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AN ULTRA-LOW-COST PHOTOACOUSTIC SYSTEM

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by
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ABSTRACT

In preclinical cancer biology studies, the combination of ultrasound and optical imaging, known as photoacoustic imaging (PAI) is an emerging approach for real-time molecular imaging in preclinical research and has recently expanded into the clinical environment. It combines the advantages of ultrasound’s high penetration depth and optic’s rich molecular contrast. Although traditional photoacoustic models have shown to perform more optimally with higher penetration depth, and near real-time imaging capability, there has only been a limited success with fabricating a fully functional miniaturized PA system, whilst such systems exist with other modalities. While two-photon and other optical microscopy systems have recently emerged in portable and wearable form, there is much less work reported on portable and wearable photoacoustic systems. To address this issue, we have made the following contributions through this work:

- Creating a Microcontroller based Data Acquisition System, which replaces expensive digitizer boards
- Fabricating a compact photoacoustic probe that consists of custom-made PMN-PT ring ultrasound transducer and a fiber-coupled laser diode
- Constructing a trigger and delay circuit to drive the laser diode
- Encapsulating the entire system in a custom-made portable casing
- Performing experiments on tissue-mimicking phantoms and successfully sensing temperature change as a primary evaluation
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Chapter 1

Introduction

Technology has enhanced, streamlined and automated multitudes of systems and processes. Examples of such technology include portable automated patient health monitoring systems. They provide easy access to patient data and hence provide the ability to deliver higher quality care at lower costs. They help patients to get better access to healthcare, by providing improved quality of care, peace of mind and improved support. Although modern miniaturized and portable imaging and sensing systems exist for the ultrasound and optical modalities, similar work on photoacoustic imaging needs to be explored as it would provide information on much better resolution, precision, and contrast. This work reports the development of a low-cost and miniaturized photoacoustic sensing and imaging device.

Background and Technologies Used

In this section, the basic concepts involved in this project are discussed. This section is divided into the following parts.

1. Biomedical imaging modalities
2. Microcontrollers and their architectures (ARM Cortex – M)
3. Analog to Digital Converters
4. Data Communication Techniques

The discussion below will also help in analyzing the drawbacks of other modalities of biomedical imaging and sensing and clarifying the motivation for our work in the next section.
Biomedical Imaging Modalities

Biomedical imaging is a modality of scientific imaging which deals with creating a visual representation of internal tissue structures, to provide additional information for clinical analysis and medical intervention. The making of a biomedical imaging device involves multiple layers of research, design development, and integration of hardware and software. Modern-day biomedical devices have also been made portable and wearable, to provide doctors a way to remotely assess organ conditions and biochemistry of cells for early detection of diseases.

Since the inception of biomedical imaging systems in 1895 – the X-Ray – this field has been developing exponentially and revolutionizing healthcare. The developments though always came with some adverse side effects. For instance, over-exposure to X-Rays could cause mutations in DNA and therefore lead to cancer later in life. These side effects have of course triggered the development of substituent systems that employed exploiting light, sound or both.

Throughout the twentieth century there have been numerous inventions in biomedical sensing and imaging systems – MRI, ultrasound and optical imaging and Positron Emission Tomography (PET) – to name a few. Optical imaging systems, which have been in existence for a while, provided a way to image in-vivo, without exposition to ionizing radiation, making it safer than X-Ray and CT-Scan systems. Ultrasound imaging has increased the penetration depth and resolution of images produced. Recent inventions suggest combining individual modalities of imaging – light, and sound – to produce images of better resolution and penetration depth. They are termed as a photoacoustic imaging system, which combines the advantages of optical and ultrasound systems.

With the widespread developments in technology, wearable healthcare devices would allow patients to be more connected than ever to healthcare. Soon would come a day where there
won’t be a need to visit doctors. The immediacy of personalized healthcare could revolutionize the availability of quality medical care worldwide.

Keeping this in mind this work focusses on the development of an ultra-cost-effective portable photoacoustic device that can be used for biomedical in-vivo sensing and imaging. In the next few sections, modalities of imaging such as optical and ultrasound have been introduced, to explain the motivation for our work.

*Optical Imaging*

Medical Optical Imaging is a non-invasive medical imaging technique that uses visible light (non-ionizing) to obtain detailed images of organs and tissues as well as smaller structures including cells and even molecules.

![Figure 1.1: Schematic of optical imaging system. [5]](image)
There are multiple types of medical optical imaging - optical microscopy, spectroscopy, endoscopy, scanning laser ophthalmoscopy, and optical coherence tomography – to name a few. It works on the principle of excitation of endogenous or exogenous chromophores within a volume of interest by an external source of light. With absorption, photons hitting a chromophore disappear and release all their energy to molecular electrons. This absorption occurs only at unique molecular-specific values of frequency (energy). The absorbed energy may then be released as a delayed emission of light, the so-called autofluorescence, and as non-radiating heat to surroundings. Images are constructed from the information received by the light-capturing cameras, complex algorithms are devised and used to form the reconstructed images.

**Ultrasound Imaging**

Ultrasound Imaging is a modality of biomedical imaging which uses high-frequency sound waves to form high-resolution images of internal structures of the human body. It is also called sonography.

![Block diagram of the optical microscope system](image-url)

**Figure 1.2:** Block diagram of the optical microscope system [6]
The procedure to acquire images uses a transducer and a gel placed directly on the skin. Ultrasound waves that emanate from the transducer are collected by it after they bounce back from internal structures. These waves are then sent to data acquisition tools such as computers to create images. Since real-time images are captured, internal structure and movement of internal organs can be shown in real-time.

Photoacoustic Imaging

Photoacoustic imaging is a new and upcoming modality of imaging, which operates on the principle of photoacoustics. The photoacoustic effect can be simply put as the vibrational effect induced in the matter by light. It was first discovered by Alexander Graham Bell over a hundred years ago, when he had the concept of telephone, to encode a sound signal in a light beam and propagate the light beam in space and convert light back into sound again by the photoacoustic effect. Although Bell was interested in the audible frequency range, we are more interested in the ultrasound frequency range.

In this paragraph, an attempt has been made to give an elaborate description of the photoacoustic effect. When incident light is absorbed by a substance, it generates a temperature rise due to the thermoelastic expansion. When pulsed light is used for irradiation on material, rapid thermoelastic expansion and contraction are caused, which in turn results in pressure waves generation. These pressure waves can also be understood as sound wave emission. These sound waves are used for imaging of biomedical tissues. An effort is made to generate sound in the ultrasound range so that the wavelength of the sound is shortened, enabling objects to be imaged
with better spatial resolution. This improves the ability to define the structures and see the tumors better.

Another emerging modality of photoacoustic imaging is photoacoustic tomography. Photoacoustic Tomography (PAT) is based on the acoustic detection of optical absorption from either endogenous chromophore, such as oxy-hemoglobin and deoxy-hemoglobin, or exogenous contrast agents, such as organic dyes and nanoparticles. Because ultrasound scatters much less than light in tissue, PAT generates high-resolution images in both the optical ballistic and diffusive regimes.

There are two types of photoacoustic imaging systems – photoacoustic/thermoacoustic tomography and photoacoustic microscopy. A typical PAT system uses an unfocussed ultrasound detector to acquire the photoacoustic signals, and images are reconstructed by inversely solving the photoacoustic equations. A photoacoustic microscope system, on the other hand, uses a spherically focused ultrasound detector with 2D point-to-point scanning. It requires no reconstruction algorithm.

In this work, an effort was made to make an ultra-portable and low-cost photoacoustic microscope. The imaging depth of this technique is limited mostly by the factor of ultrasound attenuation.

According to [2], the amplitude change of a decaying ultrasound wave can be expressed as:

\[ A = A_0 \exp(-\alpha z) \]

In this expression, \( A_0 \) is the unattenuated amplitude of the propagating wave at some location. The amplitude \( A \) is the reduced amplitude after the wave has traveled a distance \( z \) from
that initial location. The quantity ‘$\alpha$’ is the attenuation coefficient of the wave traveling in the z-direction. The dimensions of are nepers/length, where a neper is a dimensionless quantity.

Attenuation is generally proportional to the square of sound frequency. Quoted values of attenuation are often given for a single frequency, or an attenuation value averaged over many frequencies may be given. Also, the actual value of the attenuation coefficient for a given material is highly dependent on the way the material was manufactured. Therefore, quoted values of attenuation only give a rough indication of the attenuation and should not be taken to be accurate. Generally, a reliable value of attenuation can only be obtained by determining the attenuation experimentally for the material being used. Attenuation can be determined by evaluating the multiple back wall reflections seen in a typical A-scan display like the one shown in the image at the top of the page. The number of decibels between two adjacent signals is measured and this value is divided by the time interval between them. This calculation produces an attenuation coefficient in decibels per unit time $U_t$. This value can be converted to nepers/length by the following equation.

$$\alpha = \frac{0.0051}{v} U_t$$

where ‘$v$’ is the velocity of sound in meters per second and $U_t$ is in decibels per second.

The spatial (i.e. axial and lateral) resolutions depend on the ultrasonic transducer used. An ultrasonic transducer with high central frequency and broader bandwidth are chosen to obtain high axial resolution. The lateral resolution is determined by the focal diameter of the transducer. For instance, a 50 MHz ultrasonic transducer provides 15 micrometers axial and 45-micrometer lateral resolution with ~3 mm imaging depth.
Spatial resolution

According to [1], Spatial resolution is defined as the measure of the smallest object that can be resolved by the sensor, or the ground area imaged for the instantaneous field of view (IFOV) of the sensor, or the linear dimension on the ground represented by each pixel.

Microcontrollers and their architectures

A microcontroller is an integrated circuit designed to perform a specific operation in a system it is embedded in. A typical microcontroller includes a processor, memory and input/output (I/O) peripherals on a single chip. In this project we have used a microcontroller development kit
by NXP, the LPC Link – 2, which consists of the LPC 4370 microcontroller and three different processors - one main ARM Cortex M4 processor and two ARM Cortex M0 processors.

ARM is fundamentally a family of RISC (Reduced Instruction Set Computing) architectures. Cortex Microcontrollers belong to the ARM Cortex microprocessor architecture. A microcontroller unit can be visualized as a small computer that integrates a processor unit, memory, and peripherals inside the same integrated circuit and is specifically designed for embedded applications.

Figure 1-4 shows the internal structure/schematic of microcontrollers. As shown, a microcontroller comprises a microprocessor with a set of additional peripherals and memory banks, which are all interconnected by data buses. These buses provide a way for the MPU to interact with the external world and the memory bank registers.

Another important aspect of MCU architecture is memory. Program memory, or the memory that is responsible if program code allocation can either be located on the same bus as the rest of the peripherals or the data memory or have its own separate bus. Microcontrollers usually
include a variety of memory banks which include Read-Only Memory (ROM), Random Access Memory (RAM) and Electrically Erasable and Programmable Read-Only Memory (EEPROM). Some MCUs also use flash memory for program code memory allocation. This can be located inside the IC (built-in flash) or outside (flashless).

**ARM Cortex – M Microcontrollers**

As mentioned, ARM is a family of RISC (Reduced Instruction Set Computing) architectures for computer processors, and they hold a variety of designs that are configured for different environments. These processors are based on a 32-bit instruction set. Due to lesser power consumption and more basic RISC command protocols, they are better than CISC (Complex Instruction Set Computing) based devices, especially for small and less power-driven devices. ARM architecture microcontrollers have their own special bus variant, called the AMBA (Advanced Microcontroller Bus Architecture) High-Performance Busses (AHB). The MCU is designed in such a way that each peripheral is associated with its own register, which can be mapped as its reference addresses to be used to access the microprocessor through the AHB. Thus, the AHB makes it provide access to the MPU to interact with the external world and with the memory registers. The general description of this MCU is as follows: The LPC 4370 is an ARM Cortex-M4 based microcontroller for embedded applications. It includes an ARM Cortex-M0 coprocessor and an ARM Cortex-M0 subsystem for managing peripherals, 282 KB of SRAM, advanced configurable peripherals such as the State Configurable Timer (SCTimer/PWM) and the Serial General Purpose I/O (SGPIO) interface, two high-speed USB controllers, Ethernet, LCD, an external memory controller, and multiple digital and analog peripherals including a high-speed 12-
bit ADC. The LPC4370 operates at CPU frequencies of up to 204 MHz. The ARM Cortex-M4 is a 32-bit core that offers system enhancements such as low power consumption, enhanced debug features and a high level of support block integration. The ARM Cortex-M4 CPU incorporates a 3-stage pipeline, uses a Harvard architecture with separate local instruction and data buses as well as a third bus for peripherals, and includes an internal prefetch unit that supports speculative branching. The ARM Cortex-M4 supports single-cycle digital signal processing and SIMD instructions. A hardware floating-point processor is integrated into the core. The LPC4370 includes an application ARM Cortex-M0 coprocessor and a second ARM Cortex-M0 subsystem for managing the SGPIO and SPI peripherals. The ARM Cortex-M0 core is an energy-efficient and easy-to-use 32-bit core which is code- and tool-compatible with the Cortex-M4 core. Both Cortex-M0 cores offer up to 204 MHz performance with a simple instruction set and reduced code size. The Cortex-M0 coprocessor hardware multiply is implemented as a 32-cycle iterative multiplier.

[Product Description of LPC4370]

Motivation

While optical microscopes have gone through a revolution in terms of their portability and their ability to be carried in wearable forms, only minimal work has been reported for photoacoustic imaging in this direction. A low-cost portable PA system would open a wide range of applications in the field of medical sensing and imaging. It would enable easy point-of-care diagnosis applications such as cancer and tumor detection and region of care parameter sensing. This would prove to be extremely beneficial especially in resource-deficient regions on account of its easier availability and low cost.
Already existing table-top PA systems are capable of sensing and imaging biological tissues. Although PA sensing has been gaining interest within the academic and industrial groups, the state-of-the-art systems are still bulky, expensive and not sustainably powered. The reason can be the use of bulky high energy lasers and data acquisition (DAQ) systems. In spite of the fact that the PA system miniaturization has been investigated by multiple groups including work on making more portable probes [X. L. Deán-Ben et al. (2013), P. K. Upputuri et al. (2015), H. M. Heres et al. (2017), K. Park et al. (2018), S. Liu et al. (2019)], making a small footprint PA Microscope [Gao, F. et al. (2015)], a micro-ultrasound PA system [A. Needles et al. (2013)], a fingertip laser-driven PA system [H. Zhong et al. (2019)], and a minimally invasive PA system [D. Piras et al. (2013)], they are all still very expensive (> $5000), bulky (few centimeters to tens of centimeters), and not sustainably powered.

**Figure 1-5:** The system-level photograph of the high-frequency portable photoacoustic sensing system. PB: power bank; MCU: microcontroller unit; RPi: Raspberry Pi; PreA: Preamplifier; CM: communication module; LD: laser driver; PG: pulse generator; UT: ultrasound transducer; MMF: multimode fiber;
Fig. 1-5 shows a system-level architecture of the system. The dimensions of the system are 15 x 10 x 9.5 cm³, which also includes a photoacoustic probe of diameter 6 mm. The PA probe is composed of an in-house fabricated ring ultrasound transducer of 13.2 MHz center frequency, and a 905 nm multimode fiber (Laser Components Inc. 905D2S3J09FP-40/22-F-0-01) with a core diameter of 400 µm and 0.22 numerical aperture driven by a laser driver. The laser driver is driven by a lab-made pulse generator, which generates pulses of width 40 ns whenever it receives a trigger from the microcontroller unit (MCU). A microcontroller unit (MCU) then converts the generated PA signal to digital data with the help of its high-speed (80 mega samples per second) analog to digital converter. The data generated is then displayed on a connected monitor and is stored for analysis. The system is sustainably powered by a 20,000 mAh battery power bank, which was tested to run the system at full acquisition rate for over 13 hours. The total cost of construction of the system is US$1,000, much lower than any other portable PA system.

Figure 2-6: Transrectal Ultrasound and Photoacoustic Device [14]
The system is further planned to be validated by experimentally measuring temperature change. Results from the experiment will show the feasibility of real-time region of interest (ROI) temperature monitoring for applications such as thermotherapy and cryotherapy.

The aim of this project is to set a new standard for a portable PA system, which is realized by miniaturizing the individual blocks contributing to the bulkiness of such a system. This thesis reports the results obtained from the efforts to reduce the bulkiness by using smart, compact and low-cost components.
Chapter 2

Instrumentation

Description of the system

Fig. 2 describes the event flow of the system. The power control unit provides power for the individual components of the system. It uses a 20000 mAh power bank to power up to two DC-DC voltage boosters or regulators to convert the 5V output from power bank to 7.5V and 15V for the pulse generator and the voltage amplifier respectively. The microcontroller unit (MCU) acts as the master and controls the entire system by sending triggers to pulse the laser. After driving the pulse generator (which generates pulses to drive the laser driver), it also converts the analog PA signals to digital data using its Analog to Digital converter (ADC).

**Figure 2-1:** The event flow of the system. It is consisted of six units – power control, microcontroller, laser control, front end, pre-processing, and post-processing unit.

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The trigger-acquisition process is explained as follows. First, the MCU sends triggers to Arduino. The Arduino is used to provide a 160 µs delay for acquisition. A custom-made pulse generator regulates the pulse to have a pulse width of not more than 40 ns, which is further used to trigger the laser driver. The laser driver drives the laser diode to rapidly illuminate the tissue, causing a photoacoustics phenomenon that generates ultrasound signals from the tissues. The generated signals are then captured by the in-house fabricated PMN-PT ultrasound transducer. Next, the acquired photoacoustic ultrasound signals are first sent to a pre-processing unit for amplification and offset, and then to the MCU’s high-speed ADC where it is converted to digital data and is averaged. Finally, the digital signal is sent to the post-processing unit (Raspberry Pi) to collect and display the data. A detailed discussion about the power control unit, laser control unit, front end unit, pre-processing unit, microcontroller unit, and post-processing unit have been included in the subsections below.
1. The Microcontroller Data Acquisition/Synchronization Unit

A commercial MCU (NXP Semiconductors LPC4370) is used for the process of data acquisition and laser triggering. The criteria considered for the proper selection of the microcontroller are explained below.

a. **Clock frequency and instruction pipelining:** The main requirement of our system is to make an ultra-compact PA sensing device that can process and give out information at a rapid rate (>1000 Hz). This means a need for an MCU with processors running at frequencies greater than 150MHz. The chosen MCU has 3 processors (one main ARM M4, one co ARM M0, and one ARM M0 sub-processor), all capable of running at 204MHz.
Pipelining is a technique used to execute multiple instructions at the same clock. To do this, each instruction is divided into stages so that each stage from different instructions are executed at the same clock, improving the total throughput. A non-pipelined MPU might take ‘X’ ms to complete an operation, but if pipelining of 4 stages is done, effective time is reduced to $\frac{1}{4}$th the time it first took, thus increasing performance by 3 times.

b. **Memory organization:** There are two main types of memory architecture, namely, Von Neumann and Harvard. Although both architectures are used for designing MCUs, it is extremely important to know which one is being used by each microcontroller in order to optimize code execution. Although Von Neumann architecture is used in MCUs, Harvard architecture is preferred for embedded solutions.

The main difference between these architectures is that the Von Neumann memory architecture shares the same bus for data and program memory, whereas, the Harvard architecture is based on separate memories for program and data, with their own data busses. The Von Neumann architecture is used when the program code is to be run from the RAM, which in turn slows down the processor. The Harvard architecture is usually used for program memory on a ROM memory.

To preserve the potential for the future development of an ultra-low-cost portable PA imaging system, it is very important to keep the style of memory architecture modular. This processor can be used to run on both Von Neumann and Harvard processor architectures. The M4 processor has a Harvard memory architecture and the M0 implements Von Neumann, making it a suitable choice for the system.
Peripheral requirements are the single most important decision in selecting an MCU. For developing our system, three main requirements of peripherals were – an ADC with a very high sampling rate, reliable general-purpose input-output pins, and fast communication ports. This MCU turns out to satisfy all the needs - it has an 80 Mega Sample per Second (MSPS) high-speed ADC, stable GPIOs and supports the USB communication protocol.
Analog-to-Digital Converter

Post-processing and analysis can be done only in the digital domain. When a digital system, such as a microcontroller needs to process physical voltage (analog) signals, these must be digitized and transformed into digital data. Digitizing an analog signal is done by the process of sampling the input analog signal and discretizing the samples by the process of quantization. Figure 2.5 gives a schematic of how an analog signal looks like, pre and post digitization. The device that converts analog information into digital data is called an Analog to Digital Converter (ADC).

Figure 2.7: Digitizing Analog information to Digital information

An ADC, as explained above, is a device that takes analog voltage signals as its input and gives digitized data as its output. The output digitized voltage value approximates the analog input voltage, with respect to a reference voltage. The process of conversion is divided into three stages:

a. **Sampling:** Discretization of time for the given input signal is called sampling. The sampling frequency must be constant for maintaining sync and coherence; Sampling is achieved by using a sample and hold circuit. After the signal has been time sampled, the next step is to quantize the signal.
b. **Quantization:** Once sampled in time, input voltage levels are quantized. This is done because of the limitation in the resolution of digital devices. Each input analog signal can take only a limited output value. A sample input-output graph has been shown in figure 2.6.

Some important things to know about Analog to Digital Converters are:

1. The number of bits used – resolution of ADC: There are ADCs of resolutions such as 8, 12, 16 bit and so on. To understand resolutions better, let’s consider a 4-bit resolution. If we plot the input analog v. output digital data, as in figure 2.5, for minimum and maximum inputs of 0 and 5 volts, the output would be 0000 and 1111 respectively. Thus, there are 16 values that can be represented between 0 and 5 volts, which further means that the resolution of the ADC becomes (5/16 = .3125 volts). For LPC 4370, the ADC is 12 bits, and the voltage can be between 0.1 and 0.9 volts, and the resolution becomes 195 micro-volts.
2. Types of ADCs:
   a. Single Slope ADC
   b. Dual Slope ADC
   c. Successive Approximation ADC
   d. Flash ADC

Algorithm of synchronization and conversion

The following steps describe the algorithm followed by the microcontroller to digitize, acquire and communicate data. The designed system can be broadly divided into four levels: initialization, triggering, acquisition, and conversion and communication. These stages are discussed here:

1. Initialization: The board, chip, and general-purpose input-output interface components are initialized.

2. Triggering: Functions and directions (input or output) of pins are set using the pin multiplexer - one pin is used to drive the laser by providing triggers, and the other is used as the ADC input.

3. Acquisition: Clock is generated for the ADC, and sampling and quantization take place; the digitized values are stored in the Static RAM (SRAM) of the MCU.

4. Conversion and Communication: The digitized values are bit manipulated and sent to the cloud with a Raspberry Pi, using USB communication. Future modules will have a WiFi module to replace the Raspberry Pi, and post-processing manipulation will be made on the cloud.
Given below are a flowchart and a timing diagram of the algorithm followed by the MCU to drive the system and acquire the digitized data, for post-processing it and displaying it on the monitor.

The steps mentioned above are described in detail below.

Initially, the board components such as the system core clock generator, chip general-purpose inputs outputs (GPIO), USB communication interface, pin multiplexer, Internal RC Oscillator (IRC) clock generator, USB0 Phase-Locked Loop and clock dividers are configured and initialized.
Next, a trigger from the MCU turns the laser on, through the custom-made pulse circuit, generating a 40 ns pulse width of light of wavelength 905nm as described in the optics unit, and is shone onto the tissue-mimicking phantom. Due to the PA effect, there are ultrasound signals that emanate from the phantom. The photoacoustic ultrasound signals generated from the tissue phantom are received by a custom fabricated transducer at 13.2 MHz center frequency. The sampling rate - 80 MSPS satisfies the Nyquist criterion which says [The Nyquist criterion states that a bandlimited continuous-time signal can be faithfully sampled and reconstructed from its samples if the sampling rate is over twice as fast as its highest frequency component].

To sample the generated photoacoustic signal at 80 MSPS, the ADC needs to be driven at 80MHz. For this requirement, we use a component called PLLUSB0 as its clock driver: a PLL (Phase Locked Loop) is a system that generates an output signal that is related to the input in a linear way. Hence one obvious application of a PLL is synthesizing various phase-related frequencies from a known frequency.

The PLLUSB0 is a dedicated PLL for the USB0 High-speed controller. It accepts input frequencies from external oscillators in the range of 14KHz to 25MHz, which are multiplied up to higher frequencies using a current-controlled oscillator (CCO) that further gives output frequencies in the range of 4.3MHz to 550MHz. The IRC oscillator, which runs at a nominal frequency of 12MHz was given as an input to PLL0. The CCO multiplies this clock and gets an output clock of 480MHz, which is further brought down to 80MHz using the MCU’s dividers. The hence produced clock of 80MHz thus enables it to sample at 80 MSPS.

The acquired photoacoustic signal is provided into the MCU’s ADC. With this MCU, a high-speed ADC is used for this purpose. This is a flash type 12-bit ADC with a series array of
4096 comparators that process the input voltage and convert them into valid digital data. Although the digital data generated is made of 12 bits, it is stored in a 32-bit array. The USB protocol can only send 8-bits per transaction in its bulk transaction mode. The data is bit-manipulated in such a way that only the required data is sent.

2. The Power Control Unit

The power control unit (PCU) has a 20,000 mAh power bank that acts as the power source that consistently pumps out a constant supply of 5 Volts and a sum of 5 Amperes from two of its terminals to the components of the system. The PCU also consists of two DC-DC voltage boost converters (XL6009) to satiate the voltage needs of 7.5V, and 15V of the pulse generator and the laser driver respectively. A voltage divider circuit is used to generate 0.3 volts from the 5-volt input. It is used as an offset voltage for the ADC to convert the acquired and amplified PA signal into the MCU’s acquirable range.

![Figure 2.10](image_url)

Figure 2.10: (a) DC-DC converter, (b) The power distribution flow.
3. The Front-End Unit

The front end of a system refers to the integrated components that are responsible for the generation of the PA signals. It is made up of two units:

1. The laser diode
2. The transducer

Pulsed laser light is used to irradiate and excite the material on which the photoacoustic waves are supposed to be generated. This system uses a 905 nm pulsed laser diode (905D2S3J09FP-40/22-F-0-01, Laser Components Inc) with 65 W peak optical power as the optical source. The laser diode is coupled with a multimode optical fiber of 400 µm core diameter with a 0.22 mm numerical aperture. This fiber is then fed through the center hole of the in-house fabricated Holley PMN-PT ultrasound transducer. A 110 µm thick rectangular plate of lead magnesium niobate-lead titanate (PMN-PT) (TRS technologies, State College, PA) has been chosen as the piezoelectric material for the ultrasound transducer. PMN-PT has been shown to exhibit high electro-mechanical coupling and high strain energy density, making it ideal for photoacoustic imaging. The ring transducer fabrication is described in the predecessor paper from our group [3]. The ring transducer has an outer diameter of 2.5 mm and an inner diameter of 0.5 mm.

![Diagram of the Photoacoustic Transducer]

Figure 2.11: (a) The Photoacoustic Transducer. (b) Picture of the photoacoustic transducer
4. The Pre-processing Unit

Fig 2.12: (a) Signal flow diagram of the pre-processing unit.

Fig 2.12: (b) A BPF with $F_{ch} = 5$MHz and $F_{cl} = 20$ MHz, (c) Pre-Amp VCA 2615

The pre-processing unit is made up of three components:

1. A 63dB gain pre-amplifier
2. A voltage bias circuit
3. A Band Pass Filter

The stages of the pre-processing unit are used to pre-process the received PA signals to make them more usable for post-processing, using the MCU and further, the Raspberry Pi. The preamplification circuit provides a 63dB gain for reliably detecting and strengthening the signal. Further, to get the photoacoustic signal in the convertible range of the microcontroller’s ADC (100
to 900 mV), the received PA signal is amplified and centered. This is accomplished by a custom-made amplifier breakout board that uses Texas Instrument’s VCA2615 pre-amplifier and a minicircuits offset circuit. The offset circuit uses the 0.3 Volts generated by the voltage divider circuit from the power control unit, while the pre-amplifier provides 63 dB gain for the produced PA signal, which contains the working voltage range of the developed system. The added high-frequency and low-frequency noises are canceled with the help of a bandpass filter (BPF), whose circuit has been shown in Figure 2.12 (b).

5. **The Post-processing Unit**

![Signal flow diagram of the post-processing unit.](Fig 2.13)

The post-processing unit consists of a Raspberry Pi 4 module and a graphical user interface (GUI). After the process of digitization, the data was communicated to the cloud using a Raspberry Pi 4 module. A python script was written to convert the received bits into usable voltage decimal data. Further, a user interface is created to continuously plot the averaged received data, implementing it as a miniaturized system-specific oscilloscope. The signal will then be used manipulated for different applications as needed.
6. **The Laser Control Unit**

The laser control unit is made up of three different components:

1. A custom pulse generator
2. A commercial laser driver
3. A delay circuit

The MCU generates triggers that are used to drive a custom pulse generator, which in turn drives the commercial laser driver. The process flow is explained as follows. A trigger sent from the MCU is delayed for 160 µs for the operation requirement of the MCU data acquisition. This is

![Diagram](image)

**Fig 2.14:** (a) The Laser Control Unit, (b) The pulse driver circuit, (c) The pulse driver PCB, (d) The laser driver.
due to the delay in switching the MCU’s high-speed analog to digital converters, each time after getting triggered. The trigger generated by the Arduino after the delay was found to have a pulse width which was much higher than the required pulse duration – about 40 ns. Hence a customized pulse generator is constructed. The circuit diagram for this pulse generator is given in Figure 3. This allows the capacitor bank to discharge and drive the laser diode through the MOSFET of the laser diode. The laser diode is driven by a matchbox-sized capacitive discharge circuit which charges a bank of capacitors using the output of a high-volt (up to 80 V) on-board boost convertor which is triggered by the MCU. During the capacitor-discharge, a pulse of ~30 Amperes current passes through the laser diode generating an optical pulse with ~2.5 µJ energy.
Chapter 3

Results

System Performance

1. Back End Performance Evaluation

Evaluation of the back-end system is done in two different ways:

   a. System power consumption.

   b. ADC characterization.

a. System power consumption evaluation and runtime calculation:

   System power consumption evaluation is done by calculating the amount of power required to acquire each a-line and the reliable signal acquisition frequency range. The power consumption details of all units (as in the system diagram - Fig. 2) are summarized in Table 1. The system has a total power consumption of 7.5 Watts (1.5 Amperes), indicating that a 20,000 mAh power bank would be able to run for 800 minutes (>13 hours). The a-line rate of the system is more than 1000 a-lines per second, where the number of samples/a-line is 128.
System Runtime Calculations:

The calculations used for determining the runtime are as follows:

Voltage output of battery : 5V
Rated Ampere capacity of battery : 20,000 mAh
Average Discharge Current : 1.5 A
Energy Consumption : Voltage output × Rated Ampere capacity
                        = 5 × 20,000
                        = 100-Watt Hours.
Runtime : Energy Consumption ÷ Discharge Current
          = 100 ÷ 1.5
          = 13.33 hours.

Thus, the system when on its active acquisition mode, would work for more than 13 hours with a 20,000 mAh power bank.
**b. ADC characterization:**

The Analog to Digital Converter (ADC) is characterized by the process of frequency sweeping. Frequency sweeping is the process of sending a wave at multiple frequencies and capturing its peak-to-peak amplitude response. In this case, frequency sweeping is done by sending sine wave signals at different frequencies, and later capturing the peak to peak amplitude response. It should be noted that the outputs are captured after 64 averages. The function generator sends sine waves with 300 mV amplitude with frequencies range from 1 MHz to 20 MHz with 1 MHz increment and the peak to peak amplitude captured by ADC and stored in Raspberry Pi. The plotted frequency sweep results show that the system can capture 83.3% of the amplitude response at 20MHz.

![Frequency Sweep Graph](image)

Figure 3.1: System characterization by sending pulses with different frequencies and observing the peak to peak voltage change. The graph indicates that the system can pick up > 80% of the peak amplitude at ~20 MHz.
2. Front End Performance Evaluation

The front end is made up of the in-lab fabricated ring transducer. The fabricated transducer is characterized by the pulse-echo response technique. This was done by placing the transducer at 1.5 cm from a solid block of aluminum, which was placed in a water bath, and to get the ultrasound pulse-echo response from the incident ultrasound pulse. A commercial pulse-receiver (5073 pulse-receiver, Olympus Inc.) was used to provide a voltage pulse input with 2 µJ energy. The received pulse with 0 dB gain shows ~80 mV peak-to-peak voltage output. The frequency response of the received pulse shows that the ring ultrasound transducer has a center frequency of 13.12 MHz with a fractional bandwidth of ~30% at -6 dB loss (Fig. 3.2).

The lateral imaging resolution is found by sensing two white lines (low-absorbing targets) separated by a small width from each other, placed on a black-card phantom. This is shown in Fig.
3.3 (a). The transition between the black and white regions form an edge in light absorption. Photoacoustic sensing of the phantom was performed over the line shown with the dotted orange line in Fig. 3.3 (b).

The peak-to-peak photoacoustic signal amplitude along the 30 mm scan line in Fig. 3.3 (b), which clearly shows the four transitions from black to white and white to black respectively. The distance corresponding to the regions and their peak-to-peak photoacoustic signal amplitude response gave an estimated spatial resolution of ~300 µm. To summarize, the center frequency of the fabricated transducer is 13.2 MHz with a fractional bandwidth of ~30% at -6dB. The lateral resolution of the transducer is tested to be ~300 µm.

**Photoacoustic Result**

The working of the ultra-portable low-cost photoacoustic sensing device is to be tested by measuring the temperature change of a given material. The working of the device was first tested experimentally. The temperature sensing experiment is to be conducted. The process of photoacoustic signal acquisition and temperature change sensing is tested in multiple stages:
a. A signal from a function generator

b. Ultrasound from pulse-receiver

c. PA signal on an oscilloscope from black-card/blood-tube

d. Temperature sensing

### a. A signal from function generator:

The first working prototype that was constructed had only the high-speed ADC. The input was given to the MCU from a function generator, and the ADC produced digitized values from a UART terminal. This system was not very reliable, and the high speed of the ADC was hindered by the low communication throughput of the UART communication protocol. The next stage of development involved the integration of the SPI communication protocol, which was faster than the UART. After the SPI communication replaced the UART, the communication proved to be much faster, but the data that was received by the Raspberry Pi was erroneous. Hence, it was decided that there should be USB communication established. The USB protocol is a complex algorithm, heavily modular, and very fast. After integration of the USB communication, the data flowing out of the USB protocol were 8-bit integers, whereas the data produced by the high-speed acquisition module were 12-bit. This meant there had to be bit manipulation so that the outputs coming from the USB communication were usable at the Raspberry Pi end. The manipulations were done for successful USB data communication.

### b. Ultrasound from pulse-receiver

To mimic the system, a function generator was used to generate a burst signal, whenever triggered by the Microcontroller Unit. Once triggered, the signal from the function generator was sent to an in-lab fabricated ultrasound transducer and was transmitted through a water bath to another similar transducer. This experiment was made to test the working of the system. An amplifier was used to recover the received signal and was further plotted on the oscilloscope. The system setup and oscilloscope results are shown below in Fig. 3.4.
After testing the working of the system with ultrasound transducers, the laser driver was integrated with the system. It should be noted that the system is running at a trigger-rate of a little more than 1000 Hertz (1KHz). The laser driver is successfully set to work on external pulses, and this is done by a custom-built pulsing circuit, (see Fig. 3), generates trigger pulses of the required pulse width, to drive and hence generate laser pulses. The laser is tested to work at a rate of about 1 kHz and the PA signal is captured first from a black card on an oscilloscope.

The photoacoustic signals were amplified using a VGA2716 (commercial variable gain amplifier) and were acquired using the microcontroller DAQ system results were recorded on
the Raspberry Pi. The oscilloscope was used as the ground truth to compare the signal transmission quality to which from the MCU system. The next step was to power the entire system with the power bank, and this was done successfully. Figure 3.5 shows the plot obtained on the system. The circle in black shows how an averaged (64 sample averaging) PA signal looks like, from a black card. This proves to be a validation for the working of our device.

![Averaged PA Signal](image)

Figure 3.5: The photoacoustic signal seen on the WPA device.

d. Temperature Sensing Experiment

When the laser excitation is much shorter than the pressure propagation, then the fractional volume expansion \( \frac{dV}{V} \) a heated region is given by:

\[
\frac{dV}{V} = -\kappa \rho + \beta T \tag{1}
\]

where \( \kappa \) is the isothermal compressibility, \( \beta \) is the thermal coefficient of volume expansion, and \( p \) and \( T \) denote changes in pressure (measured in Pascal), and temperature (in Kelvin), respectively.

Since the fractional volume change is minimal, given the rapidity of laser excitation, the local pressure rise is given as:
The Gruenesian parameter ($\Gamma$) is given as:

$$\Gamma = \frac{\beta}{\kappa \rho C_v} = \frac{\beta v_s^2}{C_p} = f(T)$$  \hspace{1cm} (3)

where $v_s^2$ is the velocity of sound, $C_p$ denotes the specific heat capacity at constant pressure, and $T$ is the temperature of the object. Therefore,

$$p_0 = f(T) \eta_{Th} A_e$$ \hspace{1cm} (4)

Since the pressure is a direct function of the temperature, it is physically possible to measure temperature with photoacoustics.

The experimental setup of the temperature sensing photoacoustic, as of now is under construction and scrutiny. After testing the validity of the device with a blood capillary tube, the temperature change using the data acquisition and plotting system will be measured. The temperature in the blood bath will be changed by heating the blood and then letting it cool down. As in equation (2), since the pressure generated is a direct function of the temperature of the object, the amplitude of the signal is in direct variation to the temperature of the object. The received photoacoustic signal, whose amplitude is directly proportional to the object temperature as seen in equation (2), is proposed to be used to measure the temperature of the object.
Chapter 4  
Discussions and Conclusion  

In this paper, the designing of a microcontroller-based ultra-low-cost and portable photoacoustic sensing system has been designed. The validation of the system is to be done soon by a temperature sensing experiment. The entire PA sensing system measures 15 x 10 x 9.5 cm³ in volume, with an ultrasound probe of diameter 2.5 mm. Both the PA system and the probe account for more than a two-fold decrement in size, compared to the current state of the art system. The full cost of the system construction costs a total of US$1000 (Table 3).

In conclusion, this is the first PA system that utilizes the microcontroller unit for data acquisition. Although a Raspberry Pi has been used for displaying the data, a WiFi module can be integrated into the system to make the entire system wireless.

<table>
<thead>
<tr>
<th>Component</th>
<th>Cost ($)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MCU</td>
<td>25</td>
</tr>
<tr>
<td>Raspberry Pi</td>
<td>50</td>
</tr>
<tr>
<td>Pre-Amp</td>
<td>150</td>
</tr>
<tr>
<td>Laser Driver</td>
<td>200</td>
</tr>
<tr>
<td>Laser Diode</td>
<td>400</td>
</tr>
<tr>
<td>Arduino</td>
<td>20</td>
</tr>
<tr>
<td>Transducer</td>
<td>30</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td><strong>~ $1000</strong></td>
</tr>
</tbody>
</table>

As described in Fig. 4.1, the acquired amplitude of the high-frequency components decreased to 83% of the original amplitude, indicating the system may not generate enough SNR
for higher frequency transducers. New research shows a successful integration of the GRIN lens to increase the fluence of the laser, therefore increasing the absorption and hence, the amplitude of the PA signal generated. This may overcome the amplitude drop due to under-sampling. The current a-line rate of more than 1 kilohertz will be improved by tweaking the system-level design, such as making the MCU unit act as the slave of the laser drive unit to bring down the overhead on the MCU.

Besides the capability of temperature sensing, the system has the potential to be integrated with multiple laser diodes of different wavelengths, making it capable of indicating blood glucose levels. The developed system is planned to be integrated with two micro-scale motors (X-Y stages) - Tiny Ultrasonic Linear Actuators (TULA), which enables the raster scanning so it becomes an ultra-compact and ultra-low-cost PA imaging system. This would be a generate strong interest in point-of-care and wearable PA applications.

A significant amount of research is currently in progress in miniaturizing medical devices and making them more wearable and portable. The device described in this thesis submission would be exceedingly useful in providing a point of care medical care with accurate sensing capability. This device is ultra-portable and low cost, and it is less than 2.5 lbs. in weight and $15 \times 10 \times 9.5$ cm$^3$ in volume. The entire system works on a battery, further enabling a wireless portable PA system. This would be very beneficial as such devices can be easily used in resource-poor environments.
### TABLE III

Comparison of the proposed system with the state-of-the-art systems.

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>PRICE</strong></td>
<td>HIGH</td>
<td>LOW</td>
<td>HIGH</td>
<td>ULTRA-LOW</td>
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<tr>
<td><strong>SYSTEM SPECS</strong></td>
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<tr>
<td>CENTER FREQUENCY</td>
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<td>1 MHz</td>
<td>13.2 MHz</td>
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<td>SAMPLING FREQUENCY</td>
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<td>5 MHz</td>
<td>80 MHz</td>
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<tr>
<td>PENETRATION DEPTH</td>
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<td>1 MM</td>
<td>2 MM</td>
<td>*</td>
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<tr>
<td><strong>PORTABILITY</strong></td>
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<tr>
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<td>×</td>
<td>✓</td>
</tr>
<tr>
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<td>×</td>
<td>✓</td>
<td>✓</td>
</tr>
<tr>
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<td></td>
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<td></td>
</tr>
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<td>NR</td>
<td>650 g</td>
<td>900 g</td>
</tr>
<tr>
<td>SYS. VOLUME</td>
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<td>14×8×3 cm³</td>
<td>17×14×10 cm³</td>
<td>15×10×9.5 cm³</td>
</tr>
<tr>
<td>PROBE VOLUME</td>
<td>24×25×80 mm³</td>
<td>NR</td>
<td>NR</td>
<td>2.5×2.5×10 mm³</td>
</tr>
</tbody>
</table>

* - This will soon be replaced with a value.
In future, with the integration of stages, and more ADC channels, we plan on making a portable melanoma detection and monitoring device that would revolutionize point of care medical systems, by providing data that could be used in intravascular imaging of such tissues. Other plans include integration of the device with a lab-built deep-learning model that would help in identifying tissues with no prior information. This would revolutionize the portable medical sensing and imaging in the field of photoacoustics.
References


[3] Towards a Low-Cost and Portable Photoacoustic Microscope for Point-of-Care and Wearable Applications Ajay Dangi, Member, IEEE, Sumit Agrawal, Gaurav Datta, Visweshwar Srinivasan, Sri-Rajasekhar Kothapalli


