Electronics Design for a Portable, Disposable Ultrasound Guidance System for Intracranial Pressure-Relieving Surgical Procedures

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1. INTRODUCTION

Catheterization is a common procedure used during external ventricular drainage (EVD), ventriculoperitoneal shunt (VPS) placement, as well as a variety of additional procedures, such as central venus catheter (CVC) placement, hemodialysis, epidural anesthesia and lumbar puncture (Toma, 2009). However, the vast majority of catheterizations are guided using only external anatomical landmarks for guidance, resulting in shockingly high error rates. Up to 60% of VP shunts are placed incorrectly and up to 19% of CVC procedures result in placement related complications (Kakaria, 2008). Complications include arterial puncture, hematoma, pneumothorax, and infection leading to increased lengths of stay, additional procedure time and expense, and even death (Bucholz, 1995). Although catheter guidance systems exist, they are complex and time consuming making them unsuitable for ER and ICU settings where many catheterizations occur.

Dr. Mourad's lab requires a disposable and sterile piece of electronic equipment that could be attached to the guide-wire of the catheter. This wire would contain a small 10 MHz transducer (1 mm wide) at the end of the wire, within the catheter. The electronic device would power the transducer with a short pulse, and process the ultrasound information received. The lab expects that upon the insertion of the transducer into the ventricle, there would be large increase in acoustic impedance, from that of the brain, to the 'fluid sac' of the ventricle. This increase would be identified by the electronic device, and would visually signal the surgeon that the catheter is both headed in the right direction as well has reached its destination. The device would be required to run for no more than 10-30 minutes, and would be powered by a lithium ion battery. The aim of this project is to develop the electronics and software for a miniature, disposable ultrasound system that could be used to improve the precision of catheter guidance in high intracranial pressure surgical procedures.

Initially a design was made for the system in its nearly miniaturized form, and through the course of the year, it was revised, in response to challenges and new design requirements. A connection diagram was made to show all necessary components for the miniaturized device. A display format and power solution was chosen for the size, weight and power specifications. Finally, a bill of materials was created and updated throughout the process as well, which reflected not only the total cost of building a single prototype, but also the projected cost per device on an estimated production rate of 10k/year.

The system would be built in two stages: first, an initial prototype, which would be operated by power supplies and be approximately the size of a shoebox. The display for the first prototype would be in the form of a Matlab GUI that would be updated to show 1) the signal, and 2) the physical interpretation of the signal (distances from the catheter tip to the ventricle, as well as other landmarks (ex. the skull). This initial prototype could be used in animal studies (pig brain studies) to validate the claim that the device can identify when the transducer is aimed at the ventricle, and the distance to the ventricle. For simplicity, I'll be using the title VentriGuide to refer to the ultrasound guidance prototype, and EchoGuide Medical as the company name used at the Business Plan Competition.

2. REQUIREMENTS SPECIFICATION

A. Miniaturized Device Requirements

The VentriGuide device sends a high voltage pulse stream to the transducer, and analyzes the A-mode (1D) ultrasound transducer response for distance to ventricle information, which is evident from ventricle wall and tissue pulsitility information in the response signal. This information is displayed in a visual format on a small LCD screen located at the top of the device, which displays distance information to the neurosurgeon. In the third prototype iteration, a multi-element transducer could be used to indicate the direction the surgeon has to move the transducer to get line of sight with the ventricle (shown with arrows around the circle in Figure 5). In the first miniature prototype, the device would indicate placement by indicating when the ventricle/brain signal-to-noise ratio had crossed an experimentally-determined threshold; this would be displayed by the intensity of the donut on the LCD screen and the final blinking of the circle to indicate that the line of sight had been found.

The miniaturized device CAD drawing (created by a mechanical engineer at the Mourad lab) is below, and shows the eventual multi-element user interface (Figure 5). The blue area of the device represents the LCD screen, the green area is the electronics, and the dark gray area represents the batteries. The dimensions of this model were based off of my initial electronics design for the miniaturized device and each of the components' dimensions.

To use the device, the surgeon holds the device along the ergonomic handle of the device, and looks straight down the flat upper surface of the device to the target, while keeping the LCD screen in view. In this singular transducer element device, the surgeon would move the device on the surface of the brain, inside the burr hole in the skull that had been created, until the SNR threshold is crossed and the circle begins to blink.
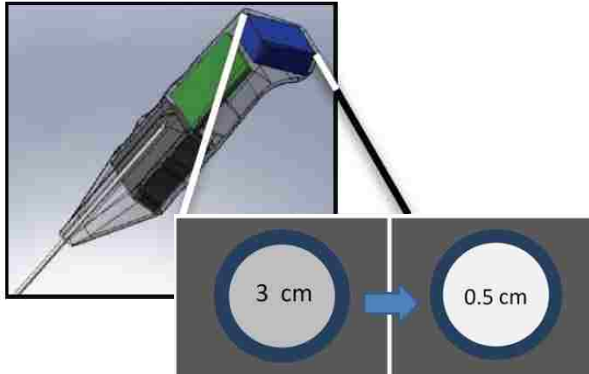

Figure 1: VentriGuide Drawing

B. *First Portable Prototype Requirements*

In the initial prototype, the requirements are to display the ultrasound signal in real time onto a laptop screen with analyzed distance information both available for debugging purposes. Also, the initial prototype has a "shoebox" size constraint with no hard weight limit, compared to the handheld, low-weight requirement of the miniaturized device. In the design process, components that could make fast prototyping possible (with evaluation boards) were favored, but each component was also evaluated and selected for its compatibility with the final device's size, weight and power requirements.

Design Requirements for VentriGuide First and Final Prototype

Design Requirement	First Prototype	Final Prototype
Technology/Data Capture	Ultrasound (10 MHz), data capture hardware part of design requirement	Same
Battery Life	N/A (power supply/wall outlet)	10-30 minutes
Battery Type	N/A	Primary (Disposable)
Size (First Prototype)	Shoebox size	Final dimensions (Figure 11)
Voltages	High (200 Vpp)	May be up to 400 Vpp
Weight (First Prototype)	No hard requirement, but needs to be portable	No hard requirement, but cannot significantly detract surgeon from performing the procedure
Sampling Rate	> 50 million samples/second (at least 5 samples/period)	Same
Type of Data	Analog from transducer	Same
Data Storage (First Prototype)	None	None
Data Display Medium	Laptop	Miniature LCD (1 in x 1 in)
Type of Data Displayed	Real-time ultrasound response	Graphical display for navigation

	waveform for debugging	(Figure 5)
Data Processing	Signal processing in hardware (bandpass filtering)	Signal processing in hardware (bandpass filtering) as well as software (IQ-demodulation, additional filtering)
On/Off Switch	'Enable' button on ultrasound pulser board	LCD pushbutton initiates the device

3. DESIGN SPECIFICATION

The portable prototype connection diagram acted as a 'living document' throughout the design process and was continuously updated based on new information and troubleshooting. The diagram below is from March 2012, and has been included because the first pcb (for the parts that weren't on evaluation boards) was designed based off of the connection diagram below (Figure 6). The function of each of the evaluation board components are explained in further detail below:

Ultrasound Pulser

The Supertex MD1822DB3 board is an ultrasound pulser board which has a variety of pre-programmed options for number of pulses in a group, and signal frequency. The input voltages to the board determine the output voltages, the board outputs each group of pulses at 5 kHz. The board also has the capability to output up to 200 Vpp. In the initial prototype, the pulser was set to 8 pulse cycles/group, 10 MHz (to match the 10 MHz resonance frequency of the transducer), and output swing voltages of +100 V and –100 V (the maximum capacity of the board).

TX/RX Chip

The TX/RX chip allows for the travel of both the ultrasound pulser output as well as the transducer response within the same channel, by ensuring that the high voltage output of the pulser doesn't reach any of the other circuitry except for the transducer (those high voltages would damage any of the other components). The TX/RX chip only allows voltages below a 2 V threshold to pass through; in this voltage range, the ultrasound response can pass through, and the remaining circuitry is protected as well.

ADA4817 Board (Used to Design Bandpass Filter)

The ADA4817 is a low noise, high gain-bandwidth-product op-amp – this allows for high amplification of the 10 MHz ultrasound response signal. An Analog Devices PCB board designed for this chip was also used, since stray capacitance at the inputs could cause suboptimal performance of the board. A 2nd order bandpass filter design that uses a single op-amp was picked, since it would require fewer chips and would decrease the overall cost of goods. Since the board that was used with the chip wasn't perfectly suited for the placement of components in this manner, Analog Devices recommended 'sky-wiring' the remaining components, which means that thin wire would be used to connect components that didn't have traces available in the board, and that in cases where open solder pads were not available, the components could be soldered to the edge of solder pads that were being used by other components. Due to the time constraints at the time, and the level of magnification and experience required to do the PCB soldering and sky-wiring, a contracted engineer pieced this component together.

AD8138 (Differential Amplifier)

The differential amplifier wasn't used to further amplify the signal, but to level shift the signal from being centered around 0 V (ac-coupled) to be centered around 1.5 V, which is well in the input voltage range for the A/D converter. This way, the entire signal could fit in the 1 Vpp input range of the ADC.

A/D Converter

The ADC is 8-bit parallel output, 80 MSPS (million samples per second), which can satisfactorily fulfill Nyquist requirements for a 10 MHz ultrasound signal. A parallel output ADC was picked because the alternative would be to 1) get a deserialization circuit that could convert a serial output ADC to data that the microcontroller could read, or 2) be restricted to FPGAs that had a specific port that could read serialized data.

Microcontroller

The microcontroller reads the output pins of the ADC (which would be tied to the microcontroller's input pins), fills a buffer, and sends the buffer over UART to MATLAB on a laptop. In the final prototype case, all signal processing would be performed on the microcontroller, and the microcontroller would display an image on the small LCD button screen.

LCD PushButton (NKK Switches) – Miniaturized Prototype

The LCD programmable pushbuttons by NKK Switches have some of the smallest LCD screens in the world. A pushbutton was chosen because the pushbutton could be used to initiate the device, acting as an 'on' switch when the surgeon is ready to begin. The device can be communicated with using SPI protocol, where the entire image to be displayed is sent to the screen to refresh the output. An example of the NKK pushbutton switch display is shown below:

Figure 2: NKK Programmable LCD Pushbutton (62 X 32 pixels) (Taken from NKK Switches Website)

Ultra Thin Primary Lithium Cells (Powerstream) – Miniaturized Prototype

Another product that came up in researching components for the miniature device was the ultra-thin, primary lithium battery, provided by Powerstream. These lithium batteries are rated at 3.0 V, and are 2.9 cm (width) x 2.9 cm (length) x 0.45 mm (thickness). This makes them some of the smallest lithium batteries available for their level of current availability. Powerstream also offers rechargeable cell options in slightly larger sizes, in case the device design changes to partially-disposable (vs disposable).

Figure 3: Ultrathin Lithium Ion Batteries, less than 1 mm thick (Taken from Powerstream Website)

All of the initial components for the first prototype were displayed in a connection diagram (Figure 8). The PCB designed to house the components that did not come with evaluation boards is shown in Figure 9.

A connection diagram was used instead of a traditional schematic mainly because of the time constraints and the intended use of the document, which was to obtain quotes from various prototyping companies that could create the

PCB as well as box up the prototype. The contracted company mentioned earlier (to solder the bandpass filter) was also contracted to solder the small components of designed PCB board as well.
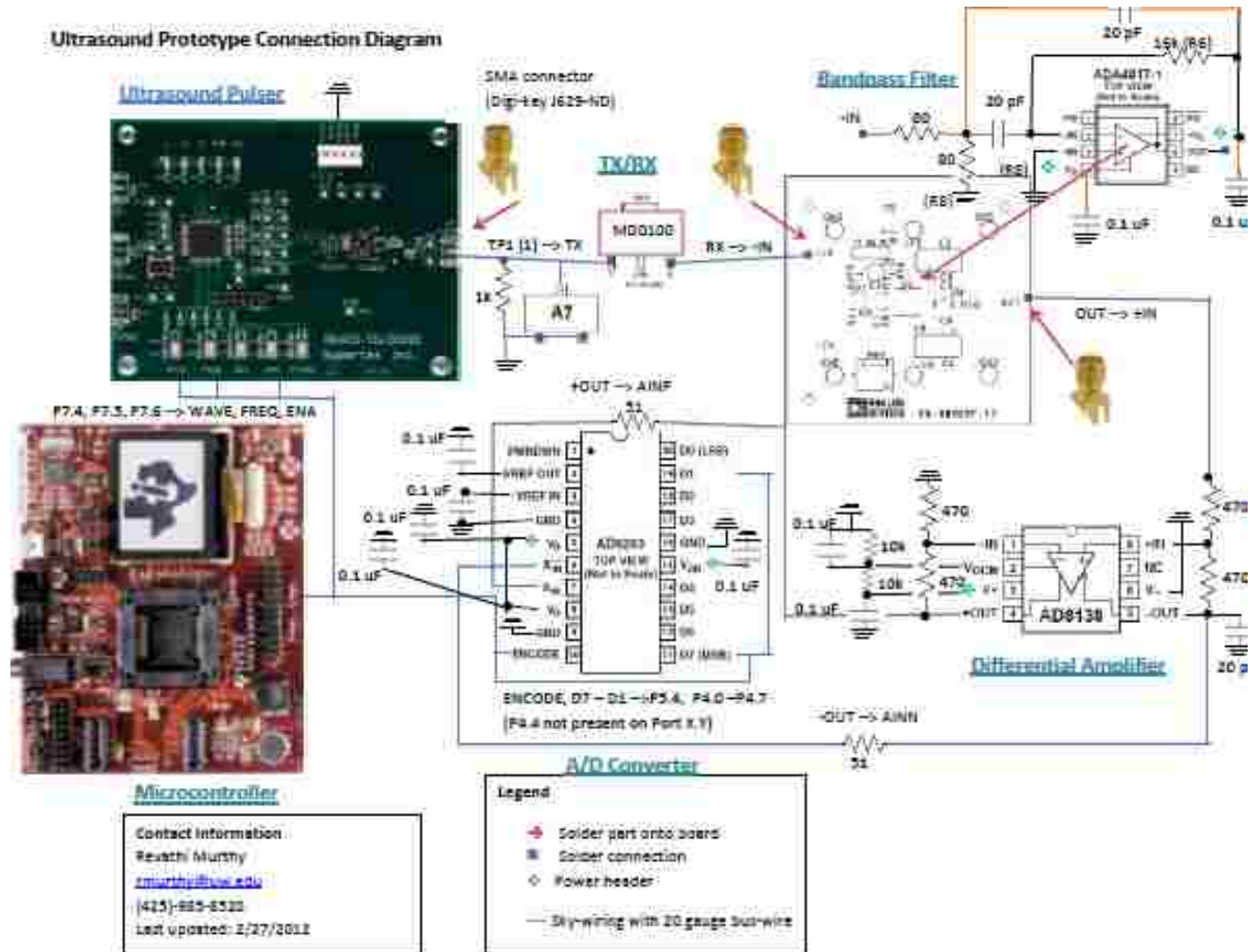

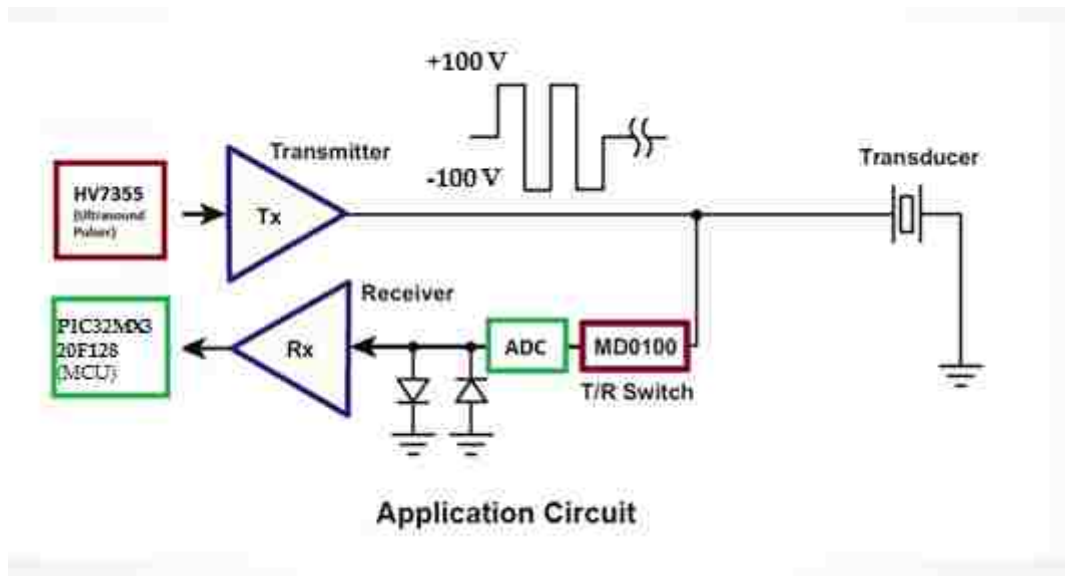**Figure 4: Prototype Diagram March 2012**
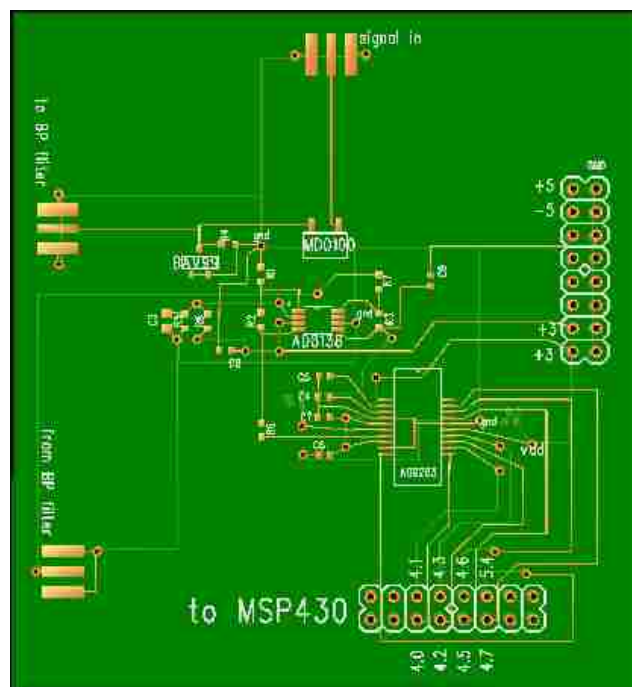
Figure 5: Final Application Circuit

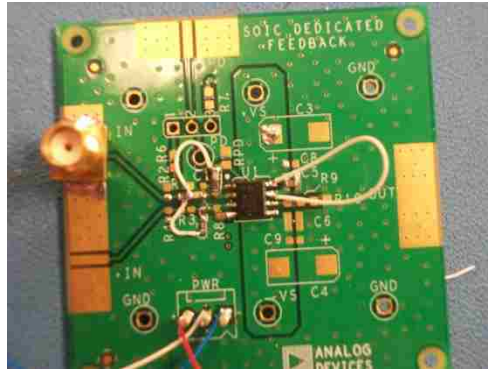Figure 6: PCB Design for Remaining Components
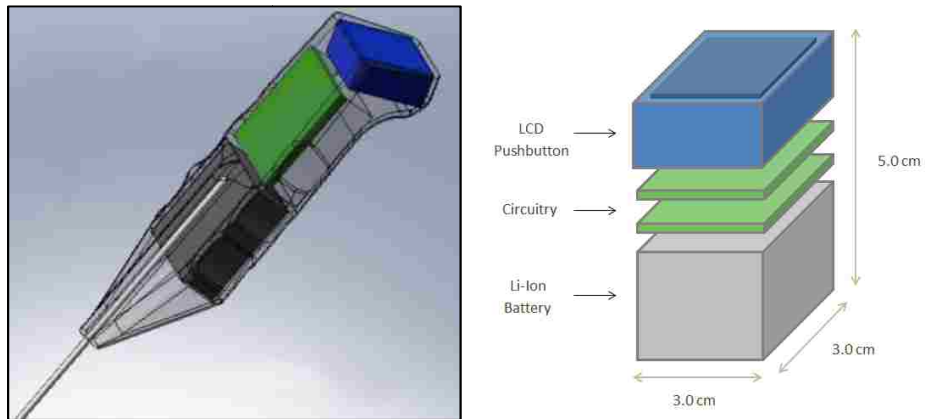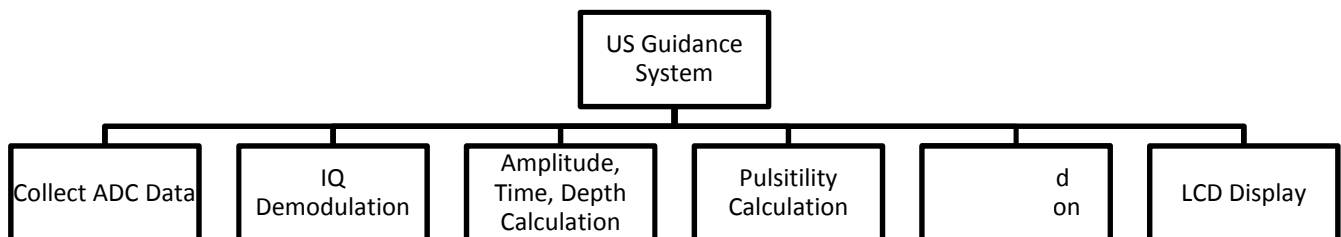
Figure 7: Bandpass Filter Design

Figure 8: Mechanical Diagram (Left), Simplified Diagram (Right)

4. SOFTWARE DESIGN

Although the software was not implemented in the device, an overall design approach was created for future incorporation. The software design approach is described below by a functional decomposition, and a state diagram:

Function	Parameters	Output	Description	Hardware Involved
Collect ADC Data	None	Ultrasound voltage data (unsigned int	Interprets ADC digital input and represents it in terms of the	MCU input pins tied to ADC outputs

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|		VoltageData[ ])	voltage range of the input signal	
IQ Demodulation	unsigned int VoltageData[ ]	unsigned int USData[ ]	Removes frequencies aside from the fundamental frequency of the transducer	None
Amplitude, Time, Depth Calculation	unsigned int USData[ ]	VentriDepth[ ] VentriAmp[ ] VentriTime[ ]	Calculates ventricular amplitude and depth from time information derived from the peaks in the USData[] buffer	None
Pulsitility Calculation	unsigned int USData[ ]	BrainPuls[ ]	Calculates pulsitility from response time-shift information derived from the USData[] buffer	None
Threshold Comparison	VentriDepth[ ] VentriAmp[ ] VentriTime[ ] BrainPuls[ ]	unsigned int ThresholdComp	Calculates whether ventricular amplitude/depth/time/pulsitility measurements are consistent over time, and fit threshold for ventricular placement	None
LCD Display	ThresholdComp	boolean YesCathTime	Converts ThresholdComp to a saturation level for the inner circle on the display. If the input ThresholdComp exceeds an additional threshold, the inner circle is highlighted strongly and made to blink.	LCD
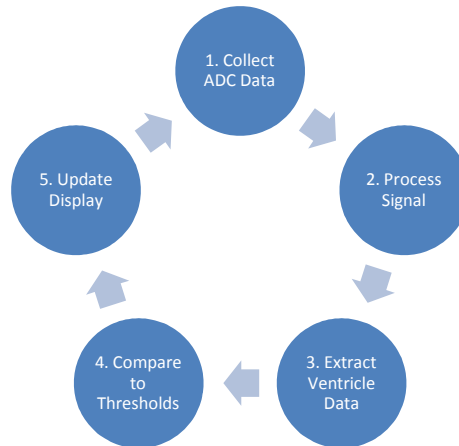
Figure 9: State Diagram for Software Flow

5. ALGORITHMS

Distance Calculation

The distance algorithm is simple; the response from the transducer is identified by evaluating if it crosses an amplitude threshold. The time value corresponding to the signal crossing the threshold is converted into a distance value, based off the speed of sound in brain tissue of 1540 m/s. Since this distance is based off of the round trip travel time between the transducer output and the received response, the final distance is divided by two to calculate the distance from the transducer tip to the ventricle.

IQ-Demodulation Calculation

IQ-demodulation can help extract pertinent response data from other frequencies, in addition to the ultrasound band pass filter. IQ-demodulation involves three steps: down-mixing, low-pass filtering and decimation.

Down-mixing involves multiplying the real signal with a complex sinusoid, which moves the frequency spectrum of the signal down to be centered around f-demod.

$$x_{IQ}(t)=x_{RF}(t) \cdot exp(-i2\pi f_{Demod} \cdot t)$$

Low-pass filtering using a Hamming window is performed to remove frequencies and noise outside the desired bandwidth of the final signal. Finally, decimation involves finding the smallest integer fraction of of 20 that's greater than twice the filter cut-off frequency; this integer indicates the nth sample that should be retained in a final buffer to reconstruct the signal from the frequency domain. The reconstruction is performed through interpolation, zero-padding, low-pass filtering, and up-mixing (the reverse of the IQ-demodulation steps).

Ventricle Identification

To identify the ventricle, an array of amplitude values is compared to a threshold. The amount of values in the array which cross the threshold determines the intensity of the circle on the LCD display. If the number of values crosses an additional threshold, the angle of the transducer is optimal for guidance, and the circle on the LCD is made to blink.

6. TEST CASES

Initial test cases included:

- Verify output functionality of the ultrasound pulser.
- Verify ultrasound response visible on the output of the TX/RX chip.
- Verify the function of the bandpass filter using the function generator as a noise input.
- Verify the functionality of the ADC using the function generator as an input.
- Verify UART data transmission.
- Identifying the ultrasound response from the filtered signal with transducer pointing at glass wall
- Moving the transducer and watching the transducer response move. Estimating distance at which the transducer response is indistinguishable from noise (5 cm).
- Using the device with a tissue phantom which mimics the ultrasonic properties of the brain, looking for the ventricle, which is 2 cm deep on one side of the phantom, and 4 cm deep on the other side.
- Use the device or a comparable device in a pig study to determine whether the ventricle can be seen.

Simulated ventricle in alginate   Ultrasound stylet directed at tissue phantom
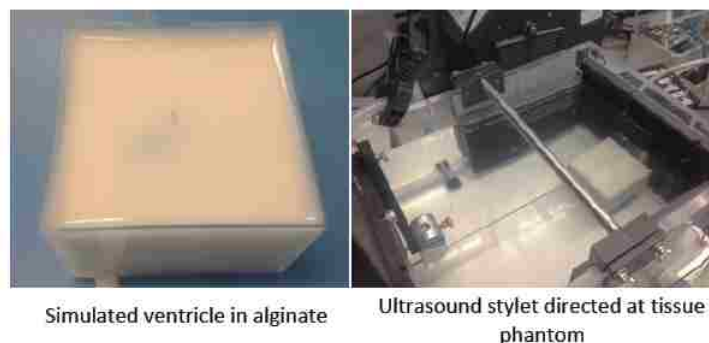
Figure 10: Tissue Phantom (Ventricle is the blue shadow on left image)
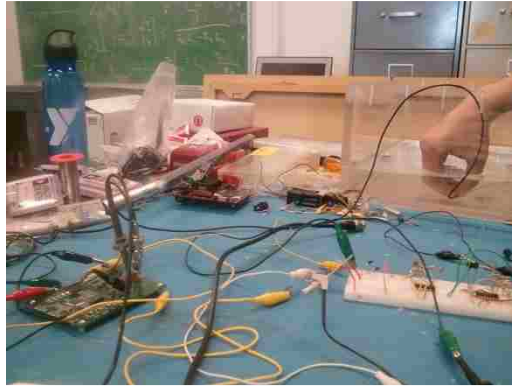
7. PROGRESS

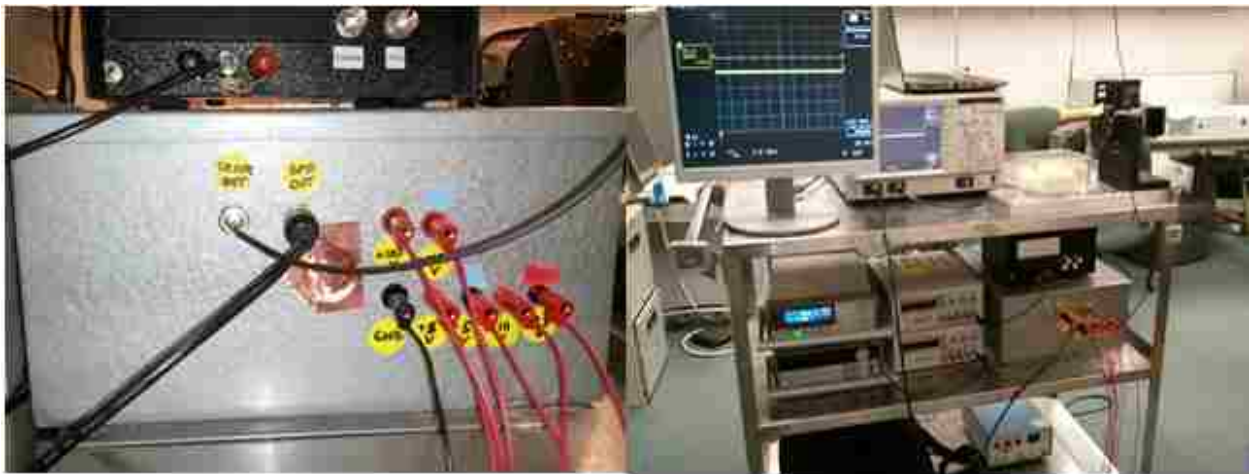Figure 11: Early stages of the prototype

Figure 12: Current Boxed Prototype
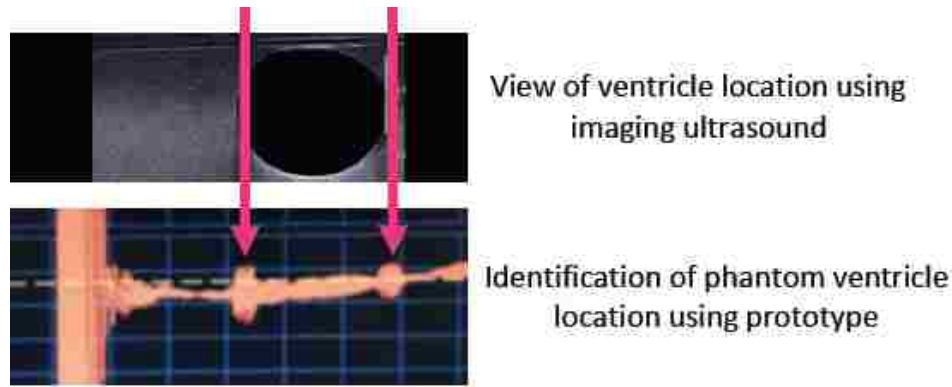
Tissue Phantom Testing – Successful at 2.5 cm

Figure 13: Tissue Phantom Results – 2.5 cm

At the time of doing the tissue phantom study, the real-time display of data hadn't yet been accomplished, so the output of the bandpass filter was directed to the oscilloscope to visualize the data. With some averaging, the front and back brain-ventricle interfaces in the line of the transducer can be seen above. An ultrasound image is also shown above it for comparison, and shows the clear distinction in impedance between the ventricle (dark) and its surroundings. The beginning of the signal in Figure 12 (the concentrated region) is the 8-pulse high voltage cycle from the pulser after being clamped by the TX/RX chip's voltage protection diodes.

The following table illustrates VentriGuide's success in passing its test cases:

Test Case	Description	Success (y/n/ partly)	Failure Points	Comments
Ultrasound Pulser Functionality	200 Vpp output, 8 pulses per cycle, 5 Khz cycle	Y		
TX/RX Functionality	Allows voltages below 1.4 V to pass through, with the two diodes clamping all higher voltages to 1.4 V	Y		
Bandpass filter functionality	10 MHz center frequency, 1 MHz bandwidth, Gain 100	Y		Using the same evaluation board, the filter components could be chosen for a smaller bandwidth and higher gain. 1 MHz was a conservative choice, largely driven by time constraints. Due to the later SNR problems with the tissue phantom, I've revised the bandpass filter for higher gains; the component values are included with the MATLAB code in the appendix (Appendix B)
Differential Amplifier functionality	Levels input signal to be centered at 1.5 V instead of 0 V	P	Added Noise	Due to either PCB design (long leads) or the chip itself, the signal was corrupted with too much noise. It couldn't be rerouted through the bandpass filter, either, because the filter would filter out the DC offset on the signal
ADC (AD9283-50 MSPS)	50 MSPS, 8 bit parallel output	P	Due to the added noise from the	A different ADC was selected (AD9238-65 MSPS) which could interpret a signal centered around 0 V, with a higher bit

			differential amplifier, the ADC's performance became irrelevant.	output. The downside of making this switch was the higher cost of the ADC.
<u>REPLACEMENT ADC</u> ADC (AD9238-65 MSPS)	65 MSPS, 12 bit parallel output	Y	This ADC has a demoboard which can accept a signal centered at 0 V, thereby circumventing the need for the AD8138 differential amplifier	The ADC successfully displays the input signal on a laptop with the VisualAnalog software. The downside of this ADC is that it is not integrated with the rest of the system, and that it adds some small noise spikes to the displayed signal (See Appendix Figure 18 and 19)
MCU (MSP430F5438)	Sample ADC output	N	System clock configuration too slow to sample the ADC at more than the Nyquist frequency	Misunderstood the datasheet; the MSP430 could take an external crystal up to 32 MHz, but the system clock can only go up to 16 MHz regardless of the external clock. This frequency isn't enough to satisfy the Nyquist frequency. The MSP430F5438 was replaced by the uno32 board, containing the PIC32MX320F128H, which has a system clock capable of going up to 80 MHz
<u>REPLACEMENT MCU</u> PIC32MX320F128H	Sample ADC output	Y		MCU can sample ADC output for slower frequencies; software has to be refined to speed up sampling for higher frequencies. The hardware is capable of the sampling speeds required.
UART Connection	Real time display of ADC sampling	P	Display of data successful at lower frequencies	Given 1) time constraints, 2) the understanding that the solution would be about finding a good compromise between buffer size and data transfer (refresh) rate, and the knowledge that the UART code would not extend to a final prototype, a decision was made to obtain a larger companion board for the ADC which already had the capability of providing real-time data display.
Functionality of UART Companion Board for Replacement ADC Board	Real time display of ADC sampling	Y	Display of data successful at relevant frequencies	See Appendix Figure 18 and 19
Tissue Phantom Testing – Part 1	Ventricle at 2 cm	Y	SNR somewhat low; can be increased with averaging in the microcontroller	See Figure 12 for image
Tissue Phantom Testing – Part 2	Ventricle at 5 cm	N	Signal too attenuated	Signal is lost in the 10 mVpp noise. Greater SNR is required through better bandpass filter design.
Tissue Phantom Testing – Part 2 with Panametrics	Ventricle at 5 cm	Y		See Appendix Figure 22 for image
Tissue Phantom Testing – Part 3 (Debugging Part 2	Looking for reflectance from glass wall through pure tissue phantom (no	N	Signal too attenuated	This step was done to troubleshoot Tissue Phantom Testing Part 2, to examine at what depth of pure brain tissue phantom

Failure of Device)	ventricle)			(no ventricle) a perfect reflector (glass wall) be identified in the first place. It was discovered that at more than about 2 cm away, the signal becomes too attenuated. Since the noise of the system is only about 5 mV, the amplified signal of < 5 mV indicates an original signal of < 1 mV. Higher amplification with the band pass filter could increase the range of depth of the device, as well as alternate transducer shape design, or lowering the frequency of the transducer.
Testing with air-backed transducer on tissue phantom	Determining whether SNR was improved or not by an air-backed design, which can increase the output power of a transducer by 2-5x	Y	SNR is higher using the panametrics commercial ultrasound pulser/receiver	SNR appears to be approximately x2 compared to the previous SNR of the other transducers. However, the air-backed transducer is much more fragile and additional mechanical design may be required for the final device as a result.
Testing panametrics pulser/receiver in vivo on a pig	Determining whether a commercial benchmark device could be used to show proof of concept with the air-backed transducer	Y	Ventricles can be identified, as well as the region between the ventricles (the fornix)	See Appendix Figure 23 for ventricle image
Increasing SNR with larger gain bandpass filter	Changed gain from 100 to 1000	P	SNR is higher, but filter appears more unstable – gives an oscillating output when particular pulser cycles/pulse settings are used	Bandpass filter formula will need to be reviewed for stability as well as gain, with greater steps during evaluation of the bandpass filter than just SNR measurements.
Increase SNR with shielding (box)	Boxing up device and connecting power with banana plugs instead of alligator clips	Y	SNR is improved; noise only decreased by about 30% though.	SNR might be further improved with using better shielded banana plugs, as well as with shielding on each individual board.

Additional Notes:

Bandpass filter Functionality

Further testing of the bandpass filter in comparing it to the panametrics device reveals that it is equivalent to a 22 dB setting on the panametrics device. This implies that revisions of the bandpass filter design, while increasing the gain, did not increase it to the levels anticipated. Currently, passing in a function generator input at 10 MHz and 6 mVpp results in an output of 60 mVpp, which indicates that the equations which represent the bandpass filter are not translating into practice. Examining the pcb board makeup for how it may be affecting the bandpass filter performance, as well as tweaking resistor/capacitor values usingthe function generator as an input could be used to refine the filter design until the desired functionality is obtained.

Replacement ADC

Due to time constraints, an ADC with a readily available demoboard and USB interface board was bought to meet deadlines for boxing up the device. The true reason behind the noise in the AD8138 chip and the resulting inefficacy of the AD9283 chip is likely due to poor PCB design rather than the chips selected. Further debugging of the PCB

design with outside contracted engineer advice can direct the project towards miniaturization without further redesign.

8. *TECHNICAL CHALLENGES*

Some of the major technical challenges involved in this project involve the high frequency, high voltage, small signal response, and small size constraints of the device. These, in conjunction with new information resulting in different technical requirements, have been the main source of technical challenges in this project. These challenges have been further explored below:

High Frequency, Small Signal

The challenge of sampling a 10 MHz ultrasound signal is that the Nyquist frequency requires an ADC with a higher sampling rate than 20 MSPS, and for true data visualization, at least five points per period would be ideal (50 MSPS). For a higher level of signal representation, I aimed to use a board with at least 50 MSPS sampling rate, and ended up picking one with 65 MSPS.

The raw, small signal response from the transducer of less than 10 mV requires bandpass filtering with a high level of gain to increase SNR as much as possible. These two requirements of high gain and high frequency combined requires ADC boards with a high gain-bandwidth product (GBP). To increase the signal response 50x at a minimum of 20 MSPS, the GBP of the chip would need to be at least 1 GHz. Within this small population of ADC evaluation boards, there are even fewer which can output data in parallel. Most can only output in parallel LVDS, which imbeds additional information and requires an additional deserialization chip to interpret the data. The benefit of picking a parallel output over a parallel LVDS output is that the speed of interpretation is faster; the downside is the larger footprint of the parallel output chip due to the larger number of output pins involved.

The fast attenuation of ultrasound in tissue makes the gain choices even more difficult. At short distances, an ultrasound response of 10 mVpp with a gain of 100 (40 dB) would become 1 Vpp, which is the breadth of input of many ADCs. However, this gain setting is insufficient for the 10 MHz system to identify a ventricle greater than 2 cm away (see Test Cases below). On the other hand, a gain of anything above 200 would result in saturation of the return signal at small distances – with ADCs with an input range of 2 Vpp. Since greater precision at farther distances is more important than saturation at shorter distances, the bandpass filter was eventually designed for a gain of 1000 (65 dB), which is similar to the gain found on some commercial pulser/receiver ultrasound systems (Panametrics).

High Voltage, Small Size

The use of high voltage components in a small form factor requires a preference for low power components, since the small size of the device (3 x 3 x 5 cm) doesn't give room for most battery-only solutions. Luckily, I was able to find extremely thin 3 V lithium ion batteries at 2.9 cm x 2.9 cm that were 0.45 mm thick. Using only batteries and some power regulation circuitry, these batteries could produce 300 Vpp and power the system in series.

If higher voltages are required or if there are problems discovered in tying the multiple batteries in series, voltage step-up circuits can be used to obtain even higher voltage outputs. The downside of these circuits is that they could add substantial cost to the device, especially in the case of incorporating transformers (range in the hundreds). A cheaper alternative would be to have multiple voltage step-up regulators in a cascade – this alternative solution could be achieved for less than $50.

9. *ADDITIONAL LEARNING*

A. **Ultrasound Simulation in Brain (MATLAB, K-wave Toolbox)**

In my proposal to the SAC committee, I mentioned Field ii being the software that I'd use to simulate transducer activity in brain tissue. Upon researching the software further, I discovered that Field ii is widely used for transducer design in general, but not for simulating biological environments. At the beginning of autumn quarter, I experimented with the k-wave toolbox in Matlab, to simulate the ventricle in the brain. Though my initial simulation wasn't precise in terms of modeling the characteristics of the transducer we would be using, it did identify the method of using both reflections from the front and back ends of the ventricle as a method for identifying the ventricle visually.
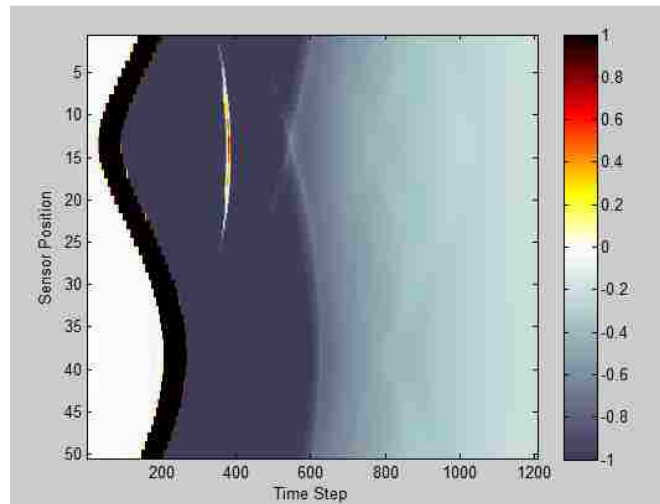
Figure 14: K-wave Matlab Toolbox Simulation

In the figure above, the two waves of normalized pressure responses can be seen at sensor position 13, which is the position closest to transducer (the black wave refers to the initial pressure wave). The reflection of the ventricle wall closest to the transducer has pressure reflection intensities of about 20-60 % of the original pressure intensity, and the intensity of the second reflection is between 0-10% of the original.

B. **UW Business Plan Competition**

During the design process, I made estimates of the device cost in bulk (> 10k units sold/yr) and continued learning about the business aspects of bioengineering at the UW Business Plan Competition. In a team composed of business students and other bioengineering students, I was able to research other applications of the VentriGuide device, such as with urinary catheter and lumbar puncture market research. At the Investment Round, I was one of the team members that spent a day pitching the the VentriGuide device to entrepreneur judges, who decided which companies they would invest in. Our team succeeded in getting into the Sweet Sixteen round, and we also won the Best Technology Idea award for $2500.

The following images illustrate some of the technological, market (and product) research that I helped facilitate:

Component List	Small Batch	In Bulk
Ultrasound Pulser	$15	$7
MAX220 Chip	$3	$3
Bandpass filter	$4	$4
Analog/Digital Converter	$25	$20
Microcontroller	$7	$7
Thin/Flexible Lithium Ion Battery	$30	$20
LCD Screen	$30	$20
Metal Stylet	$8	$8
Plastic Injection Molding	$5	$5
Miscellaneous Parts	$10	$9
Total Wholesale Cost	**$137**	**$103**
Manufacturing Labor, etc	**$69**	**$31**
Total COGS per Device	**$206**	**$134**

Figure 15: Cost of Goods (COGS) for Final Device in Small Batch and Bulk

Potential Markets
(Annual Procedures in the US)

- CVC — 5.0m
- Dialysis — 400k
- Arterial Line — 8.0m
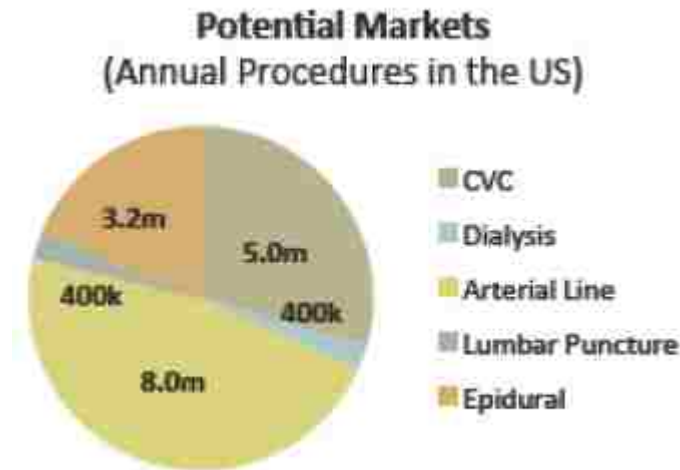- Lumbar Puncture — 400k
- Epidural — 3.2m

Figure 16: Additional Markets for Guidance Technology

10. CONCLUSION

This project successfully established the baseline design for a portable ultrasound system for relieving intracranial pressure. The system can identify ventricles in phantoms at 2.5 cm. Improvements to the bandpass filter, increasing the voltage to the system or improving transducer design, and added processing on the microcontroller/DSP chip end will improve the device' efficacy in identifying ventricles at larger depths.

11. NEXT STEPS AFTER FIRST PROTOTYPE

To begin miniaturizing the device after the first prototype is completely functional, the following steps need to be taken (assuming that the pulser and overall design of the device doesn't change significantly):

A. Developing simplified, refined circuitry for ultrasound pulser
B. Miniaturization of ADC

C. Battery + Power circuitry development

D. PCB Design for all combined components to fit into final device dimensions

12. FUTURE CHALLENGES AND IMPROVEMENTS

The following would be the next targets in improving the current VentriGuide system:

Next Challenges:

Noise Reduction/SNR Optimization
Due to the high signal frequency of this device and the low SNR of the unprocessed transducer signal, it was harder to create fast prototypes, since adding any new component involved evaluating its noise contribution, and many initial prototyping techniques (alligator clips, long wires, etc.) become significant contributors. Miniaturization onto PCB boards becomes a requirement by necessity (to achieve higher SNR), and time for evaluating each chip is lost or postponed.

Revision of Frequency/Power (Directionality and Depth compromise)
Currently based on testing the device compared to the Panametrics pulser/receiver (see Appendix), the device appears to be the equivalent of the energy 1, 22 dB setting of the panametrics pulser, which has six total energy stages and a highest gain setting of 59 dB. Considering that the current device cannot resolve the ventricle at present, the bandpass filter on the board needs to be revised for greater gain. If the panametrics pulser cannot view ventricles at even the highest gain setting at the energy 1 level, additional strategies need to be implemented to increase the voltage output of the pulser. This latter strategy has already been pursued; an ultrasound pulser board which can output 400 Vpp max (compared to the 200 Vpp maximum earlier) has been obtained from Supertex. The main challenge with increasing the voltage will be in the power circuitry that'll supply the voltage.

Power Circuitry
The power circuitry for this device may be simple or complex, depending on the stability of the ultrathin primary cell batteries when put in series. Since the device does not need to operate for more than 10-30 minutes and has less than 20 mA power draw, it seems likely that – if the device doesn't need to supply more than 200 Vpp – a simple voltage drop regulation circuitry at the most would be required. If higher voltages are required, dc to dc converters are the likely solution for achieving the voltages within budget.

Improvements/Recommendations
1. It will probably benefit the device's success in the market to include a higher resolution LCD screen if the LCD output appears too pixelated/unprofessional due to lower resolution.
2. For alternate applications, haptic feedback during a procedure (ex. epidurals, lumbar puncture) could offer better precision.
3. It might be interesting to explore a singular element transducer system with a low and high frequency component (resulting from the variation in thickness in two different dimensions) – this could offer better depth perception, as well as directionality/higher resolution at device surface.
4. Investigate obtaining a DSP FPGA with an internal ADC at more than 25 MSPS. In my research, I haven't encountered a DSP FPGA with an ADC greater than 1 MSPS, but if such a device exists at less than a $50 pricepoint, it would be worth replacing the ADC and MCU portions of the device with a single DSP FPGA, for both cost efficiency, greater signal processing power, as well as space.
5. Investigate a method for ensuring that the surgeon cannot drive the transducer in air when initializing the device, such as a small bubble of sterile liquid attached to the stylet tip that breaks when it comes in contact with the brain surface. This initiation could also prompt the device to turn on, as an alternative to the pushbutton LCD.

13. REFERENCES

Background
1. Toma et al. External Ventricular drain insertion accuracy: is there a need for change in practice? Neurosurgery 65:1197; 2009.
2. Kakaria, U.K. et al. Safety and accuracy of bedside external ventricular drain placement, Neurosurgery, 63, 1, 162-6. 2008
3. Bucholz, Richard et al. The correction of stereotactic inaccuracy caused by brain shift using an intraoperative ultrasound device. Frequency Control, 1995.

IQ Demodulation
4. Information taken from http://folk.ntnu.no/htorp/Undervisning/TTK10/IQdemodulation.pdf

K-wave Toolbox
5. www.k-wave.org

Device Websites
6. www.supertex.com (Ultrasound pulser, tx/rx chip)
7. www.analog.com (Bandpass filter chip, ADC)
8. www.ti.com (MSP430 microcontroller board)
9. www.digilent.com (Replacement PIC microcontroller board)
10. www.powerstream.com (Li-ion primary ultrathin batteries)
11. www.nkkswitches.com (LCD pushbutton)
12. www.panametrics.com (Panametrics pulser/receiver box)

14. APPENDIX

A. User Manual Instructions

1. Insert power banana jacks into the power inputs on the side of the box in the following order: GND, 3.3 V input, 10 V input, +5 V, - 5 V, + 100 V, - 100 V. When removing power, remove in reverse.

2. Insert AC power into the two small power input holes of the box

3. Insert USB into side of box

4. Connect "BPF Output" via BNC cable to oscilloscope

5. Connect output of function generator (40 MHz square wave, low level 0 V, high level 1.7 V) to "Clock Input" and enable output

6. Connect "Transducer Output" to the transducer via a bnc cable

7. Connect other end of USB to laptop, open up VisualAnalog. Select AD9238 as the device, and "Samples" as the template. Click on "continuous run" button (green triangle) on top left. There shouldn't be any output yet, but you should see some small spikes if you zoom in.

8. Open the top of the box, hit the 'enable' button on the ultrasound pulser board, close top of box

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9. Look at VisualAnalog; you should see the signal. If not, check if clock output was turned on, or try closing the software and reopening. You should also be able to see the signal on the oscilloscope by hitting the 'auto' button and changing the voltage and time increments to about 100 mV and 10 us respectively.

B. Matlab code for Bandpass Filter Component Selection

```matlab
%%BANDPASS FILTER Calculation

clear all; close all; clc;


Q = 72; %24 db fallout
G = 10000
F = 10 * 10^6; %10 Mhz
C = 20 * 10^-12; % 20 pF
k = 2*pi*F*C;
R1 = Q ./ (G*k)
R2 = Q ./ (((2*Q^2)-G)*k)
R3 = (2*Q) ./ k
B = 1/(pi*C*R3) %1 Mhz


R1_R2 = (R1*R2)/(R1+R2);
w = 1/(C*((R1_R2*R3)^.5))/(2*pi) ; %10 Mhz
Ho = R3/(2*R1); %100

%For gain of 100, Q = 10,
%R1 = 80 Ohms
%R2 = 80 Ohms
%R3 = 16 kOhms


%for gain of 330, Q = 15, B = 667 kHz
% R1 = 36 % P35.7ABCT-ND, 0.11 per
% R2 = 99 % P100ABCT-ND, 0.11 per
% R3 = 23.8 kOhms % P23.7KABCT-ND, 0.11 per


%for gain of 500, Q = 20, B = 500 kHz
% R1 = 32 % P32.4ABCT-ND , 0.11 per
% R2 = 53 % P53.6ABCT-ND, 0.11 per
% R3 = 32 kOhms % P31.6KABCT-ND, 0.11 per


%for gain of 1000, Q = 25, B = 400 kHz
% R1 = 20; %Digikey P20.0ABCT-ND, 0.11 per resistor
% R2 = 80; %already on the board
% R3 = 40 kOhms %40.2 kOhms on digikey P40.2KABCT-ND, 0.11 per


%for gain of 1500, Q = 30, B = 336 kHz
%R1 = 16.2;
%R2 = 76; (use 2 of the 35.7 ohm)
%R3 = 47.4k;


%for gain of 1950, Q = 39.7, B = 251.83 kHz
%R1 = 16.2;
```

```
%R2 = 26.26; (divide the 53.6 ohm by 2)
%R3 = 63.2 Kohms;

%for gain of 2518, Q = ....
%R1 = 10.8 (three 32.4 Ohms in parallel)
%R2 = 138.72 (100 ohm + 80 || 80 in series)
%R3 = 63.2K (two 31.6K in series)

%For GAIN OF 10000, Q = 72, B = 138.9 Khz
%R1 = 5.73;
%R2 = 155.7;
%R3 = 115.59 kOhms;
```

C.   C code for PIC microcontroller

```c
#include <plib.h>
#include <p32xxxx.h>
// Configuration Bit settings
#pragma config FPLLMUL = MUL_20, FPLLIDIV = DIV_2, FPLLODIV = DIV_1, FWDTEN = OFF
#pragma config POSCMOD = XT, FNOSC = PRIPLL, FPBDIV = DIV_8 //alteration
#define SYS_FREQ (40000000L)
#define DESIRED_BAUDRATE (9600)
// Let compile time pre-processor calculate the PR1 (period)
#define PB_DIV        8
#define PRESCALE      256
#define TOGGLES_PER_SEC  1
#define T1_TICK       (SYS_FREQ/PB_DIV/PRESCALE/TOGGLES_PER_SEC)

int pbClk;
int brd;
int main(void)
{
  //enable pins on PortB as input
  TRISB = (BIT_2 | BIT_3 | BIT_4 | BIT_5 | BIT_8 | BIT_9 | BIT_10 | BIT_11 | BIT_12 | BIT_13 | BIT_14 |
BIT_15);
  TRISD = (BIT_0 | BIT_1 | BIT_2);
  //enable pins as digital input
  AD1PCFG = (BIT_2 | BIT_3 | BIT_4 | BIT_5 | BIT_8 | BIT_9 | BIT_10 | BIT_11 | BIT_12 | BIT_13 | BIT_14 |
BIT_15);
  mPORTDSetPinsDigitalOut(BIT_4);

  // Optimize PIC32 performance and return peripheral bus frequency
  pbClk=SYSTEMConfig(SYS_FREQ, SYS_CFG_WAIT_STATES | SYS_CFG_PCACHE);
  brd = pbClk/16/DESIRED_BAUDRATE-1;
  // Enable UART2 and set baud rate to DESIRED_BAUDRATE=9600
  OpenUART1(UART_EN, UART_RX_ENABLE | UART_TX_ENABLE, pbClk/16/DESIRED_BAUDRATE-1);
  /*OpenOC5(OC_ON | OC_TIMER_MODE16 | OC_CONTINUE_PULSE | OC_TIMER2_SRC);
  OpenTimer2(T2_ON | T2_PS_1_1, )/**/

 // set variables for buffer
  unsigned int output = 0;
  unsigned int output2 = 0;
  //unsigned int outputBuf[2000];
  unsigned int indexSize = 999;
  unsigned int index = 0;
```

```
    unsigned int outputBuf1[1000];
    unsigned int outputBuf2[1000];
    int i = 0;
    int temp = 0;

    while(1)
    {
            output = 0;
            //LSB
            output = output | (PORTReadBits(IOPORT_B, BIT_15) >> 15);//J7_12
            output = output | (PORTReadBits(IOPORT_B, BIT_14) >> 13);//J7_11
            output = output | (PORTReadBits(IOPORT_D, BIT_0) << 2);//J6_07
            output = output | (PORTReadBits(IOPORT_B, BIT_12) >> 9);//J7_09
            output = output | (PORTReadBits(IOPORT_B, BIT_11) >> 7);//J7_08
            output = output | (PORTReadBits(IOPORT_B, BIT_10) >> 5);//J7_07
            //BIT6 SKIPPED
            output = output | (PORTReadBits(IOPORT_B, BIT_9) >> 3);//J7_06
            output = output | (PORTReadBits(IOPORT_B, BIT_8) >> 1);//J7_05
            output = output | (PORTReadBits(IOPORT_B, BIT_5) << 3);//J7_04
            output = output | (PORTReadBits(IOPORT_B, BIT_4) << 5);//J7_03
            output = output | (PORTReadBits(IOPORT_B, BIT_3) << 7);//J7_02
            output = output | (PORTReadBits(IOPORT_B, BIT_2) << 9);//J7_01
            //MSB
            output2 = output >> 8;
            outputBuf1[index] = output2;
            outputBuf2[index] = output;
            index++;
            if (index > indexSize)
            {
              i = 0;
              for (i = 0; i<indexSize; i++)
              {
                 while(BusyUART1());
                 putcUART1(outputBuf2[i]);
                 while(BusyUART1());
                 putcUART1(outputBuf1[i]);
              }
              index = 0;
            }
    }

}
```

D.   Matlab code for data display

```
averageout = 0;
max = 600;
first = 1;
miny = 0;
maxy = 140000;
xmax = 1000;
buffsize = 2048;
outmax = 100;
newpnt = 0;
out = (1:outmax);
```

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```
s.ReadAsyncMode = 'continuous';
s.ByteOrder = 'bigEndian';
s.InputBufferSize = buffsize;
fopen(s);
while(1)
    if(first == 1 && s.BytesAvailable == buffsize)
        out = fread(s,s.BytesAvailable/2,'uint16')';
        plot(out,'-*');
        xlim([0,xmax]);
        ylim([miny,maxy]);
        shg;
        first = 0;
    elseif(s.BytesAvailable == buffsize)
        newpnt = fread(s,s.BytesAvailable/2,'uint16')';
        out = [out(length(newpnt):length(out)) newpnt];
        plot(out,'-*');
        xlim([0,xmax]);
        ylim([miny,maxy]);
        shg;
    end
end
fclose(s);
```

E.   Photographs taken during various testing scenarios
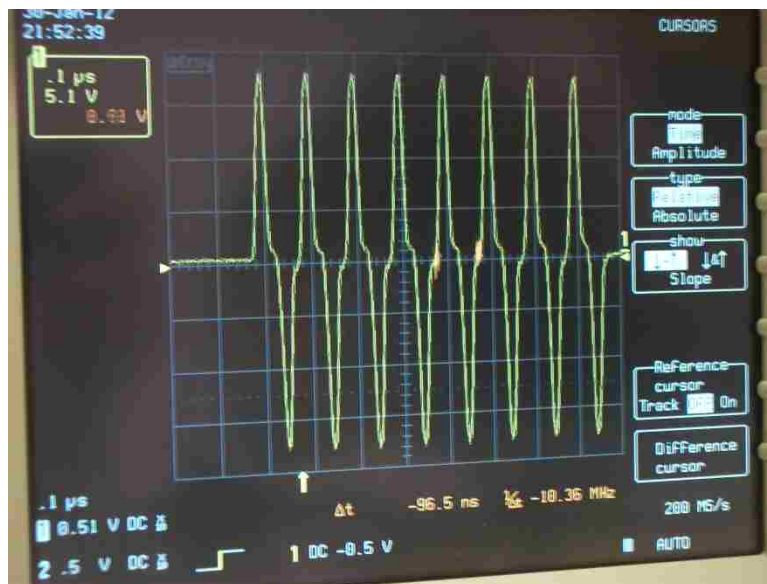
Figure 17: Pulser Output to Transducer with LED 2 lit (8 pulses/cycle) and default 10 MHz output.
Input voltage to board when picture taken was ~40 Vpp

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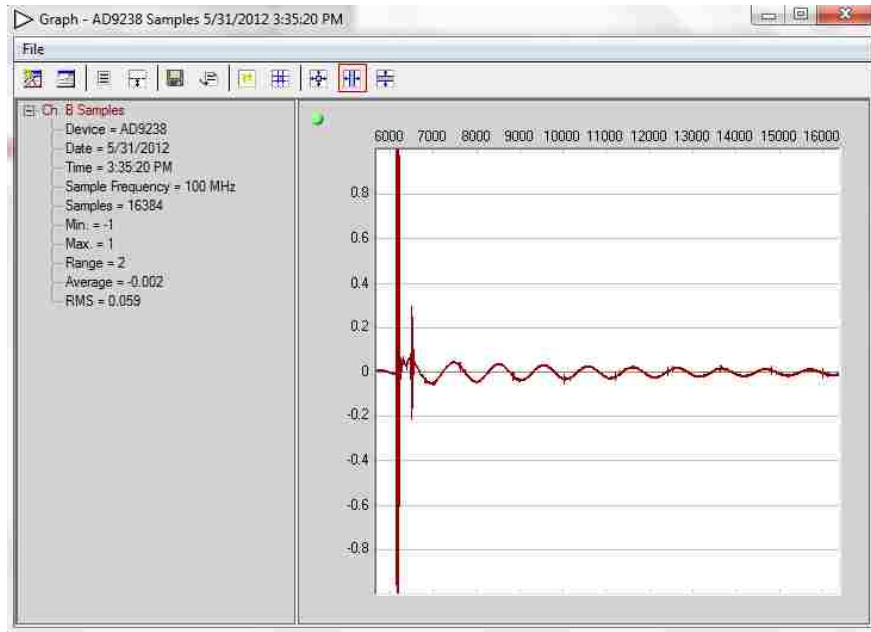
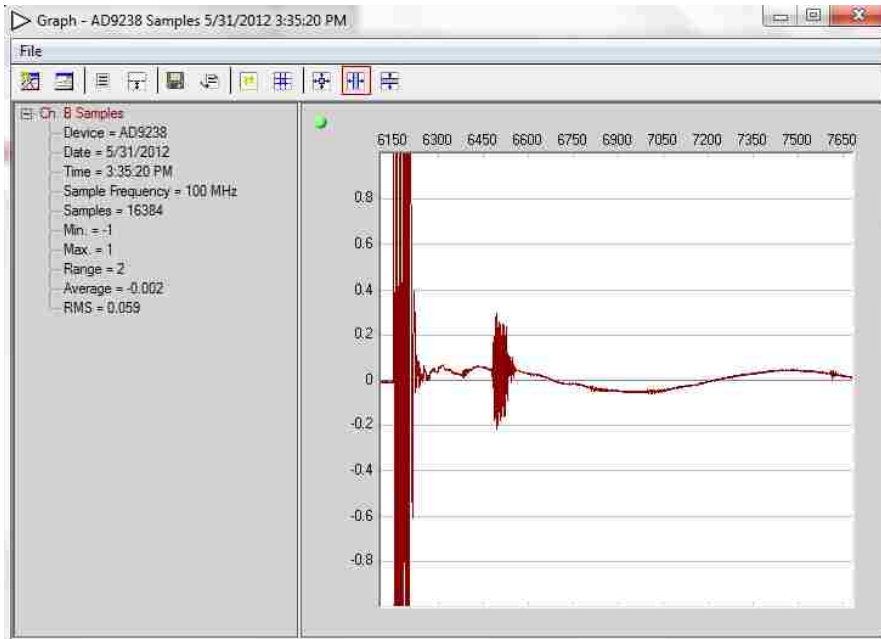Figure 18: Real-time ADC display of signal using VisualAnalog

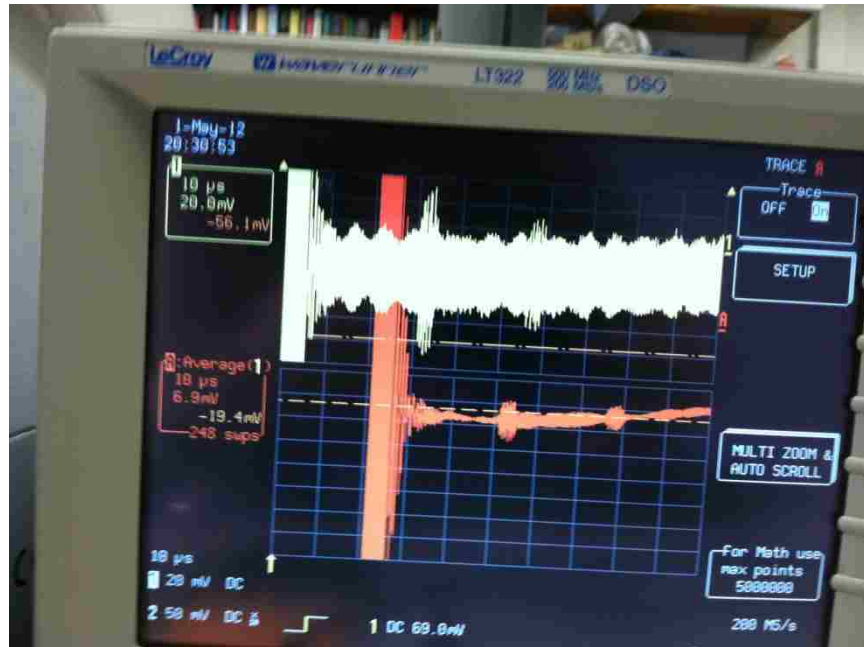Figure 19: Zoomed Ultrasound Response on VisualAnalog

Figure 20: Ventricle Identification at 2.5 cm

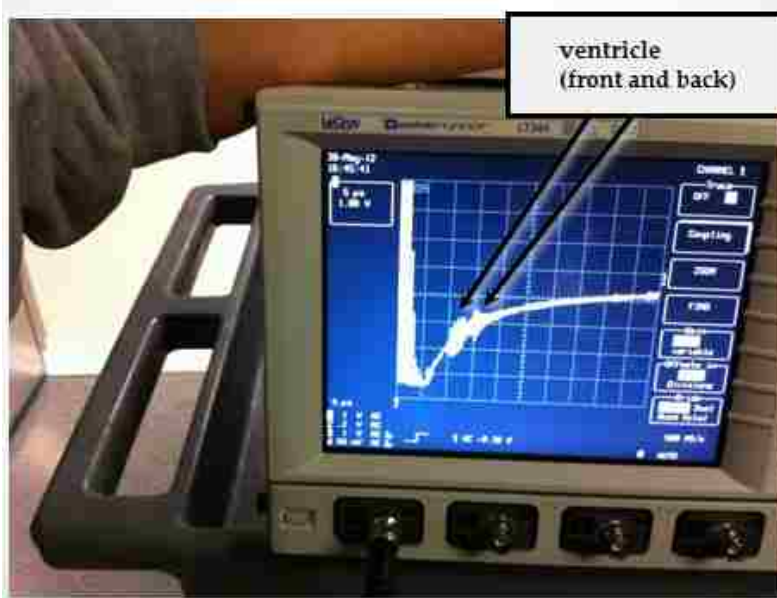Figure 21: Ventricle identification at 5 cm using Panametrics Pulser/Receiver

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Figure 22: Results of In-Vivo Testing

F.   Brain Acoustic Parameters

Brain Acoustic Properties				
Thermal Conductivity	**Temp ©**	**Conductivity (W m^-1 K^-1)**	**Reference**	
Brain,Human,78% water	5-20	.528	Cooper & Trezek 1972	
White, 71% water	5-20	.503		
Gray, 83% water	5-20	.565		
Cortex	37	.515	Valvano et al 1985	
Thermal Diffusivity, alpha	**Temp ©**	**Diffusivity cm^2 s^-1 * 10^3**	**Reference**	
Brain,Human,78% water	5-20	1.38 +- 0.11		
White, 71% water	5-20	1.34 +- 0.10		
Gray, 83% water	5-20	1.43 +- 0.09		
Cortex	37	1.47		
Thermal Properties of Tissue Sub Materials	**k (W m^-1 K^-1)**	**C (J g^-1 K^-1)**	**rho (g cm^-3)**	**alpha (cm^2 s^-1)**
Brain, 2.45 Hz	0.478 +- 0.015	3.41	0.98 +- 0.03	0.00143

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Specific Heat Capacity, C	Specific Heat (J g^-1 K^-1)	Reference	
White, 71% water	3.60	Cooper & Trezek 1972	
Gray, 83% water	3.68		
Optical Attenuation Coefficient	**lambda, nm**	**alpha, mm^-1**	**Reference**
Brain, human	630	0.45, 0.56	Muller & Wilson 1986 cont
Brain, human, in-vivo	630	0.2-0.34	Wilson et al 1986a
Brain, human, in-vitro	488	1.4-2.5	Svaasand & Ellingsen 1983
Velocity of Ultrasound through normal soft tissues	**Temp ©**	**Velocity m s^-1**	**Reference**
Brain,human	37	1562	Kremkau et al 1981
corna radiata	40	1573	
cerebellum	22	1537	
newborn, in vivo	37	1525	Willocks et al 1964
Velocity of Ultrasound through human pathalogical tissues	**Temp ©**	**Velocity m s^-1**	**Reference**
Brain Accustic neurinoma	22	1522-1547	Schiefer et al 1968
Arachnoidal sarcoma	22	1530	
Astrocytoma	27.5	1545	Van Venrooij 1971
Ependymoma	22	1537-1545	Schiefer et al 1968
Glioma	22	1525-1547	
Meningioma	22	1540-1550	Schiefer et al 1968
Menstatis	22	1535	Schiefer et al 1968
Spongioblastoma	22	1532	
Tumor cyst contents	22	1504-1524	
Dispersion of acoustic velocity in tissue, 1-10 MhZ	**Dispersion, m s^-1/MhZ**	**dc/df %/MhZ**	**Reference**
Brain	1.2	0.08	Kremkau et al 1981
Ultrasound amplitude attenuation coeff for normal tissue	**Temp ©**	**Freq MHz**	**Attennuation Coeff Np cm^-2** / **Coeff dB cm^-1**
Brain, human	25	5	0.072 / 4.5
grey matter	37	2.2	0.121 / 0.625
white matter	37	2.2	0.018 / 1.05
cerebellum	30	0.97	0.14 / 0.5
Ultrasound Absorption Coef	**Temp ©**	**Freq MHz**	**Np cm^-2 * 10^-2** / **dB cm^-1**
Brain, cat	37	0.7	1.4 +- 0.3 / .12 +- .02
Acoustic scattering cross-section for tissue	**Freq MHz**	**Backscatter cm^-1 Sr^-1**	**B** / **Reference**

					Nassiri & Hill 1986
Brain	4-7		1.27 * 10^-7	3.98	

G. K-wave code for ultrasound simulation

```matlab
% Revathi Murthy
% Bioen 700
% Ultrasound Propagation Example

clear all; close all; clc;

% create the computational grid
Nx = 256;           % 15.6 mm % number of pixels in the x (column) direction
Nz = 256;           % 15.6 mm % number of pixels in the z (row) direction
dx = .1e-3;         % pixel width [m] % .1 mm
dz = .1e-3;         % pixel height [m] % .1 mm
kgrid = makeGrid(Nx, dx, Nz, dz);

% make the cyst
cyst = 1*makeDisc(Nx, Nz, Nx/2, Nz/2, 25); %centered at the middle
%with radius of 2.5 mm

% define the properties of the propagation medium
medium.sound_speed = 1562*(ones(Nz, Nx)); %  - cyst) ; % [m/s] %brain
medium.sound_speed = medium.sound_speed ; %+ (1482 * cyst);   % [m/s] %water
medium.density = 1040*(ones(Nz, Nx)); % - cyst);      % [kg/m^3] %brain
medium.density = medium.density; % + (1007 * cyst);     % [kg/m^3]
%cerebrospinal fluid
medium.alpha_coeff = 0.6;  %human brain % [dB/(MHz^y cm)]
medium.alpha_power = 1.2; %human brain. dispersion acoustic velocity?

% create initial pressure distribution using makeDisc
disc_magnitude = 500; %500 kPa 0.5 MPa
disc_x_pos = Nz/2 - (.5*50) - 20; % 2 mm from left of cyst
disc_z_pos = Nz/2;    % pixels
disc_radius = 10;    % 1 mm radius
disc_1 = disc_magnitude*makeDisc(Nx, Nz, disc_x_pos, disc_z_pos,
disc_radius);

source.p0 = disc_1;

% define a centered circular sensor
sensor_radius = 2.5e-3;   % [m] %at the surface of the cyst
num_sensor_points = 50;
sensor.mask = makeCartCircle(sensor_radius, num_sensor_points);

%RECORD MOVIE
% define the input arguments
input_args = {'PlotPML', false, 'RecordMovie', true, 'MovieName',
'example_movie_1', 'MovieType', 'image', 'PlotFreq', 5, 'MovieArgs', {'fps',
30}};
```

```
% run the simulation with optional inputs for plotting the simulation
% layout in addition to removing the PML from the display
sensor_data = kspaceFirstOrder2D(kgrid, medium, source, sensor, 'PlotLayout',
true, 'PlotPML', false);

% plot the simulated sensor data
figure;
imagesc(sensor_data, [-1, 1]);
colormap(getColorMap);
ylabel('Sensor Position');
xlabel('Time Step');
colorbar;
```

H. Alternate Technology Application (Urinary Catheter) Evaluation

Urology Applications of Catheter Guidance Technology

Market Opportunity

Urinary catheters come in three main categories: foley (indwelling) catheters, condom catheters, and intermittent (short-term) catheters. Foley catheters are used for a few days up to a few months, condom catheters are primarily used for elderly men with dementia, and intermittent catheters are used temporarily to empty the bladder several times a day[1].

In 2010, the market for urinary catheters was evaluated to be around $230 million dollars by the Millennium Research Group (MRG). MRG expected the evaluation to double to $460 million by 2014. This increase is due to Medicare's change in policy in 2008 to reimburse 200 intermittent catheters/month from 4/month, increasing the demand (40% annual compound growth rate) for intermittent catheters through 2012. The top companies that supply urinary catheters are Coloplast, which had over 50% of the US market in 2009. Hollister, Bard, Astra Tech, Rochester and Rusch are also major contenders[2].

Approximately 90 million indwelling catheters are sold to US hospitals annually, and 4 million Americans undergo urinary catheterization in hospitals every year[3]. Indwelling catheters can be separated into two types: urethral and suprapubic catheters. Suprapubic catheters are attached to the abdomen, which has been surgically connected by a tube to the bladder. Urethral indwelling catheters (also known as Foley catheters) are the more common (25 million sold/yr in the US, with 500k procedures/yr) and do not require surgery[4]. Between 15-25% of hospitalized patients receive short-term foley catheters[5] and 5% of residents in long-term care facilities use foley catheters as well (50k residents)[6].

Excess, stagnant urine in the bladder has been linked to contributing to CAUTI (catheter associated urinary tract infection), which accounts for nearly 32% of all hospital infections -- about 800k cases every year due to indwelling catheter usage alone[7]. Hospitals spend about 340-500 million dollars every year (500-1k/patient[8]) treating the

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[1] http://www.nursingcenter.com/prodev/ce_article.asp?tid=1027118
[2] http://mrg.net/News-and-Events/Press-Releases/INTERMITTENT-CATHETER-MARKET-CURRENTLY-VALUED-AT-$.aspx
[3] http://journals.lww.com/homehealthcarenurseonline/Fulltext/2010/06000/Best_Practices_in_Urinary_Catheter_Care.5.aspx
[4] http://www.nursingcenter.com/prodev/ce_article.asp?tid=1027118
[5] http://www.annals.org/content/137/2/125.full
[6] http://enoxbiopharma.com/urinary-catheters
[7] http://www.the-hospitalist.org/details/article/186502/How_can_we_Reduce_Indwelling_Urinary_Catheter_Use_and_Complications.html
[8] http://www.ncbi.nlm.nih.gov/pmc/articles/PMC2631699/pdf/11294737.pdf

symptoms of CAUTI in the United States[9]. Finally, mortality from CAUTI ranges from 2.3% (minor infection) to 10% (bloodstream infection)[10]. Indwelling cathethers have a 25% risk of infection after 7 days of use, with a daily risk between 3-7%[11].

The market trend for urinary catheters is the larger use of intermittent catheters, which offer greater patient independence, reduce the need for extra equipment (ex. drainage bags), and for reduced lower urinary tract symptoms (frequency, urgency, incontinence) between the catheterizations. It also has a lower risk of urinary tract infection than indwelling catheters. [12] However, intermittent catheters require repeated emptying (4-5 times a day) and do not remove the risk of urinary tract infections entirely, since the risk of external bacteria introduction also increases with the higher rate of catheterization; the FDA has recommended that short-term use of indwelling catheters should only be replaced due to blockage or infection as a result[13]. Intermittent catheters are also nearly 2-3x more costly than indwelling catheters due to the cost of the medical professionals' time in repeatedly inserting/emptying the catheter during the day[14].

Another complication due to the foley catheter is the inflation of the balloon at the end of the catheter (used to keep the catheter secure) in the urethra, rather than inside the bladder. This complication has even been classified as a "Never Event" by the Centers for Medicare and Medicaid Services, defined as "a serious, largely preventable patient safety event"[15]. The risk of urethral injury is even higher due to low operator training in the use of catheters; an estimated 0.3% of hospital patients have catheter placement trauma[16], and a San Diego study found balloon inflation complications in 3.2 of 1000 male patients[17]. The most common way of evaluating inflation of the balloon in the urethra is to watch for signs of pain when inflating the balloon[18][19]. Increased training has not made a significant impact on reducing the rates of injury[20]. Urethral trauma of this kind also impacts methods of future catheterization (suprapubic catheterization is often substituted)[21] and costs hospitals thousands to ten-thousands of dollars in in treating the urethral trauma[22].

Our Technology

An ultrasound guide-wire could detect the presence of urine in the bladder and ensure that the foley catheter balloon is not inflated prematurely. The presence of post-void urine in a healthy child/adult bladder is less than 200 ml[23], in a sick bladder, greater than 200 ml [24][25], and at least 70 ml even when the bladder is emptied using a catheter – hence, the transducer will have a low risk of being exposed to air.[26]

Transducer design considerations, such as a multi-element transducer in an inverted 'U' configuration on the guide-wire could be used to estimate the amount of urine in the bladder before, and after for catheterization (to assess risk for CAUTI) as well as notify the operator upon the insertion of the catheter tip into the bladder. The reduction of significant urethral damage (ex. when the foley catheter balloon is inflated in the urethra) also results in a lowered

[9] http://www.the-hospitalist.org/details/article/186502/How_can_we_Reduce_Indwelling_Urinary_Catheter_Use_and_Complications.html
[10] http://www.cdc.gov/hicpac/cauti/005_background.html
[11] http://www.suna.org/education/2013/article3101229.pdf
[12] http://www.nursingcenter.com/prodev/ce_article.asp?tid=1027118
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[24] http://pjmhsonline.com/sonographic_assessment_of_post_v.htm
[25] http://www.bedfordwomenscare.com/services-procedures/urodynamics-urinary-incontinence-womens-care-nh-new-hampshire.php
[26] http://ccn.aacnjournals.org/content/25/2/44.full

risk of CAUTI blood infection. Finally, the urine volume measurement before and after catheterization could also provide useful clinical information for practitioners who would normally use portable ultrasound (ex. BladderScan[27]) before and after to measure the pre and post-void urine in the bladder.

As market trends move further towards self-catheterization using intermittent catheters, an ultrasound system like the above could be used in conjunction with a smart phone application to remind patients to self-catheterize, and notify them of early indications of CAUTI (high levels of stagnant urine in the bladder).

Competitors

Many of the patents that address urethral injury due to incorrect balloon inflation analyze pressure differences between that of the urethra and the bladder and give mechanical feedback or have safety pressure valves in place[28]. A large number of patents already use ultrasound to attack biofilm in urinary catheters[29] [30] [31] but these patents do not address the potential of ultrasound in identifying the foley catheter balloon's entrance to the bladder, nor do they address the potential of ultrasound in measuring urine amount in the bladder as a predictor for CAUTI.

One revolutionary urologic technology is available in Europe. The In-Flow device by SRS Medical for women uses a pump to completely void the bladder, using a hand-held device by the patient. SRS has found the UTI rate is same or reduced with the In-Flow device than intermittent catheters. They've also found that it eliminates tubes/drainage bags, and reduces the need to self-catheterize by 4-8 times a day. Finally, patients appreciate that the device allows the use of a toilet. The main downside of this technology is that it only applies to women with an atonic bladder. However, the companies' findings of lowered rates of UTI due to lower urine stagnation in the bladder shows that our technology is promising not only for catheter guidance, but also for reducing UTI through the monitoring of urine volume intermittent and indwelling catheterization[32].

Conclusion

Foley (indwelling) urinary catheters are sold in large quantities in the United States (25 million sold/yr) and the complications resulting from improper guidance (incidence rate of 0.3%, or 7.5k cases/yr) cost hospitals between 10k -100k per incident. An even more frequent complication of all urinary catheters (90 million sold/yr) is urinary tract infection (32% of all hospital infections), which costs hospitals an additional 500-1k/incident (350 million/yr), with a mortality rate of 2.3-10%. The cost of using multiple single-use intermittent catheters has been shown to be more cost-effective compared to the cost of treating UTI[33], which implies that a reusable, ultrasound-tipped guide-wire may be most cost-effective for patients that require indwelling catheters (burn ICUs, inpatient medical wards, and neurosurgical ICUs[34]), that also are the subset of patients most likely to have the highest risk of acquiring CAUTI. In these sets of patients, this device has an additional advantage over other competitor catheters, which only attempt to prevent urethral trauma, by including measures of early prediction of CAUTI. When limited to its guidance application, this technology is still at an advantage over its competitors because the disposable guide-wire can be immediately compatible with any current foley catheters available in the market. The price of the device in market would also be significantly lower than its neurosurgical counterpart (which is estimated at $400), because lower frequency piezocrystals could be sufficient in achieving the pulse-echo distance measurements which determine bladder urine volume.

[27] http://www.verathon.com/Products/BladderScan.aspx
[28] http://www.freepatentsonline.com/y2011/0082444.html
[29] http://www.freepatentsonline.com/7393501.html
[30] http://www.freepatentsonline.com/7829029.html
[31] http://www.freepatentsonline.com/7393501.html
[32] http://www.srsmedical.com/downloads/inflow/In-Flow-Background.pdf
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[34] http://www.cdc.gov/hicpac/cauti/005_background.html