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EVALUATION OF AN INTEGRATED ANALOG FRONT-END COMPONENT FOR ECG PACEMAKER PULSE DETECTION

Master’s Thesis

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ABSTRACT

Berihun Gebrehawariat Gebreselasse: Evaluation of an Integrated Analog Front-end Component for ECG Pacemaker Pulse Detection
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Accurate detection of pulses from an artificial pacemaker device could help in achieving correct automated analysis and visualization of the ECG on a monitoring device. With the development of modern artificial cardiac pacemakers and their small amplitude and duration spikes detection of their signal is becoming increasingly challenging. MAX30001 is a single-channel analog front-end designed for biopotential and bioimpedance measurements. This thesis work was aimed at evaluating MAX30001 in cardiac pacemaker pulses detection. The AFE did not validate a pacing event rather it captured edge information of the pulses and the ECG records at a sampling frequency of 128 Hz. An algorithm was then developed that converts the edge information to pulse detections. It defined consecutive opposite edges that fell within a minimum and maximum pulse duration of 10-µs and 2-ms as valid pacing pulses. The algorithm was further developed to reject narrow pulses and incorporate a refractory period.

Eight batches of square pulse trains with pulse durations ranging from 1-ms to 0.1-ms and each batch having 40 amplitude steps were introduced to the AFE. It was found that at default configurations, a complete detection was achieved to a threshold amplitude of 0.4-mV for all the available pulse widths. At ±337.5-mV detector threshold voltage the amplitude threshold rises to 1-mV. A publicly available simulator-generated ECG dataset of 624 files corrupted by tremor containing pacemaker pulses with pulse durations (from 100-µs to 2-ms) was supplied to the AFE. For this input, at the default configuration amplitude values 0.75-mV had sensitivity percentages higher than 80% for all the available pulse widths. The effect of tremor in triggering false pacing indications was found insignificant. Measurements were undertaken using an ECG dataset of 100 patients with artificial pacemakers, each patient having a maximum of 5 trials of pacemaker settings. Annotations of the pacemaker pulses were included with the data files based on eight leads, but the analysis was performed only on lead II. Of the total records, it was possible to analyze 414 records on two settings of the AFE. At the default setting, a sensitivity of 80.6 % and PPV of 97% were registered. A sensitivity of 75.7% and PPV of 98.2% were obtained with a channel gain group of 240/22.5/90 and a detector threshold of ±22.5-mV, which is the lowest available option.

The output of MAX30001 was compared to two AFE prototypes of GE HealthCare reference design by supplying similar inputs. The comparisons were not straightforward for the AFEs vary in several aspects. Using the paced ECG datasets from humans, 63.1% sensitivity for MAX30001 and 62.3% sensitivity for the reference design with basic configuration were obtained. A 2-mV peak-to-peak white noise at a sampling rate of 96-kHz and filtered at a 15-kHz low-pass filter was fed to the AFEs and resulting pulse indications were counted in a minute time window. A 30-ms refractory period was incorporated into the pace detection algorithm of MAX30001 with channel gain group set to 270/50.625/101.25 and detector threshold set at ±112.5, a count of 1715 false positive indications were registered while on the GE reference AFE prototype with basic configurations 384 pulse counts were found. From the noise test, it was seen that MAX30001 was too sensitive to random noise compared to the GE reference design prototypes. It was also found that ringing in the analog signal chain gave rise to multiple edges which could count as false detections.

In the evaluation process human human-based dataset is preferable to artificially synthesized records. Furthermore, it is recommended to include more challenging testing signals that could contain spikes that resemble actual pacemaker pulses. These can be obtained, for instance, from environments with a chance of electronic discharges and patients with LVAD.

Keywords: Pace detection, AFE, Testing signal, Sensitivity, PPV

PREFACE

This thesis work was held in GE HealthCare Finland from June 2022 to April 2023. I am very grateful to the management of GE HealthCare for this opportunity and their cooperation during the thesis. I got a lot of valuable training and experience while working in the company. It also helps me identify my strengths and weaknesses.

I am especially indebted to my supervisor, Juha Virtanen, for his enormous support in completing this thesis work and for his brotherly approach. I would like to thank Ville Vartiovaara, Heikki Vaananen, and all other members of the ECG team for their regular guidance and support. A special thank you also goes to my examiners Kim Blomqvist and Jari Viik.

I am most grateful to my wife for her unwavering support until I finished this research by shouldering the other responsibilities of the family. My kids were also the source of my energy and inspiration during the thesis process, so many thanks and love to them.

Helsinki, 12 September 2023

Berihun Gebreselasse

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1 INTRODUCTION

For regular rhythms and beats of the human heart, the electrical impulses that trigger the myocytes to contract are required to be generated and propagated smoothly throughout the wall of the heart. However, disruptions to this electro-mechanical interaction could occur leading to the implantation of cardiac pacemakers by medical intervention to keep the heart working normally. Pacing occurs by inserting the leads of the pacemaker only in the atrial or ventricular chambers or in some cases in both. The increasing number of patients with cardiac pacemaker devices and implantable cardiac defibrillators (ICDs) that can pace the heart requires reliable monitoring. Surface electrocardiogram (ECG) is a vital tool to assess and monitor the functionality of these medical devices [1].

An ECG waveform from a person with a cardiac pacemaker could include not only the usual P-QRS-T signals but also artifacts originating from the cardiac pacemaker. This is possible, however, if the ECG is recorded at high sampling rates. The presence of a pacemaker device and its synergy with the heart can be evaluated by detecting its pacing pulses [2]. Pacing spikes from a pacemaker device have narrow pulse duration and varying waveforms compared to noise and even to components of the ECG [2]. Regarding their amplitude, there are cases where the spikes rise above the QRS complexes given that they are not filtered, and the ECG is obtained at a high sampling rate. These properties provide a challenge for their detection. In addition to these, false pace pulse detections could arise from artifacts due to motion, noises from muscular activities, minute ventilation pulses, signals used fro telemetry purposes, and unwanted signals from other medical instruments attached to the patient [3]. There could be chances that these artifacts be counted as QRS complexes by patient monitoring devices – resulting in wrong high measured heart rates and actual low beat rates that can be missed complicating the heart treatment. On the other hand, when noise is detected as a pace pulse there can be an inaccurate pace pulse marker displayed on the monitor, leading to difficulty viewing and interpreting the ECG. Wrong arrhythmia alarms and inappropriate QRS classification could be routed to false pacing signals detections [1].

The electrical signal of the heart can be picked up by sensitive electrodes. Then the signal conditioning (filtering, amplification, etc.) is performed using analog front ends (AFE) that have the necessary electronics and connections. MAX30001, a product of Analog Devices Incorporated (ADI) is one of such AFEs that can be utilized for measuring ECG and other bio-signals

[4]. In addition to biopotential and bioimpedance measurement functions, this AFE is also designed to detect pacemaker signal edges. Detection of pacing pulses is possible by utilizing this edge detection information. This thesis work aimed to evaluate this electronic chip in the aspect of pacemaker pulses detection. Different scenarios were arranged to see how accurate MAX30001 was in producing the edge stamps. To explore false positive pace indications, white noise was fed as input in addition to pre-recorded paced ECG waveforms. Comparison to a reference design, other AFE from GE HealthCare, was performed on selected tests. MAX30001 came with an evaluation module that had an evaluation kit and application software for Windows. We chose and applied various configurations in the pace detection channel of the software. On the way, we were also finding ways to optimize the device in the pace pulses detection regard.

This thesis manuscript is composed of five chapters, where the basic concepts of artificial pacemakers are introduced in the first chapter followed by a second chapter describing the methods and materials applied in the project. Results and findings are explained in the next chapter and the conclusion and recommendations are the content of the last chapter.

This thesis work was done in collaboration with GE HealthCare oy in Helsinki, Finland. In this study, the author has participated in coordinating activities, organization of datasets and other resources, and processing and analyzing them.

2 THEORETICAL BACKGROUND

2.1 The Electrical Property of the Heart

As the main component of the human cardiovascular system (CVS), the heart acts as a central pump for it is involved in the circulation of both the oxygenated and deoxygenated blood. Coordinated depolarization and repolarization of myocytes in the generation as well as the propagation of action potential throughout the wall of the heart is the manifestation of its electrical property. In a healthy heart, the electrical impulse originates from the Sinoatrial (SA) node. This node is considered the natural pacemaker of the heart and is located at the top right region of the right atrium [5]. The impulse from the SA node propagates further down to the Atrioventricular (AV) node. The signal slows down at the AV node compared to the speed it has been traveling over the atria. After this delay, it enters the ventricles via the His-Purkinje system resulting in ventricular contraction [6].

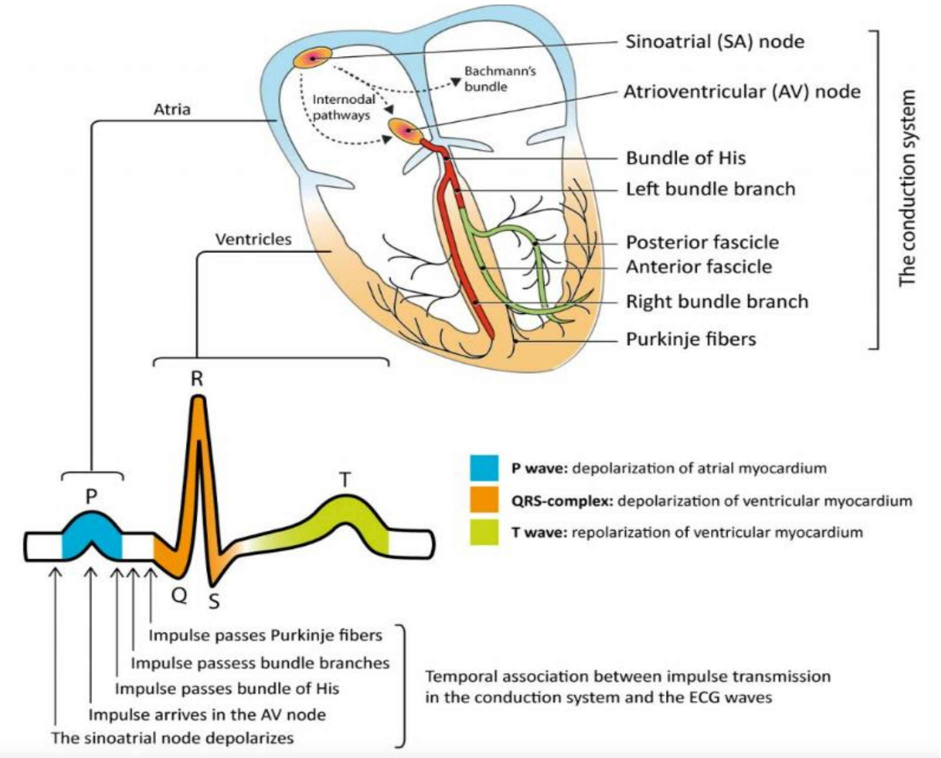


***Figure 1**. SA, AV nodes with the components of conduction system of the heart (Figure modified from figure [7]).*

For the heart to function normally, the electrical impulse from the SA node should be initiated spontaneously in a regular manner. Figure 1 illustrates the components of the conduction system of the heart with the resulting electrocardiogram (ECG) signal.

The loss of this automaticity in the generation as well as the conduction of the electrical signals can lead to various types of arrhythmias. Bradycardia, for example, can be mentioned as one of the major cardiac arrhythmias in which the beat rate of the heart falls below 60 beats per minute, or missing a beat can occur. At this condition and rate, it is difficult to fulfill the demands of the body. It is noted that the reason for the most common unhealthy pulse initiation is sinus node dysfunction, whereas AV blocks are reported to be the most common cause of defective impulse conduction [7].

2.2 Basics of Artificial Cardiac Pacemakers

When the heart fails to beat at the expected rate due to a problem in its electrical system, an external or artificial pacemaker device is implanted under the skin of the patient by surgical intervention to restore the rate. This pacemaker is programmed to evoke the cardiac muscle with precisely timed discharges of electricity that assist the heart to beat in a normal rhythm [8].

Artificial pacemakers are constructed of two components; the pulse generator and the leads. The power source, battery, control circuitry, Transmitter / Receiver, and Reed Switch (Magnet activated switch) are included in the pulse generator while the leads are sent through the vein to connect the implantable pulse generator to the heart chambers [13]. Figure 2 illustrates the connection and parts of the artificial pacemaker. The lead(s) can be unipolar or bipolar. In unipolar leads the current conduction is between the tip of the lead and the pulse generator. Whereas in bipolar leads, the current flow is between the two electrodes on the pacemaker lead [3]. Unipolar pacemakers produce larger pace spikes on ECG. On the other hand, in bipolar pacing, both electrodes are located within the heart and the resulting artifact becomes very small and is mostly invisible on ECG [7].

These pacemakers can be grouped based on the number of leads or according to their programming.

a) Based on the number of leads – single chamber pacemaker – These pacemaker types employ one electrode in the right atrium or right ventricle. On the other hand, in dual chamber pacemakers one electrode is applied in the upper chamber and another electrode in the lower chamber of the heart [3].

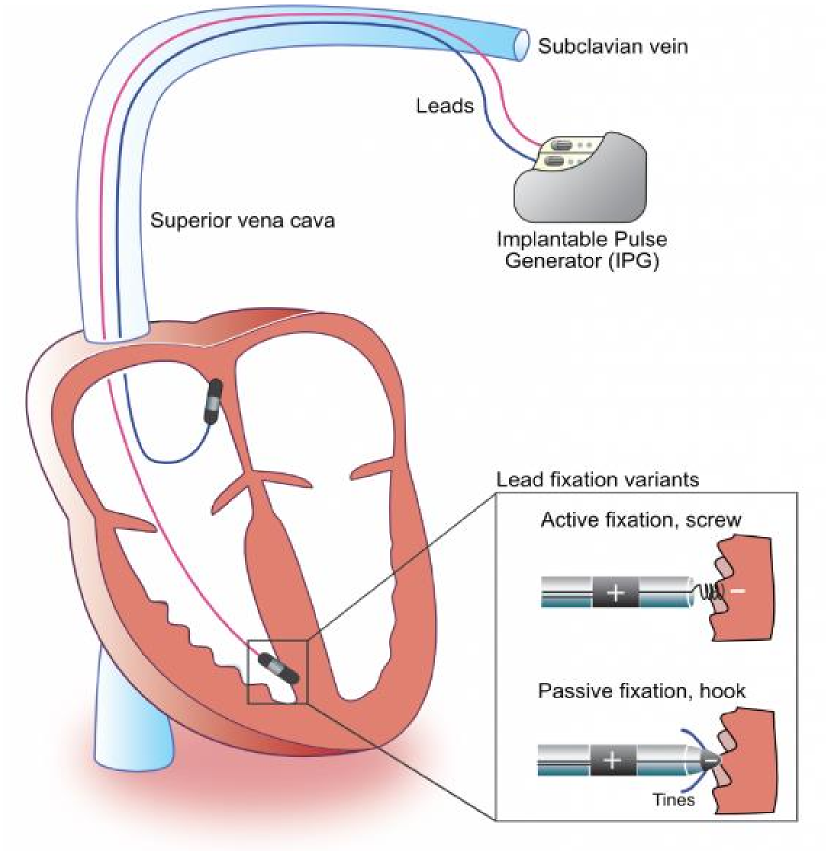


***Figure 2**. Components of an artificial pacemaker (Figure modified from figure [7]).*

b) According to the programming – Pacemakers can be programmed to evoke the heart at a fixed rate where a constant number of pulses are generated by the pacemaker. or, on-demand in which the pacing pulses are generated if the beat rate is too slow or if a beat is missed. Rate-responsive pacemaker types regulate the heart rate based on the activity of the body [3].

The Electrocardiogram (ECG) is widely used to study the electrical signals of the heart. It is possible to visualize the pulses of artificial cardiac pacemakers on the ECG, though they have narrow pulse widths and usually low amplitudes. Reliable detection of the pacing pulses requires a high sampling rate (at least 4 kHz) to capture enough energy from the narrow impulses [9]. Identification or detection of these spikes from the ECG is important for accurate evaluation of the ECG and to classify and detect heartbeats correctly. Furthermore, removing the influence of pacing pulses is vital for automatic ECG analysis [10]. Modern pacemakers generate pulses of 1 – 5 mV amplitude which last for about 0.5 – 2 ms in an ECG recording [9].

Fast-rising edges are common features seen in artificial pacemaker pulses. A typical pacemaker pulse is available in Figure 3. The positive pulse has a fast-rising edge. After the pulse reaches its maximum amplitude, a capacitive drop follows, and then the trailing edge occurs.

The artifact then changes polarity for the recharge portion of the pacing pulse. This recharge pulse is required to leave the heart tissue with a net-zero charge.

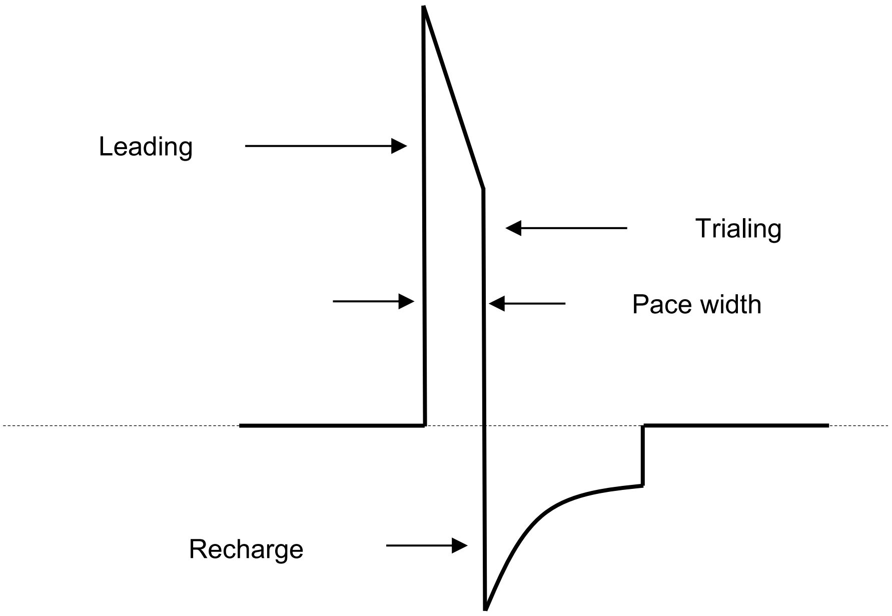


***Figure 3** Typical monophasic pacing pulse (Figure modified from [2])*

There are different medical standards with variable requirements regarding the height and width of the pacing pulse that must be captured and indicated on the screen of a patient monitoring device. According to ANSI / AAMI EC11, the features of the pacemaker pulses that should be obligatory detected are as follows.

- Total pulse duration ranges from a minimum value of 0.1-ms to 2-ms.
- Pulse amplitude starts from 2-mV to a maximum amplitude of 250-mV.
- Frequency of pulse up to a maximum of 100 impulses per minute
- The duration of the rising edge is less than 100-ms.

The IEC60601-2-27, on the other hand, states different requirements regarding the amplitude and pulse duration of pacemaker pulses. According to the standard, the acceptable pulse duration is set between 0.5-ms to 2.0-ms while for the amplitude parameter, a range of values between ±2-mV to ±700-mV is valid.

Modern artificial pacemaker devices could initiate smaller pacing pulse amplitudes that could be less than the amplitude thresholds stated in the above regulations complicating the algorithms for pacing pulse detection [11]. One approach to solve this case is to lower the detection threshold which will certainly lead to extra detections and additional logic for excluding the falsely detected pacing pulses. Analyzing multi-lead ECG leads and applying cross-checking over the detections is mentioned as another way of solving the problem [3].

2.3 Challenges in the Detection of Pacemaker Pulses

The presence of an artificial pacemaker and its interaction with the heart can be observed by identifying its pulses on surface ECG. For the pace pulses to be visible on the surface ECG, the ECG signal should be sampled at a higher rate than standard ECG as the frequency content of the pacing pulses is above the bandwidth of standard ECG [1].

It is stated by [11] that artificial cardiac pacemaker pulse detection can be approached by hardware methods, software methods, or a combination of both.

In the hardware methods, circuitry is employed to detect the slope of the leading edge of the pacemaker signal at the same time it ignores the other ECG signal components, i.e., the P-QRS-T segments of the ECG. Though it is faster it is reported to be not flexible, not simple for tuning, and only convenient for simple algorithms [11]. Despite its complexity, the software or digital method, on the other hand, includes possibilities for adjusting and adapting some parameters in the detection process, which adds room for flexibility. The general principles of both methods are illustrated in the block diagram shown in Figure 4.

Increased complexities and adjustment options in the use of implantable devices applied for cardiac treatment and therapy, for example, cardiac resynchronization therapy (CRT). These modern instruments generate pulses that are too narrow in duration and short in amplitude. They fall below the standards mentioned above. Figure 5 depicts an example of a small amplitude pace signal that is lower than the minimum amplitude threshold requirement of the AAMI standard. Detection of pacing pulses becomes challenging when they are small in amplitude and when they are abnormally shaped, among others [1].

In the small amplitude case, lowering the minimum pace pulse amplitude detection threshold can enhance the detection sensitivity. However, problems arise by including other artifacts as proper pace pulses. This should go side by side with developing complex algorithms that differentiate noise and true pace pulses.[1]. Furthermore, making use of multiple ECG leads in the identification process of true pace pulses helps avoid noise.

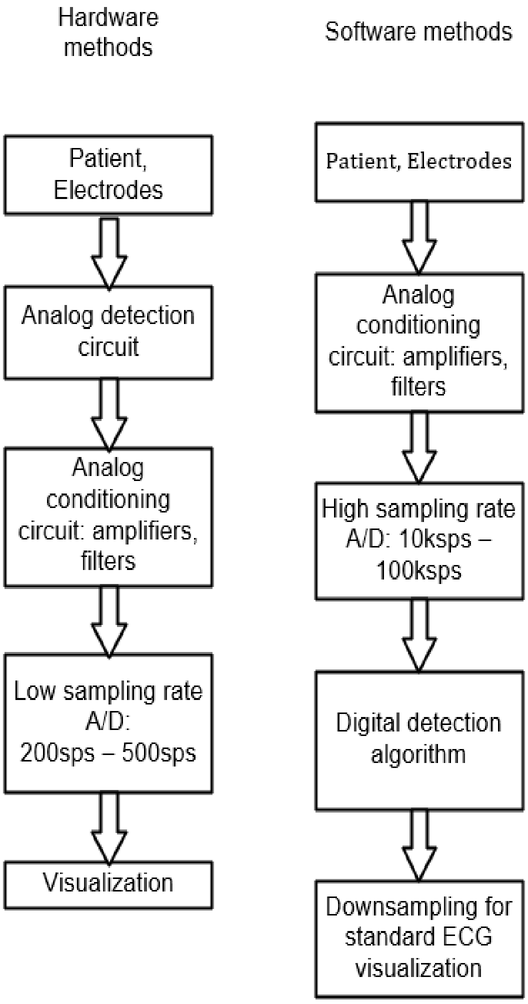


Figure 4. *Block-diagram of hardware and software pace pulse detection methods (Figure modified from [11]).*

An abnormally shaped pace signal is depicted in Figure 6. Several detection systems could miss it due to its unexpected shape. Detection algorithms that operate based on slope characteristics of the pulses anticipate a pace pulse to hold the shape of a square and perform the identification of the spikes by finding an abruptly rising initial slope. Next to this steep slope, a flat or slowly falling plateau is expected to happen followed by a vertically falling edge. This pace pulse, however, does not match these requirements and can be easily considered as a noise spike. Though it doesn’t look like a typical implanted pacemaker pulse, it will continue to show the same shape on this lead over time. It can also have common pace pulse characteristics such as a timing pattern consistent with its programming pacing mode and a corresponding pace pulse on other ECG leads. These characteristics can be utilized to identify abnormally shaped pace pulses [1].

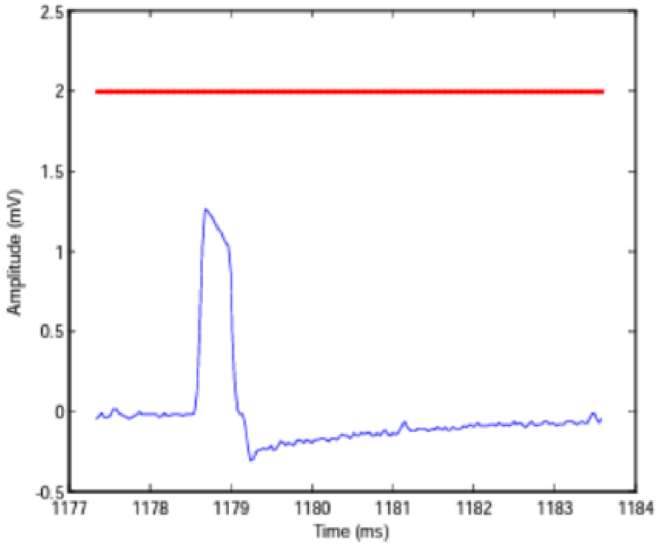


***Figure 6** AAMI amplitude standard for pace pulse detection (red) and typical pacemaker pulse on ECG (Figure modified from [1]).*

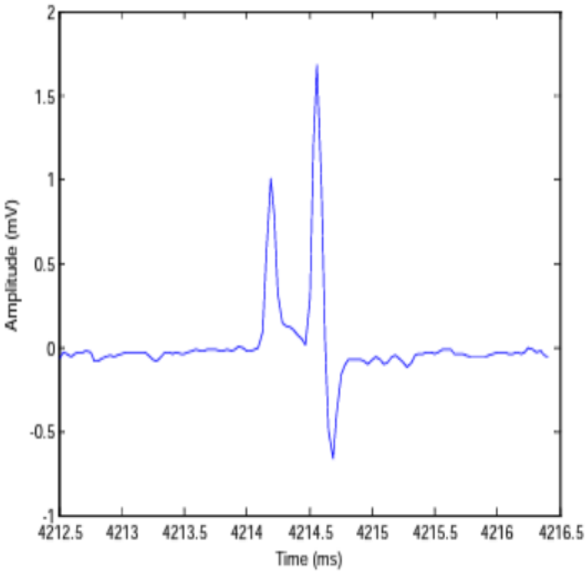


***Figure 5** Abnormally shaped pace pulse which is difficult to detect with traditional slope-based pace pulse detection methods (Figure modified from [1]).*

In the analysis of ECG, noise sources related to Electromyography (EMG) and motion artifacts are well-known reasons for errors. In rate-responsive pacemakers, the pacing rate is controlled by minute-ventilation pulses, these pulses have a pulse width in the range of 15-μs to 100-μs, which is usually less than 100-μs. These small values provide them with a vulnerability for false pace detection. The H-field telemetry scheme applied in most implantable heart devices is also mentioned as another major noise source. When noise and pace pulses are occurring simultaneously, preventing false pace detections while detecting true pace pulses becomes more complicated. In avoiding false detections, requirements can be arranged for repeatable morphology of the signal, reasonable time intervals between pulses, and the appearance of the pulses in more than one ECG lead [3].

3 MATERIALS AND METHODS

Software applications, electronic devices, and testing waveforms of different types (square pulse trains, noise, and ECG records) are used in evaluating MAX30001. In this section, the software applications are presented briefly followed by the description of electrical devices employed in this research. The testing waveforms utilized as input to the evaluation module are explained before the methods sub-section. The last part explains the algorithm developed and applied in converting the edge stamps data to pacing pulses information.

3.1 Materials

3.1.1 Software applications

Table 1 tabulates the list of software applications installed and used in the evaluation work of MAX30001. The purpose of each software is also briefly described under the purpose column of the same table.

Table 1 *List and description of software applications utilized.*

No	Description of the application software	Purpose
1	MAX30001 ECG / BioZ AFE EVSYS Software version 3.1.1- alpha 4	Acquisition, visualization, and exporting of ECG records with pace pulse edge information.
2	MATLAB R2022a update 2 May 11, 2022	• Developing scripts and generating synthetic datasets (square pulse trains and noise), • Conditioning of testing signals • Statistical analysis of results
3	Audacity 3.1.1	Visualization and studying of .wav files
4	WinSCP Version 5.21.1	Connect and manage (add, delete, or transfer) files on the removable USB drive attached to the multiparameter player.
5	National Instruments LabView WAV2SimData_PC Version 21.0.1f1	To convert .wav files to. tdms by setting the proper mid-scale voltage (0.8 V for MAX30001).

3.1.2 Multiparameter player (MultiSim_sbRIO)

This is an electrical device for playing back pre-recorded ECG as well as other waveforms at 500 kilo samples per second (ksps). It has removable USB storage for data transfer and temporary storage that has a maximum size of 4 GB. We have been using this USB stick to transfer data files from the PC to the playback device. These files shall be stored in a folder called “pbdata” that resides in the root folder of a FAT32, NTFS, EXT3, or EXT4 formatted USB mass storage device. For the files to be played back they should be saved in a National Instrument’s (NI) TDMS format. The TDMS files were created using the WAV2SimData_PC application. An infrared remote controller has been used to control the multiparameter player. The player has four single-ended playback channels.

The WAV2SimData_PC application converts a 500 ksps, signed 32-bit RIFF WAVE (.wav) file into a playable NI TDMS (.tdms) file [12]. The full scale of the .wav file needs to be ±100V, i.e., the least significant bit (LSB) is 100V/2^31 ≈4.65661nV. The output file has its bioimpedance sample values set to zero. The mid-scale voltage was set to 0.8V.

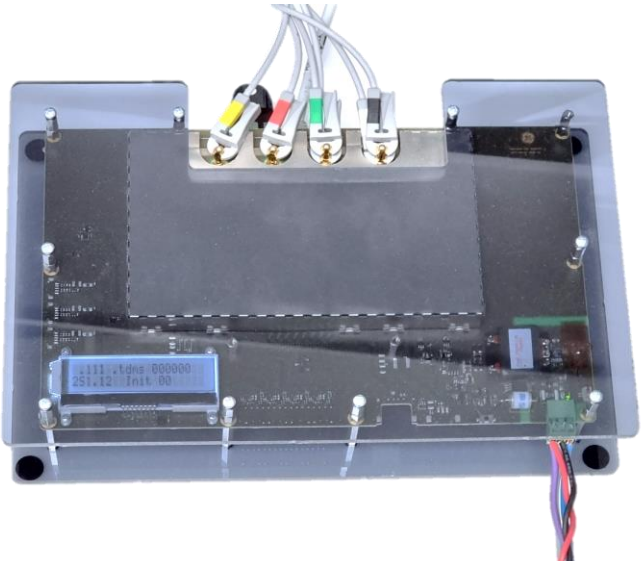


Figure 7 *Multiparameter player top view*

The diagram in Figure 8 shows the steps followed to produce a (.tdms) format file that Is ready to play on the multiparameter player. The multiparameter player plays waveforms at a rate of 500 ksps and thus the files on the USB storage should be saved in this format. This is performed using the WAV2SimData_PC application, which converts (.wav) formatted files to (.tdms) formatted ones. The (.wav) files are initially produced using MATLAB. Furthermore, a prefix (ECGData100VI32_) is required to be added before the name of the file is ready for conversion as shown in Figure 9 (a) and (b).

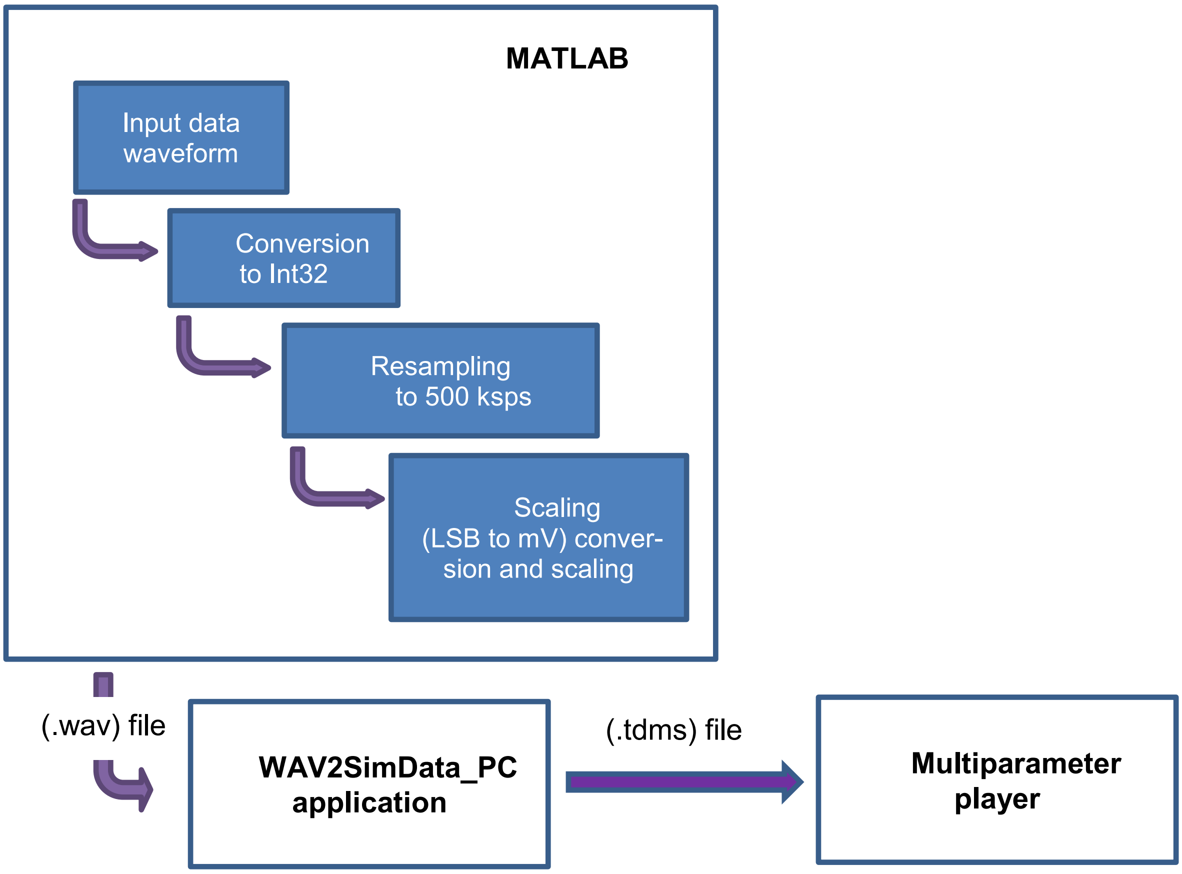
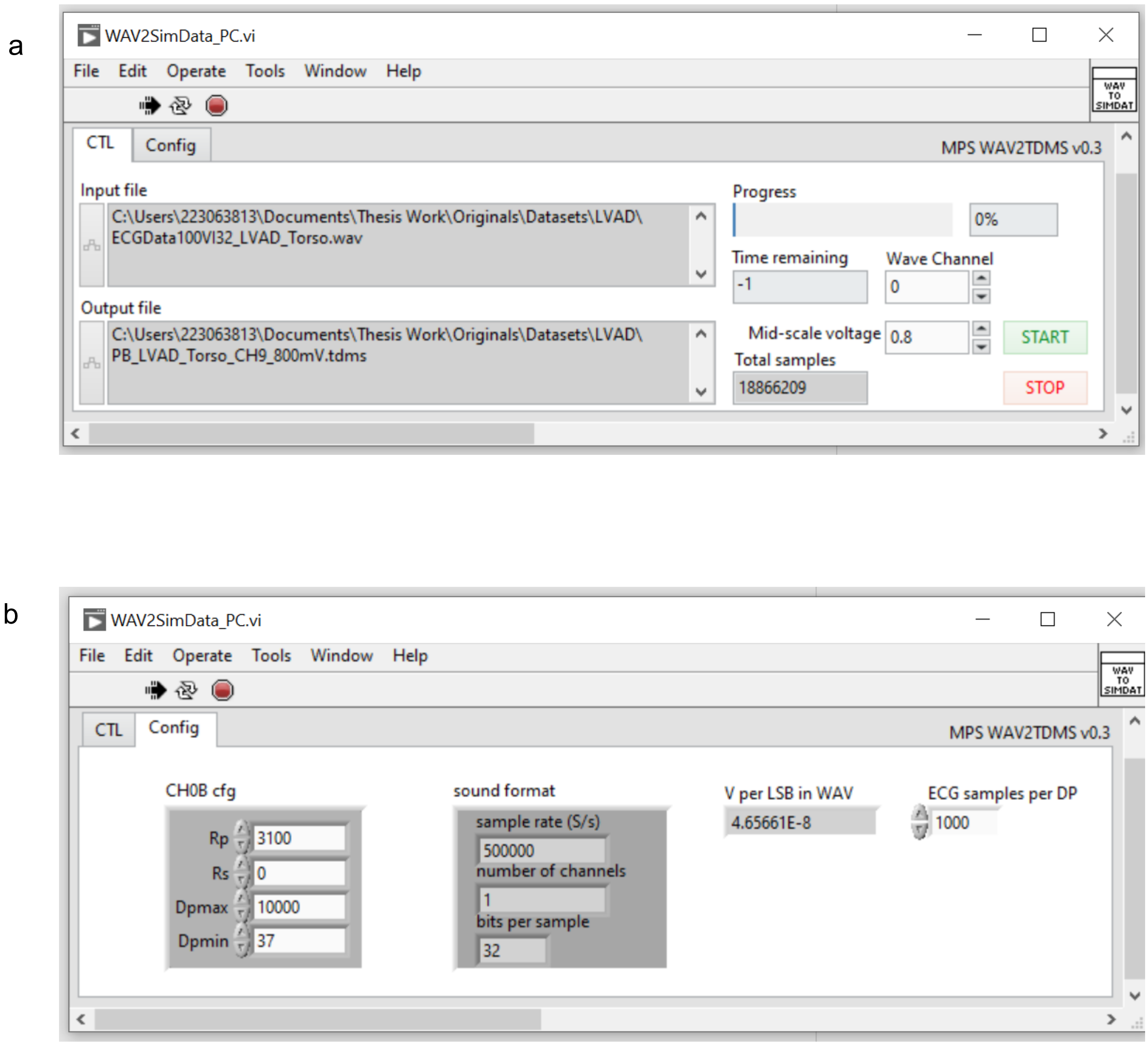


Figure 8 *Diagram showing the steps followed to convert any input data record to (.tdms) format. The output file from MATLAB is (.wav) formatted file, this file is sent as input to the WAV2SimData_PC application which results in a (.tdms) formatted file. The (.tdms) file is then playable by the multiparameter player.*



Input file: C:\Users\223063813\Documents\Thesis Work\Originals\Datasets\LVAD\ECGData100VI32_LVAD_Torso.wav

Output file: C:\Users\223063813\Documents\Thesis Work\Originals\Datasets\LVAD\PB_LVAD_Torso_CH9_800mV.tdms

Progress: 0%

Time remaining: -1

Wave Channel: 0

Mid-scale voltage: 0.8

Total samples: 18866209

CH0B cfg

Rp: 3100

Rs: 0

Dpmax: 10000

Dpmin: 37

sound format

sample rate (S/s): 500000

number of channels: 1

bits per sample: 32

V per LSB in WAV: 4.65661E-8

ECG samples per DP: 1000

Figure 9 *Example showing the conversion of a (.wav) file to (.tdms) file using the WAV2SimData_PC. The detail configuration fields are available in (a) CTL tab of the application (b) Config tab of the application.*

3.1.3 Analog Front Ends (AFE)

The MAX30001 is the main AFE evaluated in this thesis work. It has an evaluation system providing a single platform to evaluate the functionality and features of the MAX30001 with Biopotential (ECG, R-to-R, and Pace Detection) and Bioimpedance (BioZ) measurement capabilities [13]. Figure 10 illustrates the MAX30001 evaluation kit. The evaluation system comprises.

- An application software (Graphical User Interface (GUI)) installed on Windows operating system (Windows 10 Enterprise) – (MAX30001 ECG / BioZ AFE EVSYS Software version 3.1.1- alpha 4).
- Evaluation kit (EV kit) - circuit board with MAX32630FTHR cortex-M4F microcontroller for wearables.
- USB A to micro-USB cable
- Three (3) ECG cables

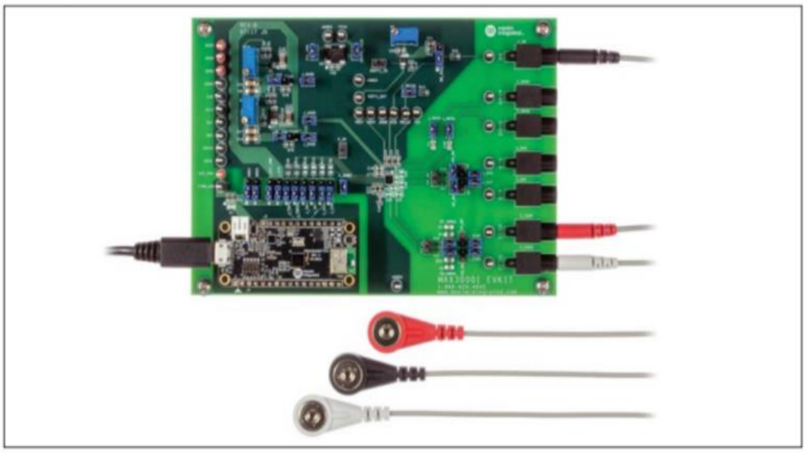


Figure 10 *A picture showing MAX30001 evaluation hardware components. Figure modified from [13].*

The evaluation board contains jumpers and optional resistors and capacitors to test the MAX30001 under several conditions. The board allows the installation of external low-pass filtering capacitors on the ECG line. Thus, we have applied external capacitors on the C26 and C28 and resistors on R26 and R27. The MAX30001 Evaluation System (EVS) was biased to 0.8V for proper functioning.

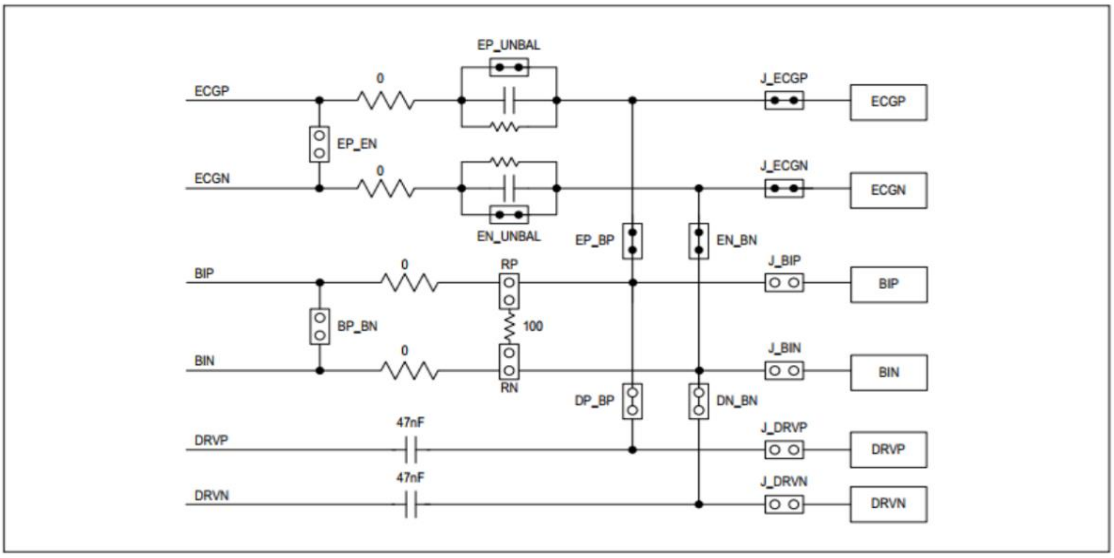


Figure 11 *Configuring the MAX30001 EV kit for pace pulses detection. Modified figure from [13]*

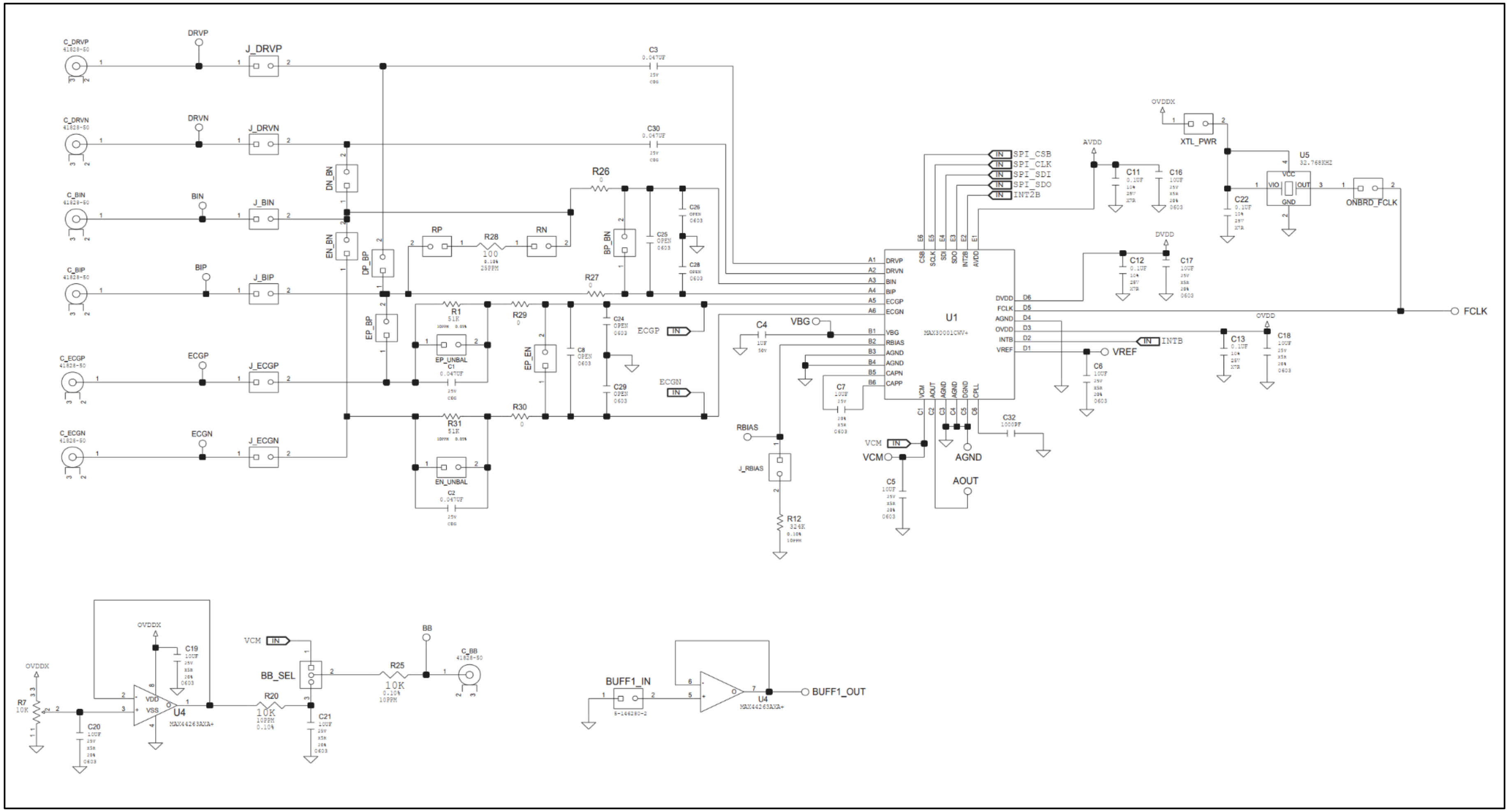
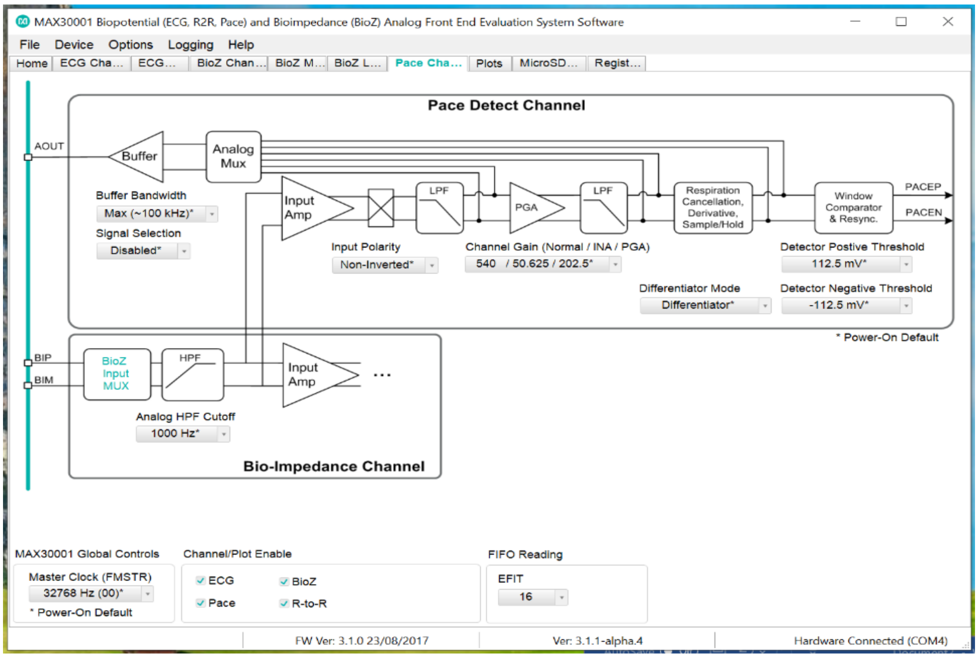


Figure 12 *Schematic diagram for MAX30001 evaluation kit (figure modified from [13])*

To enable pace detection, the ECGN and ECGP lines are connected to BIN and BIP by shunting EN_BN and EP_BP as depicted in Figure 11. The electrical schematic diagram of the MAX30001 circuit board is available in Figure 12.

The MAX30001 EVS GUI has several tabs for different functions. A snapshot of it is shown in Figure 13. When measurements are taken on the MAX30001 EVS, it is possible to capture and export data of a measurement in a comma-separated value (CSV) format. The focus of this thesis work was on cardiac pacing pulses detection. We have been saving files of ECG and pace logs from the measurements we run on the module.



Buffer Bandwidth: Max (~100 kHz)*

Signal Selection: Disabled*

Input Polarity: Non-Inverted*

Channel Gain (Normal / INA / PGA): 540 / 50.625 / 202.5*

Detector Postive Threshold: 112.5 mV*

Differentiator Mode: Differentiator*

Detector Negative Threshold: -112.5 mV*

* Power-On Default

Analog HPF Cutoff: 1000 Hz*

MAX30001 Global Controls

Master Clock (FMSTR): 32768 Hz (00)*

* Power-On Default

Channel/Plot Enable

☑ ECG ☑ BioZ

☑ Pace ☑ R-to-R

FIFO Reading

EFIT: 16

FW Ver: 3.1.0 23/08/2017

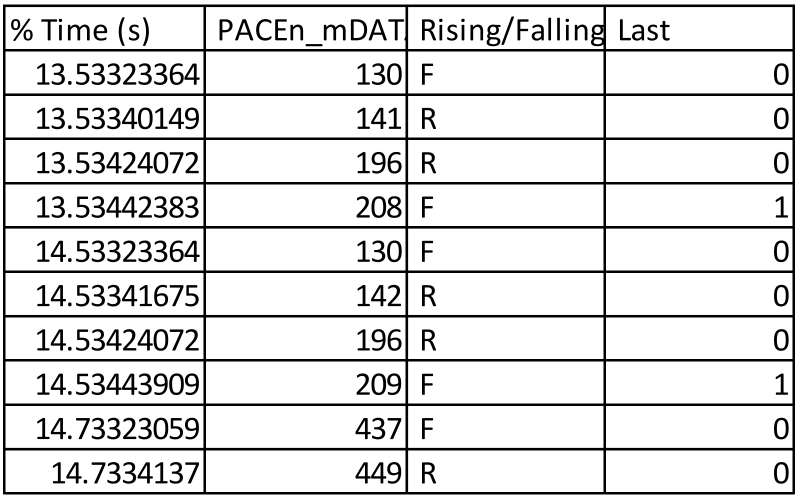
Ver: 3.1.1-alpha.4

Hardware Connected (COM4)

***Figure 13** Pace Detect Channel of MAX30001 GUI with default parameter values indicated.*

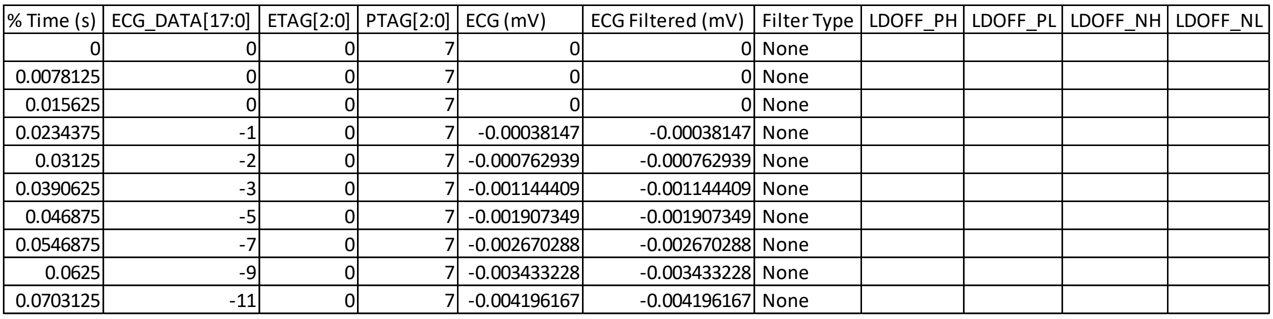
As shown in Tables 2 and 3 below, the evaluation module saves the edge information of the paces, and the ECG records at a sampling frequency of 128 Hz. A brief header is provided for each column. The first column on both Table 2 and Table 3 are real-time records of each event. The pace event on the other hand is depicted in Table 2. The third column (i.e., Rising/Falling) tells the direction of the edges found. ‘F’ stands for falling edge and ‘R’ for rising edge. The user of the AFE is expected to qualify a pair of edges as valid pace pulses.

Table 2 *An example extract showing content of the CSV output file of MAX30001 on pacing pulse edge detection.*



% Time (s)	PACEn_mDAT	Rising/Falling	Last
13.53323364	130	F	0
13.53340149	141	R	0
13.53424072	196	R	0
13.53442383	208	F	1
14.53323364	130	F	0
14.53341675	142	R	0
14.53424072	196	R	0
14.53443909	209	F	1
14.73323059	437	F	0
14.7334137	449	R	0

Table 3 *An example extract showing content of the CSV output file of MAX30001 on ECG measurement.*



% Time (s)	ECG_DATA[17:0]	ETAG[2:0]	PTAG[2:0]	ECG (mV)	ECG Filtered (mV)	Filter Type	LDOFF_PH	LDOFF_PL	LDOFF_NH	LDOFF_NL
0	0	0	7	0	0	None				
0.0078125	0	0	7	0	0	None				
0.015625	0	0	7	0	0	None				
0.0234375	-1	0	7	-0.00038147	-0.00038147	None				
0.03125	-2	0	7	-0.000762939	-0.000762939	None				
0.0390625	-3	0	7	-0.001144409	-0.001144409	None				
0.046875	-5	0	7	-0.001907349	-0.001907349	None				
0.0546875	-7	0	7	-0.002670288	-0.002670288	None				
0.0625	-9	0	7	-0.003433228	-0.003433228	None				
0.0703125	-11	0	7	-0.004196167	-0.004196167	None				

3.1.4 GE reference prototype

For this research, GE has provided an analog front end that is highly integrated and designed for the measurement of patient biopotential information. Besides ECG, impedance changes during patient respiration are recorded by a dedicated impedance measurement section using this integrated circuit (IC). The pacemaker artifact detection block has a flexible on-chip digital pacemaker spikes detection algorithm that can be applied to multiple leads. The detection algorithm is designed to detect pacing artifacts with widths from 80 µs to 2.5 ms and with amplitudes of 1 mV to 700 mV.

Two prototypes of this electronic chip were used in this thesis work for comparing MAX30001 outputs. One of the prototypes is utilized on its basic configurations, while the other one is used by adjusting or varying the configurable filter settings.

3.2 Testing signals

The testing signals in this evaluation work include pre-recorded ECG signals, square pulse trains, and random noise waveforms. The ECG signals can be categorized into two groups those obtained from humans and those generated using a simulator. In general, paced ECG signals and square pulse trains are applied to assess how accurate MAX30001 is in detecting the pacing pulses in these records, and the rest of the testing signals are employed to see the number of false positive indications arising from these waveforms.

3.2.1 Square pulse trains

Square pulses were generated and fed to the AFE to study how it behaves in producing the edge data discussed above in Table 2 and the effectiveness of the AFE in detecting them. In addition to these, it also helps us observe the amplitude and pulse duration limits the module can detect without a problem. The square pulses have a wide range of pulse widths and amplitude ranges. They were grouped into eight batches based on their pulse widths and they comprise 40 amplitude steps. Starting from the left the pulse widths are arranged as 1 ms, 0.8 ms, 0.6 ms, 0.5 ms, 0.4 ms, 0.3 ms, 0.2 ms, and 0.1 ms. Each amplitude step contains 5 pulses. In front of each pulse group, a reference line of 1 mV amplitude was added as a marker so that it is easy to identify the end and beginning of each pulse group.

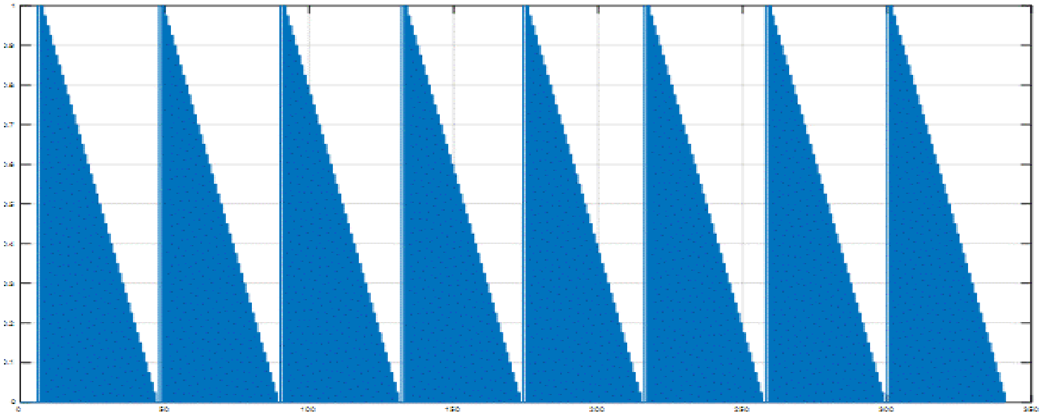


Figure 14 *Eight groups of square pulse trains with varying amplitude and pulse widths. Similar amplitude steps are applied on all pulse groups; however, each group has a pulse width that is different from others.*

3.2.2 Paced ECG Datasets

The first source of paced ECG records was GE HealthCare itself. This in-house dataset was a database of human ECG records with annotated artificial pacemaker pulses. The material was saved at the Massachusetts Institute of Technology (MIT)) / Physionet format [30] so that each data comes with a header as well as an annotation file. The digitally sampled signals were binary and saved in (.dat) files. The corresponding header files were short text files describing the contents of the associated signal files. The extension for the header files was (.hea). On the other hand, the annotation files were binary files containing annotations (labels that generally refer to specific sample numbers in associated signal files that indicate where the pace pulses are located). They were saved with the same name as the signal file, and they had a (.prf) file extension. MATLAB was used to read the data files with their corresponding header files as well as annotation files. There were eight ECG channels per case. It was data of 100 patients with implanted pacemakers, each patient having a maximum of five records.

The sampling rate for the ECG signals was 75 kHz – this preserves the steep rising and trailing edges of the pace pulses. Most of them are 13.5-s recordings, however, there are shorter recordings in between. It is observed that except for a few problematic cases, the database was noise-free.

The total number of data files (.dat) in the database was 511, however, there was not an equal number of annotation and header files. Table 4 summarizes the initial findings regarding the files and their quantities in the database.

Table 4 *Content of the Dataset found from GE HealthCare*

No	File type	File extension	Quantity	Remark
1	Data file	.dat	511	
2	Annotation file	.prf	490	444 were found with non-empty annotation files
3	Header file	.hea	511	

The number of annotation files was less than the data and header files by 21 and so the ECG data files without annotation were excluded from further processing. In addition to this, files that have readable, appropriate, non-empty, and long enough (greater than 5 samples) annotation and header files were selected for further processing. Based on this we found 414 appropriate recordings that can be processed and analyzed.

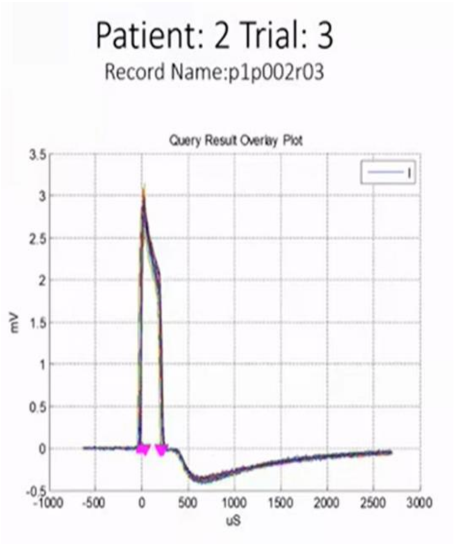


Figure 15 *Pacemaker pulses on the second patient in the dataset with a record named 'p1p002r03' showing the involved impulses overlapped. Figure modified from figure in [15]*

As shown in Figure 15, the parameters of the pacemaker varied slightly in each trial per patient by the physician. It is illustrated that the pacing impulses applied on the second patient in the database were overlapped with each other and they are highlighted in different colors. The amplitudes fell around 3-mV, and the pulse width was around 250-µs. Each record has an embedded reference flag or marker that serves as an indicator for the beginning of the signal file, depicted in Figure 16. This indicator was replaced from each data file with a marker that can easily be detected by MAX30001 EVM, see Figure 18. This was done to easily find the start of individual records and automate the splitting when the files are concatenated to a single long file for ease of measurement and handling. Otherwise, it would have been time-consuming to take individual measurements with the high number of data files.

A large glitch (shown in Figure 17) was found at the end of each record and was clipped to avoid false detections.

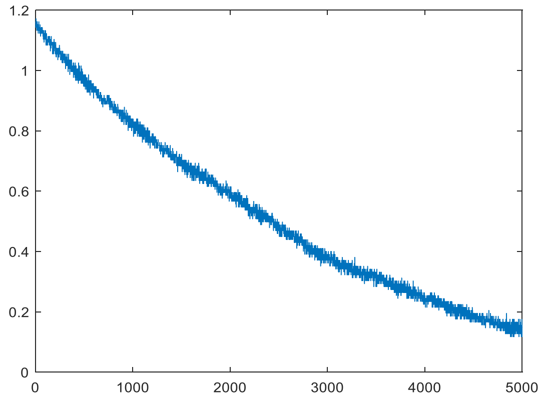


***Figure 16** Indicator found at front of each ECG signal.*

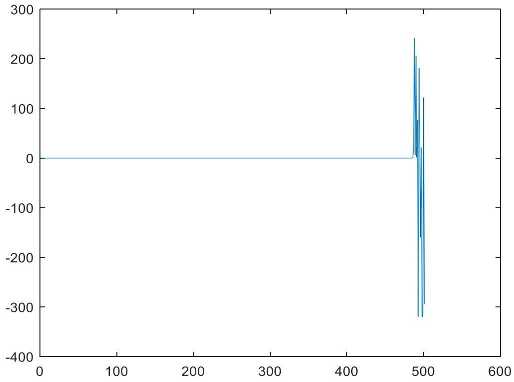


***Figure 17** Last 15 samples of the ECG waveforms was clipped to avoid false detections.*

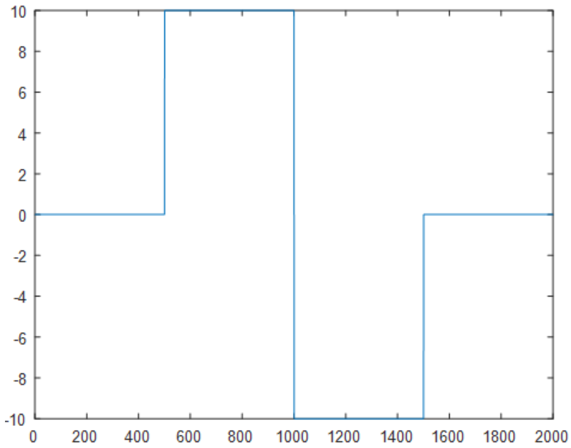


***Figure 18** Marker added in front of each record.*

The other paced ECG dataset used was generated by an ECG simulator and is publicly available. The dataset was named ‘PacedECGdb’ and was available to download from (http://biomed.bas.bg/bioautomation/2014/vol_18.4/files/PacedECGdb.zip).

MATLAB scripts are also part of the download package for opening and reading the ECG records. The ECG signals that contain different arrhythmias were produced by a simulator named Heidelberger Praxisklinik (HKP). Pacing pulses of rising edges from 10 µs to 100 µs and pulse widths from 100 µs to 2 ms were embedded in the ECG waveforms. The pacing spikes represent diversified pacemaker modes. The database was organized into two groups as ‘pure’ ECG with pacing pulses containing 780 recordings and other 624 waveforms were saved in a different group that was corrupted by tremor. Each record has a 10 s duration and 128 kHz sampling frequency. For this thesis work, the data subset that contains tremor was chosen expecting that the physiological noise (tremor) could give rise to more false pace indications.

In both datasets, as the individual records are many, the data files were concatenated to form a single long file keeping in mind the memory limit of the multiparameter player, then after capturing a long (.csv) file, it was segmented back to individual records for statistical analysis.

To summarize, the following steps were followed to undertake measurements through the multiparameter player and get files ready for statistical analysis.

- Combining a certain number of ECG records on the condition that the combined file size doesn't exceed the memory requirements of the multiparameter player (MPP) which is preferably 2 GB. This was to avoid counter overflow.
- Playing the combined file by the multiparameter player and capturing log files from the MAX30001 EVM for both ECG and Pace in CSV format.
- Splitting the ECG and edge detection files into individual records.
- Creating pace pulse detections based on edge detections in the individual edge detection files.
- Calculating the pace detection statistics

3.2.3 Random noise

We have prepared random noise generated by the MATLAB rand function and passed it to the AFE via the playback device. The goal of the noise test is to observe the reaction of the AFE to extreme input signal conditions or bursts of pulses. Falsely detected pacing pulses were expected to increase in number. This test was also good to compare the MAX30001 with the AFE in the GE reference design.

3.3 Methods

3.3.1 Pace pulses detection algorithm

The ECG and pace logs of the evaluation board have been captured as the tests were being accomplished. From these logs, it was observed that this AFE was not explicitly identifying and defining pace pulses, rather edges of the pace spikes with enough energy and slope were saved together with the real time they were happening. They were registered as falling (F) or rising (R) for stating their directional ending or polarity. For getting proper pulse detection, it was important to find and combine a pair of opposite edges that fall within a specified time duration. This time duration was set as a pulse width duration. The pulse width duration was defined to have a maximum of 2-ms and a minimum of 10- μs, however, there was also flexibility in changing these values. This was following the standards' requirements including modern artificial pacemakers that produce low amplitude and extremely narrow pulse widths compared to what is stated in the standards.

We have developed two algorithms for two purposes, one is for normal detection of pulse artifacts that fulfill the pulse width interval and contain opposite pairs of edges, see flowchart in Figure 19. The other one also contains a similar time interval for total pulse duration; however, it has the capability of rejecting narrow-pace pulses and it can incorporate a refractory period. This could be further modified for comparison with the GE reference AFE which had a dead time timer to count to 30-ms, during which period no search for new pace pulses occurred.

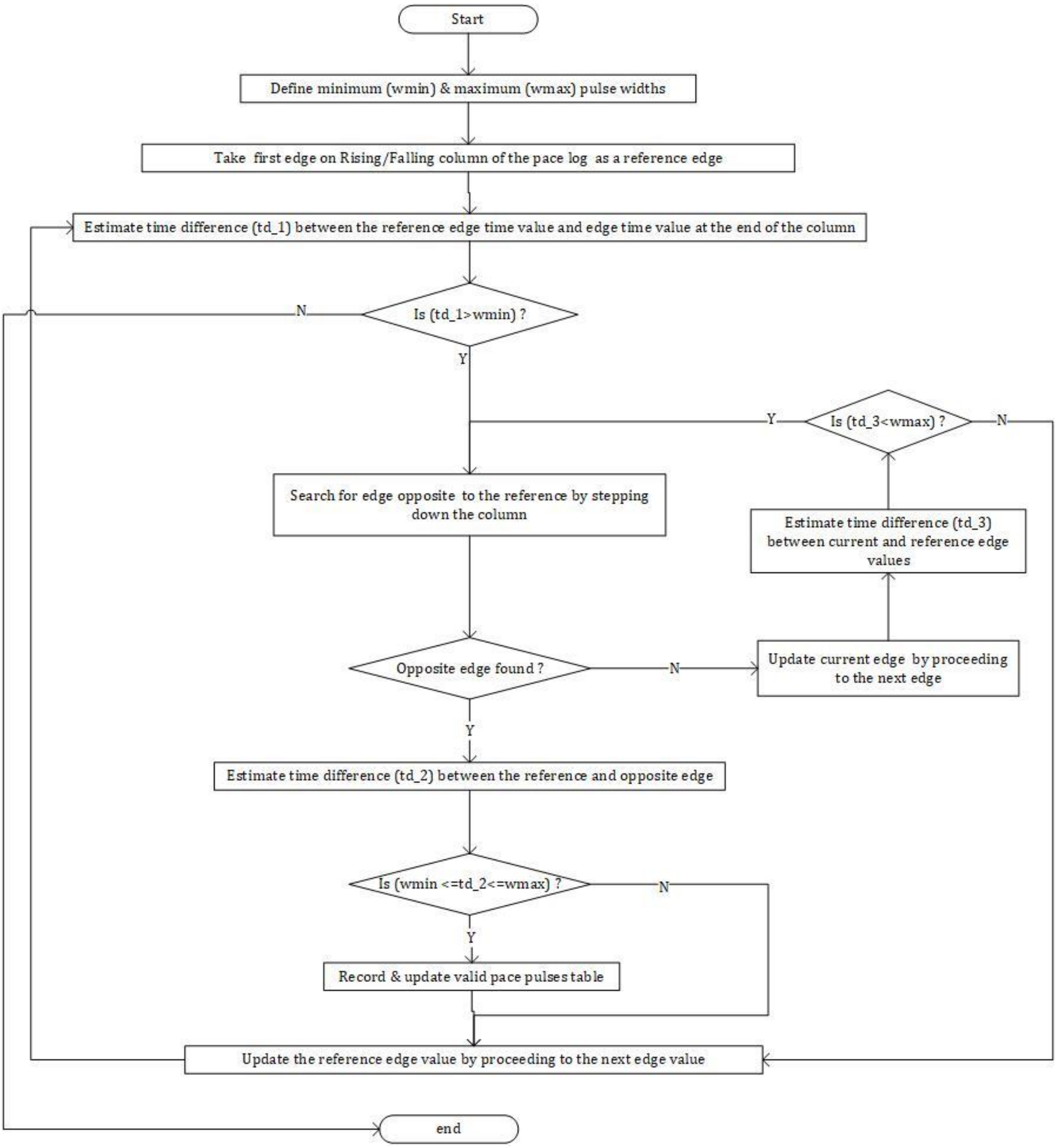


Figure 19 *Algorithm for detection of pacing pulses from MAX30001's log file.*

3.3.2 Timestamp offset definition

The pacemaker-paced ECG datasets were linked with annotation files that provided sample numbers indicating the starting point of the pacing spikes found in the ECG waveform. Based on the edge log of the MAX30001, the pace pulse detection algorithm estimated the location of the pacing impulses. These estimates were then compared to the annotation values. As shown in Table 1, it appears that a 0.006 s time difference constantly appears to exist between the two time arrays. This offset value is considered in the detection of pacing pulses in the same dataset records.

***Table 5** Example extract showing time stamp offset calculation between annotation and estimated time values.*

Pacing pulses detection time obtained using a detection algorithm (T_det)	Time stamps from annotations (T_ann)	Offset Abs(T_det – T_ann)
0.074890137	0.08132	0.00643
0.645324707	0.651667	0.006342
0.824829102	0.831267	0.006438
1.395263672	1.401627	0.006363
1.574783325	1.58124	0.006457
2.145202637	2.151573	0.006371
2.324737549	2.331187	0.006449
2.895172119	2.901533	0.006361
3.074707031	3.081147	0.00644
3.645111084	3.651493	0.006382
3.824645996	3.831107	0.006461
4.395095825	4.40144	0.006344

3.3.3 Setting up of the testing environment

Figure 21 depicts the general connection layout of the testing environment. We used two ECG leads to connect the multiparameter player and the MAX30001 evaluation (EV) kit. Each ECG cable was snapped to one output channel of the playback device, as seen in Figure 20. On the other side, as shown in Figure 22, these ECG leads were inserted into ECGN and ECGP input jacks of the MAX30001 evaluation module. An electrical grounding was established by joining the AGRND point on MAX30001 to the multiparameter player's metal frame. As the AFEs are electronically sensitive, an antistatic mat set was used with a wrist strap and grounding cable for electrostatic discharge protection and sensitive electronics operations. Following the hardware connection, the major steps followed, and tasks performed in capturing the log files (ECG and pace CSV files) are described in the diagram shown in Figure 23.

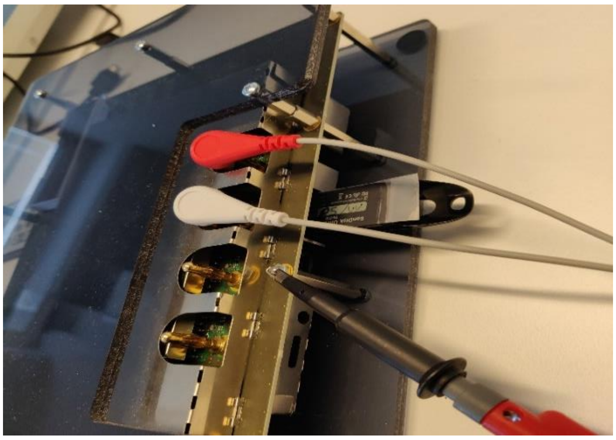


Figure 20 *Two ECG leadwires connected to two of the playback device's output points.*

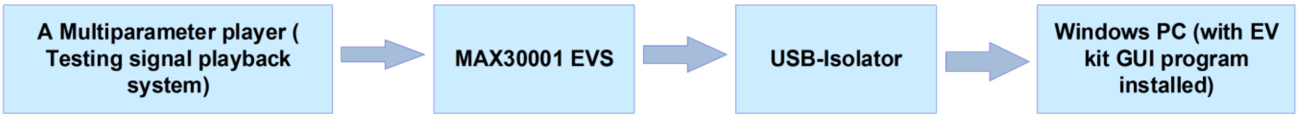


Figure 21 *Diagram showing the connected devices and the data flow direction during each test.*

The tests were performed by connecting the MAX30001 evaluation system (EVS) to a Windows PC's USB port. The EV kit gets its power via this connection. We used a USB isolator (ISOUSB – PLUS from If tools) to keep the EVS safe from electrical surges, and transient voltage spikes and to avoid ground loop currents flowing between the PC and peripherals which can cause damage and inaccurate measurements [16].

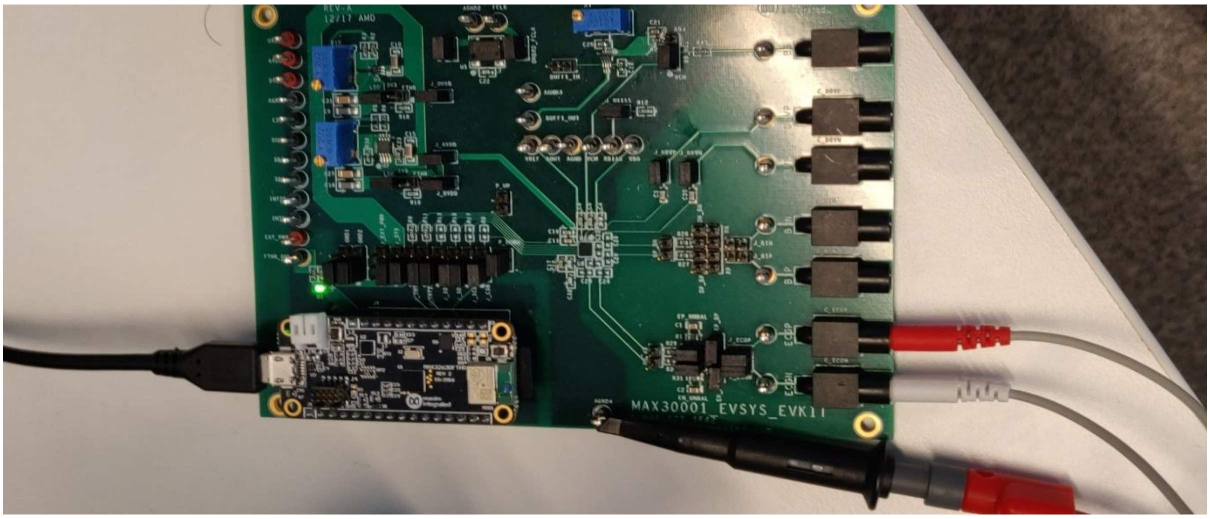


Figure 22 *MAX30001 EV kit with ECG leads and grounding connections and with appropriate configurations for ECG measurement and pace detection.*

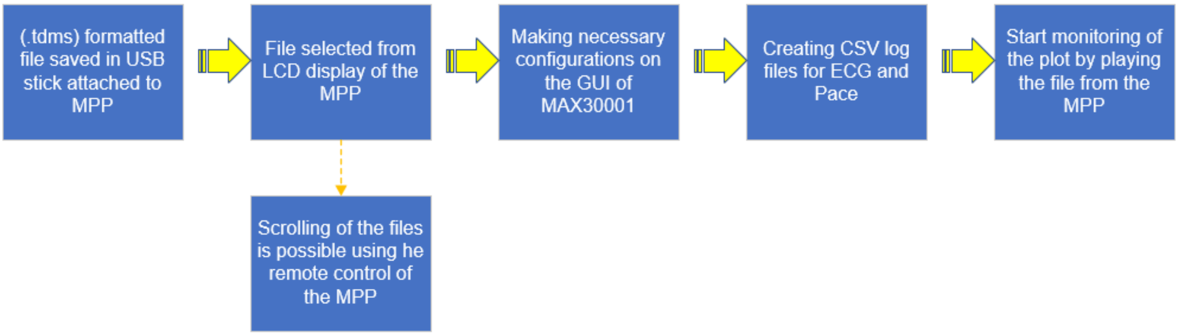


Figure 23 *A flow chart describing the steps in performing the test using the established testing environment. (.tdms) formatted files were selected by the remote control of the waveform player and (.csv) formatted log files of ECG and pace were saved.*

4 RESULTS AND FINDINGS

In this chapter, the results obtained in the evaluation process of MAX30001 AFE are detailed. The evaluation work was performed under several conditions by arranging various testing inputs. Besides MAX30001, two AFE prototypes from GE reference designs were required for comparison purposes. In this section, the comparison results are discussed following the standalone outcomes of MAX30001.

4.1 Results obtained from tests performed only on MAX30001

4.1.1 Results from square pulses

The rectangular pulses mentioned in the previous chapter were fed to MAX30001. After processing the output edge information of the square pulses, the following sensitivity plots were produced and are shown in the figures below. Figure 24 (a) displays the threshold detection amplitude for full (100%) detection at ±112.5 mV detector threshold of the AFE. Figure 24 (b) displays similar outcomes for measurements taken at ±337.5 mV detector threshold of the AFE. On the ±112.5 mV detector threshold voltage the pulse amplitude threshold is approximately 0.4 mV and on ±337.5 mV it is found to be around 1.04-mV measurement errors taken into consideration. No external high-pass or low-pass filters are applied in obtaining these results.

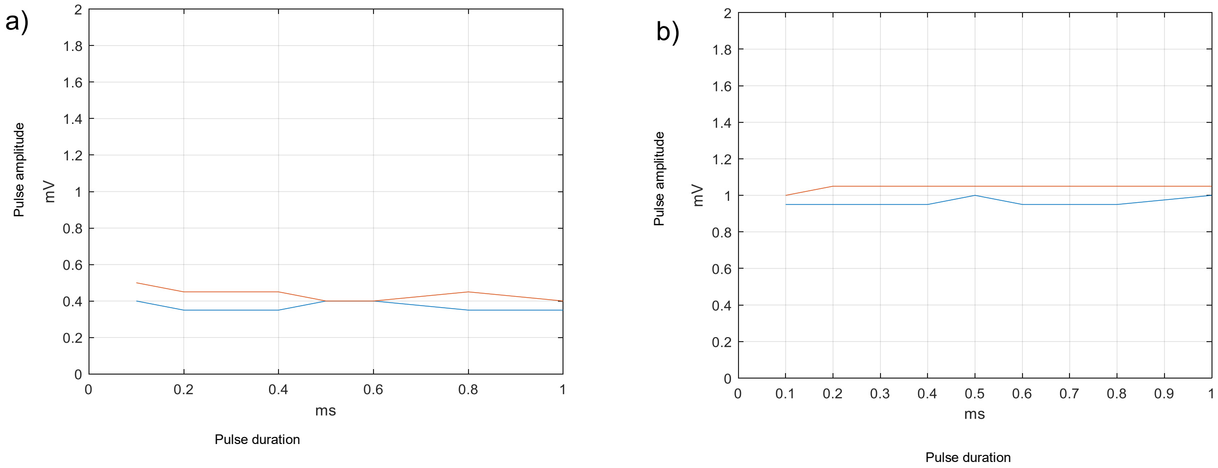


Figure 24 The red line indicates the pulse amplitude thresholds for complete (100 %) detection. The blue line represents thresholds that were partially (<100%) detected. a) Measurement taken at ±112.5-mV detector threshold b) Measurement taken at ± 337.5-mV detector threshold

According to the manual of the EV kit of MAX30001, it was possible to install low-pass filtering by adding capacitors on C26 and C28 and resistors on R26 and R27 [16]. Hardware filters were thus added by soldering capacitors and resistors on these points of the EV kit (see schematic circuit diagram and explanation in section 3.1.3). We have tested four passive low-pass filters by varying the capacitor and resistor values, and the following plots were produced by processing the measurement records.

- Filter 1 (F1) - Low pass filter 1 – Capacitor (C) = 270 pF and Resistor (R) =15 KΩ which provides cutoff frequency (fc) = 39.317 kHz.
- Filter 2 (F2) - low pass filter 2 – C = 270 pF and R = 43 KΩ which gives fc = 13.715 kHz.
- Filter 3 (F3) - low pass filter 3 – C = 680 pF and R = 43 KΩ that gives fc = 5.446 kHz.
- Filter 4 (F4) - low pass filter 4 – C = 1500 pF and R = 43 KΩ which provides fc = 2.469 kHz.

Based on the plots in Figure 25, At ±112.5 mV detection threshold value, the full or 100% detection amplitude threshold gradually increases from 0.45-mV at F1 to 0.70 mV at F4. Giving a ratio of 1.55 and similarly, at a detector threshold voltage of ±337.5-mV, an increment from 0.95 mV at F1 to 1.65 mV at F2 provides a ratio of 1.73. Within measurement accuracy, pulse detection threshold increase is similar with both ±112.5mV and ±337.5 mV comparator threshold values.

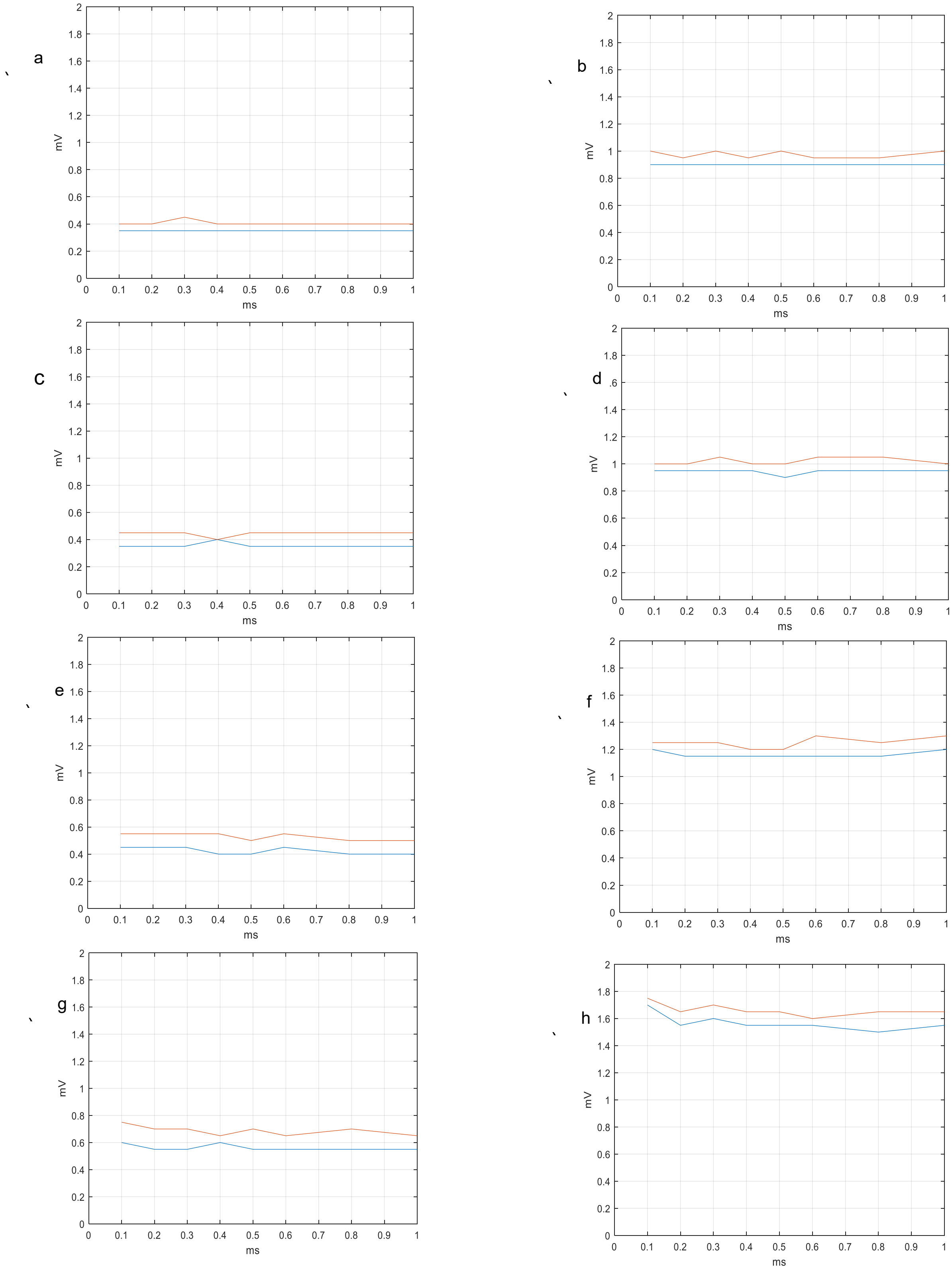


Figure 25 Plots showing amplitude threshold changes with respect to the applied hardware filter cutoff frequency changes. The figures on the left are obtained at ±112.5 mV detector threshold voltage value whereas those on the right are recorded at ±337.5 mV. a) and b) are for F1 c) and d) are for F2 e) and f) are for F3 g) and h) are for F4. In all the plots pulse amplitude is indicated in mV in y-axis while pulse duration is shown in ms in x-axis.

4.1.2 Results from paced ECG records

Paced ECG waveforms described in the previous chapter were passed to the AFE and the results obtained are presented in this section. Both datasets contained pacemaker pulses their difference lay in the source of the ECG signals. The GE database was collected from around 100 human patients with artificial pacemakers while the other database was synthesized using an ECG simulator.

We applied two sets of parameters to perform the measurements on the GE dataset. They are listed below as Setting #1 and Setting #2.

Setting #1 — Default settings – All parameters' values of the AFE were kept at their default amount.

- Analog High Pass Filter (HPF) cutoff was set to 1000 Hz.
- Channel Gain (Normal /INA/PGA) 540 /50.625/202.5
- Detector threshold ±112.5 mV

Setting #2 — The parameters and their values are set as follows.

- The Analog HPF cutoff was set to 8000 Hz.
- Channel Gain (Normal /INA/PGA) 240 /22.5/90
- Detector threshold ±22.5 mV

It is seen in Table 6 and Table 7 that the PPV (%) values remain high at all the measurements probably linked to the noise content of the dataset being low. However, the sensitivity values are shown to decrease in going from the first record categories (r01) to the last record categories (r05). This could indicate the increased challenge in the detection of all the pulses in the trials. In Setting #2, the sensitivity values are found to be even lower than those obtained from Setting #1 as shown in Table 7.

Table 6 *Sensitivity and PPV values by grouping the ECG records based on their trial number.*

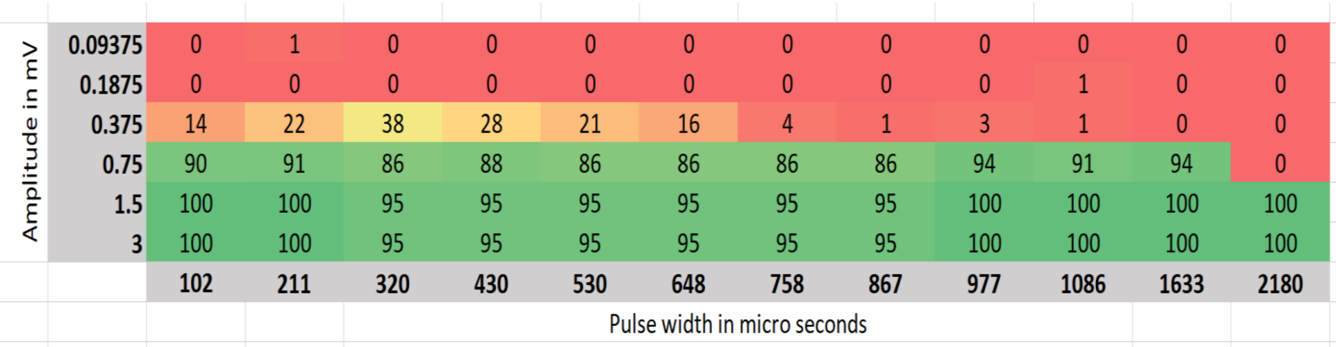
Measurement Setting	Record / Trial number	Number of records	Sensitivity (%)	PPV (%)
Setting #1	r01	50	94.5	95.3
Setting #2		51	89.7	96.7
Setting #1	r02	79	96.3	97
Setting #2		82	92.4	97
Setting #1	r03	75	85	98.1
Setting #2		71	75.4	100
Setting #1	r04	33	49.5	97.5
Setting #2		32	45.8	99.7
Setting #1	r05	79	66.8	97.2
Setting #2		76	63.3	98.8

Table 7 *Overall sensitivity and positive predictive values (PPV) and number of records with 100% sensitivity*

Measurement Setting	Overall Sensitivity (%)	Overall PPV(%)	Sensitivity <= 50%	Number of records with 100% Sensitivity
Setting #1	80.6	97.4	147	124
Setting #2	75.7	98.2	171	123

In the case of the public dataset, the results obtained are presented in Tables 8 and 9, the measurements were undertaken based on the default parameter settings of the MAX30001 AFE.

Table 8 *Sensitivity values in percentages displayed on a color gradient table. Higher sensitivity percentages are highlighted with green color whereas lower sensitivity values are highlighted with red color. Yellow and orange values represent those in between.*



Amplitude in mV	102	211	320	430	530	648	758	867	977	1086	1633	2180
0.09375	0	1	0	0	0	0	0	0	0	0	0	0
0.1875	0	0	0	0	0	0	0	0	0	1	0	0
0.375	14	22	38	28	21	16	4	1	3	1	0	0
0.75	90	91	86	88	86	86	86	86	94	91	94	0
1.5	100	100	95	95	95	95	95	95	100	100	100	100
3	100	100	95	95	95	95	95	95	100	100	100	100

Pulse width in micro seconds

Table 9 *Mean sensitivity values in percentage for various pulse amplitude values.*

Pulse amplitude (mV)	Mean sensitivity
0.09375	0
0.1875	0
0.375	13
0.75	82
1.5	98
3	98

From the above tables (Table 8 and Table 9) it was found that greater sensitivity percentages were recorded for amplitude values greater than 0.75-mV. Based on pulse width magnitudes, it is seen in Table 8 that in all pulse duration values, there are pacing pulse detections above 86% sensitivity.

4.2 Comparisons with other AFEs

Output comparisons were done between the MAX30001 and GE reference AFE by providing similar inputs. In addition to the typical GE reference AFE with basic register configurations, another prototype with modified software-based filter configurations was also included in this project. Results are presented on plots and a table is shown below.

The inputs were square pulse trains, noise, and paced ECG datasets that are discussed in the previous sections. As the two AFEs differ in their programmable parameters output formats and signal processing and handling, it is not a straightforward task to compare them. MAX30001 and the two reference AFE prototypes vary in design, and they follow different approaches to detecting pace events. The reference AFE from GE contains a digital pace detection algorithm capable of qualifying potential pacing artifacts. The internal digital detection algorithm searches for and validates pace artifacts in terms of amplitude, width, and repetition. On the other hand, MAX30001 doesn’t validate pacing pulses rather it stores every edge with enough energy (as defined by its slope and amplitude) and it is for the user to qualify a pair of edges as a valid pace. Time-stamped edges (as falling and rising) of the pace spikes with ECG are saved as output (.CSV) files.

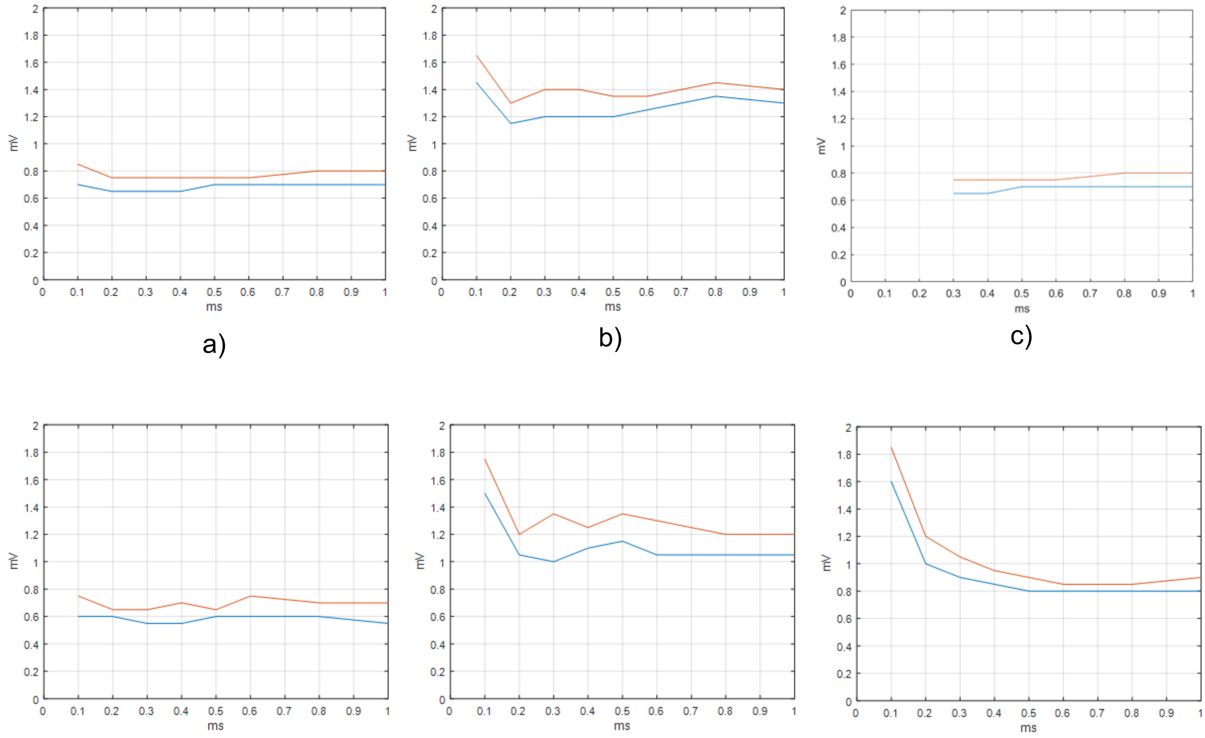
The parameters and their respective values of the MAX30001 AFE used in this comparison are shown below.

- Channel gain was set to 270 / 50.625 / 101.25 (In the order Normal / INA / PGA - as indicated on the GUI).
- Analog HPF (High Pass Filter) cutoff frequency set at 8000 Hz
- Detector threshold set at ±112.5 mV (default value).

At some of the tests, a third-order Butterworth low pass filter at 3.5 kHz cutoff frequency was introduced to the AFEs.

4.2.1 Comparison results from square pulses

The pulse trains mentioned in the last chapter (section 3.2.1) were fed as input to the MAX30001 and two prototypes of the reference AFEs as mentioned above.



a) b) c)

d) e) f)

Figure 26 *Response of the MAX30001 and GE AFEs to square pulse trains as input. The y-axis represents pulse amplitudes in mV and the x-axis represents pulse duration in ms.*

a) MAX30001 with minimum pulse width set at 0.01-ms – no rejection of narrow pulses.

b) MAX30001 with low pass hardware filter at a cutoff frequency of 3500-Hz

c) MAX30001 rejecting pulses less than a pulse width of 200-µs.

d) GE AFE with basic configuration

e) GE AFE with low pass hardware filter at a cutoff frequency of 3500-Hz

f) GE AFE prototype with modified software-based filter settings.

It is observed from the above plots. Figure 26 (a)(b)(d) and (e) that

- With measurement accuracies, in Figure 26 a) the 100% amplitude pulse detection threshold was 0.8 mV for MAX30001 while for GE's AFE it was 0.7-mV.
- With the added hardware filter both AFEs showed similar trends of decreasing sensitivity to short amplitude pulses. For GE's AFE, the full detection amplitude threshold rose to 1.2-mV and for MAX30001 it was approximately 1.4-mV. The sensitivity further dropped in both AFEs for pulses narrower than 0.2-ms.
- In Figure 26 c), it is shown that the detection of pulses narrower than 200-µs had stopped following the rejection of narrow pulses on MAX30001. Detection began from 0.3-ms for the test step was 0.1-ms.

4.2.2 Comparison results from noise

On this test, a 2-mV peak-to-peak (pp) white noise at a sampling rate of 96-kHz that was filtered at a 15-kHz low-pass filter was fed as an input and the number of pulse indications was counted for each of the AFE in one-minute time window.

The descriptions of abbreviations and notations used in presenting the outputs in Table 10 are listed below.

MAX30001 — MAX30001 with the following pace channel settings:

- Channel gain set to 270 / 50.625 / 101.25 (In the order Normal / INA / PGA - as indicated on the GUI).
- Analog HPF (High Pass Filter) cutoff frequency set at 8000 Hz
- Detector threshold set at ±112.5 mV default value.

MAX30001 LP3500 — MAX30001 with low pass hardware filter at a cutoff frequency of 3500 Hz.

MAX30001 LP3500 (200-µs reject) — MAX30001 with low pass hardware filter at a cutoff frequency of 3500 Hz and rejecting pulses narrower than 200-µs pulse width.

MAX30001 200-µs — MAX30001 rejecting pulses narrower than 200-µs pulse width.

GE AFE — GE AFE with basic configuration.

GE AFE LP3500 — GE AFE with low pass hardware filter at a cutoff frequency of 3500-Hz

GE AFE Modified — GE AFE with modified software–based filter settings.

Table 10 *Pulse counts from the noise test*

No	AFEs and Configurations	False positive rate (per minute)
1	MAX30001	1715
2	MAX30001 LP3500	975
3	MAX30001 LP3500 (200-µs reject)	734
4	MAX30001 200-µs	1674
5	GE AFE	384
6	GE AFE LP3500	183
7	GE AFE Modified	35

From Table 10, the number of false pulse indications is higher in the MAX30001 AFE compared to those in the reference prototypes. On this test, a 30-ms refractory period is incorporated into the pacing pulses detection algorithm of the MAX30001 AFE that limits the pulse count to 1715.

4.2.3 Comparison results from paced ECG datasets

As mentioned in the last chapter section 3.2.2, a dataset of paced human ECGs was employed in the evaluation process of the AFEs. These same datasets were also fed to the GE reference devices. In the database, there were records of 100 patients each with 5 trials of pacemaker settings. Although there were paced ECG data records, due to problems in opening the records, analysis was done on 414 cases. The annotations are based on all leads (8 leads), but the statistical analysis is based on lead II.

From Table 11, it is possible to conclude that the sensitivity values for MAX30001 are comparable to those in GE AFE with basic configuration. The addition of a low pass filter lowers the sensitivity values in both AFEs. Rejecting narrow pulses with pulse width less than 200-µs further decreases the sensitivity and PPV percentages in MAX30001. On the other hand, modifying the internal filters of GE AFE doesn’t bring an increment to the sensitivity value compared to the default configurations. The PPV (%) percentage remains high indicating the low noise content of the database.

Table 11 *Sensitivity and PPV values for paced ECG dataset from GE HealthCare*

No	AFEs and Configurations	Sensitivity (%)	PPV (%)
1	MAX30001	63.1	99.7
2	MAX30001 LP3500	42.2	99.5
3	MAX30001 LP3500 (200-µs reject)	34.1	64.7
4	MAX30001 200-µs	50.2	99.7
5	GE AFE	62.3	99.6
6	GE AFE LP3500	46.7	99.4
7	GE AFE Modified	54.6	99.5

4.3 Important findings

In testing the MAX30001 AFE, random noise was passed as an input and the detection thresholds were varied starting from the maximum value of ±337.5-mV. The results are shown below in Table 12. The 500 kHz random noise was generated in MATLAB using the ‘rand’ function and there was no filtering. With ±270 mV and lower thresholds, there was practically a constant row of rising edges every 7.8-ms, thus no valid pace pulses were found, but a large number of rising edges were registered.

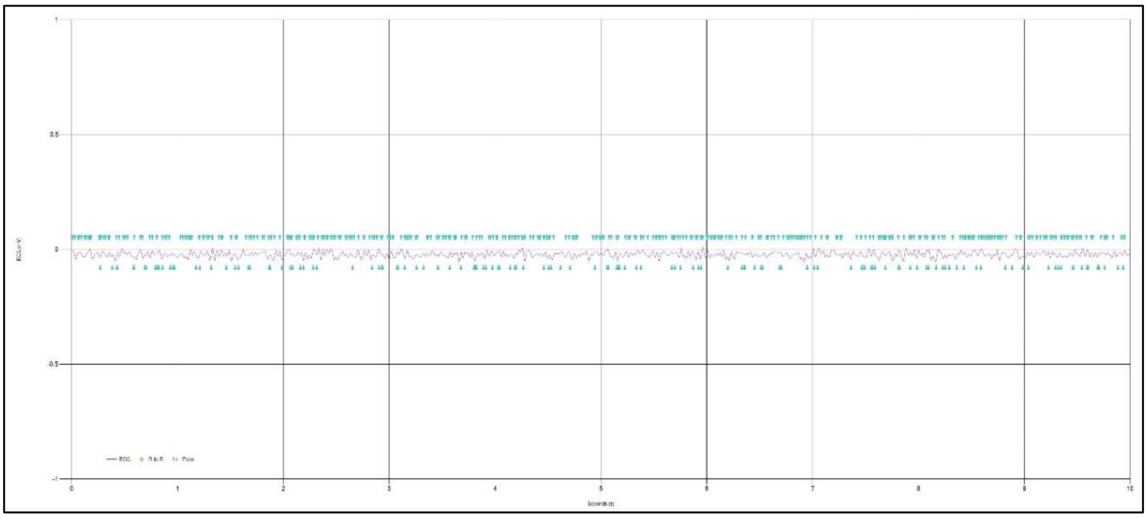


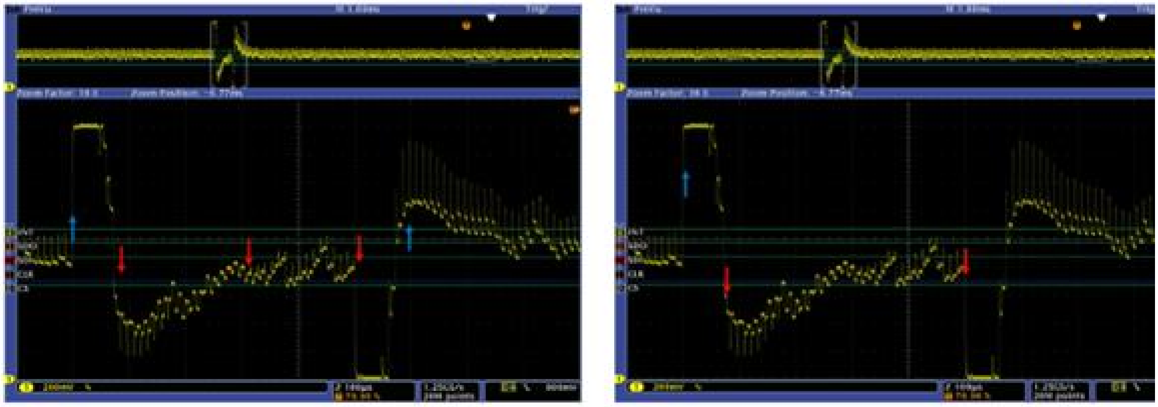
Figure 27 *A screenshot taken from on-screen streaming of the MAX30001 GUI plot showing constant rising (R) edges because of a random noise input.*

Table 12 *Quantity of rising and falling edges and number of valid pace pulses recorded obtained as random noise was provided as input to MAX30001 at various detection threshold voltages.*

Detection threshold value (mV)	Number of falling (F) edges	Number of rising (R) edges	Number of valid pace pulses
± 337.5	2391	9403	524
± 270	288	24016	5
± 202.5	1064	25442	7
± 135	2462	27274	2
± 67.5	3144	28792	0

After noticing this problem, the manufacturer was notified of the situation, and they came up with a solution by providing an alpha version of the MAX30001 GUI which allowed the user to select an EFIT different from the default 15. According to them, the problem was related to the overflow of the PACE banks/FIFO when they faced a burst of pulses.

We then started using an EFIT value of 5 for this noise test and as a result enhancements were observed.



a) b)

Figure 28 *Images showing generation of multiple (extra edges) on a single pace event due to ringing viewed on an oscilloscope. Figure courtesy of ADI - Roberto Munoz.*

a) 2 mVpp, HPF = 1 kHz, G = 540 V/V VTH = ±112 mV
b) 2 mVpp, HPF = 1 kHz, G = 540 V/V VTH = ±337 mV

Figure 28 displays oscilloscope images for a 2-mV peak-to-peak (pp) pace pulse edge detection experiment at two sets of parameter configurations of MAX30001. This shows in addition to the proper rising (R) and falling (F) edges of a single pulse event, extra edges are produced that could lead to incorrect analysis results due to false detections. In correct cases, a single pacemaker pulse must be represented by a single rising (R) edge and a single falling (F) edge. In Figure 28 (a) the system produces two extra falling (red-colored) and one extra rising (blue-colored) edge for a single pacing impulse at a detector threshold voltage of ±112 mV. In Figure 28 (b) with the ±337mV detector threshold setting, only one extra falling (F) edge is generated. This problem can be traced back to the ringing in the signal processing chain of the analog signal before the edge detector. More deflections are considered as edges at sensitive settings that can be for instance at low detector threshold voltage values.

5 CONCLUSIONS AND RECOMMENDATIONS

5.1 Conclusions

Pacing triggers originating from artificial pacemakers need to be detected firstly because they indicate the presence of cardiac pacemakers and secondly, they should be differentiated from the other components of the ECG signal. Of its other functions, MAX30001 is an AFE designed to detect the edge of these pulses. The performance assessment was focused on the MAX30001 electronic chip which is the main component of the MAX30001 EV kit. Various kinds of testing inputs were introduced at different combinations of parameters and configurations of the device. Two prototypes from reference designs of GE HealthCare were employed for comparison purposes.

A pulse train of eight batches each with a different pulse duration was constructed. They had 40 amplitude steps that were falling at a constant step. They were used to study the output of the AFE after which a pace pulse detection algorithm was developed. Based on the need of this thesis work, the AFE was configured to capture and save two (.CSV formatted) log files from the available logging options. One of the log files contained real-time stamped information on the pulse edges as falling (F) and rising (R). The other (.CSV formatted) log file detailed the ECG information in mV and real-time points, among others.

It was understood that the MAX30001 AFE doesn’t validate and store pace pulses. It is the responsibility of the application developer or user to produce an algorithm for detecting and validating pulses using the log files. A valid pace pulse was defined by a pair of opposite edges that fell within a defined pulse duration.

After this observation, the first task of the evaluation work was to draft a pace detection algorithm. The algorithm initially defined the minimum and maximum pace pulse widths and searched for opposite pairs of edges *i.e.,* falling, and rising edges that were close enough to each other, and validated those whose time gap fell within the minimum and maximum pulse widths. It ignored similar edges that could be found between these two edges or if the time gap requirement was not adhered to. The detection algorithm was flexible for further modifications and developments based on different requirements.

In the signal chain of the AFE, the test signals were subject to several signal conditioning stages. They passed through a bank of low- and high-pass filters as well as amplifications. This was accomplished before the edge detector. For this reason, the detector threshold voltages available in the dropdown list of the GUI pace channel didn’t have a straightforward relationship to the input amplitudes of the testing waveforms.

By introducing the square pulses, it was shown how sensitive and accurate the AFE was in detecting the perfectly square-angled pulses. Using the default configurations, it was possible to fully detect pulses to an amplitude threshold of 0.4-mV. At the same default settings by varying the detector threshold value to ±337.5-mV, which was the maximum available detector threshold value in the GUI of the AFE, this sensitivity decreased to a complete detection of an amplitude threshold of 1-mV. In both cases, all the available pulse widths were detected.

It was possible to add passive hardware filters on the options provided and indicated on the EV kit (C26, C28, R26, and R27) of MAX30001, four filters with different values of cutoff frequencies were applied and measurements were made at ±112.5-mV and ±337.5-mV detector threshold values. It was learned that the AFE’s sensitivity decreases to short-pace pulses as the cutoff frequency drops from 39. 3 kHz to 2.5 kHz in a proportional manner in both comparator values ±112.5-mV and ±337.5-mV.

Publicly available paced ECG datasets that were generated by Jekova *et.al* [10] using the HKP simulator were also introduced as inputs to MAX30001. The dataset included files with added physiological noise (tremor) expecting more false pace indications from the noisy waveforms. In the database, there were a total of 624 ECG waveforms. Measurements taken at the default setting of the AFE showed that pace pulses with amplitude values greater than 0.75-mV had sensitivity values higher than 80 %. Furthermore, it was able to detect all ranges of the available pulse widths and rising edge durations in the database by this default configuration. The effect of tremor in producing false pacing indications was found to be insignificant.

In the case of human-based data, we received ECG records of patients with implanted artificial pacemakers from GE HealthCare and passed them as input to the EV kit. The amplitudes and pulse widths of the pacemaker pulses were not available but using the annotation time stamps of the pacemaker pulses, it was possible to produce sensitivity and PPV percentages. The measurements were performed on two sets of parameter configurations. One of the configurations was the default setting and the other one was at a channel gain of 240 / 22.5 / 90, The analog HPF cutoff was set to 8000-Hz and the detector threshold to ±22.5-mV which is the lowest non-zero choice available in the GUI. Comparing the sensitivity and PPV values obtained from these configurations implied that if the channel gain was lowered then the detector threshold value should also be decreased to get equivalent sensitivity values to higher settings of channel gain and detector thresholds. Minimizing the channel gain by keeping the detector threshold at higher or default settings made the AFE less sensitive. Furthermore, it was also observed that the channel gain and detector threshold parameters were found to be more influential than other parameters of the AFE.

From the noise tests, it was seen that MAX30001 was too sensitive to random noise compared to the GE reference design prototypes. It was found that ringing in the analog signal chain gave rise to multiple edges. The deflections in the ringing of the signals were easily flagged as edges and a combination of these edges could lead to false detections. From these, and experiments performed later in this project, it can be concluded that the AFE's edge detector performed well but the ringing of the signals was one thing that needs improvement.

5.2 Recommendations

In addition to those waveforms mentioned in this thesis work, datasets that tend to generate false positive indications need to be involved in this kind of evaluation project. They could contain spikes or peaks that can easily be seen as pacing pulses by the detectors. Datasets that incorporate Electrostatic Discharges (ESD) can be mentioned as examples, among others. ECG waveforms from patients with Left Ventricular Assist Device (LVAD) are also potential candidates for this evaluation work in that they include high-frequency noise artifacts on the surface ECG [26].

In the case of the multi-lead ECG records from GE HealthCare that are used in this work, the statistical analysis was done based on measurements from one lead. Accurate and better sensitivity and PPV percentages can be obtained if the analysis includes all eight leads measurements and annotations.

From the accomplished tests, it was learned that MAX30001 has no issues in detecting the edges of spikes, the edge detector functions as expected. Multiple edges originated due to the ringing of the signal before the edge detector. This problem needs to be fixed. Once it is curbed, comparable results with GE reference design prototypes could be achieved by incorporating an appropriate programmable bandpass filter prior to the edge detector.

Testing datasets that are obtained from human test subjects are preferable to those generated by simulators. Synthetic datasets are prone to miss realistic conditions that could compromise their significance.

Adaptive pacing pulses detection is recommended to automate the pace pulses detection rather than manually changing the configurable parameters of the AFE. This helps the pacing algorithm adjust its parameter values based on the dynamics of the ECG waveforms.

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