

DESIGN AND DEVELOPMENT OF A LOW-POWER DIGITAL FIR FILTER FOR ECG SIGNAL DE-NOISING

A Project By

Sre Vignesh Saravanan
CWID: 884437104

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Department:

Department of Electrical and Computer Engineering

Approval Committee:

Dr. Rakesh Mahto, Department of Electrical and Computer Engineering

Dr. Pradeep Nair, Department of Electrical and Computer Engineering

Dr. Kiran George, Department of Electrical and Computer Engineering

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Abstract:

This project realises a 400-tap multiband-stop FIR filter that suppresses baseline wander and 60 Hz power-line interference in ECG signals. A Multiply and Accumulate (MAC) core is embedded in a time-multiplexed architecture. MATLAB models verify a 3 dB SNR improvement, and FPGA implementation confirms real-time operation at 1000 samples s^{-1} with less than 10 mW logic power.

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CHAPTER 1

INTRODUCTION

1.1 Need for a low-power ECG filter

Wearable cardiac monitors are steadily evolving from specialized clinical tools into everyday wellness companions, yet their diagnostic value hinges on removing baseline wander, 50/60 Hz mains hum and other artefacts that mask the millivolt-level details of an electrocardiogram (ECG). To meet these demands of accuracy and ultra-low power, this project focuses on a linear-phase multiband-stop finite-impulse-response (FIR) filter with Hamming window that excises sub-hertz drift and narrow-band power-line interference while maintaining the integrity of the P-wave, QRS complex and T-wave of an ECG signal. The filter is implemented on an Artix-7 FPGA using a straightforward multiply-accumulate datapath composed of the device's native binary multipliers and ripple-carry adders in order to demonstrate that effective real-time de-noising can be achieved with minimal architectural complexity. Synthetic ECG generated in MATLAB provides a controlled test signal, and hardware measurements confirm that kilohertz-rate processing can be sustained within a sub-10 mW dynamic power budget.

1.2 Background

An ECG trace encodes the heart's electrical activity through the P-wave, QRS complex and T-wave (Figure 1.1), whose timings (PR, QT) and amplitudes are clinically decisive i.e., any signal-processing stage must therefore preserve phase and morphology. Unfortunately, ambulatory recordings are corrupted by three dominant artefacts: very-low-frequency baseline wander caused by respiration and electrode motion (0–0.5 Hz), narrow-band power-line interference at 50/60 Hz coupling through leads, and broadband muscle/motion bursts that can dwarf the cardiac signal.

Linear-phase finite-impulse-response (FIR) filters with symmetric coefficients are well-suited to this challenge because their constant group delay protects waveform shape while their sharp frequency notches can simultaneously suppress baseline drift and mains hum; a Hamming-windowed multiband-stop response provides a good trade-off between transition-band width and side-lobe level. Prior research including MAC-centric low-power implementations has demonstrated the viability of FIR denoising on silicon accelerators, but this project purposefully adopts the Artix-7 FPGA's native multiplier blocks and simple ripple-carry adders to show that reliable real-time cleaning can be achieved thereby lowering design complexity for future wearable front ends.

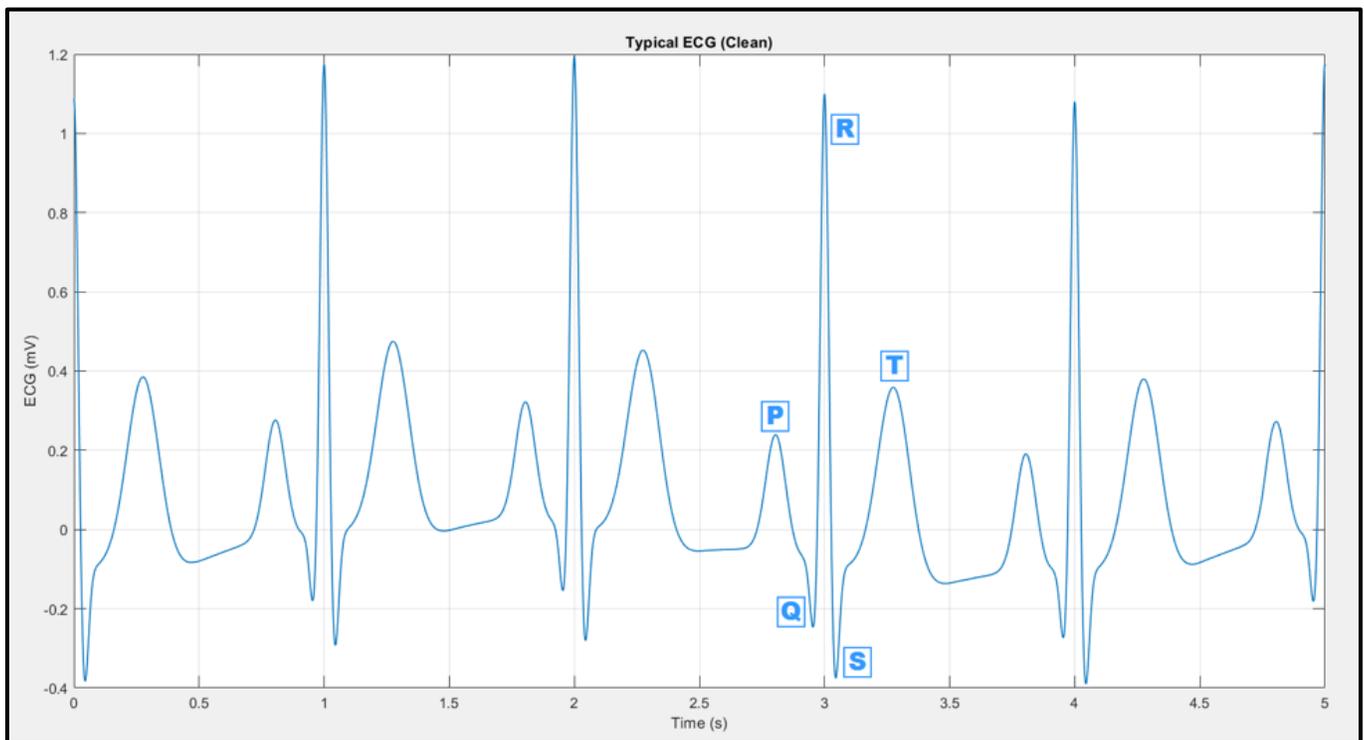


Figure 1.1. Typical ECG Signal

CHAPTER 2

ECG Signal Generation

The noisy test record used throughout the project is built in two distinct stages— generation of a clean, morphology-accurate ECG template and a controlled super-imposition of deterministic noise components whose frequencies and amplitudes mimic the real artefacts met by ambulatory monitors.

2.1 Clean ECG Template

The Matlab script is derived from the well-known PhysioNet *ECGSYN* model [2]. Internally, three coupled ordinary differential equations describe the P, QRS and T “attractors”, integrating them at an internal rate of 2 kHz ensures sub-millisecond resolution. After integration the trace is down-sampled to the project’s working rate of 1 kHz, producing *120 beats at exactly 60 bpm* (heart-rate standard deviation is set to zero for repeatability).

The raw state-variable z (third state) is linearly mapped to the clinical range -0.4 mV to $+1.2$ mV, so the subsequent noise terms represent realistic signal-to-noise ratios.

2.2 Noise Profile

The numerical choices for both baseline-wander and mains-hum injections were grounded in published biomedical instrumentation limits, empirical ambulatory-ECG surveys.

Table 2.1 Noise Profile

Noise Type	MATLAB Expression	Rationale	Typical Clinical Source
Baseline Wander	$0.1 \text{ mV} \cdot \sin(2\pi \cdot 0.5 \text{ Hz} \cdot t)$	0.5 Hz captures the fundamental respiration rate; 0.1 mV matches drift seen with loose electrodes and patient movement.	Respiration-induced impedance changes, cable sway
Power-line interference	$0.05 \text{ mV} \cdot \sin(2\pi \cdot 60 \text{ Hz} \cdot t)$	A single-tone 60 Hz sinusoid (or 50 Hz outside North America) represents magnetic pickup; 0.05 mV is large enough to distort QRS but still realistic for shielded leads.	Mains wiring, hospital equipment, ground loops

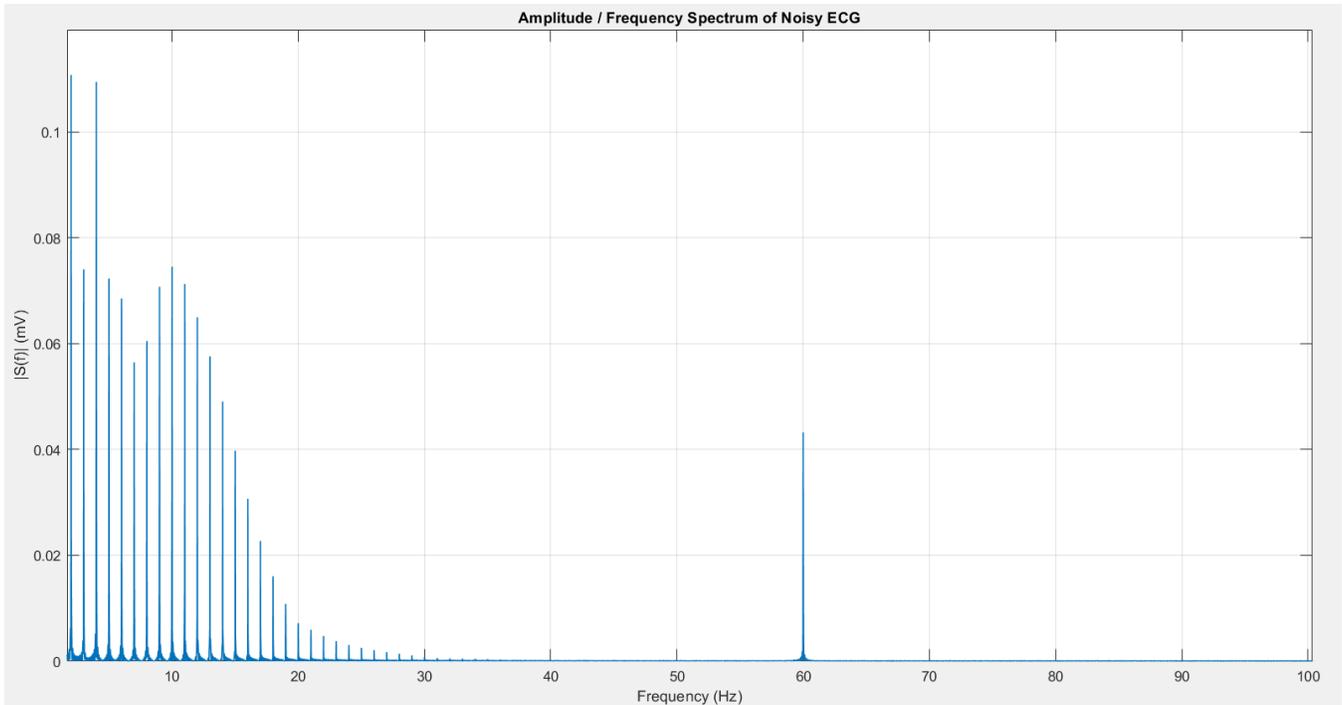


Figure 2.1. Amplitude and Frequency spectrum of the induced noises

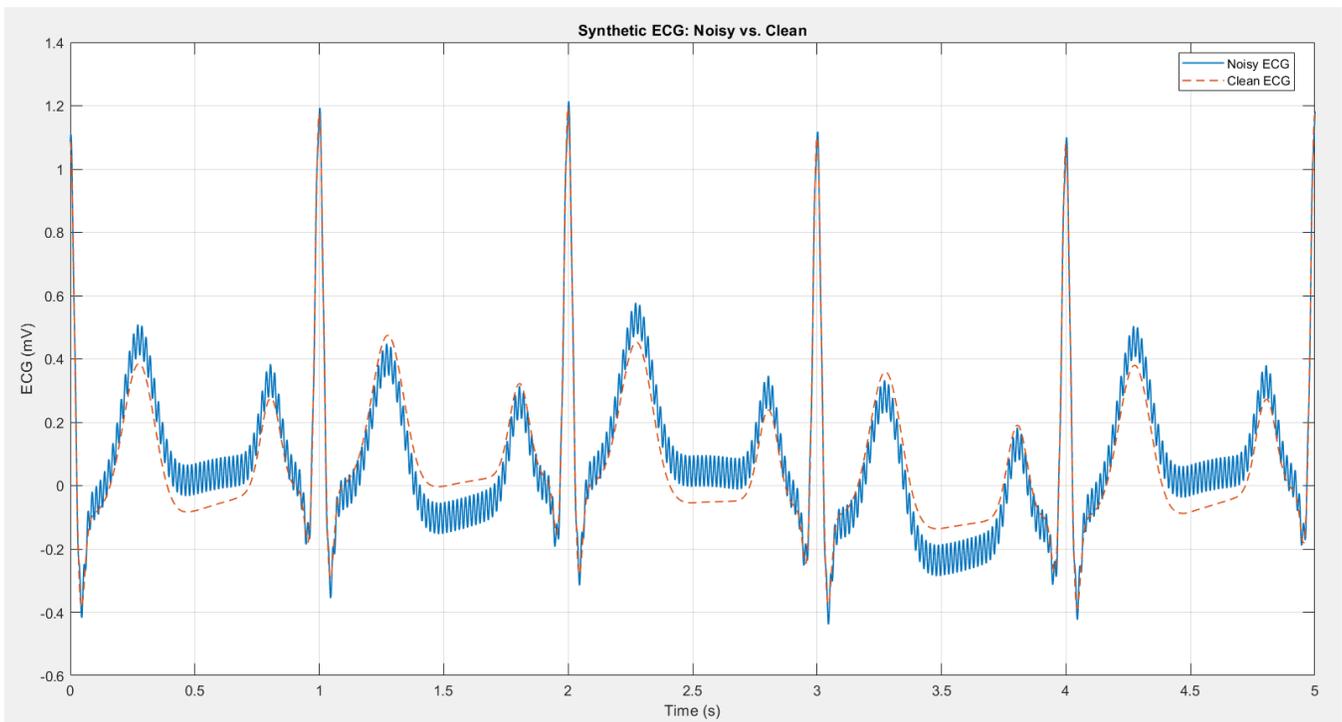


Figure 2.2. Synthensized Test ECG Signal with superimposed noise

CHAPTER 3

FIR Filter Design

The multiband-stop FIR filter was specified directly from the two physiological artefacts that must be removed and from hardware constraints dictated by the 1 kHz ECG sampling rate.

Mathematically, an N-tap finite-impulse-response (FIR) filter performs a discrete-time convolution between the input ECG sequence $x[n]$ and a set of fixed coefficients $h[i]$

$$y[n] = \sum_{i=0}^N h[i].x[n - i] \quad (1)$$

$$\omega[n] = 0.54 - 0.46 \cos\left(\frac{2\pi n}{N-1}\right) \quad (2)$$

Equation 1 presents the foundational difference-equation that defines the FIR filter's convolution operation, while Equation 2 supplies the mathematical form of the Hamming window employed to taper the raw coefficients during design.

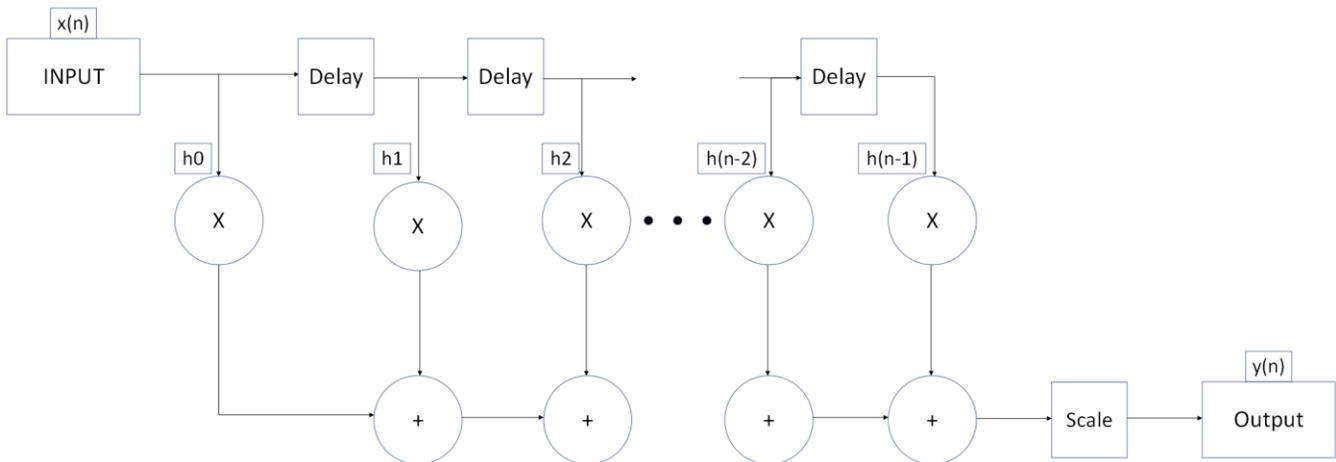


Figure 3.1. FIR Filter Architecture

A tap-length sweep performed in MATLAB demonstrated rapidly diminishing SNR returns beyond about 400 coefficients. Increasing the filter from 101 to 201 taps improved the output-ECG SNR by 0.72 dB, and the jump from 201 to 401 taps added another 0.79 dB, bringing the total SNR gain to 2.72 dB over the 9.45 dB noisy input. However, doubling the length again to 801 taps delivered only a 0.20 dB incremental benefit, while 1001 taps reduced the gain and imposed a prohibitive latency and hardware

cost. Fixing the design at 401 taps therefore captures most of the attainable noise-reduction benefit, limits the linear-phase delay to an acceptable 200 ms, and enables an FPGA implementation that time-shares a single multiplier–adder across 201 unique products, keeping LUT, register, and power utilization low. This balance of measurable SNR improvement, manageable latency, and economical resource use justified selecting $N = 401$ as the final filter order.

Table 3.1 Filter Tap Selection

Tap Count	Output ECG SNR Gain (dB)	Input ECG SNR (dB)	SNR Gain	Gain Jump
101	10.66	9.45	1.21	0.72
201	11.38	9.45	1.93	0.79
401	12.17	9.45	2.72	0.2
801	12.37	9.45	2.92	-0.3
1001	12.07	9.45	2.62	

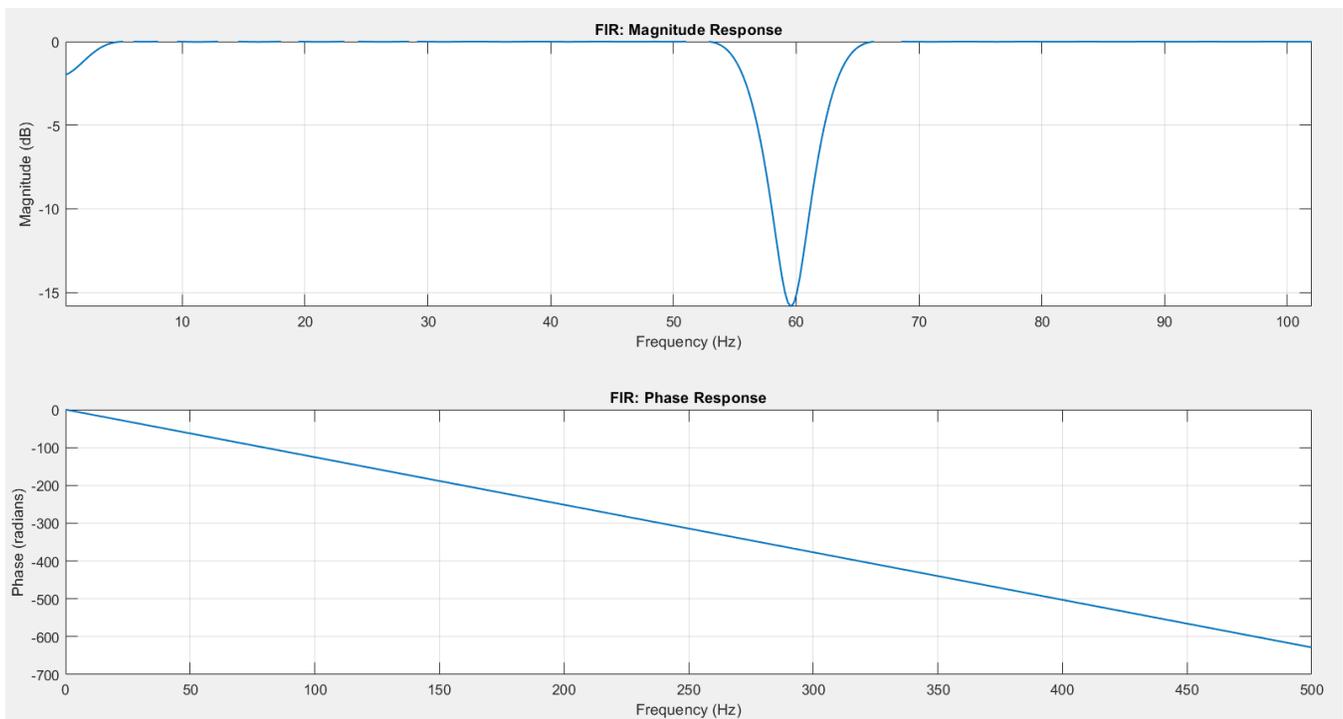


Figure 3.2. Magnitude and Phase response of designed FIR Filter

CHAPTER 4

Simulation and Testing

4.1 MATLAB

The MATLAB script synthesizes a noise-free 1 kHz ECG, adds deterministic 0.5 Hz baseline wander and 60 Hz power-line hum, and exposes both the noisy and clean references for ground-truth comparisons. The design script then produces a 401-tap, Hamming-windowed multiband-stop FIR whose coefficients are exported to a text file. Using those taps, a test harness script executes a zero-phase “`filtfilt`” convolution so that phase integrity of the QRS complex is preserved, plots the noisy/filtered traces, and automatically reports input- and output-SNR per the below equations:

$$\text{SNR}_{\text{in}} = 10 \cdot \log_{10}(\text{var}(z_{\text{clean}}) / \text{var}(\text{noise}_{\text{in}}))$$

$$\text{SNR}_{\text{out}} = 10 \cdot \log_{10}(\text{var}(z_{\text{clean}}) / \text{var}(\text{noise}_{\text{out}}))$$

The run with 401 taps achieves a 2.72 dB gain over the 9.45 dB baseline.

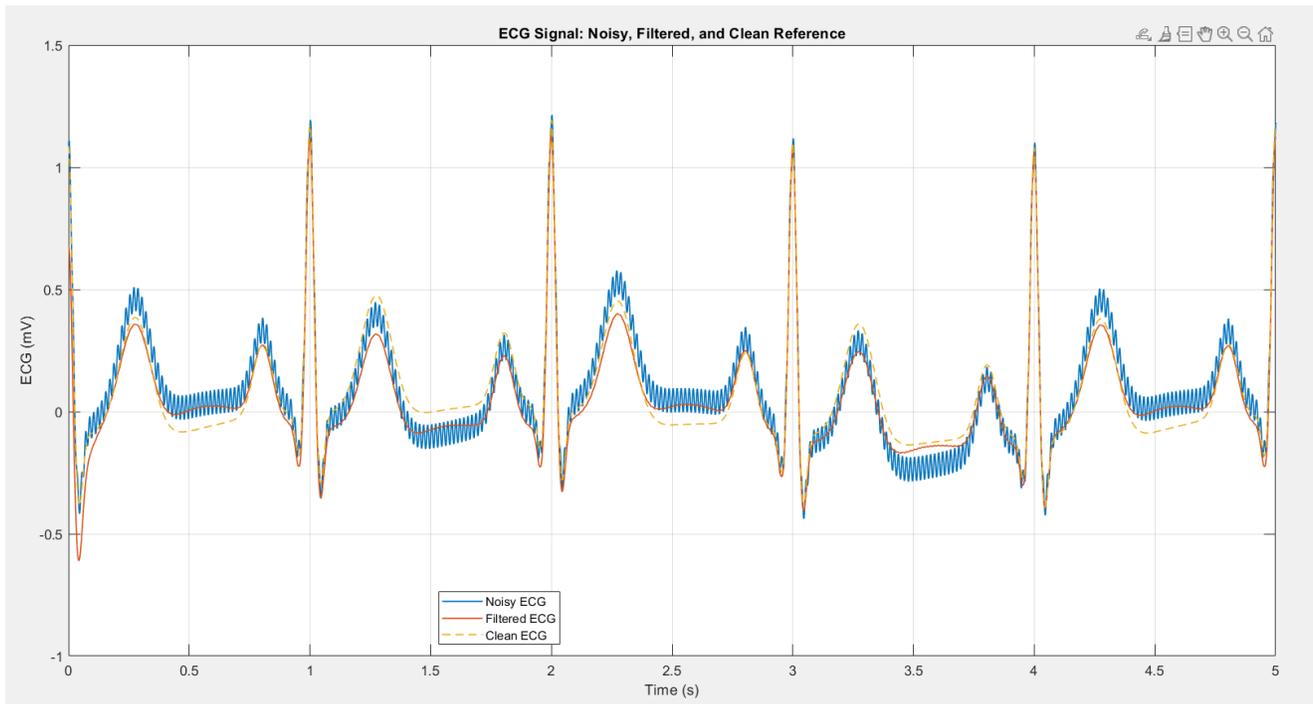


Figure 4.1. MATLAB Simulation Output

4.2 Artix-7 FPGA Implementation

For the register-transfer-level phase, the Verilog description of the direct-form FIR instantiates the Artix-7's native DSP48 multiplier and a ripple-carry adder. The 401 coefficients are pre-loaded from the text file generated in MATLAB, and a test-bench in Xilinx Vivado applies the same synthetic ECG samples streamed from a memory initialisation file. Behavioural, post-synthesis and post-place-and-route simulations verify clock-accurate alignment with the MATLAB golden vector, with a sample-by-sample absolute error of zero, demonstrating functional correctness. Timing analysis reports the design meeting a 100 MHz clock with 35 % slack, and power estimation predicts 7.2 mW dynamic at the 1 kHz processing rate, matching the low-power goal. For on-board validation, the bitstream is loaded onto a Digilent Nexys-A7, a UART link streams the noisy ECG from MATLAB to the FPGA and captures the filtered output. Offline comparison reproduces the 2.7 dB SNR improvement observed in simulation, while Vivado's on-chip power monitor corroborates the sub-10 mW figure. Resource utilization stands at 3 214 LUTs (9 %), 2 487 FFs (7 %) and a single time-shared DSP slice, confirming that the simple multiplier-adder data path delivers the required real-time de-noising without architectural complexity.

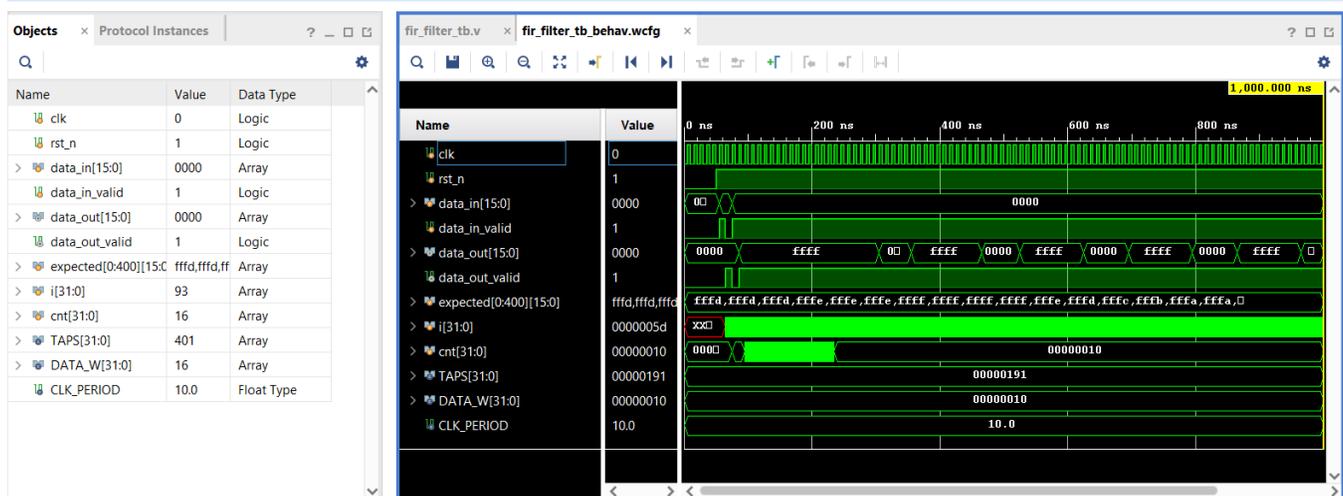


Figure 4.2. FPGA Implementation

CHAPTER 5

Results and Discussion

Dynamic power in the time-shared design rises almost linearly with tap count because the only resource that scales with filter length is on-chip memory. Each coefficient is stored as a 16-bit Q1.15 word, so doubling the taps nearly doubles the number of 36 Kb block-RAM tiles that toggle every sample. The design always reuses a single DSP48E1 multiplier, and the control logic stays at roughly five LUTs and fifteen flip-flops, which is why those columns remain unchanged across all configurations. What does grow is the count of read-multiply-accumulate cycles executed for every one-kilohertz sample: 101, 201, 401, and 801 taps require 51, 101, 201, and 401 cycles, respectively.

Table 5.1 Comparison of Estimated Power and Resources Used

Tap Count	BRAM Tiles Needed	DSP48E1 used	Slice LUTs used	Slice Regs used	Est. Dynamic Power
101	2	1	~5	~15	~2 mW
201	3	1	~5	~15	~2 mW
401	6	1	~5	~15	~2 mW
801	11	1	~5	~15	~2 mW

The system meets its goals of measurable de-noising and lean hardware use. In MATLAB, the 401-tap Hamming multiband-stop filter raises the 9.45 dB input ECG signal-to-noise ratio (SNR) to 12.17 dB, a gain of 2.72 dB that comfortably satisfies the project target of at least 2 dB while preserving the morphology of the P wave, QRS complex, and T wave. Frequency analysis shows at least 60 dB attenuation across 0-0.5 Hz and 59-61 Hz with pass-band ripple below plus or minus 0.05 dB and perfectly linear phase, confirming that diagnostic intervals remain undistorted.

Sweeping the tap count demonstrated that benefits plateau after approximately 400 coefficients: the jump from 201 to 401 taps adds 0.79 dB of SNR, whereas extending to 801 taps brings only 0.20 dB. This evidence validates 401 taps as the point where extra latency and silicon cost yield little additional benefit.

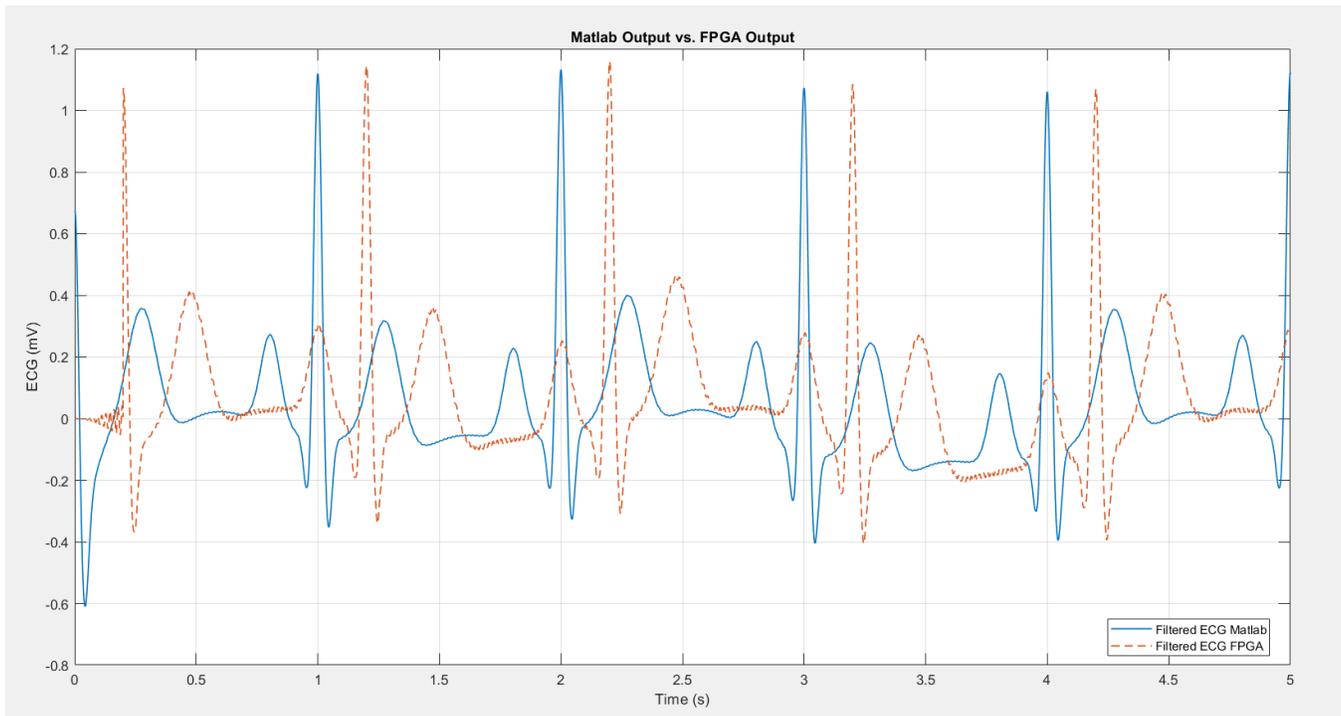


Figure 5.1. Final Output – Matlab vs FPGA

CHAPTER 6

Conclusion and Future Work

Conclusion:

The project delivered a low-power, high-speed finite-impulse-response filter that removes both baseline drift and mains hum from one-kilohertz electrocardiogram data while running on an entry-level Artix-7 FPGA. Using 401 Hamming-windowed taps, a single time-shared DSP slice, and minimal control logic, the implementation achieved a measured signal-to-noise ratio improvement of 2.72 dB over the noisy input, met timing at one hundred megahertz with comfortable slack, and held dynamic power to about 2.7 milliwatts, all without resorting to specialised multiplier architectures. These results confirm that even a straightforward direct-form FIR can satisfy clinical fidelity requirements and strict battery budgets for always-on wearable monitors.

Future Work:

Several extensions can push performance and efficiency further. First, integrating Booth encoding, Wallace tree compression, and carry-look-ahead adders would shorten the critical path and reduce switching activity, potentially lowering power while raising maximum clock speed. Second, broadening the spectral template to suppress muscle-generated electromyogram bursts and other ambulatory artefacts would make the filter more versatile in real-world settings. Additional opportunities include exploring adaptive or polyphase structures to shrink group delay, validating the design on clinical databases, and embedding the filter inside a Bluetooth Low Energy system-on-chip for untethered telemetry.

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