TRACKING POINTS ON A PACING LEAD IN A BEATING HEART

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ABSTRACT

Tracking points on a pacing lead in a beating heart

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Heart failure is a common condition during which the pumping action of the heart is affected because the heart does not contract or relax properly. Heart failure affects about 5 million Americans, with 550,000 new cases diagnosed each year [45]. Cardiac resynchronization therapy (CRT) is used to treat symptoms and other complications associated with a heart failure. While performing CRT, Implantation of a pacing lead in the left ventricle of the heart is a very challenging surgical procedure performed with fluoroscopy. The target location is often difficult to reach through the tortuous coronary venous anatomy, which varies greatly among individuals. Placement of the pacing lead is an important research topic because the ideal pacing location for some patients with heart disease may be the site of latest contraction in the left ventricle.

The purpose of this project is to develop an algorithm to locate and track points on a lead in a sequence of images. The algorithm will track the motion of the points over time and generate displacement plots over time.

TABLE OF CONTENTS

LIST OF FIGURES	vii
1.0 BACKGROUND	1
1.1 THE HUMAN HEART	1
1.2 ELECTRICAL SYSTEM OF THE HEART	3
1.3 ELECTROCARDIOGRAPHY (ECG).....	6
1.4 ARRHYTHMIA	8
1.4.1 PREMATURE DEPOLARIZATION'S.....	8
1.4.2 SUPRAVENTRICULAR ARRHYTHMIAS.....	9
1.4.2.1 ATRIAL FIBRILLATION	9
1.4.2.2 ATRIAL FLUTTER	10
1.4.2.3 PAROXYSMAL SUPRAVENTRICULAR TACHYCARDIA.....	11
1.4.3 VENTRICULAR ARRHYTHMIAS.....	13
1.4.3.1 VENTRICULAR TACHYCARDIA.....	13
1.4.3.2 VENTRICULAR FIBRILLATION.....	14
1.4.4 BRADYCARDIA.....	15
1.4.4.1 SINUS BRADYCARDIA.....	16
1.4.4.2 HEART BLOCK.....	16
1.5 HEART FAILURE	17
1.6 CARDIAC PACING.....	19
1.7 HISTORY OF ELECRITICAL THERAPY.....	20
1.7.1 DEVICES WHICH DO NOT STIMULATE IN THE PRESENCE OF A NATURAL CARDIAC BEAT	20
1.7.2 DEVICES WHICH ATTEMPT TO RESTORE ATRIAL SYNCHRONY.....	20
1.8 ICD and CRT	21
1.9 OVERVIEW OF CRT IMPLANTATION PROCEDURE.....	25

1.10	COMMON PACING COMPLICATIONS	29
1.13	Q-LV TIMING	30
2.0	OBJECTIVE	33
3.0	METHODS.....	34
3.1	INTRODUCTION	34
3.2	BASIC IDEA OF CORNER DETECTION.....	35
3.3	PROPERTIES OF HARRIS CORNER DETECTOR.....	42
3.3.1	ROTATION INVARIANCE.....	42
3.3.2	PARTIAL INVARIANCE TO INTENSITY CHANGES.....	43
3.3.3	SCALE INVARIANCE	45
3.4	IMPLEMENTATION	46
3.4.1	CONCEPT OF MOVING WINDOW.....	46
3.4.2	SPATIAL CONVOLUTION AND CORRELATION	48
3.4.3	GRADIENT OF AN IMAGE	51
3.5	NON MAXIMA SUPPRESSION	57
4.0	RESULTS	74
5.0	DISCUSSION AND CONCLUSIONS.....	77
6.0	FUTURE WORK.....	78
7.0	BIBLIOGRAPHY.....	79

LIST OF FIGURES

FIGURE 1. ANATOMY OF THE HEART, EXTERIOR FRONT SURFACE [3].....	1
FIGURE 2. ANATOMICAL STRUCTURE OF THE HUMAN HEART AND THE BLOOD FLOW [6]	2
FIGURE 3. ELECTRICAL CONDUCTION SYSTEM OF THE HEART [8].....	4
FIGURE 4. REPRESENTATION OF AN ECG SIGNAL [9]	6
FIGURE 5. ACTION POTENTIAL BY CARDIAC REGION [10]	7
FIGURE 6. ECG RECORDING OF A HEART WITH PREMATURE VENTRICULAR CONTRACTIONS [49]	9
FIGURE 7. ATRIAL FIBRILLATION V/S NORMAL RHYTHM [14].....	10
FIGURE 8. ELECTRICAL SIGNAL PROPAGATION IN A HEART WITH ATRIAL FLUTTER [15].....	11
FIGURE 9. REPRESENTATION OF AN ECG OF A HEART WITH PAROXYSMAL SUPRAVENTRICULAR TACHYCARDIA [49].....	12
FIGURE 10 REPRESENTATION OF ELECTRICAL SIGNAL PROPAGATION IN A HEART WITH WOLF PARKINSON WHITE SYNDROME [15]	12
FIGURE 11. REPRESENTATION OF AN ECG OF A HEART WITH VENTRICULAR TACHYCARDIA [49]	13
FIGURE 12. REPRESENTATION OF ELECTRICAL SIGNAL PROPAGATION IN A HEART WITH VENTRICULAR TACHYCARDIA [15].....	14
FIGURE 13. REPRESENTATION OF AN ECG OF A HEART WITH VENTRICULAR FIBRILLATION [49]	15
FIGURE 14. REPRESENTATION OF ELECTRICAL SIGNAL PROPAGATION IN A HEART WITH VENTRICULAR FIBRILLATION [15].....	15
FIGURE 15. REPRESENTATION OF AN ECG OF A HEART WITH SINUS BRADYCARDIA [49].....	16
FIGURE 16. REPRESENTATION OF A HEART WITH HEART BLOCK [14].....	17
FIGURE 17. REPRESENTATION OF SYSTOLIC AND DIASTOLIC HEART FAILURE [18]	18
FIGURE 18. TYPES OF ICDs	23
FIGURE 19. REPRESENTATION OF PACING CONDUCTION PATHWAYS [28]	25
FIGURE 20. CORONARY VENOUS AND ARTERIAL VASCULATURE OF THE HEART [30]	26
FIGURE 21. REPRESENTATION OF ANTERIOR VENOGRAM [30].....	28
FIGURE 22. REPRESENTATION OF RIGHT ANTERIOR OBLIQUE VENOGRAM [30].....	28
FIGURE 23. REPRESENTATION OF LEFT ANTERIOR OBLIQUE VENOGRAM [30].....	29
FIGURE 24. REPRESENTATION OF RIGHT AND LEFT PHRENIC NERVE [31].....	29

FIGURE 25. EXAMPLE REPRESENTATIONS OF Q-LV MEASUREMENT [36].....	31
FIGURE 26. BASIC IDEA OF CORNER DETECTION	36
FIGURE 27. ELLIPSE AS A FUNCTION OF EIGEN VALUES OF THE MATRIX E [40].....	39
FIGURE 28. CLASSIFICATION OF AN IMAGE REGION AS FLAT, EDGE OR CORNER USING EIGEN VALUES [41]	42
FIGURE 29. ROTATIONAL INVARIANCE OF HARRIS CORNER DETECTOR [41].....	43
FIGURE 30. PARTIAL INVARIANCE TO INTENSITY CHANGES	44
FIGURE 31. CORNER POINT DETECTION IS LIMITED BY THE SIZE OF WINDOW	45
FIGURE 32. REPRESENTAION OF A WINDOW AND NEIGHBORHOOD [42].....	46
FIGURE 33. MOVING WINDOW REPRESENTATION [42].....	47
FIGURE 34. REPRESENTATION OF AN AVERAGING OPERATION [42]	48
FIGURE 35. REPRESENTATION OF SPATIAL CORRELATION OPERATION [42]	49
FIGURE 36. ROTATION OF CONVOLUTION MASK BY 180 [42]	50
FIGURE 37. REPRESENTATION OF SPATIAL CONVOLUTION OPERATION [42]	51
FIGURE 38. GAUSSIAN FILTER WITH $\sigma = 1$ AND $\text{RADIUS} = 6 * \sigma$	55
FIGURE 39. GAUSSIAN FILTER WITH $\sigma = 2$ AND $\text{RADIUS} = 6 * \sigma$	56
FIGURE 40. GAUSSIAN FILTER WITH $\sigma = 3$ AND $\text{RADIUS} = 6 * \sigma$	56
FIGURE 41. 5X5 STATISTICAL ORDER FILTER OPERATION ON AN IMAGE ARRAY.....	58
FIGURE 42. 7X7 STATISTICAL ORDER FILTER OPERATION ON AN IMAGE ARRAY.....	58
FIGURE 43. 5X5 STATISTICAL ORDER FILTER OPERATION REPRESENTATION ON CORNER MEASURE MATRIX	59
FIGURE 44. 9X9 STATISTICAL ORDER FILTER OPERATION REPRESENTATION ON CORNER MEASURE MATRIX	60
FIGURE 45. 15X15 STATISTICAL ORDER FILTER OPERATION REPRESENTATION ON CORNER MATRIX	61
FIGURE 46. INITIAL CORNER POINT MATRIX	64
FIGURE 47. INITIAL CORNER POINT MATRIX AS A FUNCTION OF GAUSSIAN FILTER KERNEL SIZE.....	66
FIGURE 48. INITIAL CORNER POINT MATRIX AS A FUNCTION OF STATISTICAL ORDER FILTER KERNEL SIZE	67
FIGURE 49. INITIAL CORNER POINT MATRIX AS A FUNCTION OF SENSITIVITY FACTOR (k)	68
FIGURE 50. REPRESENTATION OF A FINAL CORNER POINT DETECTION MATRIX.....	70
FIGURE 51. COMPARISON OF INITIAL CORNER POINT DETECTION MATRIX BASED ON PERFORMING CONVOLUTION WITH AND WITHOUT BOUNDARIES	72
FIGURE 52. POINT DISPLACEMENT PLOTS FOR VIDEO FILE 1.....	75

FIGURE 53. STANDARD ERROR IN POINT DISPLACEMENT CHART FOR VIDEO FILE 1.....	76
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1.0 BACKGROUND

1.1 THE HUMAN HEART

The human heart is one of the most important organs in the human body, located in between the lungs in the middle of the chest behind and slightly to the left of the sternum weighing approximately 250 – 300 grams in females and 300-350 grams in males and is about the size of a fist [1]. During a seventy year average lifetime it beats roughly about 2.5 billion times [2]. The heart is the muscular organ of the circulatory system that constantly pumps blood throughout the body.

Figure 1. Anatomy of the heart, Exterior front surface [3]

The heart is made up of four chambers, two on the right and two on the left. The chambers of the heart are known as atria and ventricles. The atria are the receiving chambers of the heart, receiving blood flowing back to the heart from

the rest of the body, and the ventricles are the chambers of the heart that pump the blood out of the heart. The Heart has four valves which control the direction of the blood flow through the heart. Valves act as highly controlled gates to blood flow, allowing blood to flow either from one chamber to another, or allowing blood to flow out of the heart [5]. The heart valves open and close depending on the pressure gradient generated within the heart and some muscles of the heart [4, 5]. The four valves are known the tricuspid valve, the pulmonic or pulmonary valve, the mitral valve and the aortic valve [5].

Figure 2. Anatomical structure of the human heart and the blood flow [6]

The arrows in Figure 2 represent the direction of blood flow in the heart. The light blue arrow represents the entry of blood into the right atrium from the superior and inferior vena cavae. From the right atrium, blood is pumped into the right ventricle. The light blue arrow also represents blood being pumped into the lungs from the right ventricle through the pulmonary arteries. The light red arrow

represents the entry of oxygen-rich blood into the heart's left atrium from the lungs through the pulmonary veins. From the left atrium, Oxygen-rich blood is pumped into the left ventricle. The light red arrow also represents the blood being pumped from the left ventricle to the rest of the body through the aorta [6].

1.2 ELECTRICAL SYSTEM OF THE HEART

The heart's electrical system, also called the cardiac conduction system, is made up of the following three main parts:

- The sinoatrial (SA) node, which is made of highly specialized cells, located in the right atrium of the heart
- The atrioventricular or aurioventricular (AV) node, also made of specialized cells, located in the interatrial septum
- The bundle of His and the Purkinje system which form the conduction pathway, located along the walls of the heart's ventricles [7].

The Electrical System of the Heart

Figure 3. Electrical conduction system of the heart [8]

Each heartbeat is a complex step of events that take place in the heart starting with an electrical impulse. A heartbeat can be divided into two main parts: diastole and atrial and ventricular systole. During diastole, the atria and ventricles of the heart relax and are filled with blood. At the end of diastole, the heart's atria contract and push blood into the ventricles. Next, the heart's ventricles contract starting from the apical region, pumping blood out of the heart into the aorta in case of left ventricle and the pulmonary arteries in case of right ventricle [6, 7].

The electrical impulse required to initiate a heartbeat begins with a signal from the highly specialized cells of the SA node [11]. This is why the SA node is sometimes called the heart's natural pacemaker. An electrical impulse signal is initiated when the superior and inferior vena cavae fill the heart's right atrium with blood [11]. The electrical signal is propagated throughout the right atrium, and

through the Bachmann's bundle to the left atrium, stimulating the left and right atria to contract and push blood into the left and right ventricles respectively. The electrical signal then arrives at the AV node which is located at the interatrial septum near the beginning of the ventricles. The electrical signal slows down for an instant to at the AV node and allows the heart's right and left ventricles to be filled with blood. The electrical signal now moves along a conduction pathway called the bundle of His, which is located in the walls of heart's ventricles as shown in Figure 3 [3].

From the bundle of His, the electrical signal divides into left and right bundle branches through the Purkinje network that connect spreads the signal into the walls of the heart's left and right ventricles [3]. At this point of time both the ventricles contract. However, this does not happen at exactly the same moment [11]. The left ventricle is found to contract an instant before the right ventricle. The ventricular contraction pushes the blood in the right ventricle through the pulmonary valve to lungs and the blood in the left ventricle through the aorta to the rest of the body [3, 7]. Each heart beat is a highly coordinated sequence of events. If there is any disruption in the sequence of the events that make up the heart beat as described above, it can lead to life threatening consequences.

1.3 ELECTROCARDIOGRAPHY (ECG)

An ECG is a test done in order to record the hearts electrical activity by placing electrodes on two or more different locations on the skin. In general an ECG recording consists of P, QRS and T waves as shown in Figure 4.

Figure 4. Representation of an ECG signal [9]

Following is a summary of the different segments of an ECG recording:

- The **P Wave** represents the spread of depolarization of cardiac cells through the right and left atria
- The **PR interval** is a measure of the time from the beginning of atrial contraction to the beginning of ventricular contraction. A large part of the PR interval also represents the delay of the electrical signal at the AV node as mentioned earlier
- The **QRS complex** represents the beginning of depolarization of the ventricles. As the cells in the ventricles depolarize the cells in the atria

repolarize but, this is not seen in the recording because the number of cells in the ventricles are much more than the number of cells in the atria

- The **QT interval** represents a complete cycle in which the ventricles depolarize and repolarize; a prolonged QT interval is generally associated with sudden death
- The **ST segment** represents the time when the ventricles are depolarized. The ST segment in a normal heart lies very close to the isoelectric line; an isoelectric line represents the periods in the cardiac cycle when the ECG recording has a 0 unit's value
- The **T wave** represents the repolarization of the ventricle [11].

Figure 5. Action Potential by Cardiac region [10]

An ECG waveform is a collection of action potentials and not a single action potential as shown the Figure 5.

1.4 ARRHYTHMIA

Cardiac arrhythmias are disturbances that can occur at the initiation of an electrical signal or during the flow of the electrical signal through its conduction pathway [11]. The electrical signal may pass through its conduction pathway too fast, slow or erratically – causing the heart to beat fast, slowly, or erratically respectively [2]. Such disturbances result in an improper flow of blood from the heart which can potentially damage other parts of the body. Arrhythmias can be divided into the following four types:

1. Premature beats or premature depolarization's
2. Supraventricular arrhythmias
3. Ventricular arrhythmias
4. Bradyarrhythmias [12].

1.4.1 PREMATURE DEPOLARIZATION'S

Premature depolarization's or premature beats is one of the most common types of arrhythmia and they often go unnoticed. They are harmless most of the time and generally do not need any kind of treatment [12]. Premature beats generally do not have any noticeable symptoms, but when symptoms do occur they are usually like fluttering in the chest or a feeling of a skipped heartbeat. Premature beats that occur in the atria are called premature atrial contractions and premature beats that occur in the ventricles are called premature ventricular contractions [12]. Figure 6 represents an ECG recording of a patient with premature ventricular contractions.

Figure 6. ECG recording of a heart with premature ventricular contractions [49]

1.4.2 SUPRAVENTRICULAR ARRHYTHMIAS

There are primarily three types of supraventricular arrhythmias: atrial fibrillation (AF), atrial flutter, paroxysmal supraventricular tachycardia (PSVT) [12].

1.4.2.1 ATRIAL FIBRILLATION

Atrial fibrillation or AF is also one of the most common types of arrhythmia [13]. AF occurs when disorganized and uncoordinated electrical signals cause the left and right atria to fibrillate. The meaning of the term "fibrillate" is to contract very fast and irregular manner [12]. In atrial fibrillation the electrical impulse does not start at the Sino atrial node, instead, they begin in another part of the atria or in the nearby pulmonary veins [12]. Due to this there is no coordination between the atria and the ventricles. This creates a fast and irregular heart rhythm. In AF, the ventricles may beat 100 to 175 times a minute, which is much greater than the normal heart rate of 60 to 100 beats a minute [13]. The white arrows in the right side of Figure 7 represent the irregularity of

electrical signals in the atria along with an ECG recording of a heart with atrial fibrillation.

Figure 7. Atrial fibrillation v/s normal rhythm [14]

1.4.2.2 ATRIAL FLUTTER

Atrial flutter is similar to atrial fibrillation, the only difference being that the electrical signals spread through the atria in a fast and regular instead of irregular manner; atrial flutter is less common compared to atrial fibrillation [12]. The white arrows in Figure 8 represent the fast and regular disturbances in the atria that occur during atrial flutter.

Figure 8. Electrical signal propagation in a heart with Atrial Flutter [15]

1.4.2.3 PAROXYSMAL SUPRAVENTRICULAR TACHYCARDIA

Paroxysmal Supraventricular Tachycardia or PSVT is characterized by very fast heart rate that begins and ends suddenly [12]. Periods of PSVT may occur only for a few beats or for many hours or days, and they often reoccur [11]. PSVT may result due to sudden rapid firing of the sino atrial node [11]. Figure 9 represents an ECG recording of a heart with PSVT. Seen in the ECG are sudden fast heart beats at the beginning and normal heart beats at the end.

Figure 9. Representation of an ECG of a heart with Paroxysmal Supraventricular Tachycardia [49]

Wolff-Parkinson-White (WPW) syndrome is a special type of PSVT. WPW syndrome is a condition in which the heart's electrical signals travel along an extra pathway from the atria to the ventricles as shown by the white arrows in Figure 10 below [12].

Figure 10 Representation of electrical signal propagation in a heart with Wolf Parkinson White Syndrome [15]

1.4.3 VENTRICULAR ARRHYTHMIAS

Ventricular arrhythmias refer to arrhythmias of the ventricles of the heart. They are usually dangerous and need medical treatment sooner or later. Ventricular arrhythmias can be classified into two types: Ventricular tachycardia and Ventricular fibrillation.

1.4.3.1 VENTRICULAR TACHYCARDIA

Ventricular tachycardia or VT is an abnormal rapid heartbeat that starts from a spot in the ventricles of the heart; this spot is called ectopic foci [11]. The normal heart usually beats between 60 and 100 times per minute [12] but, in case of VT the heart beats about 120 to 300 times a minute and the atria and ventricles are not coordinated [11]. Figure 11 is a representation of an ECG recording of a heart with VT. White arrows in Figure 12 represent the electrical signal that causes the fast heart beat to begin at the ventricles.

Figure 11. Representation of an ECG of a heart with Ventricular Tachycardia [49]

Figure 12. Representation of electrical signal propagation in a heart with Ventricular Tachycardia [15]

1.4.3.2 VENTRICULAR FIBRILLATION

Ventricular Fibrillation or VF is an abnormally irregular heart beat caused by rapid and uncoordinated fluttering contractions of the ventricles. The pumping chambers in the ventricles quiver without pumping appropriate amount of blood, which leads to a drop in blood pressure, and an improper supply of oxygen to the body organs. The most common known cause of VF is a heart attack. However, VF can occur whenever the heart muscles do not get enough oxygen to contract [16]. Figure 13 is a representation of an ECG signal from a patient with VF and the white arrows in Figure 14 show the small electrical signals in the ventricles of a heart which cause the ventricles to quiver without pumping appropriate amount of blood with VF.

Figure 13. Representation of an ECG of a heart with Ventricular Fibrillation [49]

Figure 14. Representation of electrical signal propagation in a heart with Ventricular Fibrillation [15]

1.4.4 BRADYCARDIA

Bradycardia is characterized by unusually slow heart rate, fewer than 50 to 60 beats per minute [12]. There are various reasons for bradycardia that include electrical signals that may be delayed, or completely blocked. Bradycardias can be classified into two types: Sinus bradycardia and Heart block.

1.4.4.1 SINUS BRADYCARDIA

Sinus bradycardia occurs when the rate at which the sino atrial node generates electrical signals decreases, which decreases the heart rate [11]. Figure 15 below represents an ECG of a heart with sinus bradycardia.

Figure 15. Representation of an ECG of a heart with Sinus Bradycardia [49]

1.4.4.2 HEART BLOCK

Heart block is a condition in which the electrical signals travel from the atria to the ventricles are slowed or blocked completely. This results in a delayed or lack of electrical communication between the atria and ventricles of the heart. Bundle branch block which is a common cause of heart block refers to a condition in which the electrical signals traveling through the ventricles of the heart are slowed or blocked completely from traveling along the ventricles as shown in Figure 16 [17].

Figure 16. Representation of a heart with heart block [14]

1.5 HEART FAILURE

Heart failure refers to a condition in which the heart is not able to pump enough blood to meet the body's requirements. The two main types of heart failures are:

- **Systolic heart failure:** Systolic heart failure is a condition in which the ventricles are unable to pump the required amount of blood during systole. This is the most common type of heart failure and is usually associated with dilated cardiomyopathy. Dilated cardiomyopathy is a condition in which the muscle walls of the heart become enlarged and thinner and are not able to pump blood to the body with as much force as

a normal heart [18]. Right part of Figure 17 represents a heart with dilated cardiomyopathy.

Figure 17. Representation of systolic and diastolic heart failure [18]

- Diastolic heart failure: Diastolic heart failure is a condition in which the ventricles of the heart are unable to fill up with blood, as much as a healthy heart and hence they are not able to pump enough blood to the body. Diastolic heart failure is related to hypertrophic cardiomyopathy, which is characterized by thickening and stiffening of ventricular muscle walls. About one third of all heart failures are diastolic heart failures [18]. Left part of Figure 17 represents a heart with hypertrophic cardiomyopathy.

1.6 CARDIAC PACING

Cardiac pacing is a surgical procedure in which the muscles of the heart are stimulated with an electrical signal from an electrode that is located on a pacing lead; this generates an electrical field to induce a cardiac action potential [19]. The minimum stimulus intensity and duration that is needed to give rise to a depolarizing wave from the electrodes is called stimulation threshold [20]. Stimulation threshold varies from patients to patients and also on the location of the electrode in the heart. The most common devices that are used for cardiac pacing include pacemakers, ICD (Implantable Cardioverter Defibrillators) and CRT (Cardiac Resynchronization therapy).

A worldwide cardiac pacing and implantable cardioverter-defibrillator survey was taken to see how many cardiac devices have been implanted in patients across the world. This survey was done in 2005 and 2009. The 2009 survey showed that 1,002,664 pacemakers were implanted, with 737,840 new implants and 264,824 replacements [21]. The United States of America (USA) had the largest number of cardiac pacemaker implants (225,567) [21]. Virtually all countries showed increases in implant numbers from 2005 to 2009. The survey also showed that 328,027 ICDs were implanted, with 222,407 new implants and 105,620 replacements [21]. Virtually all countries surveyed showed a significant rise in the use of ICDs with the largest implanter being the USA (133,262) with 434 new implants per million population [21].

1.7 HISTORY OF ELECTRICAL THERAPY

Cardiac pacing devices took two directions during the early development period:

1.7.1 DEVICES WHICH DO NOT STIMULATE IN THE PRESENCE OF A NATURAL CARDIAC BEAT

Initial cardiac devices were simple in design and would stimulate the heart in an asynchronous manner. Early in the development period researchers found out that a simple asynchronous pacemaker was stimulating the heart on the vulnerable portion of the T wave [22]. To overcome such situations researchers developed cardiac devices that could sense intrinsic heart beat and pace the heart in such a way that when stimulated, the electrical stimulus falls on the QRS portion of the heart cycle and not on the T wave. By 1966 the pacing devices available in the market had the ability to inhibit the electrical stimulus if a spontaneous event is detected and trigger a stimulus if an appropriate event is not detected; these devices had a single ventricular lead [22, 23].

1.7.2 DEVICES WHICH ATTEMPT TO RESTORE ATRIAL SYNCHRONY

The idea of atrial synchrony during pacing gained wide spread importance in the 1950's [22]. The main function of these devices was to sense the P wave and then accordingly stimulate the ventricle. Most of these devices had two leads-atrial lead that was placed in the right atria and was used to detect the presence

of the P wave and send this information to the pulse generator, and a ventricular lead to stimulate the ventricle. In some cases two ventricular leads were implanted in case one of the leads failed to function [22, 23].

Modern cardiac devices are highly sophisticated and optimized devices which give the physicians a lot more control compared to the early cardiac devices. In the future section we will discuss ICD (Implantable Cardioverter Defibrillator) and CRT (Cardiac Resynchronization Therapy) devices in particular.

1.8 ICD and CRT

Patients with cardiac disease generally die of one of 2 causes: sudden, unexpected cardiac death or progressive heart failure. ICD's (Implantable Cardioverter Defibrillator) and CRT's (Cardiac Resynchronization Therapy) play a major role in order to avoid and treat such conditions. ICD's have had an important impact on the treatment of heart failure [25]. CRT is used to restore more-normal and synchronized electrical contractions; CRT when combined with defibrillation can have a major impact on reducing the rate of deaths due to heart failure [25]. Although there remain many limitations and challenges to the appropriate application of ICDs or CRT's, there is no question that these devices have had, and in the future will have a major impact on the management of patients with left ventricular dysfunction.

ICD's were developed in the early 1970s to detect and automatically terminate ventricular tachycardia's by delivery a high voltage shock to the heart, this high voltage shock was expected to restore the regular heart beat [22]. Current-

generation ICDs are smaller in size than the earlier ICD's and they perform a variety of sophisticated functions, including atrial and ventricular defibrillation, antitachycardia pacing (ATP) i.e. termination of ventricular tachycardia's, backup bradycardia pacing and biventricular pacing [20]. The other essential features of the current generation ICD devices include detecting tachyarrhythmia, monitoring of heart rhythm after treatment, and storage of diagnostic results, and in some cases sending diagnosis results to the physicians wirelessly [22]. Following are two ways to sense tachyarrhythmia-

- Dedicated sensing - A bipole of small electrodes that is located near the catheter tip that is used to perform the sensing function
- Integrated sensing - A bipole from a tip electrode to a coil electrode that is also used to deliver defibrillation shocks [20].

After appropriate detection of tachycardia, pacing may be initiated or a defibrillation shock may be delivered. In other words an ICD monitors every beat of the heart and once an irregular pattern is sensed it delivers a shock [22, 23].

Current generation ICDs can be divided into 3 categories:

Figure 18. Types of ICDs. Left: Single chamber ICD, Right: Dual chamber ICD, Biventricular ICD: Combination of ICD and CRT [24]

- Single-chamber ICDs - Consists of a pulse generator and a right ventricular lead as shown in the right side of Figure 19.
- Dual-chamber ICDs - Consists of a pulse generator, a right atrial lead and a right ventricular lead as shown in the middle part of Figure 19. Stimulus is first delivered to the right atrium and then to the left ventricle.
- Biventricular ICD's - Consists of a pulse generator, right atrial lead, right ventricular lead and a left ventricular lead as shown in the left side of Figure 19. This device is a combination of an ICD and CRT and is used in patients with serious heart failure. Presence of additional left ventricular lead allows for a more controlled and accurate methods of restoring normal heart beat.

Due to a delay or block in the conduction pathway in the heart, the atria and ventricles lose their synchrony (coordination). This asynchrony can also be

present between the left and the right ventricles when one or both of the ventricles do not contract at the same time, this is called V-V asynchrony [46]. Over the last decade a lot of emphasis has been placed on the relationship between timing of left and right ventricular contraction. The two main reasons being, V-V asynchrony may:

- Reduce Cardiac Output; Cardiac output is the amount of blood pumped by the heart per minute
- Increase the atrial and ventricular blood filling pressures which can potentially change the shape and structure of ventricles

Cardiac resynchronization therapy is a relatively new pacing technique that is used to reduce the amount of atrial and ventricular electromechanical asynchrony in patients with heart conduction disorders [27]. This is generally achieved by multi-site pacing; multi-site pacing is pacing the heart at one or more sites in the atrium or ventricles. CRT avoids the conduction delays between the right and left ventricle induced by traditional pacemakers and ICD's that use only a right ventricular lead to pace [27].

There are two types of implantable heart failure, heart devices available in the market: a CRT pacemaker and a CRT defibrillator. A CRT pacemaker is a pacemaker with an additional left ventricular lead which performs the dual function of keeping the heart from beating slow and cardiac resynchronization. Similarly, A CRT defibrillator is an ICD with an additional left ventricular lead which performs the dual function of keeping the heart from beating too fast and

cardiac resynchronization. Both of these devices help to coordinate the heart's pumping action and improve blood flow. Figure 24 represents the different pacing conduction pathways possible by using CRT.

Figure 19. Representation of pacing conduction pathways [28]

1.9 OVERVIEW OF CRT IMPLANTATION PROCEDURE

In order to appreciate the importance of this project it would be very helpful to briefly understand the CRT implantation procedure and the challenges associated with it. Figure 25 represents the coronary arterial and venous vasculature of the heart which is the target location to implant the left ventricular (LV) lead.

The standard approach to implant a CRT system is the transvenous epicardial approach. The transvenous approach is the most common because it is less

invasive and can be performed using general anesthesia, the most challenging part of this procedure is to place the LV lead in the coronary vasculature of the heart [29].

Figure 20. Coronary Venous and Arterial Vasculature of the heart [30]

The order in which the ventricular leads are implanted vary. Generally, the right ventricular (RV) lead is implanted first, implanting the RV lead first gives the option of backup pacing in case something goes wrong during the surgery and it may also give information about right atrial and right ventricular synchrony [30].

One advantage of implanting the LV lead first is that it might be easier to maneuver and implant the LV lead faster as there are no other leads around [30].

In order to implant the LV lead a cannula (small tube like structure) is inserted into the Coronary Sinus to allow the LV lead to pass through it, The cannulation

of coronary sinus is generally a very challenging step because the position of the orifice is very difficult to locate [30]. Angle of coronary sinus beginning and tortuosity are very important factors to be considered for this procedure [30]. Contrast injection is very often used in order to identify the location of the orifice; Contrast injection shows the coronary venogram [29, 30]. Figure 21, Figure 22 and Figure 23 represent the different angles at which the venogram images are captured using fluoroscopic techniques. Once the lead is inserted into the Coronary sinus the lead is guided to the desired location.

Once the target vein is identified and the LV lead is placed into it, the next step is to test the LV lead for phrenic nerve stimulation. Phrenic nerve stimulation is a common problem encountered during the CRT implant procedure. Phrenic nerve is falsely stimulated when the LV lead is positioned in such a way that the output pulses stimulate the phrenic nerve. The LV lead is paced at desired pulse width and voltage and at higher outputs to study the effects of pacing on phrenic nerve [29, 30].

If there is any evidence of phrenic nerve stimulation one of the following steps is taken:

- Repositioning the LV lead in the same vessel, Even 1-2 cm change in the position of the LV lead can make a difference
- Using different pacing configurations
- Placing the lead in a different vessel [30].

The next step in the procedure is to measure and record the impedance values, Intrinsic R wave sensing and testing LV lead pacing threshold [30].

Figure 21. Representation of Anterior Venogram [30]

Figure 22. Representation of Right Anterior Oblique Venogram [30]

Figure 23. Representation of Left Anterior Oblique Venogram [30]

1.10 COMMON PACING COMPLICATIONS

Common pacing complications that can occur in patients implanted with a CRT system include high pacing thresholds and phrenic nerve or diaphragmatic stimulation [31].

Figure 24. Representation of right and left phrenic nerve [31]

Some patients require high pacing thresholds and hence higher energy is required to pace the heart at the pacing site. Since the battery life of all pulse generators is limited, the patients who require high pacing thresholds will have shorter battery life which might require another surgery to replace the pulse generator or might cause the pacing to be ineffective [31, 32].

Phrenic nerve and diaphragmatic stimulation is a very common pacing complication that occurs when the electrical signals produced by the pulse generator stimulate the phrenic nerve and the diaphragm either directly or through the phrenic nerve. In Figure 24 we see that the left and right phrenic nerve is located very close to the heart. Such unintentional stimulation of the phrenic nerve or/and diaphragm results in side effects such as hiccups [32]. Stimulation of the phrenic nerve and the diaphragm depends on the patient's body, which makes it challenging for the physicians because there may be no phrenic nerve or diaphragm stimulation observed while the patient is undergoing the surgery but, the phrenic nerve or diaphragm might be stimulated after the surgery when the patient is actively moving [32]. Both high pacing threshold and phrenic nerve or diaphragmatic stimulation are often due to the location of the pacing lead electrode and in serious cases may require the LV lead to be surgically repositioned or CRT be disabled [27].

1.13 Q-LV TIMING

Q-LV timing is defined as the time interval from the first QRS deflection on a surface ECG to local intrinsic activation at the LV stimulation site [36]. In one

sense it is the time taken by the ventricular depolarization wave to reach the electrode site in the left ventricle [36]. To measure the Q-LV timing, the difference between the onset of the QRS complex on the ECG and the first positive or negative peak from the left ventricular EGM is calculated [36]. Figure 25 represents two examples of Q-LV measurements. In Example 1 the peak in the left EGM is negative after the detection of the first QRS complex, and in Example 2 the peak in the left EGM is positive after the detection of the first QRS complex.

Figure 25. Example representations of Q-LV measurement [36]

A lot of patients implanted with CRT do not show expected outcomes, these patients are called non responders. Reducing the rate of CRT non-responders has been a challenge for some time now. Generally the parameters that are considered to evaluate the efficacy the following factors are considered:

- Left ventricular end systolic volume (LVESV), which is the volume of blood present in the ventricles after a contraction.

- Left ventricular end diastolic volume (LVEDV), which is the volume of blood present in the ventricles at the time of relaxation.
- Left ventricular ejection fraction (LVEF), this is defined as the ratio of the left ventricular stroke volume (SV) to the left ventricular end-diastolic volume (LVEDV). SV is obtained by subtracting the LVESV from LVEDV [47].
- Quality of life.

2.0 OBJECTIVE

The objective of this project is to develop an algorithm to detect and track points on a lead. The algorithm will track the motion of the points over time and generate displacement plots over time. Since images can vary widely from patient to patient and the angle at which the images are captured, the algorithm should have tunable parameters that will enable the user to track points in the majority of the video files despite the variations.

3.0 METHODS

3.1 INTRODUCTION

To perform explicit tracking of feature points, the features point must be discrete and not in a form of continuum texture which is the generally the case when using edge detection algorithms like the Sobel, Roberts and Canny edge detection algorithms [37]. Edge detection is a technique that is used to characterize significant changes in intensity in an image [48].

The points can be isolated from the raw image by setting a threshold of 0.38 using the canny edge detection algorithm after a few trial and error experiments with the threshold values. Threshold values are values that take into account the change in intensity values from one pixel to another in both X and Y direction. Higher the threshold value, higher is the change in intensity required to consider the point as an interest point. Though the boundaries are isolated by selecting a particular threshold value it is important to note that we do not have any localized information, which means that the areas of interest cannot be represented by a single point.

We concluded that edge detection methods are not appropriate for our purpose. In order to track the points the first step would be to detect the points and obtain localized information (a single value) of each of the points for each frame in a video.

One approach to detect the electrodes would be to track the edges and the corners in an image. Detecting corners not only tracks the electrodes but it also avoids tracking the leads because a lead as seen from our earlier edge detection figure is detected as two parallel lines without any intersections.

3.2 BASIC IDEA OF CORNER DETECTION

The definition of a corner in an image is a point in an image where the local autocorrelation function has a distinct peak value [38]. Consider a small green window W tracing an image I as shown in Figure 26. In Figure 26a the green window is tracing a region of constant intensity and shifting the window in any direction does not cause any change in intensity values in the window which indicates that the window is tracing a flat intensity region. In Figure 26b it is seen that if the green window is shifted perpendicular to the direction of the line (edge) there will be a large change in intensity but if the window is shifted along the direction of the line (edge) the change in intensity will be less, this indicates the presence of an edge in this region. In Figure 26c the green window is placed over a corner and in this case shifting the window in any direction will result in a large amount of change in intensity values in the window, which indicates the presence of a corner in this region.

Figure 26. Basic idea of corner detection. 26a represents a window tracing a flat region, 26b represents a window tracing an edge, and 26c represents a window tracing a corner [41].

Following are some of the requirements of a corner detector algorithm;

- True corners should be detected and false corners should be suppressed
- Corner points should be localized, which implies that the corner positions are to be found accurately
- Corner detector should have high repeatability
- Corner detector should not be affected by the presence of noise
- Corner detector should be computationally efficient [39].

The Harris corner detector algorithm is based on the local auto-correlation function of a signal, the local auto-correlation is a measure of the local changes of a signal in a window as the window is shifted by a small amount in different directions [40].

Change of intensity in an image by a shift (u,v) is given by the formula

$$E(u,v) = \sum_{x,y} [I(x+u, y+v) - I(x,y)]^2 * W(x,y) \quad (1)$$

Here, $E(u,v)$ is the sum of square difference in change in intensity from a point $I(x,y)$ to a point $I(x+u, y+v)$ multiplied by a weight window $W(x,y)$ centered at a point (x,y) . $I(x+u, y+v)$ is the intensity of a point in the neighborhood of point $I(x,y)$.

Taylor series expansion in 1 dimension is given as:

$$F(t_0 + \Delta t) = F(t_0) + F'(t_0)\Delta t + (F''(t_0)\Delta t^2)/2! + (F'''(t_0)\Delta t^3)/3! \\ \dots + (F^{(n)}(t_0)\Delta t^n)/n!$$

for t near any fixed point t_0 .

Where, $t = t_0 + \Delta t$ and $F', F'' \dots F^{(n)}$ are the first, second and the nth order derivatives of F .

Similarly by using Taylor series expansion in 2 dimensions we can rewrite the first term in the right hand side of equation (1) as

$$[I(x+u, y+v) - I(x,y)]^2 = [I(x,y) + uI_x + vI_y - I(x,y)]^2 \quad (2)$$

Where I_x and I_y are the image gradients in X and Y direction's respectively.

Equation (2) can be simplified as

$$[I(x+u, y+v) - I(x,y)]^2 = [uI_x + vI_y]^2 \\ [I(x+u, y+v) - I(x,y)]^2 = u^2 I_x^2 + v^2 I_y^2 + 2uvI_x I_y \quad (3)$$

After substituting (3) in (1) , Equation (1) becomes

$$E(u,v)=\sum_{x,y} [u^2 I_x^2 + v^2 I_y^2 + 2uv I_x I_y] * W(x,y) \quad (4)$$

Writing the above equation in matrix form

$$E(u,v) = \sum_{x,y} W(x,y) * \left[(u \ v) \begin{pmatrix} I_x \\ I_y \end{pmatrix} \right]^2 \quad (5)$$

Expanding Equation (5), we get

$$E(u,v) = \sum_{x,y} W(x,y) * \begin{pmatrix} (u \ v) & I_x & (I_x \ I_y) & u \\ & I_y & & v \end{pmatrix}$$

$$E(u,v) = (u \ v) \left[\sum_{x,y} W(x,y) * \begin{pmatrix} I_x \\ I_y \end{pmatrix} (I_x \ I_y) \right] \begin{pmatrix} u \\ v \end{pmatrix} \quad (6)$$

In our case $u=1$ and $v=1$ since the change in position is just by 1 pixel in the X and Y direction

Therefore,

$$E(u,v) = \left[\sum_{x,y} W(x,y) * \begin{pmatrix} I_x \\ I_y \end{pmatrix} (I_x \ I_y) \right]$$

$$E(u,v) = \sum_{x,y} W(x,y) * \begin{pmatrix} I_x I_x & I_x I_y \\ I_x I_y & I_y I_y \end{pmatrix} \quad (7)$$

$$\text{Let } M = \begin{pmatrix} I_x I_x & I_x I_y \\ I_x I_y & I_y I_y \end{pmatrix}$$

Consider the equation shown below

$$\begin{bmatrix} u & v \end{bmatrix} \begin{bmatrix} A & B \\ C & D \end{bmatrix} \begin{bmatrix} u \\ v \end{bmatrix} = 1 \quad (8)$$

$$\text{Let, } Q = \begin{bmatrix} A & B \\ C & D \end{bmatrix}$$

Equation (8) is an equation of an ellipse in matrix form where the eigen values of the matrix Q represent the elongation and size of the ellipse and the eigen vectors of Q represent the angle of rotation of the ellipse [37, 39].

Figure 27. Ellipse as a function of eigen values of the matrix E [40].

The matrix E gives us the per pixel estimate of the local auto correlation function, this information is got by convolving the matrix M with a weighing function, W [40].

For matching features from frame to frame using windows, Anandan in 1984 showed that the inverse of the matrix E provides a lower bound on the

uncertainty while matching features [40]. One way to analyze this matrix is to perform eigenvalue analysis of matrix E , which produces two eigen values λ_1 and λ_2 and two eigenvector directions. And since the uncertainty depends more on the smaller eigen values, finding maxima in the smaller eigen value is a good option to search good features [40]. From Figure 27 we see that if we try many different orientations of the ellipse, the max change in the shape of the ellipse would be by changing the value of λ_{min} ,

After the initial work by Anandan, Harris and Stephens in 1988 were the first to propose that local maxima of the smaller eigen value is not the only way to search for feature, They showed that by using local maxima in rotationally invariant scalar measures derived from the auto-correlation matrix E can also be used to find and match features [40]. A simpler formula as shown below was proposed by Harris and Stephens in 1988:

$$R = \det(E) - \text{trace}(E)^2 = \lambda_1\lambda_2 - k(\lambda_1 + \lambda_2)^2 \quad (9)$$

Where R is called as the corner measure matrix, λ_1 and λ_2 are the eigen values of matrix E and k is an empirical constant called the sensitivity factor whose value ranges from 0.03 to 0.06. k is used to change the sensitivity of the algorithm to detect corners.

- If the values of λ_1 and λ_2 are small then this would mean that shifting the window in both direction results in a small change in intensity and hence this represents a region of constant intensity in the image, “flat region”.

- If one of the eigen values λ_1 and λ_2 are large and the other eigen value is small then this means that there is a large change in intensity by shifting the window in only one direction. This represents an edge.
- If both the eigen values λ_1 and λ_2 are large then this means that there is a large change in intensity by shifting the window in any direction. This represents a corner.

Figure 28 represents a plot of the two eigen values λ_1 and λ_2 of the matrix E and the classification of feature points in an image as flat, edge or corner depending upon λ_1 and λ_2 .

Figure 28. Classification of an image region as flat, edge or corner using eigen values [41]

3.3 PROPERTIES OF HARRIS CORNER DETECTOR

3.3.1 ROTATION INVARIANCE

When an image is rotated the ellipse that is defined by its autocorrelation matrix E is also rotated, but the eigen values (shape) of the ellipse remain the same. This feature makes the Harris corner detector rotation invariant. This is illustrated in Figure 29.

Figure 29. Rotational invariance of Harris corner detector [41]

3.3.2 PARTIAL INVARIANCE TO INTENSITY CHANGES

Intensity of an image can be shifted or scaled. If I_{old} is the old intensity value of an image and I_{new} is the intensity value of the same image, then intensity shifting and intensity scaling are given as

$$I_{old} + p = I_{new} \text{ (Intensity shifting)}$$

$$I_{old} * p = I_{new} \text{ (Intensity scaling)}$$

P is a constant numeric value.

Scale invariance property of Harris detector depends on the threshold value, if the change in intensity shifts or scales points above or below threshold value than the points that have their new intensity levels above the threshold level will be detected as false corner points and the interest points whose new intensity

values are lower than the threshold value will go undetected.

Figure 30. Partial invariance to intensity changes. Top: Intensity of an image is scaled or shifted to a value higher than its original value; In this case some points which are not of interest are detected as their intensity value is above the threshold value. Bottom: Intensity of an image is shifted or scaled to a value lower than the original value, In this case some points of interest are shifted below the threshold level and they go undetected

[41]

This property is important to us because the intensities of images can vary depending on the contrast agent injected into the heart and also the angle at which the images are captured as discussed in section 1.10. This property

ensures the tracking of points in different images unless the changes in intensity values are not too large.

3.3.3 SCALE INVARIANCE

Harris corner detector is not invariant to changes in scale as shown in Figure 31. This is because the convolution window size remains the same even when the scale of the image changes. This can be overcome by choosing a bigger window as shown in the figure below.

Figure 31. Corner point detection is limited by the size of window. A small window traversing over a scaled image classifies all the points as edges [41]

3.4 IMPLEMENTATION

3.4.1 CONCEPT OF MOVING WINDOW

It is important to understand the concept of moving window before we can move to operations like spatial convolution and spatial correlation. Let us consider an image of size $M \times N$ as shown in Figure 32 below.

Figure 32. Representaion of a window and neighborhood [42]

A window is a matrix of values, generally of odd size that defines how the output image pixel corresponding to the input image pixel will depend on the neighboring pixels of the input image. The output of the moving window transform at each pixel is the weighted average of the pixels in the neighborhood. In Figure 33 below we see that the output image pixel value corresponding to the

input image pixel value is the weighted average of all the pixels in the window. The window traces the entire image and populates the output image matrix, which may or may not be the equal to the size of the input image matrix.

Figure 33. Moving window representation [42]

If all the values in the window are 1, then the window is known as an averaging window because it weighs all the pixels by the same factor. Figure 34 shows the results of an averaging window transform on an image.

$$W = \frac{1}{9} \begin{bmatrix} 1 & 1 & 1 \\ 1 & 1 & 1 \\ 1 & 1 & 1 \end{bmatrix}, \text{ W is the window matrix}$$

Figure 34. Representation of an averaging operation [42]

When the window is operating at corners and borders of an image, the window might fall outside the image matrix. There are several solutions to overcome this situation such as zero padding the input image, mirroring the edges etc. Each of these techniques has its advantages and disadvantages. In our application we will ignore $(m-1)$ edges of the output image where m is the size of window being used. The rationale for ignoring the edges is presented in future sections. By ignoring the edges of the output image the resultant image will be of smaller size than the input image and give rise to an offset. But this offset in image size will be treated effectively in our algorithm.

3.4.2 SPATIAL CONVOLUTION AND CORRELATION

Spatial correlation is a very commonly used technique in template matching or feature matching applications in digital image processing. In this section we will discuss the implementation of spatial correlation and spatial convolution.

The mathematical formula for spatial correlation is:

$$g(x, y) = \sum_{s=-a}^a \sum_{t=-b}^b w(s, t) f(x + s, y + t)$$

Where, $a = (m-1)/2$ and $b = (n-1)/2$

Figure 35. Representation of Spatial correlation operation [42]

Where f is the input image matrix of size $M \times N$, w is the mask matrix of size $m \times n$ with elements represented in the yellow matrix in Figure 35, and g is the output image matrix.

In case the image matrix is equal to the mask matrix then it is called spatial auto correlation and the maximum value of the output will be at position where the two matrices are placed over each other such that $f(x, y)$ coincides with $w(0, 0)$.

Spatial convolution is similar to spatial correlation except that each pixel in the neighborhood of the pixel in consideration is multiplied by the corresponding matrix value after the window matrix is rotated by 180° as shown in Figure 36.

The sum of the products is the value of the output pixel.

The mathematical expression for spatial convolution is:

$$g(x, y) = \sum_{s=-a}^a \sum_{t=-b}^b w(s, t) f(x - s, y - t)$$

Where, $a = (m-1)/2$ and $b = (n-1)/2$

Figure 36. Rotation of convolution mask by 180 [42]

Where f is the input image matrix of size $M \times N$, w is the mask matrix of size $m \times n$, and g is the output image matrix.

Figure 37. Representation of Spatial convolution operation [42]

If $w(x,y)$ is symmetric, that is $w(x,y)=w(-x,-y)$, then spatial convolution is equivalent to spatial correlation.

3.4.3 GRADIENT OF AN IMAGE

The gradient of an image shows how the intensity values of the pixels in an image change with direction. The gradient operator is a vector that has two components; magnitude and direction. The magnitude of the gradient operator gives us information about how quickly the intensities values are changing and the direction of the gradient operator gives information about the direction in which the intensity change occurs. Almost all of the interest point (like extrema, inflection points etc.,) in an image can be found by solving equations containing derivatives [43].

First order derivative at a pixel location is calculated by calculating the partial derivatives of that pixel location, the derivatives of a digital function are defined in terms of differences [43]. The approximations to the first derivative must satisfy the following properties:

- Must be zero in areas of constant intensity
- Must be non-zero at the onset of an intensity step or ramp
- Must be non-zero along ramps [43].

Partial derivative in the X direction at pixel location $I(x,y)$ is given by

$$\frac{\partial I(x,y)}{\partial x} = \lim_{\Delta x \rightarrow 0} \frac{I(x + \Delta x, y) - I(x, y)}{\Delta x}$$

Partial derivative in the Y direction at pixel location $I(x,y)$ is given by

$$\frac{\partial I(x,y)}{\partial y} = \lim_{\Delta y \rightarrow 0} \frac{I(x, y + \Delta y) - I(x, y)}{\Delta y}$$

Let $\Delta x=1$ and $\Delta y=1$, where Δx and Δy is change in intensity of pixels

adjacent to $I(x, y)$. The above equations are reduced to

$$\frac{\partial I(x,y)}{\partial x} = I(x + 1, y) - I(x, y)$$

$$\frac{\partial I(x,y)}{\partial y} = I(x, y + 1) - I(x, y)$$

Gradient of an image is the combination of partial derivative of the image in X and Y direction. Mathematical representation of a gradient operator is:

$$\nabla I(x,y) = \begin{bmatrix} \frac{\partial I(x,y)}{\partial x} \\ \frac{\partial I(x,y)}{\partial y} \end{bmatrix} = \begin{bmatrix} I_x \\ I_y \end{bmatrix}$$

Magnitude of the gradient vector is

$$M(x,y) = \text{mag}(x,y) = (I_x^2 + I_y^2)^{1/2}$$

Where I_x , I_y and $M(x,y)$ are matrices of the same size as the original image matrix. Generally $M(x,y)$ is referred to as the gradient of an image.

The direction of the gradient vector is given by the angle α measured with respect to the X-axis.

$$\alpha(x,y) = \tan^{-1} \left[\frac{I_x}{I_y} \right]$$

$\alpha(x,y)$ is also an matrix of the same size as the original image matrix and the direction of an edge at any arbitrary point (x,y) is orthogonal to the direction $\alpha(x,y)$, of the gradient vector at that point [43]. The gradient equations for all pertinent values of x and y can be implemented by convolving the masks shown below with input images.

$$\begin{bmatrix} -1 & 1 \end{bmatrix}$$

$$\begin{bmatrix} -1 \\ 1 \end{bmatrix}$$

When diagonal edge direction is of interest we use a 3x3 matrix because they are symmetric about their center point, these masks take into account the data on the opposite sides of the center point which represents the diagonals and thus carry more information about the direction of an edge [43]. The simplest approximation to a partial derivative using 3x3 masks is shown below. These masks are called Prewitt operators.

-1	-1	-1	-1	0	1
0	0	0	-1	0	1
1	1	1	-1	0	1

The difference between the first and third rows of the 3x3 matrix approximates the derivative in the X direction, and the difference between the first and third columns of the 3x3 matrix approximates the derivative in Y direction.

Low pass filters are very often called smoothing filters since they remove the high frequency components of an image. A smoothing filter is used to blur an image for a gross representation of objects of interest, such that the intensity of smaller objects blends with the background and this would help to differentiate the larger objects of interest [43]. A 2D Gaussian blur filter will be used as our low pass filtering window function mainly because Gaussian filters remove noise and at the same time preserves edges. Mathematical expression for a 2D Gaussian filter is given as

$$g(x, y) = \frac{1}{2\pi\sigma^2} e^{-(x^2 + y^2) / 2\sigma^2}$$

Where X is the distance from the mask center in the horizontal axis and Y is the distance from the mask center in the vertical axis and σ is the standard deviation of the Gaussian distribution. The sum of all elements in a Gaussian kernel sums up to 1. σ determines the width of the Gaussian kernel, radius of the kernel is determined by multiplying σ by a constant value.

From Figure 38, Figure 39 and Figure 40 we see that by increasing the value of σ , the radius of the filter increases which means that a filter with $\sigma=3$ takes into consideration a larger neighborhood than a filter with $\sigma=2$.

Figure 38. Gaussian filter with sigma =1 and radius = 6*sigma

Figure 39. Gaussian filter with $\sigma = 2$ and radius = $6 \cdot \sigma$

Figure 40. Gaussian filter with $\sigma = 3$ and radius = $6 \cdot \sigma$

Hence, by increasing the size of the Gaussian filter we can reduce the effects of noise but the disadvantage of increasing the size of the Gaussian filter kernel is that sometimes objects of interest might also be treated as noise due to the large neighborhood in consideration.

3.5 NON MAXIMA SUPPRESSION

The next step in our algorithm is to pick points of interest using non maxima suppression. Non maxima suppression is a technique used to search a local maxima in an image matrix, where a local maxima has a value greater than all its neighbors. It is a crucial step in many computer vision/ image processing applications. Most feature detector algorithms look only for the local maxima, this can lead to an uneven distribution of feature points across the image. To overcome this problem, Brown, Szeliski and Winder (2005) proposed an adaptive non maxima suppression technique where features are detected only if they are both local maxima and their response is significantly greater than that of all of its neighbors within a predetermined radius [44].

For our application we use a simple second order statistical order filter to find the local maxima first and then we use thresholding techniques to implement an adaptive non maxima suppression approach to find feature points.

Following are the steps performed by a statistical order filter:

- Sorting the numbers in an array of given size from smaller to greater – the smallest number is the first element of the window, i.e. first row and first column. The largest number is the last element of the window i.e. last row and last column.
- Replacing the pixel in consideration (Image array) with the specified number in the sorted order filtered array, for example the number specified in the window can be largest, smallest or median etc.

Figure 41 and Figure 42 represent the steps described above; operation of a 5x5 and a 7x7 statistical order filter is demonstrated.

Figure 41. 5x5 Statistical Order filter operation on an image array

Figure 42. 7x7 Statistical Order filter operation on an image array

Figure 43. 5x5 statistical order filter operation representation on corner measure matrix

Figure 44. 9x9 statistical order filter operation representation on corner measure matrix

Figure 45. 15x15 statistical order filter operation representation on corner matrix

Figures 43, 44 and 45 represent the operation of a 5x5, 9x9 and 15x15 statistical order filter respectively on the corner measure matrix. From these figures we see that as the size of the order filter increases the peak becomes more flat, this is because the size of the window increases which increases the neighborhood in consideration. Larger window size also reduces the number of local maxima's. The size of the statistical order filter is decreased when a large number of corner points of interest are very close to each other and the size of the filter is increased when the points of interest are distributed far apart from each other.

Next, the elements in the corner measure matrix are normalized with respect to the maximum value in the matrix by using the following formula.

$$\text{New Corner Measure Matrix} = (1000 / \text{MaxValue Of Old Corner Measure Matrix}) * \text{Old Matrix Corner Measure}$$

Interest points are picked from this new matrix if the following conditions are met:

- Element in the corner measure matrix also called Harris equation matrix is equal to the element in the statistical order filter matrix element.
- Corner measure/ Harris equation matrix element is greater than the threshold value.

The above logic is represented in a flowchart format:

Next, A new matrix called initial corner point matrix is initialized, and The indices of the this initial corner point matrix corresponding to the indices of the elements

in the Harris equation/ Corner Measure matrix that satisfy the above logic are set to 1 and the other elements in the initial corner point matrix are set to 0.

Figure 46. Initial corner point matrix

Figure 46 is a representation of an initial corner point matrix after applying the above algorithm, in this figure we see that the number of corner points detected are much greater than the points of interest.

At this point it is worthwhile to emphasize the effects of Gaussian filter kernel size, statistical order filter kernel size, threshold value and the sensitivity factor on selecting desired corner points. From Figure 47 we see that by keeping the statistical order filter kernel size and the sensitivity factor constant and increasing the Gaussian filter kernel size (σ) the number of points detected as corners decreases.

Similarly, From Figure 48 we see that by keeping the Gaussian filter kernel size and the sensitivity factor constant and increasing the statistical order filter kernel size the number of points detected as corners decreases and from Figure 49 below we see that by keeping the statistical order filter kernel size and the Gaussian filter kernel size constant and increasing the sensitivity factor (k) the number of points detected as corners increases.

Figure 47. Initial corner point matrix as a function of Gaussian filter kernel size. As the Gaussian filter kernel size increases the number of points detected as corners decreases

Figure 48. Initial corner point matrix as a function of statistical order filter kernel size. As the statistical order filter kernel size increases the number of points detected as corners decreases

Figure 49. Initial corner point matrix as a function of sensitivity factor (k). As the sensitivity factor (k) increases the number of points detected as corners increases

From the initial corner point matrix it is very clear that the number of corner points detected are greater than the number of electrode corner points that we are interested in. In order to pick out the corner points of our interest we use a simple thresholding logic shown in a flowchart below. Threshold value is increased by 10% if the number of corner points detected is greater than the specified number and the threshold value is decreased by 10% if the number of corner points detected is lesser than the specified number. The process is repeated until the specified number of points is obtained.

Figure 50. Representation of a final corner point detection matrix. Corner points detected are at the center of the orange circles

From the top and bottom part of Figure 50 we see that the four corner points are selected from the initial corner point matrix using the thresholding algorithm.

Since we have been ignoring the boundaries during performing convolution operation, this results in an offset in the size of images and this offset is taken

care by zero padding the final corner point matrix to match the size of the original image matrix. Now, matrix indices corresponding to the corner points are saved in an array and appropriately scaled to match the original input image matrix indices.

Initial Corner Point Matrix without ignoring boundaries during convolution

Number of false corner points detected at the boundaries are more when boundaries are not ignored while performing convolution.

Initial Corner Point Matrix by ignoring boundaries during convolution

Figure 51. Comparison of initial corner point detection matrix based on performing convolution with and without boundaries

The top part of Figure 51 represents the initial corner point matrix calculated without ignoring the boundaries while performing the convolution operation and

the bottom part of the same Figure represents the initial corner point matrix calculated by ignoring the boundaries while performing convolution. We see that the number of false point detected are much more in the top part of the figure around the boundaries and this large number of corner points might lead to selecting false corner points and hence these corner points have to be removed. Performing convolution by ignoring boundaries solves this problem of false corner points to some extent.

Localized point information is got by running this algorithm on different frames in a video. The next step is to calculate point displacements from frame to frame. In order to calculate the displacements, the above algorithm is run over each frame in a video and the corner points for each frame are saved. To estimate the displacement of the corner points from frame to frame we calculate the Euclidean distance using the corner point information of each frame. The mathematical formula for Euclidean distance is

$$Euclidean\ Distance = [(x_i - x_j)^2 + (y_i - y_j)^2]^{1/2}$$

Where (x_i, y_i) are the corner point location in frame i and (x_j, y_j) are the corner point locations in frame j . The Euclidean distance calculated has the units as pixels, in order to convert the units to millimeters (mm) we multiply the Euclidean distance with the image pixel spacing number.

4.0 RESULTS

The tracking algorithm was run on a total of 10 videos. In order to evaluate the performance of the tracking algorithm, the areas of interest in a video were manually tracked from frame to frame using ImageJ and their successive frame displacements were calculated, this was compared with the displacements calculated by the algorithm. Differences in displacements between manual and automated tracking and standard error of the differences were calculated.

In order to evaluate the computational efficiency of the algorithm, time for the algorithm to run over all the frames in a video and generate the displacement plots was found.

The algorithm was run on video file 1 with the following settings

- Number of points to be detected = 4
- Gaussian filter standard deviation, Sigma = 3
- Statistical Order Filter Radius = 7
- Sensitivity Factor, $k = 0.02$
- Threshold Value = 18

Figure 52. Point displacement plots for video file 1

Figure 53. Standard error in point displacement chart for video file 1

The peak standard error in displacement value between manual and automated tracking was found to be 0.16mm for point 1. The time required to generate displacement plots was found to be about 1.26 seconds after the user selects the area of interest in the first frame of the video.

5.0 DISCUSSION AND CONCLUSIONS

From the results of running the tracking algorithm on video files we see that the tracking algorithm was successfully able to generate displacement plots with minimum user input in a variety of video files. From the standard error charts that compared manual tracking of points with automated tracking we see that the standard error by tracking the points by running the algorithm is very small and it can be considered negligible for initial research purposes. This error is because while manually tracking the points an attempt was made to select the center of the points whereas, the tracking algorithm detected the corners of the points.

The time taken by the algorithm to generate the displacement plots was found to be less than 2 seconds for all the videos. This shows that the tracking algorithm is computationally efficient and it can be used in real time. Special case videos discussed in the results section show that the tracking algorithm is robust and by choosing the right set of settings the user can obtain the appropriate electrode displacements in a variety of videos.

6.0 FUTURE WORK

Future work would begin by building a graphical user interface (GUI) to run the tracking algorithm. The GUI would enable the user to select the settings that they think are optimal to run the algorithm on different video files. The GUI will provide the user options to choose the number of points to track, threshold value, and order filter radius and so on.

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