Chapter 1

Introduction

1.1. CARDIOLOGY

Cardiovascular disease (CVD), a common term for various heart disorders, is the foremost reason of death globally when compared with other diseases. The overall leading global cause of mortality due to CVD accounts for 17.3 million for every year and this would produce more than 23.6 million in the year 2030 (Laslett et al., 2012). According to the information given by World Health Organization 2014, on an average 17.7 million people died from CVDs out of which 7.4 million die because of coronary heart diseases and 6.7 million owing to stroke (Bhatnagar et al., 2015). The Global burden disease statistics has also estimated that CVD causes 30% death. The number of people in United Kingdom having heart disorders is estimated to be around 2.3 million suffering from coronary heart disease and 1.5 million due to stroke. The cause of CVDs include hypertension, diabetes, unhealthy diet, tobacco use, obesity, etc. (Dariush et al., 2015). Assessment of heart functioning therefore remains a significant research area in the medical arena.

Nowadays there are numerous imaging modalities such as Magnetic Resonance Imaging (MRI), Computerized Tomography (CT) and Single Photon Emission Computerized Tomography (SPECT) which are available for cardiac imaging which provide qualitative and quantitative information on anatomy and morphology of the heart (Frangi et al., 2001). Among different modalities, cardiac ultrasound or Echocardiography is considered to be quickest and most cost effective as it permits real-time imaging of the heart, using portable equipment and is a radiation-free technique. The method is used on a routine for diagnosis, management and follow-up care of patients for any suspected heart diseases. Thus the echocardiogram provides ample amount of information such as size and shape of the heart chambers, pumping capacity, tissue damage, etc. Apart from this, the heart functioning can also be understood by calculating the universal clinical indices for instance Cardiac Output (CO), Ejection Fraction (EF), and Stroke Volume (SV) from the volume during the relaxed and contracted phase of the heart.

The Transthoracic Echocardiogram (TTE) is the standard method of viewing the heart by movable transducer toward diverse locations on the chest or abdominal wall.
A handheld device called transducer is positioned on the ribcage and it transfers high-frequency sound waves (ultrasound). The sound waves bounced off the heart and reach the transducer in the form of echoes and turn, get converted to images and are displayed on the monitor as moving pictures of the heart. During the process transducer captures several anatomical sections of the heart from different perspectives. The analysis of these images requires user intervention in both imaging and interpretation sector.

1.1.1. ANATOMY OF THE HEART

The heart is a muscular structure, positioned in the chest and protected by the rib cage. The foremost purpose of the heart is to pump oxygenated blood throughout the body. The heart consists of four main cavities- right and left auricles, right and left ventricles as shown in figure 1.1. The function of left side of the heart is to pump oxygenated or purified blood from the lungs all over the body, i.e. the systemic circuit (Marieb et al., 1994). The right side pushes the deoxygenated blood toward lungs and is termed as a pulmonary circuit. Subsequently, the systemic circuit has a greater blood resistance than the pulmonary circuit, the left adjacent of the heart is bigger and stronger than the right side. For both systems, the atria handle inflow to the heart and pump blood into the ventricles. Consecutively, the ventricles pump blood out from the heart, into the body or the lungs.

![Figure 1.1: Schematic diagram of the heart (Marieb et al., 1994).](image)

The atria and ventricles are divided by directional valves which allow blood to drift from the atria into the ventricles, but not in the reverse path. The mitral valve is
positioned in the middle of the left atrium and left ventricle, while the tricuspid valve is located between the right atrium and the right ventricle. These valves can hold the high ventricular blood pressure because they are anchored by thin strings, or chordae, to papillary muscles which are connected to the inside of the ventricular wall. Reflux of blood back into the ventricles is prevented by a separate set of directional valves. The pulmonary valve is positioned at the outflow region of the right ventricle, whereas the aortic valve is sited at the outflow area of LV.

The heart wall consists mainly of muscular tissue called the myocardium, which is composed of specialised cardiac muscle cells bundled into muscle fibers. These fibres are organised in multiple layers, each having a different orientation. The interior of the heart chambers has a folded structure called trabeculae. The endocardium is the deep cover of tissue that protects the chambers of the heart and separates the myocardium from blood (Braunwald et al., 1992). This layer mainly consists of endothelial cells and connective tissue. The outer covering of the heart is termed as epicardium and contains mainly of connective tissue. The layer epicardium forms the innermost part of the heart sac named pericardium, which contains the heart and the origin of the great vessels.

1.1.2. REGION OF INTEREST: LEFT VENTRICLE

The cardiac cycle refers to the physiological processes that initiate from start of the heartbeat till the commencement of next heartbeat. A cardiac cycle consists of numerous stages such as ventricular contraction (systole), ventricular relaxation and filling (diastole). The driving action of the heart sources the left ventricular blood volume to alter during a heartbeat. The heart undergoes seven different phases during each beat. The different phases includes (1) atrial systole (2) isovolumetric contraction (3) rapid ejection (4) reduced ejection (5) isovolumetric relaxation (6) rapid filling and (7) reduced filling. LV is the left ventricle region, ECG the electrocardiogram, AP the aortic pressure, LVP the left ventricle pressure, LAP the left atrial pressure, LVEDV the left ventricle end diastolic volume, LVESV the left ventricle end systolic volume and S1 to S4 are the four heart sound which is represented in figure 1.2 (Berne et al., 2001).

The systole commences once the left ventricular pressure exceeds that of atrial pressure and approximates with closure time of the mitral valve. The first phase of systole is known as isovolumic contraction because the volume within the left ventricle
is fixed as the mitral and aortic valves are closed. The moment, left ventricular pressure
tops aortic pressure, aortic valve unlocks and systolic ejection period follows. The
pressure gradient finds the volume of blood expelled through aortic valve and also by
elastic effects of aorta and arterial branch.

Figure 1.2: The seven phases of a cardiac cycle (Berne et al., 2001).

The volume of blood that is ejected through the aortic valve during cardiac
systole is expressed as a percentage of the entire volume of blood present in the left
ventricular cavity towards the end of ventricular filling (diastole) and is called the
ejection fraction. The amount of blood that is expelled by the LV throughout each
heartbeat is known as stroke volume. Therefore in systolic contraction, blood is ejected
from the ventricle above the aortic valve, and the ventricular volume is reduced. The
mitral valve closes, preventing blood flow towards the atrium.

The left ventricle diastole has four phases such as isovolumic relaxation, early
passive left ventricular filling, diastasis and late active left ventricular filling associated
with atrial contraction. As the left ventricle actively relaxes, the left ventricular pressure
drops without variation in left ventricular volume. Once the left ventricular pressure
decreases lower than the left atrial pressure, the mitral valve opens, indicating the
completion of isovolumic relaxation phase (Anderson, 2002). After the mitral valve
opens, early passive left ventricular filling stage starts. Early diastolic filling depends
on the magnitude of the pressure gradient between the Left Atrium (LA) and also Left Ventricle (LV) which pumps blood into the left ventricular chamber.

In brief during the beginning of diastole, the ventricle relaxes that means, the volume of the chamber will be more, and the reduced ventricular pressure allows blood to drift through atrium into ventricle via the mitral valve. This situation is known as rapid filling. Diastasis occurs when the pressure gradient over the mitral valve has been equalised, and volume is relatively constant. The duration of the heartbeat varies from stroke to stroke, mainly caused by the different duration of this stage. In late diastole, the atrium contracts, causing the ventricle to get filled.

The left ventricle chamber of the coronary heart pumps oxygenated blood towards different portions of the body. The contraction of left ventricle pushes blood into the aorta, the largest artery in the body. From the aorta, a variety of smaller constituent arteries arise, and these arteries bring blood to nourish various components of the body. Cardiologists consider left ventricle structure with high importance as it supplies the oxygen contained blood to several portions of the body and provides valuable information on its pumping efficiency and plays a major role in certain heart disorders such as cardiomyopathies and myocardial infarction (Zhou et al., 2010).

Heart ailments like coronary artery disorders, myocardial infarction, heart failure and ischemia are often manifested through reduced contractility of the heart muscle, leading to lowered pumping capacity or Ejection Fraction (EF), and changes in ventricular volume (Braunwald, 1992). Therefore the left ventricular volume and EF are important measurements for diagnosis, prognosis, and treatment of patients with cardiovascular disease and reliable determination of these parameters is of high clinical interest (Taylor et al., 1980).

1.1.3. MEASURING CARDIAC FUNCTION

Cardiac function can be measured in numerous ways but left ventricular volumes, stroke volume, and ejection fraction are a few informative parameters (Liu et al., 2008). Stroke volume (SV) can be defined as the difference between the End Diastolic Volume (EDV) and the End Systolic Volume (ESV) and is a measure of the amount of blood the heart pumps in each stroke. Ejection fraction (EF) is the ratio of stroke volume to end diastolic volume, measuring the percentage of the end-diastolic volume being emptied in each stroke. EF is a good indicator of general heart state and
a useful predictor of clinical outcome (Taylor et al., 1980), as cardiac diseases often lead to reduced heart contractility. SV and EF are often called global parameters since they consider the total volume of the left ventricle.

Global parameters can therefore be used to express the general state of the heart, but they alone do not serve the purpose of indicating the part of cardiac muscle that is affected. Ventricular volumes and EF are assessed by calculating the volume enclosed by the endocardium and the mitral annulus. Clinical recommendations state that papillary muscles should be regarded as part of the blood volume (Lang et al., 2015). Different protocols for volume computations from manual tracing exist, but the most common one is to use the biplane method of disks where the endocardial boundary is traced manually in two nearly orthogonal images, whereby a stack of ellipses are fit to the traced contours (Schiller et al., 1979).

1.1.4. ECHOCARDIOGRAPHY: PRINCIPLE, ACQUISITION AND DISPLAY

The basic principle behind echocardiography is transmitting high-frequency ultrasound pulses from a transducer through the body, and then processing the backscattered echo received by the transducer. The pulse frequencies used are typically in the range of 1.5 -10 MHz, depending on the depth of the organ being imaged. Different organs like blood and tissues show different acoustical properties, which in turn render distinct echo responses that can be used to create images.

Imaging is obtained by transmitting and receiving ultrasound beams steered in clear directions so that they together form a plane or image volume (Luigi, 2014). The resulting volume can then be visualised on display. The model of echocardiography machine is shown in figure 1.3. There are five basic essential parts for an ultrasound scanner which consist of generation, display and storage units of the ultrasound picture. These include pulse generator which provides high amplitude voltage to trigger the crystals, transducer which transforms electrical energy into mechanical energy and vice versa. The receiver which distinguishes and then amplifies very weak signals, a display unit where ultrasound signals in diverse ways is displayed and finally memory which stores video display. For the easy transmission of the sound waves from the transducer into the skin, ultrasound gel is applied. The transducer converts echo into an electrical signal which is processed and displayed as a picture on the monitor. Thus the translation of sound to electrical energy is termed as the piezoelectric effect.
The sound wave generated by way of electrical stimulation of a piezoelectric crystal, travels via an interface between two tissues of varied acoustic density, for instance, myocardium and blood case, part of the energy is returned (the reflected wave), and the rest travels ahead through the subsequent tissue (the refracted wave). The transducer acquires the reflected wave, converts back into electrical energy, amplifies, and displays. If there is an excessive amount of variance among the acoustic density of the tissues imaged (as in between air-occupied lung and myocardium or between bone and myocardium), the entire ultrasound wave is reflected and the heart structure cannot be imaged.

The measure of reflected wave identified during ultrasound imaging depends not only on the acoustic attributes of the interface but also depends on the angle of incidence. An ultrasound beam which is having a flat surface perpendicular to the beam
will reflect a wave in the way of the transmitted sound. Similarly, a beam corresponding to a structure or that which come across an irregularly shaped structure, as is common in tissue imaging, will be reflected with a grade of scattering that is comparative to the angle of incidence.

The rays in 2DE move in a sector fashion which uses a tomographic technique to provide flat views of the heart. A 2D image is generated from a set of data either mechanically or electronically (phased array transducer). Thus a 2DE will provide flat views which is a kind of tomographic technique and for 3DE real-time volumetric imaging that acquires pyramidal data sets shown in figure 1.4 (Luigi, 2014).

A conventional 2D phased array transducer is comprised of numerous piezoelectric elements, electrically isolated from one another, organised in a single row. The wave fronts of ultrasound are separately produced by firing individual elements in an exact sequence with an interruption in phase about the transmit initiation time. Every element adds and deducts pulses to produce a single ultrasound wave with a precise direction that establishes a radially circulating scan line. The linear array is often steered in two dimensions (vertical (axial) and lateral (azimuthal), whereas resolution within the z-axis (elevation) is mounted by the thickness of the tomographic slice, which in turn is expounded to the vertical dimension of piezoelectric elements. In 3DE fully sampled matrix array transducers are used and consist of 3000 piezoelectric crystals which are operated in the range of 2-7 MHz. Therefore the piezoelectric components are organised in the form of matrix arrangement inside the transducer, and the ultrasound beam is electronically steered.

The electronically precise phasic firing of the components in that matrix produces a scan line that spreads radially and can be driven both laterally (azimuth) and in the elevation appropriate to attain a volumetric pyramid of data. The foremost technological innovation which permitted manufacturers to advance the matrix transducers by miniaturising the electronics and allowed the progress of individual electrical interconnections for every single piezoelectric element which could be individually controlled, both in transmission and in reception. The micro beamforming capability allows the same size of the 2D cable which can be used along with 3D probes despite a large number of digital networks for the fully sampled components which are associated.
Figure 1.4: a) Two-dimensional and b) Three-dimensional echocardiography probe (Luigi, 2014).

For 3DE acquisition there are two approaches which include real time (or live 3D) and multiple beat 3DE imaging (Lang et al., 2012). Real-time imaging obtains several pyramidal data sets in a single heartbeat. The various acquisition modes for real-time imaging include simultaneous multiplane, slender volume, zoom, wide angle (full volume) and colour Doppler. The use of simultaneous multiplane mode allows the dual screen to show two real-time images at the same time where one will be the reference image and the other is the lateral plane of the same image. The narrow sector mode depicts the pyramidal volume. The zoom mode focuses the wide sector view of different structures of the heart, and full volume acquires the largest possible sector which captures specific structures like a mitral valve or aortic root.

Although this acquirement mode overwhelms rhythm turbulences or respiratory motion limitation, still suffers from reduced temporal and spatial resolution. The multiple beat 3DE is acquired over entire sequential volumes of data over two to five cardiac cycles, but the error occurs due to artefacts created by cardiac rhythm irregularities. Real time imaging is acquired using 60 by 60 degrees volume in figure 1.5(a), multi-beat imaging showing ECG-gated stitched sub-volumes of several cardiac cycles in figure 1.5(b) and a real-time 3D zoom where data is acquired with subtle region of 30 by 30 degrees which are illustrated in figure 1.5(c). In real time 3D zoom the frame rate can be increased by decreasing the volume size (Badano et al., 2011).
A noteworthy convenience in using the 3DE method is that it provides a volumetric information thus covering the important region of interest. This provides the examiner with the choice of some outlooks for the cardiac data available in the data. The three existing methods to present the volume data are: (1) The complete volume can be cut into single or several slices by the system and then exhibited as normal 2D images (2) A 3D appearance of the object can be generated by displaying volume rendered images (3) Presenting a structure rendered prototype in a 3D scene The volume samples which are interconnected by a cross-sectional plane of the heart structure are used to produce two-dimensional images, represented by a slice display and is shown in figure 1.6 (a). Slice display is beneficial in generating a 2D image which is not obtainable in transthoracic 2D echocardiography, irrespective of the cross section being chosen. This allows the examiner to find precise standard views of the object and avoid foreshortening.

Surface rendering is another technique used to display purpose (Lang et al., 2012). Since the computer is 2D and the examiner might desire to view the cardiac structures with 3D insight. Then, volume rendering best suites in generating images with a depth insight as shown in figure 1.6 (b). This technique makes the arrangement to cast rays from the sight plane through the volume and records the sample values along with the rays. All the samples along the ray path from the viewpoint till the backplane are grouped together, and that represents the last ray value recorded behind the object. Thus, the name volume rendering. Usually, the combination will be the addition of all the sample values, multiplied by the weight termed as opacity. The grey level of the sample controls opacity. A pixel is rendered opaque (i.e. tissue) when the ray strikes the high grey level samples, whereas the pixels are rendered crystal clear.
(i.e., blood pool) when the ray possesses shining through low grey level samples. Opacity function specifies the relation between the sample grey level and opacity.

![Image](image_url)

**Figure 1.6:** a) Slice principle b) Volume rendering principle c) Surface rendering principle (Rabben, 2000)

There are necessities to be satisfied by the opacity function to create high-class rendering: (1) opacity function needs to correspond to the image data (2) opacity for the remaining samples get allocated so that it reduces the formation of misleading artefacts.

Along with the above-explained technique of rendering, a light source can also be complemented into the technique of ray-casting. This creates a shading of the structure that is rendered. It is achieved by merging information present around the tissue blood contour path and along with that of the light source. Volume rendering poses one common difficulty in that the concerned structure is present at the back of the other structures and thus may not be visualised. As an example, if it is wished to split the left atrium to have a surgeon’s view of the mitral valve, acknowledged as cropping, and is attained by using volume rendering technique to detach the volume on one side of a plane that is defined within the volume. The 3D scanners provide unusual instruments to modify the understanding of the view and also the crop planes. Navigation is made easier by first letting the alignment of suite slices to the traditional views of the apical or parasternal imaging window (Badano et al., 2010). The auto cropping option allows the user to make use of push buttons to select the structure to be rendered. It is to be noted that the pyramidal structure 3D ultrasound scan needs a change of coordinate system from the 3d polar to Cartesian system in both the slice and volume rendering displays, using a 3D scan converter. This conversion can be carried out by either applying the procedures of slice or volume rendering or directly as a part
of the algorithms for volume rendering. Voxels are the box-shaped samples obtained from the Cartesian information. The term voxels are conjointly used with non-box shaped information by some authors.

Surface rendering models offer a replacement way of exhibiting the cardiac structures without disturbing the depth perception, in 3D scenes as shown in fig. 1.6(c). The necessity to apply this method is that the system already knows the blood-to-tissue contour of the cardiac structures. Hence, forming the outline of the structure both manually and automatically is possible before considering surface rendering. Surface rendered pictures can be created after obtaining a geometrical description.

![Images](image1.jpg)

Figure 1.7: a) Slice rendering of short axis view of LV b) Volume rendering of mitral valve c) Surface rendering of LV in apical four chamber view in 3D scene (Badano et al., 2010)

To enhance the depth perception, it is normal to complement shading before creating the final surface rendering. Usually, shading depends on the estimation of the possible directions of the object surface and connecting these directions with that of the light source. The perpendicularity of the light direction to the rendered surface decides the ultimate brightness and shade of the surface. If the light direction is perpendicular to the surface, then there is shine to the surface and parallelizing leads to the darker surface (Lang et al., 2012). Surface renderings might also be termed, geometrical models. A slice rendering in a short axis view of LV shown in figure 1.7(a), volume rendered for mitral valve opening in figure 1.7(b) and surface rendering for LV in apical four chamber view as illustrated in figure 1.7(c).

1.1.5. ULTRASOUND INTERACTION WITH TISSUE

Ultrasound is mechanical stress waves with frequencies inaudible to the human ear. In medical applications, the typical frequency range is 2-10 MHz, similar to
electromagnetic radio frequency (RF) waves. In soft tissue, the speed of sound is approximately as in water; \( c = 1540 \text{ m/s} \), giving wavelengths \( \lambda \) in the range of 0.15-0.75 mm. Ultrasound wave pulses are typically generated using an ultrasound transducer consisting of piezoelectric crystals that vibrate when exposed to a high-frequency electric potential. As a transmitted ultrasound pulse traverses through the body, it interacts with the tissue, causing scattering, reflection, and absorption of the wave energy (Finn et al., 2011). Absorption is caused by the conversion of kinematic wave energy to heat and is frequency dependent. Scattering occurs when the ultrasound wave interacts with objects of size less than the wavelength, whereas reflection occurs at tissue interfaces. Scattering and reflection causes a fraction of the transmitted energy to be echoed back to the transducer. These echoes can be measured as electric signals generated by the same piezoelectric crystals used to transmit the pulse. The time from the pulse is transmitted, to the echo is received and is about the depth of the scatters or tissue interfaces. During the transmit process, an electrical voltage is applied to the piezoelectric elements, causing them to vibrate and generate ultrasonic waves. Then, during the receive process, acoustical vibrations from the received echoes induce electrical signals across the crystals that are sampled and used for image formation. The distance \( r \) between the transducer and the origin of an echo signal is determined using the two-way transit time. By assuming a constant wave propagation speed \( c \) the distance is given as a function of a two-way transit time \( t \) and shown in Equation 1.1.

\[
r(t) = \frac{ct}{2}
\]

where \( c = 1540 \text{ m/s} \) for soft body tissue. Transmitted ultrasound waves are partially transmitted to deeper structures, moderately reflected back to the transducer as echoes, partially scattered, and somewhat transformed to heat. The extent of echo returned subsequently hitting a tissue interface is found by (1) the angle of incidence, (2) dimensions of the reflecting structure, and (3) a tissue property named acoustic impedance \( Z \) was given by \( Z = \rho \times c \), where \( \rho \) is the density of the tissue. Structures that are even and greater than the wavelength because a part of the echo gets reflected into the transducer and the incident pulse travels deep into the tissue body, a kind of reflection called specular reflection. The amount of reflected echo therefore depends on the angle of incident and difference in acoustic impedances among two mediums. For non-perpendicular incidence at an interface between tissues with dissimilar acoustic
impedances, the transmitted pulse is detected by an angle described by Snell's law. If
the ultrasound pulse come across reflectors whose dimensions are lesser than the
ultrasound wavelength, then omnidirectional scattering occurs, which does not rely
upon angle of incidence.

![Beam focusing diagram](image)

Figure 1.8: Beam focusing by adding time delays across the array of piezoelectric elements.

Sector (2D) and volumetric (3D) images can be acquired by transmitting beams
that are steered in different directions to cover a region of interest. An array of
transducer elements can be focused by controlling the phase of a signal associated with
each element. Application of discrete time delays stimulating each element emphases
the transmitted beam at a specific range. The amount of delay for a given element
depends on the distance between the element and the focal point illustrated in figure
1.8. On reception, the focus can be removed along with the beam by dynamically
allocating the time delays linked with the elements; therefore, their position coincides
uninterruptedly with that of the instantaneous origin of the echoes.

1.1.6 TRANSTHORACIC ECHOCARDIOGRAM (TTE)

The advancement of technology has led from 2D (two dimensions) to 4D (three
dimensions and time) which captures the moving picture of the heart. The ultrasound
images in 2D mode, and 3D format was first reported in the 1960s and the
reconstruction of 3D from 2D echocardiograms were all reported. In 1990 the first real-
time 3D echocardiography scanner had come into the role, and from 2008 onwards the
real-time 3DE has come into the role (Hung et al., 2007). The transthoracic
Echocardiogram has been a largely standardised method in clinical routine by the cardiologist. A standard 2D phased array transducer consist of multiple piezoelectric components and electrically isolated from each other and organised in a single row. There are diverse views acquired for the case of 2DE.

The images are taken from standard locations of transducer that permit to picture various parts of the heart. The first standard locations are the parasternal position, which has both long axis and short axis visions, which is placed by the transducer approximately on the 3rd intercostal space on the left of the parasternal. The second standard position is the subcostal where the patient is placed at on their back. In this ultrasound beam is further perpendicular to the atrial and ventricular septum (Horton, 2010). The third standard view is the suprasternal view. The patient lies completely at without anything under their head. Then the transducer is placed in the suprasternal notch through the probe pointed at the left shoulder at about 2 o'clock position. This site usually provides a good look at the arch, from which the three major vessels may be understood to arise. The last standard view is the apical position, which is acquired by placing the probe at the point of maximal impulse. The patient is positioned on their left side, but not in the case of the parasternal window.

The apical window is typically found in fifth intercostal space of the ribs so that the probe can be positioned more accurately for obtaining the apical view. There are several views that can be acquired from this position, an apical 2 chamber views were left ventricle (LV) and left atrium (LA) can be viewed in figure 1.9(a), an apical three-chamber view where LV, LA and aortic valve can also be viewed in figure 1.9(b). Another view is apical four chambers where all the chambers of the heart can be visualised in figure 1.9(c).

Figure 1.9: Apical Views in 2DE  a) two chamber view b) three-chamber view c) four chamber view (Horton, 2011)
On the routine clinical basis, 2D is used by the cardiologist but has drawbacks in computing the volume of structures of the heart. The real-time three-dimensional echocardiography (3DE) imaging shows better performance when equated with two-dimensional echocardiography (2DE) regarding geometric assumptions for the chambers (Dicki et al., 2010). Thus 3D ultrasound imaging has drawn undue attention in current years as this procedure displays the volumetric data sets in the beating heart (Badano, 2014). The breakthrough within the history of 3DE has been the progress of absolutely sampled matrix array transthoracic transducers created on advanced digital process and improved image formation algorithms that allowed the operators to obtain transthoracic echocardiogram (TTE) real-time volumetric imaging with shorter acquisition time, higher spatial and temporal resolution (Luigi, 2014).

![Figure 1.10: a) Multiple beat 3DE data acquirement from TEE apical window b) Real-time 3DE single beat of entire heart c) Left ventricle from transthoracic window (Lang et al., 2012)](image)

Currently, 3DE matrix array transducers include approximately 3000 impartial piezoelectric elements with operational frequencies ranging from 2 to 4 MHZ within the situation of TTE. Three-dimensional TTE full-volume attainment mode can accommodate a maximum of the coronary heart structures surrounded with the aid of a single 3D information set. However, with current era, there is a reduction in each
spatiotemporal resolution and penetration that could effect from increasing the volume angle to attain the entire heart from a certain acoustic window which is practically difficult. For overwhelming these bounds, 3DE data sets should be achieved from multiple transthoracic transducer locations (Lang et al., 2012). A focused 3DE investigation usually includes relatively few 3DE data sets. Few example such as gated 3DE full volume dataset achieved from apical window to compute LV volumes, EF, and heart failure situations. The data sets attained from both parasternal and apical is completed to visualise mitral valve for mitral stenosis cases. The 3D zoom type acquisition, with high density from the parasternal frame to visualise the aortic valve in a patient suspected bicuspid valve. In figure 1.10 (a) depicts multi-beat 3DE data acquirement from transthoracic apical window where pyramidal volumes starting from four cardiac cycles and is stitched together to form single volume dataset, figure 1.10 (b) real-time 3DE single beat acquisition of complete heart and figure 1.10 (c) left ventricle portion alone is highlighted in transthoracic apical window.

![Coronal and Sagittal](image)

![Transverse and Full volume data set](image)

Figure 1.11: Real-time three-dimensional dataset cropped into coronal, sagittal, transverse and full volume dataset (Hung et al., 2008)

A full 3D TTE exam involves multiple acquisitions from the parasternal, apical, subcostal, and suprasternal transducer locations. The volume rendered 3D data set will also be cropped to display a range of intracardiac structures by selecting different cut
planes as a substitute to view anatomical planes which can be used to define the image orientation. The most customarily used cropping planes are (1) the transverse plane, a horizontal plane that goes perpendicular to the long axis of the body separating the heart into superior and inferior sections; (2) the sagittal plane, a vertical plane that separates the heart into right and left segments; and (3) the coronal plane, a vertical plane that splits the heart into anterior and posterior segments. The live 3DE can be cropped in different planes such a) Coronal b) Sagittal c) Transverse d) Complete volume dataset as shown in figure 1.11.

1.2 CHALLENGES

Ultrasonography is an appealing modality for imaging soft tissue. Few advantages of ultrasonography include cost effectiveness, possibility of real time imaging, absence of side effects and invasion and ease of usage. Although diagnostic ultrasound offers advantages, there is a necessity for trade-off between resolution and attenuation, presence of speckle noise and artefacts.

The conventional 2DE technique is widely used for assessing the geometry and function of LV in clinical routines. However, the accuracy and reproducibility are somewhat suboptimal when compared to the other methods such as Magnetic Resonance Imaging (MRI) and Computerized Tomography (CT).

Few general problems which the researchers experience during the ventricular segmentation of echocardiographic images are:

- Variation in the size, shape and orientation of ventricles among patients
- The edge of the left ventricle becomes invisible due to a low contrast between the blood pool and wall of the heart
- The presence of noise, particularly for echocardiographic images
- Papillary muscles overlap left ventricle and have the same order of magnitude as the wall of the LV

Researchers have been developing different approaches towards LV segmentation and there are constant efforts to enhance the image detection. For instance, the work carried out by (Santiago et al., 2015) presents a novel technique based on the probabilistic data association filter (PDAF) for segmentation of 3DE images, proposing different initializing procedures.
Few other challenges are the interpretation and visualization of 3D ultrasound images due to the presence of speckle noise which makes the spatial visibility of clear contours unlikely. The presence of artefacts adds up to this problem (Mazhaferi et al., 2013). This demands removal of speckle noise from the data without losing the necessary information before segmentation.

Another important parameter in pixel-based segmentation of an ultrasound image is the pixel intensity which is a result of reflection and interference patterns and there is a complex relation between pixel intensity and the physical property of the tissue under consideration. The tissue types are differentiated based on the subtle variations in the speckle patterns. This impacts the intensity distribution in the ultrasound images giving a non-Gaussian density (Chalana et al., 1996).

Echocardiographic image information is highly anisotropic and position dependent, since reflection intensity, spatial resolution, and signal-to-noise ratio depend on the depth and the angle of incidence of the echocardiography beam as well as of user-controlled depth gain settings.

The speckle present in an image could be random and a settled interference pattern in an image formed with coherent radiation of a medium containing several sub-resolution scatterers. It results from the overlying scattering echoes (Finn et al., 2011). The local intensity of the speckle pattern, nevertheless, reflect the local echogenicity of the underlying scatterers. The removal of this speckle noise is therefore proven important.

There is a possibility that the cardiac ultrasound imaging can contain missing information at the edge of the cardiac wall, in the areas of the heart cavity where the wall is absent or hidden by the speckle noise. These regions may be found near the septum and these areas might be difficult to distinguish due to the absence of a difference in intensity and may cause issues in the segmentation process.

Papillary muscles are an important entity in echocardiograms having acoustic properties similar to that of the myocardium. For volume measurements, papillary muscles are considered to be a part of the LV cavity (Lang et al., 2015). The Figure 1.12 shows a) speckle pattern b) drop out problem and c) similar intensity pattern.

Among the interesting ongoing research on segmentation issues such as tissue inhomogeneity, reverberations, and acoustic shadowing, LV segmentation remains to
be an important topic. Accuracy in the quantification of LV volumes and EF is prominent in diagnostic, prognostic and therapeutic applications (Shahgaldi et al., 2010). Therefore, accurate volume measurements are needed which involve precise delineation of the endocardial border (Morales et al., 2002). Since the manual method for extracting the LV portion is time consuming and tedious during the clinical routine, automatic procedures are a good solution.

![Echocardiographic imaging challenges](image)

Figure 1.12: Echocardiographic imaging challenges (A) Strong speckle pattern with low contrast (B) drop out of the anterior wall due to long shadow and (C) similar intensity myocardial wall with trabeculation, and the yellow line is the ground truth segmentation.

To achieve further reduction in user variability which is a common problem with semi-automatic segmentation method, automatic segmentation approach is preferred. This method ensures regularity in final segmentations for a specific volume.

Therefore, my work focuses on the development of semi-automatic and automatic algorithms for LV segmentation in 2DE and 3DE and validation of the calculated clinical parameters.

1.3 MOTIVATION AND OBJECTIVES

An automatic segmentation approach of 3DE data using the B- spline explicit method by considering the darker appearance of blood, a work by (Barbosa et al., 2012) prompted the idea to develop an automatic algorithm using ellipse detection method, a comparatively simple technique.

A novel method by (Xie et al., 2002) to detect an ellipse which clearly mentions the parameters for detection and the detailed steps served to be an important work which helped the current analysis.
A work by (Junior et al., 2010) applies image processing techniques to a semi-
automatic algorithm to segment the LV over a complete cardiac cycle, the application
of cellular automata method for noise removal and detection of border for digital images
by (Popovici et al., 2002) include some of the interesting works. The Von Neumann
neighbourhood detailed in (Popovici et al., 2002) for noise removal is applied for the
current work.

Boochieng et al., (2004) detailed the procedures of edge detection and
segmentation method of a 2DE image. Image improvement algorithms of noise
suppression, histogram, brightness adjustment, threshold and median filtering were
used. This was followed by edge detection algorithm with sobel compass gradient mask
to show the endocardium border edge. Performance of the method was compared with
manual track by evaluation through Pearson correlation co-efficient. This work served
as another motivating factor for the current work.

The current study is aimed at achieving the following objectives:

- To develop a semi-automatic algorithm for segmentation of the left ventricle in
  2DE and calculate the clinical parameters
- To develop an automatic algorithm for segmentation of left ventricle in 3DE by
  removing the speckle noise
- Detection of the edge voxels from the borer of the left ventricular cavity
  followed by surface mesh generation
- Validation of clinical parameters such as stroke volume and ejection fraction

1.4. OUTLINE OF THE THESIS

This thesis is organised into five chapters. In the second chapter, a brief literature
review on the various segmentation approaches in 2DE and 3DE. Chapter 3 covers the
methodology adopted for segmentation of left ventricle in two-dimensional and three-
dimensional echocardiography data. Chapter 4 summarises the results obtained from
proposed algorithms and a discussion about the contributions of the work. Chapter 5
concludes this thesis and discusses the possible directions for future work.