [21]. There are several plausible causes of AF which can either be cardiac or non-cardiac diseases which may cause abnormalities in cardiac electrophysiology thus serving as a substrate for the development of arrhythmia[22].

Another atrial disease is the atrial stunning (AS) which is basically the loss of mechanical atrial contraction following a successful cardioversion which usually take up to about six (6) weeks to regain normal atrial contraction[23]. A longer period of AS may result in an increased rate or risk of thromboembolism. AS rarely happens following spontaneous cardioversion in paroxysmal arrhythmias. The duration of the atrial fibrillation, atrial pressures and size as well as the presence of structural heart diseases are the main factors delaying the return of the normal atrial mechanical functioning according to research. The main mechanisms generating the impaired atrial mechanics in AS are however unknown[24], [25].

The intrinsic electrophysiological attributes of the atria are transformed during chronic AF due to the atrial fibrillation induced electrical remodelling(AFER)[26]-[30]. According to research and various clinical electrophysiological and experimental studies have shown that electrical remodelling is categorised by an abbreviated atrial action potential(AP) morphology which is usually associated with the underlying changes to dnesity as well as kinetics of some membrane ionic currents[27]-[29], [31]-[33].

1.5 - Existing treatment techniques of atrial associated diseases

Rate and rhythm control drugs are the first means of controlling atrial diseases such as AF but when this is ineffective[17], then access to the left atrium is required to resolve the defect. The initial methods such as transbronchial, supraclavicular [34] of measuring the LA pressure were first developed in the 1950s and they were mainly a direct method of access. Despite the acceptable feasibility and safety of these methods [35], they were replaced by the indirect measurements of the LA pressure with the Swan-Ganz catheter [13]. Percutaneous access to the LA is mostly required in various minimally invasive procedures which includes catheter ablation for AF, percutaneous mitral valve replacement as well as left atrial appendage closure[13], [36].

Percutaneous cardiac interventions (PCI) covers the minimally invasive procedures to access the left atrium done with the help of the blood circulation system. The open chest which was initially used required a longer time frame, a long surgical cut in the chest organ whereas the minimally invasive procedures require only a small puncture through the skin to have access to the vascular system and then to the portion of interest (LA). Minimally invasive technique has a less post-operative pain, faster, involves less blood loss, less expensive and has a fewer procedure complications[10].

It is however impossible to have a direct visualization of the target when employing the PCI due to the use of the vascular access and manipulating surgical instruments is sometimes awkward. In view of this, special transcatheter surgical tools and imaging modalities such as ultrasound, flouroscopy
and computer tomography (CT) are employed to ensure a safe medical procedure and also ease the difficulty of manipulating the instruments.

1.6 - Accessing the left atrium

Evaluation of heart diseases are mostly done by the measurement of the LA pressure hence the need to be able to get access to the LA. Two main techniques are usually employed in accessing the LA namely; transaortic (TA) and transseptal puncture (TSP). A research performed to compare the two main techniques using a magnetic navigation system (MNS) in the USA revealed that they are equal in success rate and procedure time using the MNS for the left-sided AP ablation[37].

When using the TA technique, a catheter inserted in the femoral artery is retrogradely advanced towards the left ventricle through the aortic valve. The catheter is then rotated about an angle of 180º and moved through the mitral valve (MV) into the LA chamber as shown in the figure 1.6.1(a). Due to the 180º rotation associated with this technique, it becomes difficult to manipulate the catheter thus hindering the effective performance of this procedure. This has led to the frequent use of the TSP in recent times[38].

TSP however requires a catheter being inserted into the right atrium (RA) through the venous system. A needle is then moved through it (RA) to puncture the interatrial septum (IAS) and ultimately to get access to the LA. For ease of accessing the LA using the TSP, it is advisable to perform the puncture through the thinnest region of the IAS wall known as the fossa ovalis (FO).

In view of this, the TSP technique is noted to give a more direct access pathway to the LA compared to the TA technique[15] (figure 1.6.1 (b)). In as much as this technique has been used for several years, it is still associated with a number of serious complications and/or failures mainly due to abnormal anatomy structure of certain patient’s atrium as well as complex circumstances as perforating large vessels[38].

![Figure 1.6.1: Schematic of transaortic route (a) and transseptal puncture technique (b) (adapted by Pedro André Gonçalves Morais et al from [39])]
1.7 - Transseptal Puncture Technique (TSP)

With the introduction of new procedures for percutaneous structural heart disease therapy [13], [40] and atrial fibrillation ablation [41] in the last five (5) years has necessitated the need for TSP. There has been only little modifications of the existing since the initially proposed one in 1959 [42]. In this procedure, a needle is inserted into a catheter and the catheter is thus guided into the left atrium through a transseptal puncture from the RA septum through the FO. It is worth noting that the transseptal puncture reduces the manouevrability of the catheter making it difficult in reaching the desired location of the left atrium[17]. As stated earlier, the FO is the thinnest portion of the IAS, it is the best route for the transseptal access to the LA. A thorough understanding of the anatomy of the FO and the surrounding landmarks play a vital role in having a successful procedure. Localization of the FO for TSP has traditionally been done or guided using the bi-dimensional fluoroscopy imaging. Auxillary catheters are sometimes used to prevent puncture of vital structures and the time duration of this practice usually lasts for about fifteen (15) minutes [43], [44], [45].

TSP begins with an insertion of the needle delivery catheter (guidewire) (0.81-0.89 mm) and dilator to the level of the SVC through the right common femoral vein with the help of the imaging techniques mentioned earlier. The guidewire is to define a safe route from the femoral vein to the SVC and also to position the dilator and sheath. The needle is then inserted into the dilator and allowed to freely rotate as it is advanced. Fluoroscopy is used to visualize the needle as it is advanced up to the tip of the dilator within the sheath to avoid unintentional passage out of the dilator and the sheath. Passage of the needle is easier if the sheath and dilator is introduced from the right other than the left femoral vein due to tortuosity (many twists).

Brockenbrough needle is the commonly used needle (Medtronic) [13] and it is an 18-guage hollow tube that tapers distally to 21-guage. Its proximal end has a flange with an arrow that points to the position of the tip of the needle. Now that that the needle tip has been inserted into the dilator as shown in figure 1.7.2, the entire assembly (needle, dilator, sheath) is rotated such that the needle points to a 4 o’clock orientation (the ceiling of the room is classified as 12 o’clock orientation where as the floor is a 6 o’clock orientation) and withdrawn into the mid RA. With the help of the fluoroscopic projections (right anterior oblique (RAO)), the dilator is then advanced to meet the limbus of the FO (echocardiographic guidance is employed for patients without prominent limbus). The entire assembly is then advanced as a unit to tent the fossa. Two movements are usually detected: the first one indicates the entrance of the assembly into the RA and the second which indicates the puncture into the FO region is less perceptible. The needle is then rotated to 3 o’clock to avoid posterior puncture and fully advanced. Due to the high risk of vital structures being perforated and limitation in the maneuverability of the catheter in the LA if the puncture is done outside the FO region requires that a comprehensive confirmation of the needle position should be considered [9] not forgetting the aorta, CS and His Bundle positions in order to have a safe route for the puncture. Pressure transduction through the needle is an indication that the needle has been able to puncture the LA as shown in figure 1.7.1.
In about 20% to 25% of adult patients, the FO is probe patent and may not require needle puncture [11], [46]. In about two-thirds (2/3) of patients, the FO is paper thin and the catheter can be passed into the LA with just a gentle pressure and rotation of the dilator [11]. Generally speaking, the overall serious complications associated with TSP are ≤1% [34], [47], [48]. Physicians should note that: aortic root puncture, arterial air embolism, pericardial tamponade, right or left atrial wall puncture, transient ST-segment elevation, pleuritic chest pain, persistence of atrial septal defect and death are possible complications that can be caused by this procedure [38], [44], [49]-[54]. After a successful puncture of the atrial septum, the patient should be anticoagulated immediately to minimize the risk of thromboembolism.

Figure 1.7.2: TSP technique. (a) A dilator and sheath placed into the SVC using the guidewire; (b) A needle is inserted into the dilator until (c) the SVC; the needle is pulled down and two movements are detected: (d) entrance into the RA and (e) entrance into the FO; (f) after the FO identification, the puncture is performed (adapted from: http://www.baylismedical.com/physicians/nrg-rf-transseptal-needle/ (23/05/2016) )
Chapter 2

Literature review

Access to the LA using the TSP technique was initially proposed in 1959 and has ever since been the standard approach for several left heart interventions [13], [55]. Due to the use of a transcatheter, non-invasive image acquisition of the patient anatomy is very vital coupled with pre-procedural planning based on high-resolution imaging techniques to assist in having a safe puncture site and define the best puncture route.

This section is aimed at presenting an overview of pre-planning procedure techniques mainly generation of patient-specific anatomical models as well as biomechanical simulation techniques.

2.1 Biomechanical Simulation

Novel computational or biomechanical models relevant to clinical practice has been achieved due to the recent advances in non-invasive high-resolution image acquisition techniques making them feasible to generate accurate patient-specific anatomical models.

Initially these novel image techniques were used to develop generic simulation models capable to provide useful anatomical information with respect to the atrial conduction system [56]-[58], motion pattern characterization [59], [60], study of different pathologies [61] and hemodynamics of the heart [62]. Since generic models were used without taking into account the individual information of each patient limited the application of this proposed solution in the clinical practice. With time, a suggestion to develop complex methodologies resulting in the generation of accurate patient-specific models [63]-[67]. This model is aimed at effectively combining the above mentioned generic models with the patient-specific data acquired from the either multiple imaging techniques or clinical measurements.

Some notable instances are the proposed framework to simulate the procedure used to treat mitral valve (MV) regurgitation using a minimally invasive cardiac procedure as proposed by Mansi et al [67]. A patient-specific model of the MV was generated from an ultrasound image of which a finite element method was used to simulate the closure of the valve as well as the correction procedures. This patient-specific generated MV model was then tested on eleven (11) patients and there was an average point-to-mesh error of 1.47± 0.27 mm. The entire simulation framework was however tested on only one patient with results qualitatively very similar to the actual surgical outcome [67]. Furthermore, Stevanella et al. proposed a finite element model of the MV to predict the results of
mitral annuloplasty procedures using an MRI data after which it was tested on two patients (one healthy patient and one sick patient) giving similar results compared with the real procedure being performed [68].

Soon, Wang et al. developed a patient-specific model to quantify and characterize the interaction between the trans-catheter stent and the stenotic aortic valve (SAV) [69]. The anatomic model of the patient-specific finite element model was generated from a CT data applied as anisotropic hyperelastic material to simulate the tissue mechanical properties. An eight-node hexahedral was used to generate the solid element of the trans-catheter whereas a four-node quadrilateral element was employed to depict the balloon portion. This proposed method was implemented on one patient by simulating the entire valve-replacement procedure which was a success. The authors were able to prove by this that it was possible to use this approach to extract relevant pre-planning information and determining the stent position.

Furthermore, Morganti et al. also proposed a new framework to effectively simulate the trans-catheter aortic valve implantation [70]. A CT imaging technique was used to obtain the aortic model including the native structure of the valve (leaflet and calcific plaques). In the definition of the mechanical properties of the valve in the simulation programme, isotropic and homogeneous materials were used. Commercially available Edwards SAPIEN valve were used by the authors to represent the prosthesis and generating its structure from a micro-CT image. Also, Von Mises plasticity model with isotropic hardening was used to represent the mechanical parameters of the prosthesis being simulated. This was tested on two (2) patients and the obtained results suggested that the proposed simulator could be used effectively for realistic minimal invasive procedure simulation, associated risks of post-procedural complications mainly paravalvular leakage and the presentation of the patient-specific stress distribution of the aortic wall [70].

Lastly, Jayender et al. proposed an approach to estimate the optimal puncture location by successfully combining the pre-interventional models with a mechanical model of the catheter to be used for the LA intervention [17]. The pre-interventional models were obtained from CT image datasets through a semi-automatic strategy. The model of the catheter was pictured to be made of negligible rigid links along a backbone curve and in view of this, the optimal puncturing site could easily be predicted based on the thickness of the septal wall and the mechanical maneuverability of the catheter at all the positions of the LA. This model lacked exhaustive validation due to the fact that there are numerous types of catheters that can be used in this procedure (puncture the septal wall) hence the need for different catheter models.

### 2.2 Patient-specific anatomical models

Researchers have done so much to assist physicians throughout the pre-procedural planning by proposing automatic and semi-automatic strategies to generate patient-specific anatomical models
from imaging techniques such as CT, MRI and ultrasound. This section is aimed at investigating the patient-specific anatomical models of the atria.

Daoudi et al. presented a deformable model to segment the LA in 2D CT [71]. It begins with contrast enhancement based on adaptive histogram equalization followed by morphological operators and a region growing technique to generate a coarse contour to the LA chamber. Finally, a refining stage is done using gradient vector flow technique. Visual assessment of this method was tested on 20 CT datasets. It must also be noted that relevant clinical indicators were not extracted from these images due to the bi-dimensional segmentation performed.

Haak et al. also suggested a framework based on active shape models (ASM) for segmenting multiple heart chambers in 3D ultrasound imaging [72]. Ideally, ultrasound images show limited field-of-view (FOV) hence the wide-view ultrasound images were manually generated using several individual ultrasound records. These wide-view images were posteriorly semi-automatically segmented using the following steps: 1) estimation of the heart pose; 2) heart pose and shape estimation; 3) refinement of the contours obtained in each chamber. Each step required a gamma mixture model to generate a blood-tissue probability map which was subsequently fitted with an ASM model of the heart chambers eventually generating the optimal contour. With regards to validation, the single ultrasound image and the wide-view ultrasound images were segmented using the abovementioned techniques which showed a reasonable improvement of Dice coefficient for the merged data.

Also, Almeida et al. proposed a semi-automatic strategy to define the LA chamber in the ultrasound image [73]. This framework was focused on the B-spline Explicit Active Surface (BEAS) which was initially proposed for the left ventricle (LV) segmentation in ultrasound and MRI data [74], [75] which resulted in a reduced computational time as well as high accuracy. Due to the LA not having a less regular shape compared with the LV an adaptation of the BEAS was required and implemented. This was then tested on twenty (20) volumetric sequences and the results proved that the LA functional parameters can be derived from the semi-automatic contours. It is worth noting that this technique less validation.

Zheng et al. in a Siemens Corporate Research proposed a novel machine learning strategy to automatically segment the whole-heart in 3D CT volumes [76]. The proposed technique known as marginal space learning identifies the optimal heart pose through a classification approach. A 9-dimensional vector with position, orientation and scaling was used to train the classifier method and generate, consequently a full parameter space with a large number of hypotheses (Figure 2.2.1). During the testing stage, a multi-stage strategy was used to identify the optimal pose for each chamber. This technique starts with a restricted number of possibilities and increases the dimensionality of the problem in each stage. Finally, a mean shaped model of the heart was deformed until the estimated optimal heart pose was achieved and the final segmentation was generated as shown in figure 2.2.1. A total of 323 3D-CT volumes were used in the experimental validation presenting surface-to-surface errors lower than 1.6mm for all the structures. A total of 2 seconds per volume were required with regards to the computational time [76].
Margeta et al. presented a supervised learning method for fully automatic left atrium segmentation from 3D cardiac MRI datasets based on machine learning techniques [77]. Blood-pool region extraction through a simple threshold commences this procedure but because intensity homogeneity was observed between the different chambers, multiple cardiac structures were detected. This technique was tested on 10 different databases using a leave-one-out approach and the results obtained showed an unsatisfactory dice coefficient lower than 70% indicating that this procedure was unable to delineate the entire cavity.

Zuluaga et al. focused on a multi-atlas approach to accurately segment the heart from 3D MRI and 3D computer tomography angiography (CTA) sequences [78] as shown in figure 2.2.2. The method uses two (2) steps: 1) localization of the region of interest (ROI) using affine alignment between the unseen image and the atlas as shown in (1) and (2) of figure 2.2.2. 2) The second step is the whole-heart segmentation based on non-rigid registration between the resulting ROI and the atlas as shown in (3-5) of figure 2.2.2. The obtained deformation fields were posteriorly combined to transform the labels from the atlas to the unknown image and consequently generate the final contour. This procedure requires a high computational time due to the set of non-rigid alignments required to delineate the final contour. It was tested on 23 and 8 MRI and CTA datasets respectively and a resulting dice scores of 90.8% and 89% respectively for the whole-heart chambers were obtained making this procedure well valid. A total of 30 and 60 minutes were required for the magnetic resonance (MR) and CTA datasets regarding the computational time respectively.

Similarly, Kirisili et al. proposed a multi-atlas-based approach to segment the whole heart from CTA data. The current strategy was validated into a large-scale multicenter and multivendor study with a total of 1380 datasets [79]. A total of 8 labeled CTA datasets were used to generate the atlas. Evaluation of the results per the experts showed 49% of the cases were very accurately segmented with associated error below 1mm whiles 29% were accurately segmented with an error between 2 and 3 mm. These results demonstrated the accuracy and robustness of this technique. Eight (8) fully segmented datasets were used to estimate the surface-to-surface error and the error obtained was
lower than 1.5mm for the various chambers even though a computational time of approximately 20 minutes per volume was used.

Ecabert et al. presented a new strategy for fully segment the whole heart automatically in a 3D CT [80]. The proposed technique has two (2) main stages namely: 1) heart localization (figure 2.2.3a); and 2) segmentation refinement using a deformable model (figure 2.2.3 b-d).

Figure 2.2.2: Schematic used by atlas-based approaches to segment MR images. It uses an affine registration (1) to roughly align the unseen image with the atlas. The obtained transformations are then posteriorly used to map the label images (2) into the unseen image and generate a region-of-interest through a majority vote strategy (3). Using only this region, a non-rigid registration (4) is used to align the different datasets. The resulting deformation fields are used to transform the labels and generate the final contour (5) [78].

Figure 2.2.3: Ecabert et al. methodology: a) A rough contour is initially estimated through a generalized Hough transform approach. In (b) and (c) a global and local adaptation of the model is performed, respectively. Finally, in (d) a deformable model strategy is used to refine the contour [80].

The first phase which is the heart localization method was performed using an adapted 3D generalized Hough transform due to its high robustness and versatility to detect any arbitrary shape in the target image presented by using this procedure (figure 2.2.3 a). The second phase requires a shape-constrained deformable model being applied to refine the whole-heart mesh. Two alternating steps (parametric and deformable adaptation) were used to perform the refinement stage.
Parametric adaptation starts with a global resizing (figure 2.2.3 b) of the mesh followed by a local adaptation of each contour (figure 2.2.3 c). The deformable adaptation was applied to basically guarantee optimal fit between the resulting mesh and the patient anatomy (figure 2.2.3 d). This approach was tested on 37 3D-CT datasets with surface-to-surface errors less than 1 mm in all the cardiac structures with a computational time of about 22 seconds per dataset.

### 2.3 Image-fusion and catheter tracking techniques

The pre-procedural planning data needed to be combined with intra-procedural information such as image acquisition in order to safely guide the physician during the entire procedure. This section gives an overview of image-fusion and catheter tracking techniques.

Cleary et al. in their research on image-guided interventions suggested the use of an external tracking hardware such as electromagnetic sensor or optical infrared systems [81]. These systems are coupled with the surgical catheter being consequently used to combine pre-procedural and intra-procedural data through series of markers. This procedure has a complex initial setup and the surrounding surgical equipment may generate electromagnetic interference.

Jeevan et al. suggested the integration of electromagnetic (EM) sensors on the tip of the catheter based on the above described technique to assist and guide the TSP procedure [82]. The catheter was rigidly aligned with a patient-specific atria geometry obtained from a pre-interventional MRI. To validate this technique, it was tested on a phantom model and the results showed that it (procedure) reduces procedure time, has no learning curve and less complicated. It must however be noted that this testing was performed in static models without any real-time image acquisition such as x-ray or ultrasound imaging hence it cannot be implemented in real life situations.

Also, Hatt et al. researched into an EM-tracking technique to effectively fuse X-ray, ultrasound and MRI images [83]. MRI is acquired during the planning stage whereas the X-ray and ultrasound are real-time imaging modalities that are required throughout the intervention. This proposed technique has two main pre-intervention calibrations which are firstly, the calibration between the EM and the ultrasound probe and secondly, the calibration between the EM world and X-ray image. In calibrating the EM and the ultrasound probe, two (2) EM sensors were coupled with the ultrasound probe and a surgical needle respectively. Ultrasound image acquisition of the surgical needle posteriorly was performed and the calibration was done using the difference in distance between the tool position into the ultrasound image and the actual distance measured by the EM sensor. The second calibration was done using two-custom-built phantoms with metal beads to calibrate the fluoroscopy image with the EM sensor. The difference between the metal beads measured through the fluoroscopy and EM sensor was used to determine the optimal transformation between these two systems. Fiducial markers were used to perform the intra-operative calibration of the MRI and EM system using a set of markers externally placed on the patient and they are easily detected in the MRI image. The EM sensor was used to generate a 3D-world with the different position of these markers. The position
differences measured through MRI and EM sensor were used to align the different worlds. This procedure was tested on phantom and animal models with maximum accuracy error less than 5mm [83].

Lang et al. proposed a new image strategy to register ultrasound, CT and fluoroscopy imaging without any tracking device in their research to assess accuracy [84]. This procedure involves: 1) alignment between the probe position and the transesophageal echocardiography (TEE) world using a calibration cage (Figure 2.3.1(a)); and 2) alignment between the probe position and the fluoroscopy world using a set of tracking beads coupled with ultrasound probe as shown in Figure 2.3.1(b). This allowed a transformation that mapped the ultrasound world into the fluoroscopy world.

Figure 2.3.1: (a) Calibration cage used to align ultrasound world with probe position and (b) radiopaque markers to fuse X-ray and ultrasound world [84].

A semi-automatic segmentation is applied to delineate the relevant structure from the 3D CT data. Using an iterative closest point approach, a set of manual landmarks are used to register the obtained mesh. Since a transformation between the ultrasound and fluoroscopy worlds were obtained during the calibration stage, the patient-specific data acquired from the CT can be also transferred for the fluoroscopy world [84]. It was tested on excised porcine hearts datasets with an appreciable accuracy of 2.6 mm for tracked ultrasound to CT. It is difficult to detect the required landmarks when using the ultrasound imaging hence high registration errors between CT and ultrasound can be attained which indicates the fact that the proposed method is dependent on the user input.

Housden et al. also proposed a technique to align CT, ultrasound and fluoroscopy images [85]. An initial calibration between the probe and the TEE data through a calibration cage (Figure 2.3.2(1)). A semi-automatic strategy was then proposed to align the fluoroscopy with the ultrasound worlds. A high-resolution surface model of the target structure was generated (Figure 9) after initial alignment. A manual segmentation of the esophagus centerline was created and included into the abovementioned high-resolution model. A simple downhill iterative optimization algorithm was used to align the high-resolution model with the ultrasound data (Figure 2.3.2(2)) using the esophagus position was used to constrain the optimization.
Figure 2.3.2: Workflow used to align MRI or CT, X-ray and ultrasound images. Pre-procedure phase; a CT or MRI image is acquired, automatically segmented and converted to a surface model. The esophagus centreline is manually segmented. During the procedure, an ultrasound volume is first acquired and registered to X-ray space via the TEE probe. The pre-procedure surface model is then manually positioned in the X-ray view before being automatically registered via the ultrasound image. The experiments in this paper evaluate intra-procedure stage 2 [85].

Furthermore, Huang et al. proposed a different solution regarding the alignment of pre-operative data, 3D CT or 3D MRI with 2D ultrasound imaging [86]. A rigid transformation was used to spatially register both datasets while their time alignment was obtained using a simultaneously recorded ECG. A periodic heart motion is presumed neglecting the heart deformations caused by the respiration and surgical procedures because the procedure uses only rigid transformations. An accuracy of about 1.7±0.4 mm was obtained when this technique was tested on a beating heart phantom as well as animal models.

A strategy to align a 3D TEE data with fluoroscopy imaging was proposed by Gao et al [87]. This procedure commences with the automatic ultrasound probe identification on the X-ray image using a graphics processing unit (GPU)-based image registration between a 3D virtual model of the probe and the 2D fluoroscopy image. After this stage, a pre-interventional calibration of the ultrasound probe and the ultrasound image, a transformation map between the fluoroscopy world and the ultrasound world can be obtained. A real heart phantom was used in testing this proposed method and the target registration error lower than 2 mm. Later, an offline patient dataset was used and the measured mean registration error was between 1.5 to 4.2 mm. The probe estimation is a very time-consuming process of about 2 to 15 seconds hindering their application in real-time procedures and hence failing to track the cardiac structure throughout the respiratory cycle. This procedure also does not have exhaustive online and offline validation [87].

Lang et al. in their research to improve their previous proposal came up with a new method to align ultrasound and CT or MRI data. To perform this procedure, two different registration techniques were described and compared. These techniques are the surface-based registration and image
registration [88]. Surface-based registration relies on segmentation technique based on continuous max-flow algorithm to delineate the relevant structure in CT or MRI and ultrasound, followed by a mesh alignment based on ICP method. The image registration technique, a non-rigid registration with mutual information metric was used to develop the image-alignment. Also, a GPU implementation of the tracking method was used. Registration errors lower than 2.5 mm, dice coefficients higher than 80% and a low computational time were obtained with surface and image registration approaches proving that both approaches have the potential to be used in image-guidance procedures after 18 datasets were tested. Catheter tracking techniques are required to identify the surgical equipment into the ultrasound imaging since fluoroscopy imaging was not image [88].

A different technique was proposed by Grbic et al. to align pre-operative 3D-CT and intra-operative rotational-angiography based on anatomical structure position [89]. The trachea bifurcation which is visible in both modalities without contrast is used to select surrogate anatomical structure. In view of this, a probabilistic boosting classifier is used to estimate the global position of the trachea bifurcation of each image modality. A rigid-registration between the two meshes is performed in order to align the CT and rotational-angiography worlds. After testing the procedure on 28 patient datasets an accuracy of 7.57±3.22 mm.

A C-Arm technology can be employed to acquire 3D CT volumes and 2D fluoroscopy images that are intrinsically registered instead of using a multi-machine image acquisitions. 3D CT imaging is only acquired in one temporal moment hence variations in the patient anatomy due to the insertion of the catheter are not accounted for. To resolve this deficiency, Liao et al. proposed a fully-automated solution to align the bi-dimensional X-ray images with the 3D datasets during the entire procedure [90] specifically procedures that emits agent contrasts. This procedure easily detects the contrast injection on the X-ray images based on histogram analysis as well as a possibility ratio test (likelihood ratio test). An optimized alignment on the basis of rigid transformation is subsequently performed between the contrast-based fluoroscopy image and a set of bi-dimensional (2D) digitally reconstructed radiographs (DRR) extracted from the 3D CT volumes using diverse plane orientations. After testing this technique on 34 datasets, a mean registration error of 0.66±0.47 mm and computational time of 2.5 seconds per alignment were obtained. It is still far off from being used in real-time procedures because it did not cater for the dynamics of the heart due to use of rigidly aligned static 3D volume.

All the above mentioned proposed techniques aimed at combining the pre-interventional and intra-interventional data did not take into account the surgical instruments used in this procedures making it difficult to implement them in real-time image-guidance procedures. Therefore, Brost et al. based their research on solutions to identify and track surgical catheter during fluoroscopy imaging to effectively assist the expert in numerous minimally invasive interventions such as catheter ablation and TSP [91]. A machine learning technique was used to segment the catheter structure and the authors presented a real-time strategy to identify the catheter through a cascade of boosted classifier combined with haar-features. A consecutive segmentation approach was employed in tracking the catheter after which the procedure was tested on 12 offline datasets and the obtained results showed a tracking error less than 0.7 mm [91].
In Buck et al. research, they proposed a novel technique to track the target catheter from X-ray images [92]. This procedure relies on a bi-dimensional template-matching obtained from a virtual 3D cylindrical model with a rounded tip projection coupled with Kalman filters to estimate the catheter tip in order to reduce the search space. This was also tested on about 14 fluoroscopy sequences and a maximum tracking error of 1.7 mm was obtained. Further clinical validations were performed on this technique and it was well accepted by the experts. However, this was short lived due to the introduction of multiple catheters which introduced more tracking errors.

Wu et al. solved the challenges faced by using ultrasound imaging to track and identify catheters by proposing a four-phase technique which includes:

1. Automatic or semi-automatic catheter identification in X-ray imaging based on speeded up robust features (SURF) and Frangi vesselness filter
2. Catheter tracking in X-ray imaging using Kalman filters
3. Fast registration of the X-ray and ultrasound imaging based on ultrasound probe position in the X-ray images
4. Catheter segmentation/tracking in ultrasound images using the displacement field estimated in phase 2 [93].

This was tested on 5 porcine models and 4 patient datasets and the associated catheter tracking error was lower than 2 mm. The computational time was estimated to be 1.3 seconds per frame making it difficult to implement it in real-time procedures [93].
Chapter 3

Materials and Methods

Needle insertion is a very important part of numerous clinical procedures [94] and in view of this, several researches have been made to develop ways to effectively guide the tip of a needle into the target region during a number of percutaneous interventions in anesthesia, biopsy and brachytherapy [95]. This can be classified into modeling insertion forces [96], [97], needle deflection [98], [99] and tissue deformation [100], [101] as well as developing instruments to guide and steer the needle. The success of these procedures mostly depend on the ability to maneuver the tip of the needle to reach the target thus avoiding bones and other critical organs [102]. A combination of empirical and analytical modeling precisely Finite element analysis (FEA) can be employed to effectively generate a more complete model [95]. A few models have been generated but has never been used clinically for human tissues [103] chiefly due to complex physical phenomena (example high deformation in biological tissues [95]) making it almost impossible to accurately estimate associated displacements.

The Coupled Eulerian-Lagrangian (CEL) analysis was used in this practical work due to its ability to analyze deformation or flow of a continuum. If a continuum deforms or flows, the position of the small volumetric elements changes with time. CEL describes these positions basically in two (2) ways:

1. The movement of the continuum specified as a function of its original coordinates and time known as the Lagrangian description.
2. The movement of the continuum specified as a function of its instantaneous position and time known as the Eulerian description.

In Lagrangian formulation simulations, the interface between two points or parts are precisely defined and tracked thus large deformations may sometimes result in severe mesh and element distortions whereas Eulerian simulations have an Eulerian reference mesh which remains undistorted and unmoved used to trace movement of the particles or material being analyzed. This method (CEL) was used due to the fact that no element distortions occur during simulations resulting in a more accurate displacement prediction. The main disadvantage of using this method is that the interface between the two parts can not be described precisely if only the Lagrangian formulation was used [104].

Two main simulations were done using this analytical method. The first was to model the moulding process of a polymeric bottle (generating a plastic bottle by moulding melt polymer within rigid frames) and secondly, a simulation of the deformation of a simple block under pressure.
### 3.1 Moulding of a polymeric bottle

Three different parts (geometry) were generated to depict the Eulerian region (green), the internal (blue) and external moulds (white) and the initial region where the polymer is defined at the beginning of the analysis (red) as shown in figure 3.1.1 [105].

![Figure 3.1.1: Assembly of the model comprehending the internal (blue) and external (white) moulds, the Eulerian region (green) and the initial state of the polymer (red) [105].](image)

I then defined two material models called polymer for the bottle and steel for the internal and external moulds. Due to the rigid body deformations, the mechanical properties of the steel with regards to the Young's modulus and Poisson's ratio will not interfere with the solution obtained and the selected density will drive the computational cost of analysis due to the explicit solver. Using consistent units, the lengths were in millimeters (mm), force in Newton (N), mass in tonne, time in seconds, stress in Mega Pascal (MPa), and density in tonne/cubic millimeter (tonne/mm$^3$).

The polymer was created to be elastic with a Young's Modulus of 0.21 MPa and a Poisson ratio of 0.35. It also had a density of $1.455 \times 10^{-9}$ tonne/mm$^3$. The steel was created to be elastic with a Young's Modulus of 210000 MPa and a Poisson ratio of 0.35. It also had a density of $7.8 \times 10^{-9}$ tonne/mm$^3$.

The internal and external moulds were assigned the steel whereas the Eulerian section to the Eulerian part. The initial part was however not assigned any section since it will not be part of the actual simulation but used only to indicate the initial conditions.

The individual parts were then assembled together as shown above (figure 3.1.1) and the I created a dynamic explicit step using two (2) seconds as the time step. The interval of the field output requests was changed to 50 and the Eulerian volume fraction (EVF) was selected followed by the selection of volume/thickness/coordinates.

At the mesh module, I generated the mesh using a global element size of 0.075 and a local element size of 0.04 for the edges. The number of elements in the thickness of the module was set to 2. The generated mesh is shown in figure 3.1.2 below.

After the generation of the mesh, the general contact interaction was created using the interaction module by selecting the “All* with self” as the contact domain and also a “hard contact” normal behaviour. Two rigid body constraints were also created at the two extreme ends depicted RP in figure 3.1.2 to be the reference points.

The boundary conditions were then created for the external and internal moulds which was applied to the reference points. The U1 (x-axis) degree of freedom of the internal mould was given a value
of 1.45 whiles the others had zero (0) indicating no degree of freedom. Boundary conditions were also created for the Eulerian region or part.

![Figure 3.1.2: Generated mesh of the entire assembly](image)

The next stage was to create the job so in the job module, two processors were selected for the parallelization for effective simulation of the program on my labtop to avoid computational errors. The job was then submitted and I obtained results after a period of time.

### 3.2 Simulation of a simple block under pressure

This simulation was aimed at denoting the penetration of an anvil inside an elastic-isotropic plastic block. Three different parts (geometry) were generated to depict the Eulerian region (green), the internal (blue) and external moulds (white) and the initial region where the polymer is defined at the beginning of the analysis (red) as shown in figure 3.2.1. The dimensions of the Eulerian was given as 0,0,0 and 2,2.1,0 mm, the Lagragian part had dimensions 0,0,0 and 2,1,0 mm and the punch has radius of 0.1 mm and length of 0.05 mm. All had an extrusion depth of 0.2.

I then defined an elastic-perfectly plastic material with a very high density in order to increase the time increment. An isotropic elastic material with Young’s Modulus 210000 MPa and Poisson ratio 0.3 was created. The plastic portion of the material was set to have a yield stress of 217 MPa and a plastic strain of zero (0). The density of the material was estimated to be 7.80 tonne/mm³.

The Lagrangian section was created to be an homogeneous solid whereas the Eulerian section was created as a solid Eulerian type. The same material definition was used for the different physical behaviours (an advantage of using sections and materials separately). The sections were then
assigned to the various parts: Eulerian section to the Eulerian part, Lagrangian section to the Lagrangian and punch parts.

Figure 3.2.1: The three generated parts (https://simplifiedfem.wordpress.com/about/coupled-eulerian-lagrange-cel-analysis-with-abaqus/)

The individual parts were then assembled together as shown below (figure 3.2.2) and I created a dynamic explicit step using two (2) seconds as the time step. The interval of the field output requests was changed to 50 and the Eulerian volume fraction (EVF) was selected followed by the selection of volume/thickness/coordinates.

Figure 3.2.2: Assembly of the punch model

At the mesh module, I generated the mesh using a global element size of 0.075 and a local element size of 0.04 for the edges. The number of elements in the thickness of the module was set to 2. The generated mesh is shown in figure 3.2.3 below.

After the generation of the mesh, the general contact interaction was created using the interaction module by selecting the “All” with self” as the contact domain and also a “hard contact” normal behaviour. Two rigid body constraints were also created one on the punch and the other at the middle button of the model.
The boundary conditions were then created for the external and internal moulds which was applied to the reference points. The U1 (x-axis) degree of freedom of the internal mould was given a value of 1.45 whiles the others had zero (0) indicating no degree of freedom. Boundary conditions were also created for the Eulerian region or part.

The next stage was to create the job so in the job module, two processors were selected for the parallelization for effective simulation of the program on my labtop to avoid computational errors. The job was then submitted and I obtained results after a period of time.
Chapter 4

Results and Discussion

4.1 Results of moulding of a polymeric bottle

Figure 4.1.1: Initial State before the moulding begins

Figure 4.1.2: After first impact of the bottle and the rigid steel
Figure 4.1.3: After third impact of the bottle and the rigid steel

Figure 4.1.4: Tenth impact of the bottle and steel

Figure 4.1.5: Final impact of the bottle and steel
4.2 Results of simple block under pressure

Figure 4.2.1: Initial State before the moulding begins

Figure 4.2.2: After first impact of the punch

Figure 4.2.3: After third impact of the punch
4.3 Discussion

With respect to the bottle simulation, it was noticed that the maximum pressure is experienced at the tip of the bottle when it is fully pressed against the Eulerian material. This shows that it is important to avoid obstacles when guiding the needle to the target to avoid further damage. Also, it was noticed that the Mises stress generally kept increasing from zero at the initial state until a point (about 67% of the total movement i.e. 10 out of 15 steps) when the Mises stress begins to concentrated at the tip as the bottle is pressed hard against the Eulerian material. The punching also
revealed that the maximum pressure is felt around the region where the punch is made and the Mises stress is also mostly greatest at the point or region of interaction.
Chapter 5

Conclusion

A thorough study of the anatomy of the heart, right and left atrium, the superior vena cava, inferior vena cava and the atrial septal wall has been performed. A survey about the finite element models applied to the human atria and the atrial wall has been conducted as well as the study and quantification of the mechanical properties of the different tissues of the human atria.

This project is aimed at developing an integrated framework to effectively guide the physician throughout the TSP technique based on the creation of a patient-specific biomechanical model of the atrium. This novel framework is expected to be incorporated into the intra-procedural planning of the expert, with the real-time data acquired from the three-dimensional ultrasound imaging to successfully guide the transseptal needle position with the optimal puncture site.

The proposed technique is expected to increase the efficiency of using the TSP techniques as well as reducing complications associated with this procedure. Also, it will help reduce the use of ionizing radiation due to the use of volumetric ultrasound imaging.

5.1 Future works

1. Definition of the mechanical properties of the different atria tissues; Manual delineation of the relevant structures in CT/MR images.
2. Development of a new approach for the generation of the meshes for the FEM model; Finite element analysis of the model built using the Abaqus/Ansys software.
3. Validation of the proposed FEM model in patients with and without dysfunctional regions in the atria.
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