Application of Ultrasonic Techniques for Brain Injury Diagnosis

Joel Mobley, Paul M. Kasili, Stephen J. Norton, and Tuan Vo-Dinh(*)

Oak Ridge National Laboratory. PO Box 2008 MS 6101, Oak Ridge, TN 37831-6101

ABSTRACT

In this work, we evaluate methods for detecting brain injury using ultrasound. We have used simulations of ultrasonic fields in the head to model the phase distortion of the skull. In addition we present experimental data from the crania of large animals. The experimental data help us understand and evaluate the performance of different transducers in acquiring the backscatter data from the brain through the skull. Both the simulations and acquired data illustrate the superiority of lower-frequency (≤ 1 MHz) ultrasonic fields for transcranial acquisition of signals from inside the brain. Additionally, the experimental work shows that the higher-frequency (5 MHz) ultrasound can also be useful in acquiring clean nearfield data to help detect the position of the inner boundary of the skull.

I. Introduction

In a previous study[1] we reported measurements of the ultrasonic properties of brain and skull tissue and presented some simulations of ultrasonic fields in the head. In this work we show new simulations of ultrasonic fields and experimental data from the crania of large animals. The simulations are used to model the phase distortions incurred by ultrasound due to propagation across the skull and are concerned with a different aspect of the problem than the simulations presented in the earlier work. The experimental data help us understand and evaluate the performance of different transducers in acquiring the backscatter data from the brain through the skull.

Researchers have been investigating the application of ultrasound to the identification of anatomy and the detection of pathology in the brain since the late 1940's. In spite of much promising work over the following three decades, an objective clinical procedure involving ultrasound for detecting disease or injury in the adult brain has failed to emerge [2]. However, in the 1980's Doppler ultrasound was found to be a useful modality for assessing the circulation of blood in the brain through the intact skull. Since then, progress continues in the clinical use of ultrasound employing both Doppler-based and imaging-based techniques [3,4].

The difficulty in applying ultrasound transcranially to investigate the anatomy of the adult brain is due to the presence of the skull bone. The skull bone is heterogeneous, having a single layer of compact bone in some regions and having three layers in other regions. In the three-layer regions, compact bone forms the inner and outer layers and surrounds a section of cancellous bone known as the diploe (see Figure 1). The diploe consists of a network of bone, spongy in appearance, with the spaces between the bone filled with softer marrow (fatty tissue and blood). Because of the large differences in mechanical properties between the bone and marrow, this layer is a strong scatterer of ultrasonic radiation. The non-uniformity in size, shape and position of these marrow-filled regions, causes unpredictable scattering and can have significant impact on the ability of ultrasound to penetrate the skull and maintain phase coherence. Other properties effect the entire skull, both the single layer and tri-layer sections. First, the compact bone attenuates MHz-range ultrasonic waves very strongly and this attenuation is much larger than that typically found in soft tissues and biological fluids. The skull is also not of uniform thickness over areas typical of the apertures of ultrasonic transducers. This variation is largely due to the inner table of skull. Because of the non-uniformity of the inner table, together with the difference of sound speed in compact bone ($c_{bone} \approx 3000$ m/s) and fluids/soft tissue ($c_{brain} \approx 1560$ m/s), the phase fronts of propagating ultrasonic waves get distorted. This can cause the beam to be scattered out over a wide region and lose its directivity. Therefore, an ultrasonic system for transcranial measurements must contend with the problems associated with the diploe, the large attenuation of bone, and the roughness of the inner table of the skull.

Our approach to the problem of brain injury assessment is to utilize low frequency (i.e., sub-MHz) ultrasound together with a differential detection scheme with signals captured from two transducers placed

(*) Author to whom correspondence should be sent

symmetrically on the head. Typical medical imaging applications use ultrasound in the 1 to 7 MHz region of the spectrum. The frequency spectrum used in imaging devices is a trade-off between **depth-of-penetration** into the tissue (lower-frequencies can propagate deeper) and lateral/axial resolution (tighter, more-narrow beams at higher-frequencies). However, our approach is not one of imaging but of detection: the spatial resolution needed to form a useful image is not necessary to simply detect an asymmetric condition in the propagation medium. By using lower frequencies (in the 100-900 kHz range) we can transmit ultrasound across the skull more efficiently and achieve a greater depth of penetration into the brain.

As mentioned above, the strategy for the detection of injury is to exploit the left/right symmetry of normal brain. If an injury gives rise to a space-filling lesion in the brain, it will then lead to a compression of nearby structures and break the symmetry. The general approach is to detect asymmetries in the positions of structures in the brain and in the ultrasonic properties of the tissue. This scheme is accomplished by recording signals from two acquisition sites that are symmetric with respect to the centerline of the head. A straightforward method of detection builds on the well-known midline shift technique. When insonifying from the temporal region of the head, the junction of the two hemispheres, as well as other central features such as the third ventricle and **pineal gland**, provide ultrasonically reflective anatomical markers. A condition which gives rise to a left-right asymmetry in the brain is detected by the shift of the midline of the brain or other central structures from the sagittal plane of the head. The difference in the time of travel for ultrasonic pulses taken from different sides of the head is used to measure the shift. A significant difference in the travel times could indicate the presence of an injury. In addition to measuring pulse-echo travel times, other information may be derived from the captured ultrasonic signals. Using appropriate signal acquisition and processing techniques, differences in attenuation, interface reflections and backscatter may be resolvable which can provide further indications of the state of left/right symmetry in the brain. These measures can improve the sensitivity and specificity for injury detection beyond that of the original midline shift alone. Such a device may also be useful as a serial monitoring device by detecting changes in anatomy or scattering and propagation properties of the brain over time.

II. Background

IIa. Head Injury

Trauma is the leading cause of death for people under 44 years of age in the United States. More than half of these trauma-related deaths are due to head **injury**[5]. Brain injuries fall under two classifications: primary and secondary. Primary injuries are directly related to the contact or inertial forces that occur at the moment of impact. Secondary injuries occur after the accident (typically on the order of hours but can be as long as a few days) often due to the body's response to the sustained injuries. A common secondary injury is cerebral swelling which can lead to ischemia (tissue death due to insufficient blood flow) and other pathologies. Damage due to primary injuries is considered irreversible and the medical care is concentrated on managing/preventing the secondary injuries. A patient with potential head injuries is usually imaged with some non-invasive means such as computed tomography (CT). However these machines are not portable and often it is difficult to move a critically injured patient. A non-invasive means of monitoring the state of the brain at the bedside would be an important advance in the management of these patients.

IIb. The Skull

The main obstacle to using ultrasound as a non-invasive means of interrogating the brain is getting the ultrasound in and out of the head across the skull. As discussed earlier, the basic difficulties are related to several properties of the cranium: (1) the skull bone has a very high intrinsic attenuation, (2) the diploe layer of the skull is a strong scatterer of ultrasound and (3) the inner table of the skull is rough. The inhomogeneous mechanical properties (stiffness, density, velocity of sound, etc...) makes the diploe layer a strong scatterer and phase aberrator of ultrasonic waves. Additional problems arise due to the roughness of the inner surface of the skull. The uneven inner table combined with the velocity difference (about a factor of 2) between the compact bone and the soft tissue/fluid inside the skull give rise to phase aberrations as well. Figure 2 illustrates the phase distortion incurred by plane waves that pass through the skull.

III. Simulations

We calculated the ultrasonic fields radiating from three aperture-size/ultrasonic-frequency combinations to assess the impact the phase distortion of the skull will have on the spatial coherence and directivity of the fields. The fields were calculated using the angular spectrum technique [6]. This method generates the spatial distribution of monochromatic fields from the Fourier transform of the source. Each of the three source fields were spherically-focused (5 cm focal length) by phase shifting a uniformly distributed field over a circular aperture. The following frequency/aperture-diameters were used: (1) **500-kHz field/2.5 cm** diameter, (2) **1-MHz field/2.5 cm** diameter, and (3) **5-MHz/1.25 cm** diameter. The fields were propagated through a uniform medium with no attenuation and a velocity of sound (longitudinal mode only) of 1500 m/s. Only longitudinal mode waves were considered since shear waves are **strongly** attenuated by soft tissues. The phase distortion effects were modeled by placing a random phase screen at 1 cm from the source. At each point on the plane of the screen, the phase of the field was randomly shifted using a Gaussian distribution. The half-widths of the distributions were determined from the following expression,

$$\delta\phi_{\text{half width}} = \left| 2\pi f \Delta z \left(\frac{1}{c_{\text{bone}}} - \frac{1}{c_{\text{soft tissue}}} \right) \right|$$

where f is the ultrasonic frequency (500 kHz, 1 MHz, or 5 MHz)

Δz is the variation in the thickness of the skull (due to inner table roughness, taken to be 1 mm)

c_{bone} approximate speed of ultrasound in compact bone (3000 m/s),

$c_{\text{soft tissue}}$ approximate speed of ultrasound in soft tissue (1500 m/s).

The relation between the transducer and the fields in our model is shown in Figure 3. The lower panel of this figure shows the plane of symmetry that is displayed in the field plots that follow. Figures 4-7 display the magnitude of the field at each point in the plane. Figures 4 and 5 show the unfiltered fields (i.e., no phase screen is present). The field magnitude is represented by the z-axis of the plot in Figure 4 and by grayscale (white = maximum) in Figure 5. Each of the fields shows a maximum near 5 cm which is the radius of curvature of the acoustic focusing lens.

The fields calculated with phase screen in place are shown in Figures 6 and 7. The **500-kHz** and **1-MHz** fields are both able to maintain their directivity and coherence in spite of the phase screen. The main impact of the screen on these two fields is an overall reduction in amplitude. The **5-MHz** field, however, is strongly scattered by the screen. The directivity and strength of the beam are both largely eliminated by the phase distortion. The directivity of the ultrasound beam is important to maintain in order to properly interpret the signals that are backscattered from inside the cranium. At the two lower frequencies, the beams maintain their directionality. For these fields one can be reasonably certain that detected signals are coming from scattering events occurring near the axis of the transducer. However, the **5-MHz** field is so highly dispersed that it is unlikely to generate detectable scattering events, and even if it does it will be difficult to know from what region in the brain they originated. For techniques that rely on anatomical markers to deduce the state of the brain, the lower frequency fields show some promise, but the **5-MHz** field is clearly inappropriate.

IV. Experiment

Using a portable ultrasonic acquisition system that we built in our lab (see Figure 8), we captured pulse-echo signals from the intact heads of recently slaughtered pigs at a local meat packing facility. These signals were acquired anywhere from a few minutes to a few hours after the deaths of the animals. We acquired pulse-echo data with the transducer positioned about 5 cm behind the eye. We used three different circular-aperture transducers: (1) a planar (flat-faced) **2.5-cm** diameter transducer with a center frequency near 500 kHz, (2) a planar **2.5-cm** diameter transducer with a center frequency near 900 kHz, and (3) a focused (spherically curved face) **1.25-cm** diameter transducer with a center frequency near 5 MHz. Each of these transducers was internally damped to achieve broadband ultrasonic response. A captured backscatter signal is displayed in Figure 9 where some of its features are identified. Typical waveform> acquired with the different transducers are shown in Figure 10.

Our goal for this study is to assess what frequency ranges would permit us to get a large amount of ultrasonic energy into and out of the brain across the skull, while permitting the identification of the inner table of the skull to use as a reference marker. Since the skull is not uniform in thickness, being able to resolve the echo from the inner table of the brain is essential in getting accurate estimates of the positions of the reflective anatomical markers within the brain. As seen in Figure 10, the **500-kHz** and **900-kHz** transducers produce large echoes from inside the brain. From the time-of-flight, the structures responsible for these echoes are about 7.5 cm below the skull. However, the near-field information is not as clear since there is saturation. The lower attenuation at these lower frequencies also permits some reverberations in the skull which further obscure the picture. In contrast, the **5-MHz** transducer only captures a weak signal from deep in the brain but its near field is much clearer. The large attenuation prevents reverberations and keeps the echo from the inner table of the skull from saturating.

To get a clearer view of the large scale echoes, we applied **wavelet** transform filtering to the signals. This helps to isolate the large-amplitude echoes from the small-scale scattering signals. (These small-scale signals are not noise but contain information about the properties of the tissues themselves. We are interested in exploiting these signals and this work is ongoing but not discussed further at this time). The results of the **wavelet** filtering are shown in Figure 11. From these results, it seems that a **dual-frequency** modality using a high-frequency/low-frequency approach could meet our demands. The **low-frequency** pulses can give us the desired information from inside the brain while the higher-frequency ultrasound can be used to get a clean estimate of the position of the inner surface of the skull.

V. Conclusion

In this paper, we presented simulations of ultrasonic fields to model the phase distorting properties of the skull. This modeling showed that **500-kHz** and **1-MHz** fields were able to maintain their directivity and coherence in spite of the phase distortion. However, the **5 MHz** field was strongly scattered by the phase screen. This demonstrates the effectiveness of lower frequency fields for getting signals from inside the brain across the skull. We also looked at pulse echo data taken from the intact crania of large animals. These data show that the lower frequency (**500-** and **900-kHz** center frequencies) fields are able to successfully capture signals from deep in the brain through the skull. We also observed that the higher frequency (**5 MHz**) field may be useful since it can cleanly resolve the echo from the inner table of the skull.

VI. Acknowledgements

This research is jointly sponsored by the U.S. Medical Research and Materiel Command (ARMY No. **96MM6699**), the ORNL SEED Funding (Project No. **321 1-0844**), and by the Office of Biological and Environmental Research, U.S. Department of Energy under contract **DE-AC05-96OR22464** with Lockheed Martin Energy Research Corporation. Dr. Mobley is also supported in part by an appointment to the Oak Ridge Laboratory Postdoctoral Research Associates Program administered jointly by the Oak Ridge Institute for Science and Education and Oak Ridge National Laboratory.

VII. References

1. J. Mobley, P. M. Kasili, S. N. Norton, and T. Vo-Dinh, "Evaluation of Ultrasonic Techniques for Brain Injury Detection", **BIOS '98 International Biomedical Optics Symposium**, January 24-30, 1998, San Jose, CA. Proceedings of the SPIE- The International Society for Optical Engineering, 1998.
2. D. N. White, "The early development of neurosonology: I. Echoencephalography in adults." *Ultrasound in Med. & Biol.* 18 (2). pp. 115-165 (1992).
3. R. W. Baumgartner and I. B. Baumgartner, "Transcranial Doppler and color-coded duplex sonography: Established and new applications," *Ultraschall in der Medizin* 17 (2), pp.50-54 (1996).
4. M. Kaps, G. Seidel, and W. Dorndorf, "Present status of transcranial color-coded sonography in adults," *Ultraschall in der Medizin* 16 (2), pp.50-59 (1995).

5. A. D. Gean, *Imaging Of Head Trauma*. (Raven Press, 1994).

6. J. W. Goodman, *Introduction to **Fourier** Optics, 1st ed.* (McGraw-Hill, 1968).

Figure 1 The three layers of the skull bone.

Figure 2 The distortion of the planes waves due to propagation through the skull.

Figure 6 The ultrasonic fields from the three continuous wave sources with the random phase screen at 1 cm from the source plane. The z-axis is the magnitude of the pressure.

Figure 7 The ultrasonic fields from the three continuous wave sources with the random phase screen at 1 cm from the source plane. White represents maximum pressure.

Figure 8 The ultrasonic data acquisition system. The pulser excites the transducer. The transducer generates the ultrasonic wave and detects the backscattered ultrasound. The received signal is amplified and captured by a digital oscilloscope. The digitized waveform is then downloaded to the computer and stored.

Figure 9 An ultrasonic signal captured in pulse/echo mode from the head of a test animal. This figure identifies the probable sources of the features in the waveform.

Figure 10 Representative backscattered waveforms from an intact head taken by three different transducers.

Figure 11 The three waveforms shown in Figure 10 after undergoing wavelet transform filtering