# BILATERAL EARLOBE PULSE TIMING MEASUREMNT DEVICE DEVICE

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## Abstract

Our PCB sensing system measures bilateral pulse arrival times by synchronizing a single-lead ECG with dual-channel earlobe PPG sensors. The hardware architecture utilizes low-noise analog front ends and a shared-clock data acquisition strategy to maintain multichannel timing jitters. The system is designed to compute ECG-referenced PPT across varying physical orientations, including neutral and head-tilt positions (or special cases like stroke and other medical symptoms). Success is defined by the simultaneous acquisition of clear ECG R-peaks and stable PPG waveforms from both earlobes. By achieving sub millisecond synchronization, this platform provides a dedicated hardware tool for quantifying bilateral timing variances introduced by body posture.

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# 1 Introduction

## 1.1 Problem

Pulse arrival time (PAT) and pulse transit time (PTT) are widely studied as a non-invasive physiological metric that reflects cardiovascular dynamics, vascular stiffness, and autonomic regulation. Conventional PTT systems typically measure the time delay between an ECG R-peak and a single peripheral photoplethysmography (PPG) waveform, often at the finger or ear. While these systems provide useful global timing information, they do not enable synchronized bilateral comparisons of pulse arrival times, which are important for detecting asymmetric blood flow conditions that may indicate vascular abnormalities such as stroke.

Currently, there is a lack of low-cost hardware platforms capable of acquiring multiple physiological signals with precise time synchronization. Accurate bilateral PTT comparison requires sub-millisecond alignment between ECG and multiple PPG channels to ensure that measured timing differences reflect true physiological variations rather than measurement error. Many existing research systems rely on independent sensors or software-based synchronization, which can introduce timing uncertainty and limit the reliability of bilateral timing analysis. As a result, no readily available measurement tools allow controlled bilateral pulse timing comparison between the left and right earlobes with sufficient precision. Without synchronized acquisition hardware, it is difficult to investigate how factors such as posture, head orientation, or asymmetric vascular conditions influence bilateral pulse timing.

From a societal perspective, improved non-invasive cardiovascular sensing tools contribute to public health and preventive medicine. Cardiovascular disease and stroke remain leading causes of mortality worldwide. Systems capable of accurately measuring bilateral pulse timing differences could support future research into vascular asymmetry, autonomic regulation, and early indicators of circulatory abnormalities. Although this project does not aim to produce a clinical diagnostic device, it provides foundational measurement capabilities for investigating bilateral cardiovascular dynamics and developing next-generation wearable health monitoring technologies. Additionally, affordable synchronized physiological measurement systems can broaden access to research tools in educational and low-resource environments.

## 1.2 Solution

This project proposes a custom PCB-based multi-channel physiological sensing platform capable of simultaneously acquiring:

- One ECG channel (cardiac timing reference)
- Two synchronized PPG channels (left and right earlobes)

The ECG channel provides a reliable R-peak reference for cardiac cycle timing. Two identical PPG sensing channels measure pulse waveforms at both earlobes. By computing pulse arrival times relative to the ECG R-peak, the system enables bilateral pulse timing comparison under controlled experimental conditions such as neutral posture, stroke, head tilt, or side-lying orientation.

The design emphasizes:

- Low-noise analog front-end circuitry
- Hardware-level time synchronization
- Shared sampling clock architecture
- Precise multi-channel ADC acquisition

All channels are sampled using a shared clock source to minimize relative timing jitter. Bluetooth, if implemented, is used strictly for data transmission and not for timing synchronization.

The system is positioned as a measurement and validation tool rather than a medical diagnostic device. Its primary purpose is to provide synchronized physiological waveform acquisition with sufficient timing precision to analyze bilateral pulse transit differences.

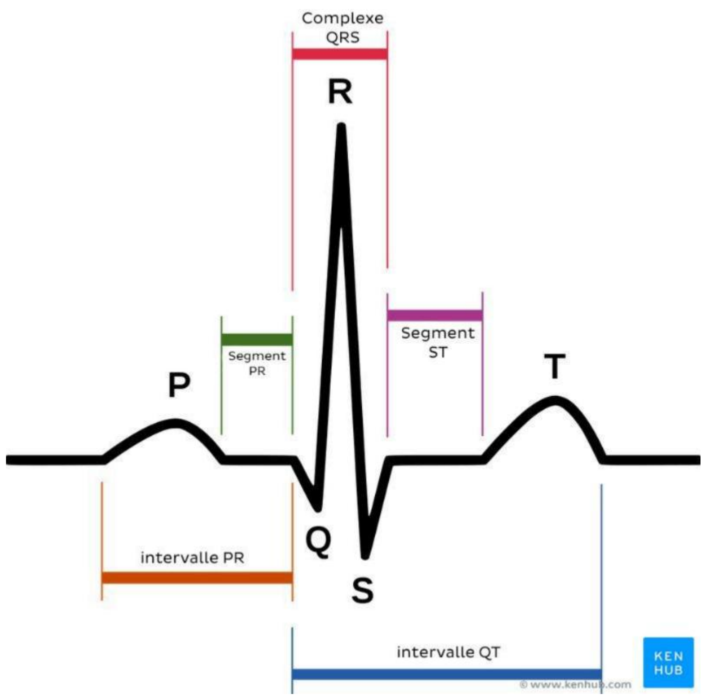


**Figure 1. The different parts of a typical EKG wave**

## 1.3 High-level Requirements:

- The microcontroller (nRF52840 chip) basic functions are working normally, including Bluetooth Low Energy wireless connection, data writing and reading, SPI communication, and the basic charging system is functioning properly.
- The MAX30003 driver is functioning properly, able to collect ECG signals and display any person's electrocardiogram dynamically via Bluetooth wireless transmission (ECG board fully working).

- Both MAX86141 drivers are working properly: they can drive the LEDs to flash, collect signals generated by the photodiodes, correctly plot the PPG waveform, and calculate the PTT time difference between both sides.

## 1.4 Visual Aid

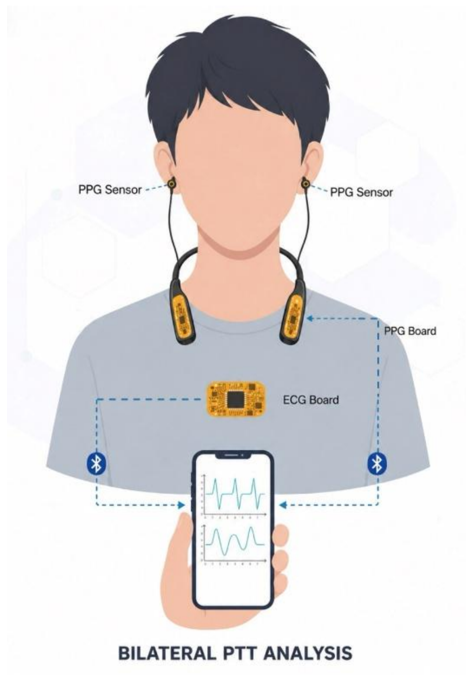


**Figure 2. A small, animated design that is not to scale of what are design looks like in terms of a physical model**

# 2 Design

## 2.1 Block Diagram

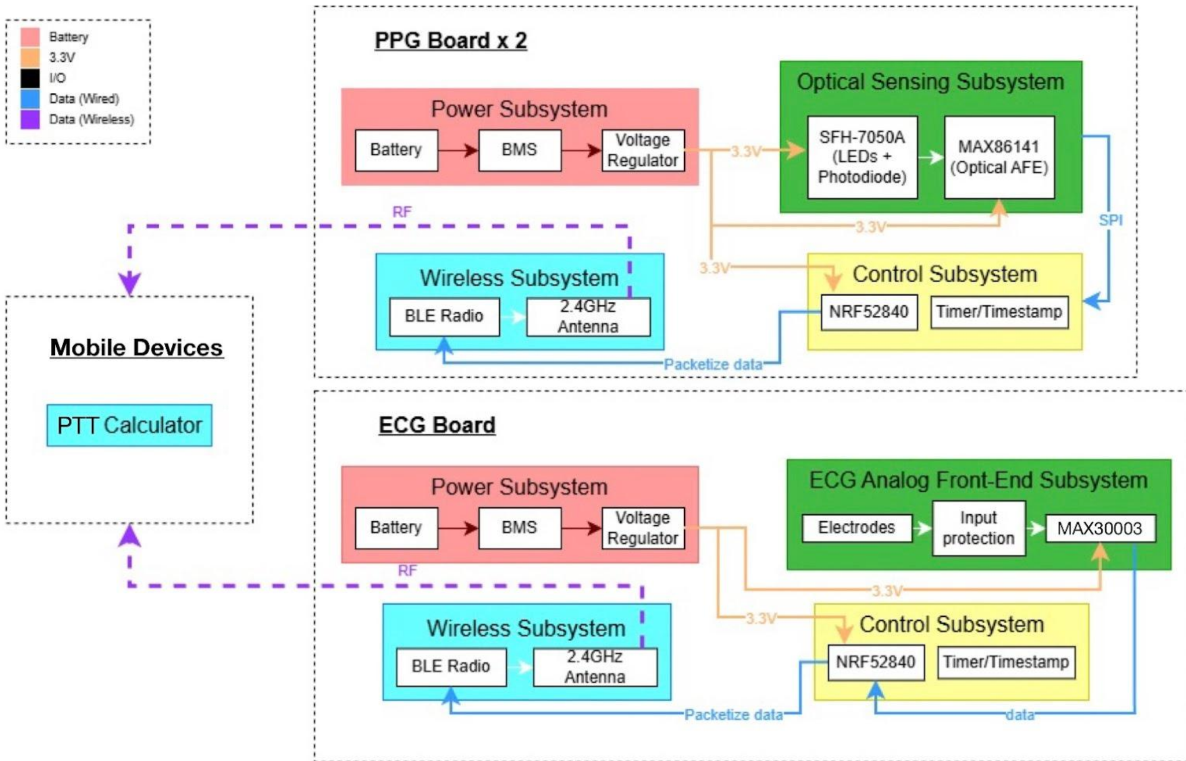


**Figure 3. Block diagram of bilateral earlobe pulse timing measurement system.**

## 2.2 Physical Design

The system consists of a central wearable hub and three peripheral sensor leads: two earlobe clips containing the SFH-7050A sensors and one 3-lead ECG cable for chest placement. The central hub measures approximately 60mm x 40mm x 15mm. Earlobe clips are spring-loaded to ensure consistent contact.

## 2.3 Software Subsystem

The software subsystem, hosted on the nRF52840, acts as the brain for the platform's data acquisition and wireless communication. Upon initialization, the firmware configures the MAX30003 and MAX86141 via the SPI bus, setting optimal sampling rates and gain stages while enabling hardware interrupts for data readiness. To meet the high-level requirement of sub-millisecond timing precision, the software utilizes the nRF52840's Programmable Peripheral Interconnect and hardware timers to capture timestamps immediately upon receiving trigger signals from the sensors' interrupt pins, effectively bypassing CPU interrupt latency. The embedded application manages synchronized ring buffers to align the ECG and bilateral PPG streams, calculates the R-peak to pulse arrival time using fixed arithmetic and encapsulates this data into custom Bluetooth Low Energy. This ensures that the mobile application receives a continuous, time-aligned stream of physiological data for visualization on cardiovascular analysis.

## 2.4 Power Subsystem

Each of the two PCB modules is powered independently by a lithium polymer battery managed by a dedicated battery management system IC. The BMS provides protection functions including overcharge cutoff, over-discharge protection and short-circuit current limiting, all of which are critical for a wearable device that operates in close contact with the user's skin during extended recording sessions. The raw battery voltage, which varies between approximately 3.0 V at minimum charge and 4.2 V at full charge with a nominal value of 3.7 V, is stepped down to a stable 3.3 V supply rail by a linear voltage regulator. The LDO topology was deliberately chosen over a switching regulator for both boards, because switching regulators introduce high-frequency ripple at their switching frequency that couples into sensitive analog front-end circuitry. On the ECG board, the MAX30003 analog front-end amplifies biopotential signals in the microvolt-to-millivolt range, meaning even a few millivolts of supply ripple can introduce artifacts that distort the ECG waveform or corrupt R-peak timing. By using an LDO, the noise supply presented to the analog circuitry is reduced to the LDO's own output noise floor, which is lower than that of any switching converter operating at the same output power.

Decoupling and filtering are implemented at the point of load for both the analog and digital domains. A combination of electrolytic capacitance and high-frequency ceramic bypass capacitance is placed at the LDO output and at the power supply pins of each IC to suppress broadband noise and prevent digital switching transients from the NRF52840 from coupling back into the analog supply.

## 2.5 ECG Analog Front-End System

The ECG analog front-end subsystem is responsible for acquiring the cardiac biopotential signal from the body surface, amplifying it to a level suitable for analog-to-digital conversion and rejecting the common mode interference and motion artifacts that are inherently present in electrode measurements. This subsystem is built around the MAX30003 single-channel biopotential IC, which integrates a high-gain instrumentation amplifier, programmable bandpass filtering, a driven circuit for common-mode rejection, lead-off detection, and an 18-bit ADC into a single package optimized for wearable ECG acquisition.

The raw biopotential signal present at the electrode-skin interface is in the range of 0.5 mV to 1.6 mV peak-to-peak for a standard Lead configuration, riding on a common-mode voltage that can be several hundred millivolts in amplitude due to electrode potentials and motion. The MAX30003 applies a fixed gain of 20 V/V through its internal instrumentation amplifier stage before the signal is presented to the ADC. The resulting ADC input range spans ±1.6 mV referred to the input, with an 18-bit resolution corresponding to approximately 12.2 nV per LSB. This resolution is more than sufficient to resolve the fine morphological features of the ECG waveform, including the P-wave, QRS complex, and T-wave, as well as subtle baseline variations that carry information. The ADC operates at a programmable sample rate; for this project it is configured at 512 Hz, which provides adequate temporal resolution to identify R-peak timing with sub-millisecond precision when interpolation is applied.

The conditioned and digitized ECG data is read from the MAX30003 over an SPI interface by the NRF52840 microcontroller. The MAX30003 asserts an interrupt line when its internal FIFO contains new samples, which the NRF52840 firmware services to retrieve data with minimal latency. During live demonstration, the ECG board successfully produced recognizable ECG waveforms on the companion laptop application at a stable 70 BPM, with clearly identifiable R-peaks that could be used as timing references for PTT computation.

## 2.6 PPG System

The PPG subsystem is responsible for measuring the volumetric pulse waveform at both earlobes simultaneously using reflectance photoplethysmography. The fundamental operating principle relies on the differential optical absorption of oxygenated hemoglobin relative to the surrounding tissue at near infrared and red wavelengths. When an LED illuminates the earlobe tissue, a fraction of the emitted light is reflected to a photodiode after scattering through the superficial vasculature. As a cardiac pulse wave arrives at the earlobe, the local blood volume transiently increases, causing a proportional increase in optical absorption and a corresponding decrease in the intensity of light reaching the photodiode. This produces a periodic photocurrent waveform whose timing reflects the moment of pulse arrival.

Each optical sensing site uses the SFH-7050A multi-wavelength LED and photodiode module paired with the MAX86141 optical analog front-end IC. The SFH-7050A integrates red, infrared, and green LED emitters alongside a broadband photodiode sensitive across the 600–1000 nm wavelength range in a compact surface-mount package designed for skin-contact applications. The MAX86141 provides programmable LED drive current control, ambient light cancellation, and a 19-bit internal ADC with a photocurrent input range of 78 nA to 4096 nA and a resolution of 7.8 pA per LSB. The high bit depth and low input-referred noise floor of the MAX86141 are necessary to resolve the small AC component of the PPG signal typically less than 2% of the total received light intensity against the much larger DC background. The device is configured to sample at 1024 Hz with 8 samples averaged, effectively presenting 128 output samples per second to the NRF52840 with the noise floor reduced through averaging.

A central design decision in the PPG subsystem was the consolidation of both earlobe sensing channels onto a single PCB controlled by one NRF52840 microcontroller. The original design used two independent PPG boards, each with its own microcontroller and crystal oscillator. This approach was abandoned after identifying a synchronization limitation: because BLE operates on a time-division multiple-access scheme, a central host cannot broadcast a signal to two peripheral devices simultaneously. The resulting variable delivery latency would introduce timing error directly into the bilateral PTT measurement. By placing both MAX86141 sensors on a single board controlled by one shared clock, both channels are guaranteed to be sampled from the same time base with no inter-board jitter.

The physical factor of the PPG board underwent significant iteration. The first version used a snaked flexible PCB trace between the sensor modules at the earlobes and the MAX86141 and NRF52840 at a central hub. While this configuration improved wearability by distributing the electronics around the head, the long analog traces between the SFH-7050A photodiode output and the MAX86141 input acted as antennas for ambient electromagnetic interference. The resulting noise dominated by 60 Hz from nearby digital circuitry overwhelmed the low-amplitude photodiode photocurrent. Although the photodiode could still respond to a direct flashlight stimulus, the noise floor was too high to extract a cardiac pulse signal. The final design revision addressed this by relocating the MAX86141 IC to a small PCB tab immediately adjacent to the SFH-7050A module at each earlobe, minimizing the length of the unshielded analog signal path to a few millimeters. The digital SPI bus, which is far more immune to noise than the photodiode photocurrent signal, carries data from the MAX86141 back to the central NRF52840 over the longer inter-board distance. Despite this redesign, the PPG subsystem ultimately returned all-zero data values from the MAX86141 FIFO during final testing. LED illumination, SPI register access, and BLE transmission were each individually verified as functional, leading to the hypothesis that either the MAX86141 optical AFE was damaged or that a PCB layout deficiency in the optical input path, potentially exacerbated by poor solder mask registration on the fabricated boards.

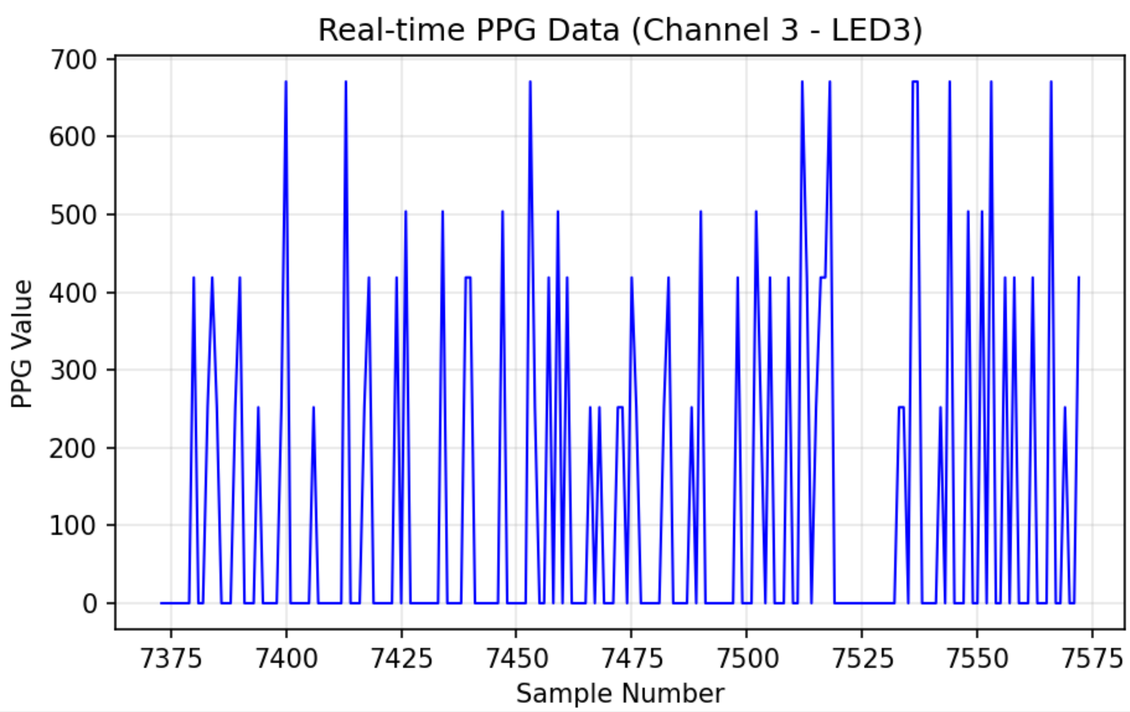


**Figure 4. high-amplitude spike pattern shown in the noisy PPG data captured during intermediate testing.**

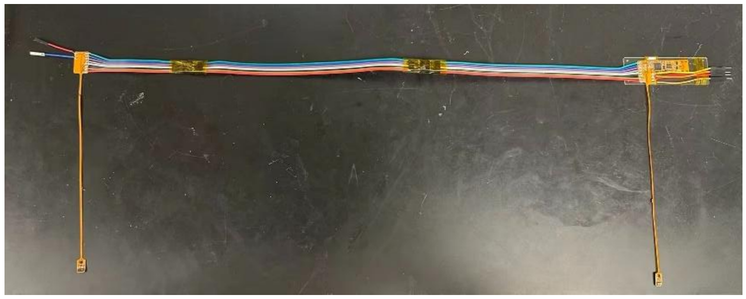

**Figure 5. A complete picture of our last iteration of our PPG system.**

## 2.7 Data Acquisition and Control Systems

The data acquisition and control subsystem coordinates all sampling, timestamping, packetization, and wireless transmission tasks on each board. Both the ECG board and the PPG headband board are built around the NRF52840 system-on-chip, which integrates an ARM Cortex-M4F processor running up to 64 MHz, a 12-bit successive approximation ADC, multiple SPI and I2C peripheral instances, a hardware timer with microsecond resolution, and a Bluetooth radio with integrated 2.4 GHz antenna matching network. All firmware is developed under the Zephyr real-time operating system using the nRF Connect SDK, which provides a mature BLE stack, hardware abstraction layer drivers for the SPI and ADC peripherals, and a thread scheduler that allows the sensor sampling loop and the BLE notification stack to execute concurrently without interfering with each other's timing.

On the ECG board, the NRF52840 services an interrupt from the MAX30003 each time new samples are available in the ECG FIFO. The firmware reads the available samples over SPI, appends a hardware timestamp generated by the NRF52840's internal RTC counter, and assembles a BLE notification packet for transmission. On the PPG board, the NRF52840 similarly polls or interrupts on both MAX86141 FIFOs in sequence at the configured output data rate, reading left and right channel samples and timestamping them from the same RTC counter. Because both PPG channels are read by the same microcontroller from the same clock source, the left and right sample timestamps are directly comparable with no inter-board synchronization overhead.

Data is transmitted wirelessly using the Nordic UART Service, a standardized BLE GATT profile that exposes a notify characteristic over which arbitrary byte streams can be sent to a connected central device. Each BLE notification packet carries a board identifier byte, a monotonically incrementing sequence number, a timestamp in milliseconds derived from the NRF52840 uptime counter, and a payload of raw sensor samples. The sequence number is the primary mechanism for aligning the ECG and PPG data streams on the receiving side: because BLE introduces variable packet delivery latency, wall-clock timestamps alone are insufficient to reconstruct the relative timing between the two boards. By matching packets with identical sequence numbers, the receiving application can reconstruct a synchronized view of the ECG and PPG waveforms and compute PTT as the difference between corresponding feature timestamps. The ECG board successfully demonstrated this pipeline end-to-end, streaming waveform data to the companion web application at 512 Hz with stable BLE connectivity and correct heart rate computation throughout extended recording sessions.

## 2.8 Tolerance Analysis

**Inter-Channel Timing Accuracy**

The primary technical risk of this design is achieving sub-millisecond synchronization between the ECG and bilateral PPG channels. Since pulse arrival time differences are the key measurement outcome, timing uncertainty must remain below 1 ms.

**Sampling Resolution**

The system may sample all channels at:

$$f_s = 1000\,\mathrm{Hz}$$

Thus,

$$T_s = 1\,\mathrm{ms}$$

The maximum timing quantization uncertainty is:

$$\Delta t_{quant} = \pm\frac{T_s}{2} = \pm 0.5\,\mathrm{ms}$$

**Oscillator Drift**

The crystal oscillator stability is ±20 ppm.
Over a 10-second recording:

$$\Delta t_{drift} = 10\,\mathrm{s} \times 20 \times 10^{-6}$$

$$\Delta t_{drift} = 0.2\,\mathrm{ms}$$

Because all channels share the same clock, drift mainly affects absolute time but is included for conservative estimation.

**Inter-Channel Skew**

ADC channel switching delay is approximately:

$$\Delta t_{skew} \approx 10\,\mu\mathrm{s} = 0.01\,\mathrm{ms}$$

This contribution is negligible.

**Total Timing Uncertainty**

$$\Delta t_{total} = 0.5\,\mathrm{ms} + 0.2\,\mathrm{ms} + 0.01\,\mathrm{ms} \approx 0.71\,\mathrm{ms}$$

# 3 Costs

## 3.1 Labor Cost

For the project if we take the standard rate for an ECE graduate student of $40 an hour. We work for roughly 2.5 hours a day for the duration of this course, which is 60 days and we have $6000 per teammate therefore total labor costs are $18000. Now to account for the parts for the board.

## 3.2 Part List

| Description | Manufacturer | Quantity | Cost | Link |
|---|---|---|---|---|
| BIOFY Optical Sensor (SFH 7050A) | ams-OSRAM | 2 | $4.42 | Link |
| Biometric AFE-PPG (MAX86140) | Analog Devices | 2 | $18.48 | Link |
| Biopotential AFE-ECG (MAX30003CTI+) | Analog Devices | 1 | $14.80 | Link |
| Bluetooth 5.0 MCU (nRF52840-QFAA) | Nordic Semiconductor ASA | 3 | $5.27 | Link |
| DC to DC converter (NTHD3100CT1G) | Onsemi | 3 | $1.56 | Link |
| Linear Voltage Regulator (TLV73318PDBVT) | Texas Instruments | 3 | $0.83 | Link |
| Power management (NPM1100-QDAA-R7) | Nordic Semiconductor ASA | 3 | $1.05 | Link |
| Linear Voltage Regulator (TLV76033DBZR) | Texas Instruments | 3 | $0.44 | Link |
| Type c connector (USB4125-GF-A) | GCT | 3 | $0.59 | Link |
| P-Channel MOSFET (SI2301CDS-T1-GE3) | Vishay Siliconix | 3 | $0.69 | Link |
| Crystal 1 for nRF (ECS-320-10-37B2-CKY-TR) | ECS Inc. | 3 | $0.35 | Link |

| | | | | |
|---|---|---|---|---|
| Oscillator for MAX30003 (ASAKMPD1-32.768KHZ-T3) | Abracon LLC | 1 | $3.09 | Link |
| Crystal 2 for nRF (CM9V-T1A-32.768KHZ-9PF-20PPM-TA-QC) | Micro Crystal AG | 3 | $0.52 | Link |
| RF Antenna (AMCA72-2R470G-S1F-T4) | Abracon LLC | 3 | $0.54 | Link |
| Other parts: (capacitors, resistors, inductors…) | | | neglect | |
| Earlobe Clips/Leads | 3D Printed | 4 | $16.00 | Link |
| Total Cost: | | | $18115.21 | |

# 4 Conclusion

## 4.1 Accomplishments

The ECG subsystem represents the most successful component of the project. The MAX30003-based ECG board was fully designed, fabricated, and verified. The board reliably acquires cardiac electrical signals from chest electrodes, digitizes them and streams the resulting waveform wirelessly over BLE to a companion laptop application. The web-based ECG monitor identifies and displays real-time R-peaks, reports heart rate in beats per minute, and maintains a stable BLE connection throughout extended recording sessions. The NRF52840 microcontroller was successfully programmed under the Zephyr RTOS framework on both the ECG and PPG boards. BLE connectivity, SPI communication with the MAX86141, and LED drive control were all confirmed to be functional on the PPG board with the LEDs illuminating on command and the microcontroller correctly reads and writes MAX86141 registers over SPI. This confirms that the digital and wireless infrastructure of the PPG system is sound and that the failure is isolated to the optical signal path rather than the control architecture.

The redesign from three independent boards to a single unified PPG headband board was also a meaningful engineering accomplishment. The original three board design was abandoned after identifying a fundamental BLE synchronization limitation: because BLE cannot broadcast to multiple devices simultaneously, a time-sync signal sent from a central host would arrive at each board at different times, introducing variable inter-board latency that would corrupt PTT measurements. By consolidating both PPG sensors onto a single PCB controlled by one NRF52840 and one crystal oscillator, the redesign eliminated this source of error entirely. Both earlobe sensors are now guaranteed to be sampled from the same clock domain with no inter-board jitter.

We also identified and partially resolved a significant noise problem during the intermediate headband design phase. The original snaked-wire configuration used to improve wearability introduced severe EMF-induced noise on the analog photodiode signal due to the long distance between the photodiode and the MAX86141 ADC input. This manifested as high-amplitude spikes in the PPG data that made pulse detection impossible, even though the photodiode could respond to a direct flashlight stimulus. The solution we adopted in the final design was to move the MAX86141 driver chip physically adjacent to the SFH-7050A sensor module at each earlobe, minimizing the length of the sensitive analog signal path.

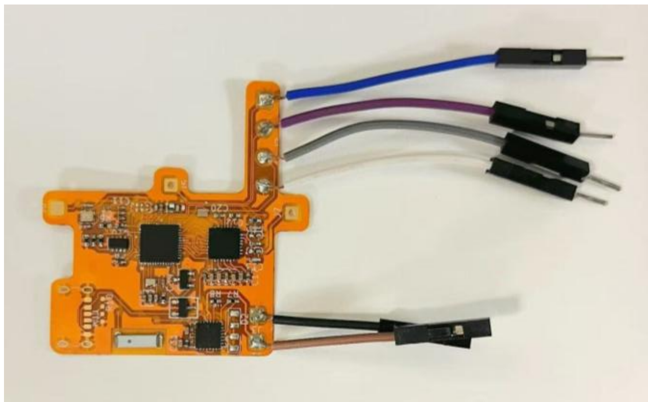

**Figure 6. A picture of our entire ECG PCB that works.**

## 4.2 Uncertainties

The PPG subsystem was unable to produce valid physiological data by the project deadline. All sampled values returned from the MAX86141 FIFO read as zero, and this condition persisted after replacing the photodiode, checking all solder joints and probing accessible board nodes. Because BLE, the NRF52840, SPI communication, and LED drive were each confirmed to operate correctly in isolation, the root cause is believed to lie either within the MAX86141 chip itself or within the PCB layout in the immediate vicinity of the optical front-end.

One leading hypothesis is a fault in the MAX86141 internal analog front-end, possibly caused by an electrostatic discharge event during handling or assembly. The MAX86141 is a highly sensitive optical AFE with no exposed ESD protection structures on its photodiode input pins, making it vulnerable to damage from ungrounded handling. A damaged AFE would explain why register communication succeeds and LEDs respond correctly.

A second hypothesis concerns the PCB layout around the photodiode input trace. As documented in the MAX86141 datasheet and application notes, the PD_A and PD_B pins must be routed with a continuous ground guard ring surrounding the trace from the sensor pad to the IC input. If this guard is incomplete or if the solder mask quality left any exposed copper near this trace, leakage currents or parasitic capacitance could be suppressing the photodiode signal below the ADC's detection threshold. The poor solder mask registration observed on the fabricated PPG boards makes this a credible failure mode, as it increased the risk of unintended shorts and made it difficult to inspect the critical optical input path visually.

A third uncertainty concerns whether the MAX86141 configuration register sequence used in firmware correctly initializes the device for reflectance PPG mode with the SFH-7050A. While the SPI communication itself was verified, it remains possible that a register write is incorrect for instance, setting the wrong LED channel, leaving the ambient light cancellation in a state that zeros the output or not correctly enabling the FIFO output and that the resulting all-zeros output is a firmware configuration artifact rather than a hardware fault. Without test points directly on the MAX86141 photodiode input pins, it has not been possible to conclusively distinguish between these failure modes using the current board revision.

## 4.3 Ethics

In developing the dual-earlobe PPG system, we will follow the IEEE Code of Ethics by ensuring honesty and realism in reporting performance and limitations. Since the device is intended for research use, we must accurately present measurement accuracy, synchronization error, and noise limitations without exaggeration. We will clearly state that the system is a research prototype and not a medical diagnostic device to prevent misuse.

Because the system collects physiological data, privacy protection is also an ethical responsibility. We will anonymize collected data, limit storage access, and use secure Bluetooth transmission to reduce the risk of data breaches.

## 4.4 Future work

The highest priority for future work is a subsequent revision resolving the PPG optical front-end failure. The first step would be to replace the MAX86141 ICs and rework the existing boards under ESD-controlled conditions, replacing the suspect chip and re-testing FIFO output immediately. If replacement chips yield the same all-zeros result, attention should shift to the PCB layout, confirming that the photodiode input trace is fully enclosed by a ground guard ring on all layers and that no solder bridges or mask voids exist near the PD pins. Adding exposed test points on the PD_A, PD_B, and INT pins of the MAX86141 in the next board revision would allow an oscilloscope to directly observe the photodiode photocurrent and interrupt behavior during operation, which would make future debugging significantly faster and more conclusive.

The second priority is improving the PCB manufacturing quality of the PPG headband. The solder mask registration on the fabricated boards was poor, resulting in exposed copper in areas adjacent to fine-pitch components and creating elevated risk of solder bridging on the MAX86141's 0.4mm pitch pads. A revised manufacturing specification should include a tighter solder mask expansion rule, a minimum mask web width of 75 µm between adjacent pads, and an electrical continuity test as a fabrication acceptance criterion.

Once valid PPG waveforms are confirmed, the remaining software work is the PTT computation pipeline in the laptop or phone application. The BLE packet structure already carries timestamps from both the

ECG and PPG boards. The outstanding implementation tasks are peak detection on the PPG waveform followed by the bilateral PTT computation as the difference between each PPG peak timestamp and the preceding ECG R-peak timestamp. Left and right PTT values should be displayed in real time alongside their difference, which is the primary measurement output of the system.

Finally, once PTT data is available, the experimental validation of measuring bilateral PTT under neutral posture, left head tilt, right head tilt, and supine positions should be conducted across multiple subjects to determine whether statistically significant and consistent bilateral differences are observable. This would be the first experimental result demonstrating the scientific value of the measurement platform and would provide the foundation for any future research application of the device.

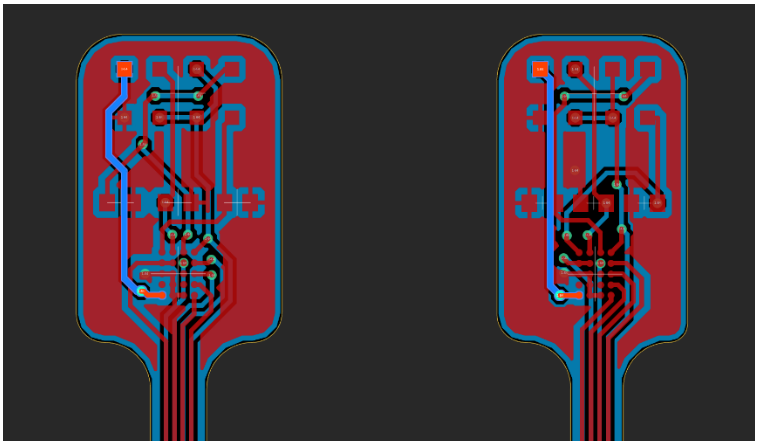

**Figure 7.A detailed picture of our trace outlines on our board for our latest PPG iteration.**

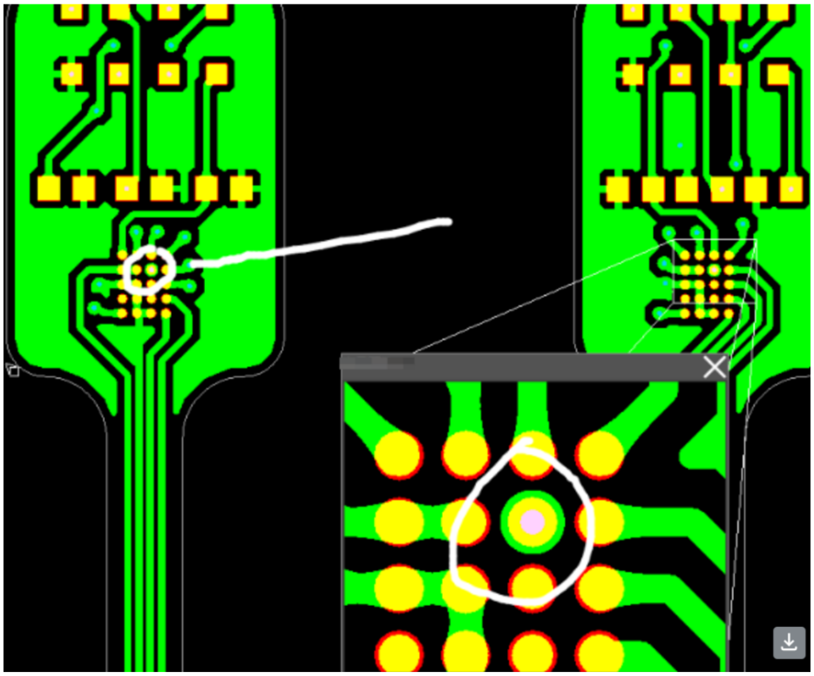

**Figure 8.You can see that because the MAX chip is a wafer package it was very difficult to solder onto the board.**

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## 6 Requirements and Verifications Table

| Requirements | Verifications |
|---|---|
| 1. Provide a regulated 1.8V ± 0.2V DC rail to all subsystems from a 3.7V LiPo battery. | 1. Measure the LDO output using a DMM under peak load. Verify voltage stays within 1.6V – 2V. |
| 2. Maintain voltage ripple below 10 mV on the MAX30003 AVDD pin to ensure high ECG signal-to-noise ratio. | 2. Use an oscilloscope to measure ripple at the decoupling capacitor closest to Pin 6. |
| 3. Limit total system current draw to < 50mA during active bilateral sensing to ensure >4 hours of operation on a 200mAh battery. | 3. Connect a DMM in series with the battery and measure the average current while the device is in Active Sensing Mode. |

| | |
|---|---|
| 4. Successfully trigger a hardware interrupt on the nRF52840 within 5 ms of an anatomical R-peak occurrence. | 4. Probe the raw ECG signal (Ch1) and the INT2B pin (Ch2) on an oscilloscope. Measure the time delta between the peak and the logic low trigger. This requirement is also met if good ECG data is read |
| 5. Distinguish R-peaks from T-waves and baseline noise on the ECG waveform. | 5. Record 5-10 seconds of data. R peaks are clearly higher than T waves which are clearly higher than baseline noise. |
| 6. Proper SPI communications between microcontroller and MAX86140 chips | 6. Able to read/write to MAX86140 registers. Either show logs of read registers or ability to turn on the LED using SPI. |
| 7. Sensors are reading proper PPG data that is not overly noisy | 7. Clearly shown plots of data that have PPG peaks clearly distinguishable from noise. |
| 8. Bluetooth Low Energy (BLE) works properly and can transmit and receive data | 8. Ability to operate the device over BLE: led turns on, or sensor data can be read. |
| 9. Sustain a stable Bluetooth Low Energy connection at a distance of 3 meters with zero packet loss. | 9. Move the receiver 3m away. Run the program and ensure that waveform data is still being communicated and plotted. |
| 10. Ability to charge battery using USB connection | 10. LED that indicates charging lights up. Measure battery voltage with Digital Multimeter while charging and see the battery voltage increase |