AUTOMATED DETECTION OF OBSTRUCTIVE SLEEP APNEA USING ULTRASOUND IMAGING

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ABSTRACT

Obstructive sleep apnea affects more than eighteen million Americans and sufferers experience frequent interruptions to restorative sleep which often leads to motor vehicle accidents, missed days of work, overnight hospital stays, and chronic fatigue. Obstructive sleep apnea (OSA) is characterized by obstructions of the air passage in the vicinity of the throat during sleep that continue until the person wakes him or herself up. The gold standard diagnostic modality, the polysomnograph, is expensive and requires the patient to do undergo uncomfortable, overnight sleep study. OSA remains a disease where 80-90% of sufferers are undiagnosed and untreated, thus intensifying the need for an improved diagnostic technique.

This thesis presents an initial study on the potential of medical ultrasound imaging techniques for the diagnosis of OSA. Ultrasound is a safe (non-invasive and radiation free), relatively inexpensive, clinically acceptable imaging modality that can be used to create real time images of the area of the throat involved in OSA. Therefore, it should be feasible to *directly* detect apneic events by utilizing ultrasound images to identify obstructions of the airway caused by OSA.

A standard ultrasound system was employed to obtain images of the area of the pharynx involved in OSA and image processing algorithms were utilized to detect obstructions. First, active contours (snakes) were shown to accurately detect the state of the airway (open or obstructed) in two dimensional axial (transverse) images of the pharynx. Subsequently, the minimum sum absolute difference (MSAD) motion detection algorithm was used to quantify tongue base movement in midsagittal ultrasound videos of posterior movement of the tongue. These experiments lead us to conclude that ultrasound
detection of apneic events necessitates an ultrasound transducer designed to image the volume of the pharynx involved in OSA.

Most current ultrasound methods utilize linear (1xN) array ultrasound transducers to obtain two dimensional images. Real-time 3D ultrasound systems typically employ low channel counts (<512) with sparse 2D arrays to facilitate reasonable system and transducer complexity. This thesis describes a fabrication technique for fully populated, 32x32 element 2D arrays operating at 5MHz. Initial prototyping, in which a blank PCB is used and in which only the side elements are available for testing, provides positive results (finite impedance response indicative of resonant activity and without an open or short circuit). As size limitations in the manufacturing of PCBs are overcome, this approach will allow for the fabrication of 2D arrays with greater density and higher frequency.

This thesis serves as the foundation for future research on the role of ultrasound imaging as a diagnostic tool for OSA. Clinical acceptance of ultrasound imaging as a diagnostic modality for OSA will require human study investigations to validate this method in comparison to the polysomnograph.
ACKNOWLEDGEMENTS

Apparently four years at the University of Virginia was not enough for me—nor was a simple major in electrical engineering. I wanted to extend my undergraduate studies in electrical engineering into the medical field to combine engineering with the pursuit for healthier living. During my third year at the University of Virginia I was accepted into the accelerated bachelors and masters program and Dr. John Hossack introduced me to the world of ultrasound. At that point, the doors opened for me to pursue the omnibus faculty of biomedical engineering.

Working on the obstructive sleep apnea project for the last two years has been a delightful experience. I wish to thank Carilion Biomedical Institute for providing project funding, and specifically Andre Muelenaer and Sam English for their assistance. I would also like to thank Dr. R James Hawley and Dr. Paul Suratt for their insight and knowledge about the clinical and diagnostic facets of obstructive sleep apnea. Furthermore, I am indebted to the following people for their support, patience in teaching me, and contributions to this project: Dr. William Walker, Dr. Travis Blalock, Dr. Scott Acton, Yongjian Yu, Ned Light, Chris Fabian, Bill Bishop, Matthew Hunt, Jim Mabry, Mike Fuller, and Clement Song.

Most importantly, I am grateful to Dr. John Hossack for giving me the opportunity to work on this project and expounding on me his knowledge, support, and guidance.

Lastly, I would like to thank my family and friends for all of your love and support throughout my undergraduate and graduate years at the University of Virginia:
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GLOSSARY OF TERMS

Acoustic impedance – The property of tissue causing resistance to the propagation of ultrasound; mathematically defined by the product of density ($p$) and acoustic velocity ($c$) of a material. The letter $Z$ is used for the impedance and is expressed in [kg/s m²] = 1 Rayl. It plays an important role in the determination of acoustic transmission and reflection at the boundary of two materials having different acoustic impedance.

Active contours (snakes) – Curves defined within an image domain that can move under the influence of internal forces coming from within the curve itself and external forces computed from the image data; they are extensively used in computer vision and image processing application to locate object boundaries.

Airway patency – The state or quality of the airway being open, expanded, or unblocked.

Apnea/Hypopnea Index - (AHI) (RDI or Respiratory Disturbance Index) The frequency of abnormal respiratory events per hour of sleep. These events are classified as Apneas or Hypopneas. Apnea is when breathing (airflow) stops for 10 seconds or more. Hypopnea is a partial blockage of airflow resulting in arousal and a possible drop in oxygen level. An AHI of 45 would indicate that the patient is experiencing complete or partial airflow blockage 45 times per hour.

Apneic event – Cessation of breathing for 10 or more seconds during sleep. In Obstructive sleep Apnea they are caused by a closure of the air passage despite efforts to breathe.

B-Scan image – A 2D image perpendicular to the face of the transducer.

C-Scan image – A 2D image parallel to the face of the transducer.

Continuous Positive Airway Pressure (CPAP) Machine – Medical device used to treat sleep apnea. This apparatus provides a highly effective, non-invasive therapy that eliminates blockages and prevents collapse of the upper airway by generating a prescribed level of air pressure that maintains airway patency during sleep. Air pressure is delivered through a hose to a mask that fits over the nose, or both nose and mouth. The mask is secured on the face by headgear that is worn over the head. The appropriate air pressure level is determined during a "CPAP titration" sleep study. The complete system consists of a programmable pressure generator, tubing, mask and headgear. Sometimes referred to as nCPAP (nasal Continuous Positive Airway Pressure).

Epidemiological – pertaining to the study of the causes, distribution, and control of disease in populations.

Epworth Sleepiness Scale - An index of sleep propensity during the day as perceived by
patients, and derived from the answers to 8 questions.

Geniohyoid muscle – Muscle that originates in the mandible and inserts in the hyoid, and lies beneath the mylohyoid muscle. This muscle elevates and protracts the hyoid bone and depresses the mandible.

Mylohyoid muscle – Muscle that spans the bottom of the jaw, originating in the mandible and inserts at the upper border of the hyoid bone. This muscle assists the geniohyoid and the digastric muscles in moving the hyoid bone and the tongue upward and forward and then upward and backward during swallowing.

Obstructive sleep apnea (OSA) – Repetitive cessation of breathing during sleep for 10 seconds or more due to complete closure (collapse) of the throat. Usually characterized by snoring, excessive daytime sleepiness, and other symptoms of fatigue.

Pharynx – The area at the back of the mouth behind the nostrils and the oral cavity. It functions as a passageway for air from the nostrils and the mouth to the lungs; and for food and liquids from the mouth to the esophagus.

Polysomnograph (PSG) – A biomedical instrument for the measurement of multiple physiological variables of sleep. It records the sleep physiological parameters of oxyhemoglobin saturation, electroencephalography (EEG), electromyogram (EMG), oronasal airflow, respiratory sound by a microphone, electrocardiogram (ECG), electrooculogram (EOG), chest wall effort, and body movements.

Speckle reducing anisotropic diffusion (SRAD) – An edge-sensitive diffusion technique for image processing geared towards ultrasonic and radar imaging.

Transducer – A material or device that converts input energy of one form to output energy of another form. For ultrasound, this material is typically a piezoelectric crystal that changes electrical energy into mechanical.

Ultrasonic event (with respect to obstructive sleep apnea) – relaxation of the geniohyoid and mylohyoid muscles followed by posterior or inferior movement of the tongue base to obstruct the airway.

Ultrasonic event (ultrasonography) – A medical imaging technique that uses high frequency sound waves and their echoes. An ultrasound machine transmits high frequency (>1MHz) sound waves into a subject using a transducer. At tissue boundaries, some of the sound waves are reflected due to the differences in acoustic impedance of differing tissues. The reflections are received by the transducer and signal processed such that a gray level image is formed based on the intensity of the received signal and the time delay between the transmitted pulse and its echo.

Via – An electrical connection between one or more layers (vertically) in a printed circuit board. They are typically formed by drilling a round hole into the layers that are to be connected and then filling or plating them with conductive metal.
NOMENCLATURE

CHAPTER 3

$I_0$ initial intensity image
$I$ output image
$div$ divergence
$var$ variance
$\Omega$ image support
$\nabla$ gradient
$\vec{n}$ outward normal to $\partial\Omega$
$X(s)$ initial parameterized snake
$\alpha$ non-negative weighting parameter for snake tension
$\beta$ non-negative weighting parameter for snake rigidity
$E_{\text{ext}}$ negative of gradient magnitude
$x^i, y^i$ n-dimensional column vectors denoting snake position at time $t$
$u(x^i, y^i)$ external energy of snake based on normalized gradient of the gradient magnitude
$b(x^i, y^i)$ internal pressure force
$\gamma$ weighting parameter for original snake
$\kappa$ weighting parameter for the external snake energy
$\kappa_p$ weighting parameter for the pressure force

CHAPTER 4

$s_n$ kernel within the image at frame n

CHAPTER 5

$k_t$ electromechanical coupling coefficient
$Z$ acoustic impedance
$\varepsilon^S$ dielectric constant
$\tan\delta$ proportionality factor for power loss, where $\delta$ is the phase retardation of current
$Q_m$ mechanical quality factor
$\lambda$ wavelength (frequency/speed of sound)
$k$ wave number, $2\pi f /$(velocity of sound)
w ceramic width
$\phi$ uniform potential
$\theta$ angle (radians)
1 INTRODUCTION

1.1 Rationale and Objectives

Sleep apnea is characterized by instances of periodic loss of breath or choking sensations throughout the night causing frequent interruptions to restorative sleep. Each apneic event (involuntary loss of breath) may last for up to one minute and more than 30 apneas per hour may occur during sleep. According to the National Institutes of Health, sleep apnea is as common as adult diabetes, affecting as many as eighteen million Americans, but an estimated 80-90% of sufferers remain undiagnosed and untreated. There are three types of sleep apnea: obstructive, central, and mixed. Of the three, obstructive sleep apnea (OSA) is the most prevalent and will be the focus of this thesis.

The effects of OSA include, but are not limited to, memory loss, weight gain, cardiovascular disease, and falling asleep during daytime activities. When left untreated, OSA often results in motor vehicle accidents, missed days of work, overnight hospital stays, and chronic fatigue. Early studies typified OSA as affecting overweight males over the age of forty. As public and clinical knowledge of the disease increased and more diagnosis were made, OSA has been found to affect people regardless of age, sex, and weight. Furthermore, obstructions of the airway due to anatomical abnormalities may affect anyone and may be the result of genetic factors.

OSA is diagnosable and treatable but unfortunately, current diagnostic technology is inadequate. Some of the available diagnostic modalities include polysomnograph
(PSG), awake imaging techniques (e.g. computed tomography (CT) and magnetic resonance (MR) imaging), and nocturnal testing involving oximetry and cardiorespiratory monitoring. The 12-channel PSG, the gold standard for OSA diagnosis, requires an overnight stay in a sleep lab and a specially trained sleep technician to do an offline interpretation of the results. This may cost upwards of $1000- $2000, and furthermore, some insurance companies may not cover these expenses. PSG is further limited because it indirectly detects apneic events using a combination of the following measurements: oxyhemoglobin saturation, electroencephalography (EEG), electromyogram (EMG), oronasal airflow, respiratory sound by a microphone, electrocardiogram (ECG), electro-oculogram (EOC), chest wall effort, and body position. Patients tend to experience discomfort due to the extensive monitoring equipment that is applied or worn for the duration of the study (approximately 6 hours).

A new diagnostic technology is desirable to circumvent the complications involved in diagnosis with PSG and to determine the location of the obstruction. While CT and MR imaging have been valuable modalities in studies of the biomechanics and pathophysiology of OSA, both modalities are relatively expensive and CT exposes the patient to radiation. According to Schwab et al., the ideal upper airway imaging modality for patients with OSA should be inexpensive, noninvasive, and allow for supine imaging in the absence of radiation as well as providing high-resolution anatomical images. Ultrasound is a safe (non-invasive and radiation free), relatively inexpensive, clinically acceptable imaging modality that can be used to create real time images of the area of the throat involved in OSA. Ultrasound imaging is an established method in the evaluation of
swallowing (normal and abnormal swallowing) and was recently utilized by Siegel et al. to evaluate OSA\(^3\).

The approach taken by Siegel et al. was to simultaneously record a PSG and ultrasound images of the pharynx and to compare apneic events identified by ultrasound and PSG. They found that each apneic event identified by the PSG corresponded to an ultrasonic event. An ultrasonic event is characterized by relaxation of the geniohyoid and mylohyoid muscles followed by posterior or inferior movement of the tongue base to obstruct the airway. Therefore, it should be feasible to *directly* detect apneic events by utilizing ultrasound images of the airway to identify movements of the tongue base to a position that obstructs the airway. Furthermore, the process of identifying apneic events could be computer-automated using novel image processing techniques.

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<tr>
<td>• Expensive</td>
<td>• Inexpensive</td>
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<td>• Indirectly detects apneic events</td>
<td>• Directly view the location of the obstruction</td>
</tr>
<tr>
<td>• Subjective and labor intensive analysis of results</td>
<td>• Doctors can see the occurrence of an apneic event; automated detection using image processing techniques</td>
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<tr>
<td>• Patient discomfort due to monitoring equipment</td>
<td>• Only one probe</td>
</tr>
<tr>
<td>• Overnight stay in a sleep lab</td>
<td>• Ultrasound is readily available and can be administered at many hospitals and clinics</td>
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Current ultrasound imaging techniques allow the sonographer to view images in a two-dimensional (2D) imaging plane. As applied to imaging the airway for detection of apneic events, this forces the sonographer to choose a single plane (in any direction) to
image. This poses a problem in that the obstruction may occur at a number of different planes in the pharynx. Siegel concluded that a probe engineered to image a volume of the pharynx would provide a more accurate account as to the nature of the obstruction. Determining the location of the obstruction will not only aid in diagnosis of OSA but also in surgical planning for the treatment of the disease.

The objective of this thesis is to present a preliminary study on the viability of medical ultrasound imaging as an alternative diagnostic modality for obstructive sleep apnea. This will entail determining the feasibility of various image processing algorithms to detect apneic events in ultrasound images of the pharynx and to prototype an ultrasound transducer designed to acquire the volume of data that encompasses the entire pharynx. An overview of the current literature on the pathogenesis and diagnosis of OSA is given in Chapter 2. Chapter 3 evaluates differentiating an open versus obstructed airway using computer automated edge detection techniques of ultrasound images. Automated motion detection of the tongue base is another potential method to detect apneic events using ultrasound images and is discussed in Chapter 4. Chapter 5 discusses the fabrication and testing of a prototype 2D ultrasound transducer that could be capable of volumetric or C-scan imaging. Finally, concluding remarks and recommendations for future work are presented in Chapter 6.

1.2 Summery of Objectives

- Develop image processing algorithms for computer-aided detection of apneic events
  - Implement an edge detection algorithm
- Implement a motion detection algorithm
- Analyze accuracy and complexity of each algorithm

➤ Design and build a 2D ultrasound transducer
  - Prototype a 2D transducer without a system interconnect
  - Test and characterize the initial prototype
  - Design the interconnect using printed circuit board techniques


2 THE PATHOGENESIS AND DIAGNOSIS OF OBSTRUCTIVE SLEEP APNEA

2.1 Introduction

Design considerations for an alternative modality to diagnose obstructive sleep apnea should be in the context of the pathogenesis and current diagnostic methods of OSA. The following sections describe previous and current research efforts in the pathogenesis and diagnosis of OSA and how ultrasound imaging plays a potential role in its diagnosis.

2.2 Pathogenesis of Obstructive Sleep Apnea

This section describes the current understanding of the pathogenesis of OSA, which involves the pharyngeal anatomy, airway patency, and epidemiological factors.

The airway of a person with obstructive sleep apnea is smaller and has different geometric configuration compared to a normal subject. Figure 2.1(a) shows the midsagittal anatomy of a normal subject compared to Figure 2.1(b), which illustrates an apneic patient. Both images were obtained using magnetic resonance imaging (MRI). Figure 2.1 also defines the four levels of interest in the airway of apneic patients: nasopharynx, retropalatal (RP), retroglossal (RG), and hypopharynx regions as well as the tongue, soft palate, airway, and epiglottis. These midsagittal images demonstrate that
apneic subjects tend to have a larger soft palate area and larger tongue area compared to normal or snorer/mild apneic subjects. (Note: In some texts the retropalatal region is further broken into retropalatal high and retropalatal low regions.)

A study by Schwab of MR images of various levels of the pharynx discusses the changes in size and structure of the airway during inspiration and expiration of normal and sleep-disordered subjects. Figure 2.2 shows a typical axial (transverse) image of the retropalatal (RP) level of a normal (a) and apneic (b) subject. The figure also illustrates the location of the airway, pharyngeal wall, parapharyngeal fat pad, subcutaneous fat, and tongue. The minimum airway area is approximately two times smaller in apneic compared to normal subjects in the RP region. Larger lateral pharyngeal walls in apneic patients (but not the size or location of the parapharyngeal fat pads) and an increase in subcutaneous fat in the lateral and posterior areas may explain these results.

Furthermore, the dilator muscle activity and distensibility of the airway differs for apneic
subjects. In all subjects there is little airway narrowing during inspiration, possibly due to
dilator muscles balancing the effects of negative intraluminal pressure. In fact, increased
dilator activity may result in an enlargement of the airway during early inspiration in
apneic subjects. In contrast, during expiration, the airway of apneic subjects narrows
significantly and heads towards a closed position. Also during expiration, when positive
airway pressure is applied there is a greater expansion of the airway in apneic subjects
compared to normal subjects. These findings indicate that airway distensibility in sleep
disordered subjects may play a role in their inability to maintain an open airway$^5$.

Another significant change in the airway of apneic patients is the anterior-
posterior (AP) and lateral dimensions. In a normal airway, the major axis is in the lateral
direction whereas in an apneic airway, the lateral dimension narrows and the AP
dimension becomes the major axis (Figure 2.3)$^6$. The shape of the airway (similar to the
area) may change as a result of larger pharyngeal walls and an increase in subcutaneous

Figure 2.2 Axial image of retropalatal level in normal (a) and apneic (b) subject. (From Schwab RJ.
Upper airway and soft tissue anatomy... Am J Respir Crit Care Med 1995; 152:1673-89.)
fat in apneic subjects. The continuous positive airway pressure device (CPAP), the most widely used treatment for OSA, tends to increase airway caliber in the lateral direction and the area and the volume of the airway while decreasing the lateral pharyngeal wall thickness (in normal subjects)\(^6\).

Figure 2.3 Respiratory changes in upper airway dimensions. (From Schwab RJ, et al. Dynamic airway imaging during awake respiration in normal subjects and patients with sleep disordered breathing. Am Rev Respir Dis, 1993; 148:1385-1400.)

The studies just discussed compare normal and apneic patients in an awake state. One study by Trudo\(^7\) describes the changes that occur in the upper airway of normal subjects during sleep, which leads to further understanding of the pathogenesis of OSA. It has been found that during sleep the volume of the RP region is reduced by 19% and the cross-sectional area is reduced by 228%. The RG region does not exhibit the same reduction, which explains why OSA affects primarily the RP region. Furthermore, there is significant posterior movement of the tongue during sleep. These factors, which occur in normal subjects, may be magnified in apneic subjects who tend to have larger tongues, pharyngeal walls, and soft palates, more fat in the subcutaneous regions, and more
distensible airways. Figure 2.4 illustrates the change of position of the soft palate and dimensions of the RP region during sleep compared to wakefulness.

Unfortunately, the pathogenesis of OSA is complex and cannot simply be characterized by the pharyngeal anatomy. In addition to anatomy, it is important to consider the factors involved in maintaining patency (openness) of the airway. During inspiration the upper airway pressure becomes negative yielding a tendency for the airway to collapse. Therefore, the dilator muscles surrounding the airway must become more active to counteract the upsurge of negative pressure. In contrast, during expiration, the upper airway pressure becomes positive and no muscle activity is necessary to prevent upper airway collapse. It has been shown that a patient undergoing repetitive hypoxia also experiences a decrease in genioglossal (the muscle of the tongue) activity,
which may lead to airway collapse. Besides muscle activity, lung volume, vascular effects, and neuromuscular activation are all important to maintaining upper airway patency.\textsuperscript{8}

Lastly, there are many epidemiological factors involved in OSA. These factors include obesity, male sex, and age. Obesity has an effect on pharyngeal anatomy and consequentially obese people are more vulnerable to upper airway obstruction. The mechanisms to explain why gender and age effect the predisposition of a person to OSA are unclear, but studies have shown a higher prevalence of OSA in males and elderly\textsuperscript{8}.

### 2.3 Diagnosis of Obstructive Sleep Apnea

As stated above, the gold standard in diagnosis of obstructive sleep apnea is polysomnography (PSG). PSG records the following parameters: oxyhemoglobin saturation, electroencephalography (EEG), electromyogram (EMG), oronasal airflow, respiratory sound by a microphone, electrocardiogram (ECG), electro-oculogram (EOC), chest wall effort, and body position. After the polysomnogram has been performed, registered polysomnographic technologists determine the Respiratory Disturbance Index (RDI) of subjects by finding the average number of hypopneas and apneas per hour of sleep. A hypopnea is defined as a 50% reduction in airflow for > 10s associated with a > 4% fall in oxygen saturation and/or arousal. Apneas are defined as cessation of airflow for > 10s. Furthermore, the following definitions are generally accepted: (1) normal subjects have an RDI < 2 events/hour; (2) snorer/mildly apneic subjects demonstrate snoring and/or RDI < 15 events/hour; and (3) apneic patients have an RDI > 15
One major issue with PSG is the lack of consistency between technologists on the definition of what constitutes a hypopnea, although the reproducibility of PSG is generally good for patients with mild or severe OSA. The analysis of PSG is an extremely labor intensive process and therefore, numerous people have tried to utilize computer algorithms to interpret the results, although the algorithms have not been standardized.

Clinicians have also attempted to simplify diagnosis by limiting the number of monitoring devices used; the most simplified being a 1-channel recording of oximetry. Unfortunately, a negative predication of OSA based simply on oximetry does not rule out the diagnosis. Home diagnosis systems have been developed that also employ fewer measurements common to PSG. This allows patients to be tested in the comfort of their own home, but many times connections become loose and data becomes unusable and the study must be repeated. The home diagnosis systems have been most successful in their ability to evaluate response to therapy.

Since PSG tends to be uncomfortable, time consuming, and costly, many clinicians employ some form of screening before recommending diagnosis by PSG. The best predictors of OSA include body mass index, age, pharyngeal examination, and bed-partner observed apneas, although these subjective predictions only have a 50% success rate. Many predictive models have been developed to increase the predictive probability. The most common patient complaint is excessive daytime sleepiness, but as a predictor, it is difficult to quantify. Some clinicians choose to use a simple questionnaire called the Epworth Sleepiness Scale (figure 5), which attempts to quantify a person’s sleepiness based on his or her likelihood to fall asleep in certain active situations. The
Epworth Sleepiness Scale runs from 0 – 24, where 0 – 6 indicates that a person is getting adequate sleep, 7 – 8 is about the average score, and persons who score above 9 should seek advice and further testing from a sleep specialist. Obstructive sleep apnea subjects tend to score in the range of 7 – 16. Nevertheless, there is no way that the Epworth Sleepiness Scale can definitively distinguish between OSA and other sleep disorders such as narcolepsy.

<table>
<thead>
<tr>
<th>EPWORTH SLEEPINESS SCALE</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 = would never doze</td>
</tr>
<tr>
<td>1 = slight chance of dozing</td>
</tr>
<tr>
<td>2 = moderate chance of dozing</td>
</tr>
<tr>
<td>3 = high chance of dozing</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Situation</th>
<th>Chance of dozing</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sitting and reading</td>
<td></td>
</tr>
<tr>
<td>Watching TV</td>
<td></td>
</tr>
<tr>
<td>Sitting inactive in a public place (e.g. a theater or a meeting)</td>
<td></td>
</tr>
<tr>
<td>As a passenger in a car for an hour without a break</td>
<td></td>
</tr>
<tr>
<td>Lying down to rest in the afternoon when circumstances permit</td>
<td></td>
</tr>
<tr>
<td>Sitting and talking to someone</td>
<td></td>
</tr>
<tr>
<td>Sitting quietly after a lunch without alcohol</td>
<td></td>
</tr>
<tr>
<td>In a car, while stopped for a few minutes in the traffic</td>
<td></td>
</tr>
</tbody>
</table>

Figure 2.5 Epworth Sleepiness Scale

In a recent study by Virkkula et al., sleep-disordered breathing was detected using an esophageal catheter. An increased esophageal pressure correlated significantly to oxygen desaturation index and obstructive sleep apnea\textsuperscript{11}. The study concluded that employing esophageal pressure monitoring with a limited PSG to determine candidates
for a complete PSG was a successful and cost effective method to diagnose OSA. The negative aspect of this technique is the invasive nature of an esophageal catheter.

Lastly, various imaging techniques have been used to study the pathogenesis of OSA, but few studies\textsuperscript{12} have been done on their usefulness as a diagnostic modality. Primarily, the difficulty lies in the expense of CT and MRI screenings. Furthermore, the radiation of CT is a deterrent. Ultrasonic imaging has recently surfaced as a potential diagnostic tool because of its ability to image the soft tissue structures of the airway, tongue and muscles at low cost\textsuperscript{3}. 
3 DETECTION OF OBSTRUCTIVE SLEEP APNEA USING EDGE DETECTION OF ULTRASOUND IMAGES

3.1 Rationale for Employing Edge Detection Algorithms

Using medical ultrasound imaging, it is possible to obtain images of the area of the pharynx involved in obstructive sleep apnea. Given that an image displays a portion of the airway, the simplest way to determine if a patient has an obstruction is to establish if the image presents an open or closed airway. It is a simple task for the human eye to look at an image and segment it according to its boundaries, and therefore, one could easily differentiate between an image of the airway that is obstructed and one that is open. In extension, it could be possible to develop a computer algorithm to perform the same task as the human eye.

One method to automate the detection of an image that contains an open airway is to use an algorithm that detects the boundaries (edges) of the airway within the image. Normally the airway is open and provides a practically perfect acoustic reflector, which is characterized by significant signal dropout beyond the tissue/air interface. The signal dropout is due to the strong reflection of acoustic signals at the tissue/air interface. The reflection coefficient is defined by $R = (Z_1-Z_2)/(Z_1+Z_2)$, where $Z$ is the acoustic impedance of the substrate. At a tissue/air interface, the reflection coefficient is approximately 0.99, implying that nearly all of the sound energy is reflected and very little energy is left to propagate further into the substrate. In addition, the signal
immediately above the air passage is enhanced due to the strong reflection coefficient at
the tissue/air interface. Therefore, each image should contain a reliable image signature
when the airway is in a normal, open state.

In the broad field of image processing, the goal of edge detection is to create an
algorithm that can segment or discern the borders that define a particular image. The key
to successful edge detection algorithms is an image with ‘good energy.’ An image has
good energy if the objects of interest within the image are defined by homogenous
regions that are separated by discernable boundaries from a background of different
intensity. Unfortunately, the multiplicative nature of the noise in ultrasound images,
known as speckle, yields an image without strong object borders. Therefore, it is
necessary to perform nonlinear filtering of the image to smooth the speckle and
accentuate the boundaries. There are many nonlinear filtering methods available, but
Speckle Reducing Anisotropic Diffusion (SRAD)\textsuperscript{13} is specifically targeted to smooth the
speckle in ultrasound images. Yongjian Yu and Scott Acton developed SRAD\textsuperscript{13}. Once the
ultrasound images have been preprocessed, then a common edge detection algorithm can
be employed.

3.2 The Methodology of Active Contours

3.2.1 Preprocessing Ultrasound Images Using SRAD

As mentioned above, it is imperative to use a non-linear, image-processing filter
to reduce the appearance of speckle and strengthen the borders in ultrasound images
before edge detection is employed. SRAD is a superior filter for ultrasound images
because it excels in mean preservation, variance reduction, and edge localization. Other speckle reducing filters are inferior because 1) they are sensitive to the shape and size of a filter window, which causes over-smoothing or under-smoothing to occur, 2) they only inhibit smoothing near edges, whereas SRAD enhances edges, and 3) all smoothing is inhibited in the vicinity of an edge instead of simply inhibiting smoothing in the directions perpendicular to the edge while encouraging smoothing in directions parallel to the edge\textsuperscript{13}.

Given an initial intensity image $I_0(x, y)$ with no zero values over the image support $\Omega$, the output image $I(x, y; t)$ is found using the following PDE\textsuperscript{13}:

$$\frac{\partial I(x, y; t)}{\partial t} = \text{div}[c(q)\nabla I(x, y; t)]$$

(1)

$$I(x, y; 0) = I_0(x, y)$$

(2)

$$\left[\frac{\partial I(x, y; t)}{\partial \vec{n}}\right]_{\partial \Omega} = 0$$

(3)

where $\partial \Omega$ denotes the border of $\Omega$ and $\vec{n}$ is the outer normal to the $\partial \Omega$. For this specific application, $c(q)$ is given by:

$$c(q) = \frac{1}{1 + \frac{q^2(x, y; t) - q_0^2(t)}{q_0^2(t)(1 + q_0^2(t))}}$$

(4)

where

$$q(x, y; t) = \sqrt{\frac{1}{2} \left( \frac{\nabla I}{I} \right)^2 - \frac{1}{16} \left( \frac{\nabla^2 I}{I} \right)^2}$$

(5)
\[
q_0(t) = \frac{\sqrt{\text{var}[z(t)]}}{z(t)}
\]  

(6)

\(\text{var}[z(t)]\) and \(\overline{z(t)}\) are the intensity variance and mean over a homogeneous area at \(t\), respectively. A discretized version\(^{13}\) of this PDE was implemented in Matlab (Mathworks, Natick, MA) and used to preprocess all ultrasound images of the pharynx before edge detection was employed.

### 3.2.2 Implementation of Active Contours

One established digital image processing method for edge detection and segmentation is to apply parametric active contours. Parametric active contours, also known as snakes, are parametric curves defined in an image that can move under the influence of internal or external forces. Internal forces are a result of a set of constraints placed on the curve itself whereas external forces are computed from the image data. In this application, the goal of the snake is to lie down onto the boundaries of the tissue/air interface in the ultrasound images of the airway\(^{14}\).

The initial snake is a parametric curve defined within the image region by 
\[X(s) = [x(s), y(s)], s \in [0,1].\]  

The snake’s movement throughout the image seeks to minimize the energy functional:

\[
E = \int_0^1 \left[ \frac{1}{2} \alpha |X'(s)|^2 + \beta |X''(s)|^2 \right] + E_{\text{ext}}(X(s)) ds
\]  

(7)

where \(\alpha\) and \(\beta\) are non-negative weighting parameters that control the snake’s tension and rigidity, respectively, and are internal energy parameters. For this specific image
processing design, the external energy function is formulated so that it is small at the boundaries defining the airway\(^1\):

\[
E_{\text{ext}}(x, y) = -\|\nabla I(x, y)\|^2 \tag{8}
\]

where \(\nabla\) is the gradient operator. In order to minimize the energy functional, \(E\), the snake must satisfy the Euler equation\(^1\):

\[
\alpha X^{\prime\prime}(s) - \beta X^{\prime\prime\prime}(s) - \nabla E_{\text{ext}} = 0. \tag{9}
\]

The Euler equation can be broken into a force balance equation:

\[
F_{\text{int}} + F_{\text{ext}}^{(p)} = 0 \tag{10}
\]

where

\[
F_{\text{int}} = \alpha X^{\prime\prime}(s) - \beta X^{\prime\prime\prime}(s) \tag{11}
\]

\[
F_{\text{ext}}^{(p)} = -\nabla E_{\text{ext}}. \tag{12}
\]

A solution to the Euler equation can be found by making the snake dynamic. This requires making the snake a function of time \(t\) as well as \(s\)- i.e., \(X(s, t)\). By setting the partial derivative of \(X\) with respect to \(t\) equal to the left hand side of (9), a solution for (13) is reached when the solution \(X(s, t)\) stabilizes and the term \(X_t(s, t)\) goes to zero.

\[
X_t(s, t) = \alpha X^{\prime\prime}(s) - \beta X^{\prime\prime\prime}(s) - \nabla E_{\text{ext}} \tag{13}
\]

The above equations can be discretized and then implemented in Matlab based on the following update equations\(^1\):

\[
x^{t+1} = (A + I)^{-1} \left\{ \gamma x^{t} + \kappa u_x(x^{t}, y^{t}) + \kappa_p b_x(x^{t}, y^{t}) \right\} \tag{14}
\]

and

\[
y^{t+1} = (A + I)^{-1} \left\{ \gamma y^{t} + \kappa u_y(x^{t}, y^{t}) + \kappa_p b_y(x^{t}, y^{t}) \right\}, \tag{15}
\]

where
\[
A = \begin{bmatrix}
c & b & a & a & b \\
b & c & b & a & a \\
a & b & c & b & a \\
& & & & \ddots \\
a & b & c & b & a \\
a & a & b & c & b \\
b & a & a & b & c \\
\end{bmatrix}
\] (16)

\(x^t\) and \(y^t\) are \(n\)-dimensional column vectors denoting the snake’s position at time \(t\) and \(x^{t+1}\) and \(y^{t+1}\) are the updated snake positions at time \(t+1\). \(u(x^t, y^t)\) is the external energy based on the normalized gradient of the gradient magnitude (equations 8 and 12) of the image at time \(t\). \(b(x^t, y^t)\) is an internal pressure force that controls the expansion of the snake like a balloon under high pressure. \(\gamma, \kappa, \kappa_p\) are weighting parameters. \(A\) is an \(n \times n\) symmetric matrix where \(a = \beta\), \(b = -(4\beta + \alpha)\), and \(c = 6\beta + 2\alpha\).

Iterations of equations (14) and (15) are continued until they stabilize. The number of iterations before termination and the values for \(\alpha, \beta, \gamma, \kappa, \) and \(\kappa_p\) are application specific and are determined empirically.

### 3.3 Results of Edge Detection of the Airway

Edge detection was performed on ultrasound images of a tissue mimicking phantom and the human pharynx. In this section, a few of the results are presented. All ultrasound scans were performed on the Sequoia 512 manufactured by Acuson. The data was transferred from the ultrasound system to a PC in JPEG format.
Prior to running SRAD and the snake algorithm on ultrasound images of the human pharyngeal region, a test case was created using a tissue mimicking phantom. The ultrasound image in Figure 3.1(a) simulates an airway in the open state. Edge detection on this image is less complicated than a human airway because the shape of the object is regular and the speckle is homogeneous throughout. The image insufficiently mimics an airway because the acoustic impedance of the ball is significantly larger than that of air, and therefore, the reflection coefficient at the boundary is much smaller. Consequently, the simulated image does not demonstrate the characteristic signal dropout and strong reflection at the tissue/air interface that can be expected in a real ultrasound image of the airway. Figure 3.1(b) is the image after preprocessing with SRAD and Figure 3.1(c) shows the initialized snake (yellow) and the snake after evolution (red). The parameters used in SRAD and the snake algorithm are given in Table 3.1.

![Figure 3.1](image)

*Figure 3.1 Ultrasound image of a tissue mimicking phantom (a). The image after preprocessing with SRAD (b) and after edge detection (c). The initial snake is shown in yellow and the final snake is shown in red.*
After obtaining satisfactory results of edge detection of the simulated airway, ultrasound images of a human (non-OSA subject) pharynx in the open and obstructed state were obtained and processed. Figures 3.2 and 3.3 are examples of the airway in the open and obstructed states, respectively, where (a) is the original image, (b) is the image after preprocessing with SRAD, and (c) shows the initialized snake (yellow) and final snake (red). To obtain the images, the ultrasound transducer was placed below the chin, using the hyoid bone as a reference (Figure 3.4). The pharyngeal region can be imaged by angling the transducer in the plane just above the hyoid bone. We were advised by Dr. Suratt to simulate the obstruction by moving the tongue rostral and posterior in order to occlude the pharynx such that a snoring noise could be made. These figures qualitatively demonstrate the success of the snake algorithm in detecting the boundary of the airway. Table 3.2 gives the parameters used in SRAD and the snake algorithm.

<table>
<thead>
<tr>
<th>SRAD</th>
<th>SNAKE ALGORITHM</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\Delta \tau$</td>
<td>$\alpha$ $\beta$ $\gamma$ $\kappa$ $\kappa_p$</td>
</tr>
<tr>
<td>0.05</td>
<td>200</td>
</tr>
</tbody>
</table>

*Table 3.1 Parameters used for SRAD and the snake algorithm applied to the tissue phantom.*

<table>
<thead>
<tr>
<th>SRAD</th>
<th>SNAKE ALGORITHM</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\Delta \tau$</td>
<td>$\alpha$ $\beta$ $\gamma$ $\kappa$ $\kappa_p$</td>
</tr>
<tr>
<td>0.05</td>
<td>200</td>
</tr>
</tbody>
</table>

*Table 3.2 Parameters used for SRAD and the snake algorithm applied to ultrasound images of the pharynx.*
Figure 3.2 (a) Ultrasound image of an obstructed airway of a healthy subject; (b) after SRAD; (c) the initial snake (yellow) and the final snake (red).

Figure 3.3 (a) Ultrasound image of an open airway of a healthy subject; (b) after SRAD; (c) the initial snake (yellow) and the final snake (red).
3.4 Discussion

The results presented above demonstrate that the method of active contours (snakes) effectively detects the boundaries defining the anterior portion of the airway in ultrasound images of an open and obstructed airway of a healthy control subject. Due to signal dropout at the tissue/air interface, it is not possible to detect the posterior boundaries of the airway because it is not visible in the ultrasound images. It is interesting to note that the snake did not expand all the way to the image bottom in either example. I will put forward two explanations. First, there is no internal energy force (boundary) drawing the snake downwards to the bottom of the image. The snake’s movement is partially dependent on an internal energy term that is minimized when the snake lies on an image boundary, and since there is no object border defined in the bottom of the image, the snake is not drawn towards the bottom. Furthermore, although
there is a pressure force that causes the snake to expand, the expansion is limited by the number of iterations of the overall algorithm. Therefore, the snake never reached the bottom of the image.

We believe that these results could easily be extended to subjects that have obstructive sleep apnea when ultrasound data becomes available. While the results are positive, it is worth further discussing the issue of snake initialization and the definition of an algorithmic benchmark that would determine the state of the airway (open or obstructed).

The primary limitation of the snake algorithm is its limited capture range, which requires the initialized snake to be in close proximity to the true border that it is seeking to detect. One method commonly used to overcome this limitation is to add an outward pressure force that drives the snake, which is initialized on the inside of the true border, outwards towards the desired border. This method was chosen for this application because of its simplicity and because it does not increase the computation time of the algorithm. However, the issue becomes determining, empirically, the strength of the pressure force. Should the pressure force be too weak, the snake will not be drawn towards the boundaries. Conversely, if the pressure force is too strong, the snake will balloon over weak edges. In this particular implementation, the weighting parameter for the pressure force was $\kappa_P = 0.68$, and was chosen by observing the final snake after multiple trials.

Gradient vector flow (GVF) is an alternative method to increase the snake’s capture range and its ability to progress into boundary concavities. GVF is calculated as a diffusion of the gradient vectors of a gray-level edge-map and has considerable flexibility
in snake initialization, which increases the capture range of the snake. The tradeoff for increased capture range is a highly complex and computationally intensive algorithm. The improvement in capture range did not merit the increased complexity and computation time for this application at this time, but in the future, full automation of border detection may necessitate it.

After obtaining the final snake, which represents the detected border of the airway, it is desirable to automate the process of determining whether the snake is representative of an open or obstructed airway. One possible method could be to determine a threshold based on the relative area encompassed by the snake such that values above the threshold would be representative of an open airway and values below the threshold would be representative of an obstructed airway. Although the area of the airway is not currently a determinant in the diagnosis of obstructive sleep apnea, it has been proposed to use it as such and it is routinely measured in studies on the pathogenesis of OSA. Unfortunately, edge detection of the airway, as illustrated, was successful only along the boundary of the airway on the side of the ultrasound transducer and consequently, it is not possible to obtain an accurate measurement of the area of the airway. In this case, the transducer was placed on the anterior side of the airway and therefore, only the anterior border of the airway was imaged. This is due to complete signal dropout beyond the tissue/air interface.

Although an accurate measurement of the area of the airway cannot be obtained from the ultrasound images, the lateral width of the airway could be measured accurately. In a study by Schwab et. al.,\(^6\) it was concluded that the lateral dimensions of the airway, specifically at the nasopharynx, retropalatal, and retroglossal levels, are significant in
differentiating an OSA subject from a normal, healthy subject (Figure 2.3). Therefore, determining the state of the airway (open or obstructed) could be automated based on a threshold of the lateral dimension of the airway.

After determining the state of the airway, another concurrent issue is to determine whether the obstruction is due to an apneic event or some other cause, e.g. swallowing. It is hypothesized that this problem can be eliminated with a transducer designed to image a volume of the pharynx, and hence the work on the 2D ultrasound transducer presented in chapter 5.
4 DETECTING TONGUE MUSCLE MOVEMENT IN ULTRASOUND IMAGES

4.1 Rationale for Employing Motion Detection Algorithms

In a more advanced application, as suggested by the Siegel article, we will try to detect the tongue motion that precedes the obstruction of the airway.

The objective of motion detection algorithms is to perform mathematical cross correlation to determine the magnitude and direction of the movement of a selected window within an image. There are many implementations of motion detection algorithms, so it is important to determine which algorithm will perform the fastest in a particular application. In a study by Friemel et al.\textsuperscript{17}, computer simulations compared the performance of three motion detection algorithms for two-dimensional speckle-tracking application in ultrasound. The three algorithms that were tested are normalized correlation, non-normalized correlation, and the Sum of Absolute Differences (SAD). It was concluded that normalized correlation and SAD performed similarly and were able to track with a window as small as one pixel (or speckle). The following section describes a template tracking algorithm based on the minimum SAD to determine the magnitude and direction of tongue movement in a sequence of ultrasound images.

4.2 Method of Motion Estimation Using MSAD
Siegel\textsuperscript{3} noted that for every apneic event there is a corresponding ultrasonic event that can be characterized by relaxation of the geniohyoid and mylohyoid muscles followed by posterior or inferior movement of the tongue base to obstruct the airway. Therefore, the tongue follows a distinct pattern of motion during an apneic event. If we can obtain a sequence of ultrasound images of the muscles of the tongue and the tongue base during apneic events, it will be possible, using an image-registration-based motion sensing technique, to track the magnitude and direction of motion. The magnitude and direction of tongue motion can then be quantifiably compared to the characteristic motion of the tongue observed during an apneic event.

Image-registration-based position techniques have been widely used in the ultrasound field as a means to expand the field of view\textsuperscript{18} or reconstruct 3D volumes of data. In this application, the movement of a specific feature of the image (the tongue) from frame to frame will be tracked. In a sequence of images, image features in motion appear in sequential image frames as slightly displaced from the previous frame. In this technique, motion is sensed by matching similar image features in successive frames.

Figure 7 illustrates two successive frames with an object in motion. Frame \( n-1 \) is used as a reference frame and is divided into one or more window. Each window in frame \( n-1 \) is then translated pixel by pixel about frame \( n \) in search of the location that best matches that image window. The position of the best match is found when the computed sum absolute difference (SAD) of the two windows reaches a minimum. The SAD is based on the following equation:

\[
\epsilon_{m,n} = \sum_{i,j} \left| s_{n-1}(i,j) - s_n(i-m, j-n) \right|
\]  

(17)
A local motion vector can then be assigned, beginning at the center of the window in frame \( n-1 \) and ending where a minimum is found in frame \( n \) (i.e. the best match). For this application, in order to limit processing time, only the windows in frame \( n-1 \) containing a feature of the tongue will be searched. Furthermore, the search area in frame \( n \) will not encompass every pixel because, at a given image acquisition rate, the maximum target velocity sets an upper bound on the maximum distance a feature can move between successive frames. This maximum distance will determine the search area.

![Image Registration Technique](image.png)

Figure 4.1 Image Registration Technique. Frame \( n-1 \) is divided into windows. These windows are translated pixel by pixel through frame \( n \) in search of the best match. Image registration is most successful when the window is chosen to have distinctive features.

After the searching process is applied, a vector map, such as the one illustrated in frame \( n \), is acquired. The average motion is used to approximate the overall tongue motion since there will be discrepancies between the motion of different blocks. The movement of the tongue was recorded.
4.3 Results of Motion Estimation

In this section, the results of motion estimation of the tongue are presented. As before, all ultrasound scans were performed on the Sequoia 512 manufactured by Acuson and the data was transferred from the ultrasound system to a PC in JPEG format.

Figure 4.2 illustrates the placement of the transducer used to acquire midsagittal images of the tongue in motion. Figure 4.3 shows the initial (a) and final frames (b) of a thirteen-frame sequence of ultrasound images of the tongue. In Figure 4.3(a), the yellow circle represents an interpretation of the initial position of the ventral surface of the tongue. The yellow arrow in Figure 4.3(b) illustrates the total magnitude and direction of the movement of the tongue over 13 frames. For this particular sequence, the tongue’s displacement was –5.09 mm in the Z-direction and 0.08mm in the X-direction.

The size of the tracking kernel was 40 pixels (6.06mm) in the Z-direction and 60 pixels (9.09mm) in the X-direction. These dimensions were chosen to encompass a significant portion of the length of the tongue (X-direction) and to fully encompass the width of the tongue (Z-direction) while remaining small enough to exclude other image features. In frame n, the kernel moved through 15 different search areas evenly spaced around the center position of the tongue defined in frame n-1. A search area was defined by ±30 pixels (±4.55 mm) in the z-direction and ±4 pixels (±0.61 mm)
in the x-direction about the center of the search area. The tongue displacement illustrated in Figure 4.3 is the average of the calculated movement from all 15 search areas. The standard deviation of the movement was calculated based on the results from the 15 search areas. The average movement and standard deviation from one frame to the next in both the X and Z directions are presented in Table 4.1. In all cases the standard deviation for movement in the X and Z direction was below 0.51mm.

<table>
<thead>
<tr>
<th>Frames</th>
<th>Average Motion Z (mm)</th>
<th>Average Motion X (mm)</th>
<th>Std Dev Z (mm)</th>
<th>Std Dev X (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 to 2</td>
<td>-1.161</td>
<td>0.087</td>
<td>0.489</td>
<td>0.337</td>
</tr>
<tr>
<td>2 to 3</td>
<td>-1.375</td>
<td>0.069</td>
<td>0.476</td>
<td>0.505</td>
</tr>
<tr>
<td>3 to 4</td>
<td>-0.929</td>
<td>0.179</td>
<td>0.410</td>
<td>0.450</td>
</tr>
<tr>
<td>4 to 5</td>
<td>-1.222</td>
<td>-0.118</td>
<td>0.415</td>
<td>0.366</td>
</tr>
<tr>
<td>5 to 6</td>
<td>-0.559</td>
<td>-0.055</td>
<td>0.288</td>
<td>0.109</td>
</tr>
<tr>
<td>6 to 7</td>
<td>-0.319</td>
<td>-0.004</td>
<td>0.177</td>
<td>0.047</td>
</tr>
<tr>
<td>7 to 8</td>
<td>-0.187</td>
<td>0.023</td>
<td>0.110</td>
<td>0.047</td>
</tr>
<tr>
<td>8 to 9</td>
<td>-0.033</td>
<td>-0.019</td>
<td>0.019</td>
<td>0.023</td>
</tr>
<tr>
<td>9 to 10</td>
<td>-0.006</td>
<td>-0.036</td>
<td>0.005</td>
<td>0.016</td>
</tr>
<tr>
<td>10 to 11</td>
<td>0.023</td>
<td>0.049</td>
<td>0.009</td>
<td>0.013</td>
</tr>
<tr>
<td>11 to 12</td>
<td>0.085</td>
<td>0.019</td>
<td>0.052</td>
<td>0.032</td>
</tr>
<tr>
<td>12 to 13</td>
<td>0.023</td>
<td>-0.024</td>
<td>0.016</td>
<td>0.005</td>
</tr>
</tbody>
</table>

*Table 4.1*
4.4 Discussion

The results presented in this section have positively demonstrated that it is possible to track the motion of the tongue in sequential ultrasound images using the MSAD algorithm. This method has proven to be computationally simple and the standard deviation gives a measure of the accuracy of the results. Methods to improve the tracking ability of this algorithm include obtaining ultrasound images with higher resolution and

![Ultrasound images of the tongue](image)

*Figure 4.3 Ultrasound images of the tongue. (a) initial frame and (b) final frame in a 10 frame sequence. The circle illustrates the initial position of the tongue’s surface and the arrow illustrates the magnitude and direction of the tongue motion.*
greater contrast of image features. If necessary, in future work it may be desirable to filter
the ultrasound images, for example with a median filter, in order decrease the amount of
speckle and improve the tracking accuracy. Lastly, should computational time become an
issue, determining the minimal kernel size and minimum search size within the image
will improve the computational efficiency.

As mentioned previously, Siegel noted a characteristic posterior or inferior
movement of the tongue for each apneic event seen in ultrasound video. This movement
of the tongue has not yet been quantified in current literature, and therefore there is no
standard to compare these results to in the case of OSA subjects.
5.1 The Ultrasound System and its Transducer

Figure 5.1 illustrates the basic components of an ultrasound system. The purpose of the transducer is to convert electrical energy into mechanical energy and vice versa. Typically, a piezoelectric material is used for the transducer because it has an asymmetric atomic lattice that will change its mechanical dimensions when an electric field is applied. Similarly, when the piezoelectric material is strained, an electric field is generated. An input signal is applied as a potential across the piezoelectric material and a longitudinal acoustic wave is excited and transmitted into the imaging medium. A portion of the transmitted acoustic wave may be reflected off the objects in the imaging medium with differing acoustic impedance, and some of the wave will continue to transmit further into the medium. The reflections are then received by the transducer, which excites mechanical resonance within the piezoelectric material that can be transformed into an electrical signal.

Figure 5.1 Ultrasound System
voltage across the element. This voltage signal is then amplified and digitized for further processing. Beamforming is applied and finally the signals are envelope detected and sent through a scan conversion, which matches the strength of the received signal to a corresponding brightness level on a gray scale. The received data is then combined and displayed as a gray scale image on a monitor.

The limiting factor affecting the quality of ultrasound images in an ultrasound system is the transducer. Current transducers are made of either a linear array (N x 1) of elements or a 1.5D (N x M, M < 8) array of elements. These transducers are limited to the formation of B-mode images—2D image perpendicular to the transducer’s face. It is thought that C-Scan images—2D images parallel to the transducer’s face—may be more intuitive to the clinical sonographer. The desire for volumetric and C-Scan images, focusing in the elevation plane, and correcting phase aberrations has lead to the development of 2D transducer arrays. 2D arrays are capable of acquiring a volume of data or a C-Scan image along any parallel imaging plane. Unfortunately, 2D transducer arrays are difficult to fabricate because each element can be as small as 0.2mm on a side, which leads to difficulty in making electrical connections to each element, small clamped capacitance and large electrical impedance resulting in poor sensitivity.

The fundamental component of the ultrasound transducer is the piezoelectric element. Typically, piezoceramics are chosen; specifically lead zirconate titanate (PZT). The industry standard is PZT-5H. Ideally, transducers need to possess five basic properties: 1) electromechanical conversion efficiency, 2) close acoustic match to tissue (Z ~ 1.5 Mrayls), 3) effective energy coupling, 4) minimal electrical and mechanical losses, and 5) the ability to be formed into various shapes and be subdivided
into acoustically and electrically isolated subsections so that the acoustic beam can be both focused and steered. Piezoceramics are usually chosen because they have a very high coupling coefficient \( k_t \approx 40 - 50\% \) and therefore have the ability to be fairly efficient when converting electromechanical energy, they possess a very wide range of dielectric constants \( \varepsilon_S \approx 100 - 2400 \), and low electrical and mechanical losses \( \tan \delta \leq 3\% \) and \( Q_m \geq 50 \). However the ceramics are a very poor acoustic match to tissue \( Z \approx 20 - 30 \text{ Mrayls} \) and do not easily conform to different shapes or allow for subdivisions without dicing. Usually a dicing saw or laser cutting techniques are employed to divide the piezoceramic into elements of a desired pattern.

Each piezoelectric element in a transducer array must be held rigidly and electric connections must be made to each element. In typical linear arrays \((1 \times N \text{ elements})\), while connections are tightly constrained in azimuthally, they are unconstrained in the elevation (perpendicular) direction. The inherent challenge in fabricating 2D arrays is fitting the electrical connection within the footprint of the element. Usually a matching and a backing layer are also added to the transducer. The purpose of the matching layer is to acoustically match the impedance of the piezoceramic \((\sim 30 \text{ Mrayls})\) and tissue \((\sim 1.5 \text{ Mrayls})\) to enhance transmission of the acoustic signal into the test region. The thickness of the matching layer should be \( \lambda/4 \) \((\lambda = \text{frequency/speed of sound}, \sim 0.1\text{mm for 5MHz})\) and it is applied to the top of the piezoceramic. The requisite acoustic impedance of the matching layer follows \( Z = \sqrt{(Z_c*Z_t)} \) where \( Z_c \) is the acoustic impedance of the ceramic and \( Z_t \) is the acoustic impedance of tissue (the desired \( Z \) for a matching layer is approximately 7.1Mrayls). The matching layer is preferably made with silver loaded epoxy \((Z \approx 5.9\text{Mrayls})\) or some other conductive material. The backing layer
is applied to the back of the piezoceramic to provide a \( \lambda/4 \)-mismatching layer, which reduces acoustic transmission into the backing substrate.

### 5.2 2D Transducer Design

The vast majority of ultrasound phased arrays that have been designed and used in commercial practice have been 1D (a single row of parallel elements spaced in the azimuthal direction). In recent years, there has been some growth in terms of developments involving a 1.5D array comprising a small number (\( \leq 8 \)) of elements spaced in the elevation direction. Although some very early preliminary work was performed more than twenty years ago on 2D arrays, progress has proven to be extremely challenging. This results from a combination of fabrication difficulties with the transducer (especially the electrical connections) and the cost and bulk of the required beamforming hardware. The Duke University group, led by S. W. Smith, has been the leading research group in 2D array technology over the past fifteen years \( ^{20,25} \). The center frequency of their devices has increased up to approximately 7 MHz, and the scale of their devices has reduced to the extent that some are fitted within intracardiac catheters \( ^{26} \). It is significant that their arrays have been exclusively sparsely populated (approximately 3% of their elements are fully functioning and electro-acoustically active). In order to obtain electrical connections, Duke has employed thin multilayer flex circuit, which consists of several thin films of polyimide with printed electrical traces routing signals horizontally and plated through vias routing signals vertically \( ^{27} \).
Sparse arrays offer significant reduction in system complexity as compared to fully populated arrays. The tradeoff is costly in terms of beam width, prominent sidelobes, and SNR. Greenstein\textsuperscript{28,30} has demonstrated fully populated 50 x 50 element arrays utilizing a conductive Z-axis electrical backing connected via a demountable Pad Grid Array to flexible printed circuits that attach to the cabling system. Also, preliminary results have been obtained from Erikson\textsuperscript{29} on a 128 x 128 fully sampled hybrid array where a 2D composite piezoelectric receiver array is bonded directly to four large custom integrated circuits. We believe that using printed circuit board techniques for the interconnect will overcome some of the manufacturing complexities of the above-mentioned 2D arrays.

5.3 2D Transducer Fabrication

As mentioned earlier, volumetric imaging of the airway is the motivation for designing a 2D array ultrasound transducer. As beamformers become increasingly integrated, channel counts are increasing and fully populated 2D arrays are becoming feasible. This section describes a fabrication technique for fully populated, 32x32 element arrays operating at 5MHz and further details of the technique may be found in Appendix A.

Simply stated, building ultrasound transducers amounts to isolating N x N elements of piezoelectric material through electrically conductive pathways to the transmit and receive circuitry of the ultrasound system. Making electrical connections to the extremely high density of elements is the primary challenge in the fabrication of 2D
arrays. The challenge is to obtain a complete circuit to every element’s signal and ground electrodes without risking a short circuit either between an element’s signal and ground or between the signal connections of adjacent elements. Using printed circuit board techniques, an interconnect can be designed that fans out an electrical connection from each element. The following is a summery of the steps taken to develop the transducer:

- Prototype a transducer on top of blank FR-4 board
- Design and manufacture the interconnect between the transducer elements and the transmit/receive circuitry
- Build the transducer on top of the manufactured interconnect

Figure 5.2 The design for a 2D array using a 16 layer printed circuit board.

Figure 5.2 illustrates a 2D array affixed to a printed circuit board (PCB) interconnect. Beginning at the top, the figure shows; 1) a common electric layer made from gold-plated polyester; 2) the PZT elements; 3) an adhesive layer made from conductive silver epoxy; 4) the top layer of the PCB consisting of one conductive pad aligned with each element; 5) silver filled vias extending through all layers of the PCB.
from which electrical connections are fanned out to connections elsewhere on the top layer of the board.

The fabrication process of a prototype 2D ultrasound transducer is illustrated in Figure 5.3. First, the surfaces of the FR-4 (the material used to build up printed circuit boards) and piezoelectric PZT are cleaned with acetone and primed (Chemlok AP-131). The PZT (17mm x 17mm) is then connected to the FR-4 using conductive silver loaded epoxy (Chomerics) (Figure 5.3(a)). The conductive silver epoxy has multiple purposes: 1) to bond the PZT elements to the FR-4; 2) to electrically connect the PZT to the FR-4; and 3) to serve as a λ/4 mismatching layer (which reduces acoustic transmission into the FR-4). PZT ceramic is not amenable to microlithographic etching or any other advanced processing technique but, the ceramic must be divided in order to minimize gross interelement coupling. Therefore, the PZT/FR-4 assembly is diced with a diamond blade (Disco, NBC-ZH2040) 0.15mm into the FR-4 to create the 32 x 32 element PZT pattern (Figure 5.3(b)). The PZT is further sub-diced 70% into the PZT in order to achieve an adequate ceramic post aspect ratio so
that the frequency of lateral resonant mode (perpendicular to thickness mode) is well separated from the designed fundamental thickness operating center frequency. The dicing saw blade creates 50 \( \mu m \) kerfs and the resulting elements are 0.48 mm x 0.48 mm with a pitch of 0.53 mm. Figure 5.4 shows the size of the elements and the kerfs after sub-dicing.

Once the PZT is diced on top of the PCB, it is necessary to form a continuous electrical layer on the top surfaces of the 2D array of elements (Figure 5.3(c)). Conductive silver epoxy is used to bond the gold plated PZT electrodes to a gold plated 12 mm polyester sheet. This bond process risks the possibility of silver epoxy wicking down the side of the element and causing a short circuit either between adjacent element signal connections or, more likely, a short between element signal and element ground. A process has been developed to backfill the kerfs with nonconductive epoxy, which mitigates a short circuit, and furthermore, mechanically stabilizes the elements (at the cost of reduced directivity and increased acoustic cross talk\(^3\)). A moat is formed around the array periphery using modeling clay. Low viscosity epoxy is mixed (Hysol RE2039 and HD3561; mixed using 100 part resin epoxy to 30 part hardener, by weight) and warmed to 40°C to further reduce viscosity until it is close to that of water. The epoxy is dripped into the gap at the sides of the array and allowed to wick under capillary forces into the arrays kerfs. Care is
taken to partially fill the kerfs so as to perfectly insulate the bottom of the kerfs. It is important to ensure that no epoxy is allowed to migrate to the surface of any PZT element where it would form an insulating layer that would destroy our ability to form a ground return connection. Once the kerf filler is cured, the gold plated polyester is bonded and allowed to cure. Bond line thickness is minimized using a rubber pad sandwiched between the array and the jaws of a machine vice.

An initial prototype of the 2D array transducer was built on to a blank piece of FR-4 and tested to ensure that electrical isolation and good acoustic performance of the 32 x 32 elements could be obtained.

5.4 Characterization of the 2D Transducer

It is possible to test the fabrication process for the 2D array using a blank circuit board containing a continuous metal surface. In this case, only the elements along the edge of the 2D array are electrically accessible because dicing through the metal surface of the blank circuit board makes the return signal pathway inaccessible for each internal element. The results of electrical impedance, pulse-echo waveform and frequency spectrum, and directivity are presented in the subsequent sections.

5.4.1 Electrical Impedance

To initially characterize the 2D array, the air-loaded electrical impedance was measured using an HP 4192A Impedance Analyzer. The impedance profile detects open or short circuits, determines whether the device is piezoelectrically active, and also
estimates whether the PZT element is damped or acoustically isolated. A schematic view of the test setup for the impedance measurements is shown in Figure 5.5. A 50Ω probe (HP 16047A) was used to contact the rear of the element via the metal surface of the blank circuit board and the ground path was connected to the Mylar foil. Figure 5.6 illustrates the electrical impedance spectra for four elements. Although the results of only four elements are displayed, each element measured was active. The two peaks are a result of the layer of silver epoxy and the Mylar foil broadens each peak. For one element, the second notch in the impedance magnitude was 3.2kΩ at 3.3MHz.
5.4.2 Pulse-Echo

The prototype ultrasound transducer was sealed for use in fluid using a low viscosity silicon (Dow Corning, Sylgard 184). Figure 5.7 displays the pulse-echo response from a single element tested in water using a Lucite plate reflector, a Panametrics transmitter (100V pulsed voltage, 500Hz PRF) and ultrasonic preamplifier. The oscilloscope (LeCroy, LC334AL 500MHz) was used to average (1000x) the output waveform to reduce noise. Figure 5.8 shows that the frequency spectrum of the single element has a 3.25MHz center frequency and a 2.3MHz, –6dB bandwidth.
Figure 5.7 Waveform from the 2D array.

Figure 5.8 Frequency spectrum from a 2D array at 3.8MHz center frequency.
Based on the thickness of the PZT (0.43mm), the expected center frequency is approximately 4.4MHz. This is a deviation from the 3.25MHz center frequency illustrated in Figure 5.8. This can be explained by the change in elastic modulus of the PZT due to the dicing process. As the PZT is diced to form the elements, the equivalent elastic modulus reduces causing a decrease in resonant frequency of the PZT element.

Using the pulse-echo water tank setup and a wire target, an attempt was made to acquire a 2D scan beam plot. A LabView program was used to sweep the transducer attached to a motion controller system (Newport model MM3000) through the scan lines necessary to obtain a beam plot. Due to the number of scans required, acquiring a beam plot was unsuccessful because water eventually permeated through the Sylgard seal and shorted out the elements. Using a single pulse-echo response, a simulated beam plot was generated and is illustrated in Figure 5.10.

Figure 5.9 Wire frame plot of a simulated 2D scan of a wire target.
5.4.3 Directivity

The theoretical angular response for a rectangular transducer element in the far field, assuming a soft baffle, is given by\(^\text{19}\):

\[
A(\theta) = k_w \frac{\varphi}{2\pi} \frac{\sin[(k_w \sin \theta)/2]}{(k_w \sin \theta)/2} \cdot \cos \theta
\]  

where \(k\) is the wave number, defined as \(2\pi f/(\text{velocity of sound})\), and \(w\) is the ceramic width or effective aperture. Figure 5.9 shows a comparison of the normalized directivity of a single element and the above theory. The experimental data was obtained by transmitting off a single element and translating a hydrophone (PVDF-GL-0200, and preamplifier: A17DB-S/N 238; Specialty Engineering Assoc., Soquel CA) in a water tank.

The experimental results show a large deviation from theory. Based on the theoretical formulation, the effective aperture of the single element is two times wider than expected. The deviation is also telling of the significant amount of energy going into the backing of the transducer and the high level of cross talk between elements. Furthermore, the experimental data shows that the beam plot levels off at a normalized amplitude of 0.2 instead of continuing towards zero. This characteristic could be explained by a noise issue in the transmitting equipment and could be solved by increasing the transmit voltage and thereby increasing the signal to noise ratio.
Figure 5.10 Experimental and theoretical results for single-element directivity.

### 5.5 Design of the Interconnect

Once the process of the transducer fabrication was established and verified, an interconnect using printed circuit board (PCB) techniques was developed. The schematic and layout for the PCB was designed using Cadence’s Orcad Capture and Orcad Layout CAD tools. The interconnect consists of an array of 32 x 32 silver filled vias (wires created by drilling a hole through the FR-4 and then filling the hole with conductive material) extending up through all 16 layers of the board and finishing on the top layer with metal, square, pads corresponding to the pattern of the elements of the transducer. The three layers of the PCB directly below where the transducer will be fabricated are
sacrificial blank layers (i.e. these layers do not have any electrical fanout traces but do have the thru hole vias extending through them) to accommodate dicing saw penetration without dicing the signals traces. The remaining 13 layers have various fanout patterns of electrical traces running from the vias below each element out to termination pads (test points) on the top layer. The termination pad provides isolated electrical access to every PZT element. The vias below each element are 0.15 mm in diameter with a 0.53 mm pitch whereas the test points are 0.8 mm in diameter and have a pitch of 1mm. The electrical traces are 75 µm wide throughout the whole board. Figure 5.10 shows the top view of all 16 layers of the PCB. Dynamic Details Inc (CA) has agreed to manufacture the board design for approximately $500 per board.

The high density array of vias corresponding to the footprint of the transducer’s elements pushes current PCB fabrication technology to the limit. The diameter of the vias was kept as small as possible while accounting for the aspect ratio of a 16 layer printed circuit board. Employing state of the art laser-via technology (capable of machining 0.05-0.075mm vias) along with buried vias was considered for this design. Unfortunately, this would only decrease the number of layers in the board required for fanout to 7-8 layers. The manufacturing complexity involved in an 8-layer board, employing buried and laser-drilled vias is enormous, and the cost is characteristic of the complexity. The benefits associated with decreasing the number of layers in the board did not warrant the increase of cost and complexity attributed to the use of buried and laser-drilled vias.

Once the interconnect is manufactured, the final 2D array can be fabricated. The same processes described earlier for building the prototype 2D array would be followed to build the final 2D array onto the PCB. The top surface of the PCB has markers to
precisely align the PZT and the dicing saw. The test points on the PCB are soldered to wires for electrical and acoustic testing.
Figure 5.11 The fanout from the vias below the transducer elements to the testpoints. Layers TOP, INNER 1, and INNER 2 will be diced into when cutting the PZT.
5.6 Discussion

We have demonstrated successful fabrication techniques for a 32x32 element array ultrasound transducer. The initial prototype array demonstrated electrical and acoustic functionality with minimal limitations. In future array iterations, it will be important to resolve the acoustic limitations of the transducer. It is understood that glass reinforced plastic, the material used in PCBs, will couple a degree of acoustic energy from adjacent elements. One suggestion is to design a backing and matching layer for the PZT elements to dampen the acoustic energy lost into the backing block and maximize energy transfer from the transducer into the body. Both layers will be $\lambda/4$ in thickness and made using conductive silver epoxy. Figure 5.12 illustrates the intended acoustic enhancements.

The current 2D transducer was designed to be minimally complex so as to demonstrate proof of concept and a methodical approach will be followed to improve acoustic performance. The results of the prototype 2D array give us significant confidence in our ability to scale our process up for a full 2D array using the fully featured PCB design that is being completed.
Although conceptually simple, the PCB design for the interconnect pushes state-of-the-art PCB manufacturing technology to its limits. For this reason, the PCB design for the interconnect has not yet been manufactured and tested. In recent years, PCB manufacturers have shown significant yearly improvement in minimum trace width and minimum via diameter. We believe that in the years to come, the capabilities of PCB manufacturers to fabricate high density boards with minimal feature size will improve to the extent that the design for the 2D array interconnect will be of nominal complexity and cost.
This thesis has discussed the feasibility of using ultrasound imaging as a diagnostic tool for obstructive sleep apnea. Although the image processing algorithms, active contours and MSAD, have shown the capability to detect apneic events, in all cases an ultrasound transducer engineered to image the volume of the pharynx involved in OSA is necessary. Thus far, we have demonstrated the ability of a snake based edge detection algorithm to define the borders of the airway in open and obstructed images of the pharynx. Furthermore, the MSAD algorithm was successful in detecting posterior motion of the tongue base that is comparable to the motion of the tongue observed in ultrasound video of apneic events. In order to validate the image processing algorithms, it is essential to obtain ultrasound data from OSA subjects. Currently, a human studies protocol is being submitted to the Human Investigations Committee at the University of Virginia to obtain such data.

The electrical and acoustic characterization of the 2D prototype ultrasound array has demonstrated a successful fabrication process. We are currently in the process of completing the design for the PCB interconnect, and upon completion, the PCB will be manufactured and the completed 2D array will be tested in conjunction with the ultrasound preamplifier integrated circuits. As PCB manufacturers overcome limitations on minimum feature size and high density fabrication, the complexities of building the proposed interconnect will become trivial. In any event, once a ‘proof of principle’ device is demonstrated, increasing channel count and center frequency will involve
relatively incremental design effort but will continue to be limited by the capabilities (minimum trace width and via size) of PCB manufacturers.

In the world of microelectromechanical system (MEMS) devices, feature size is no longer a severe limitation, and therefore, MEMS may be a potential solution to the fabrication difficulties of 2D array transducers. Ladabaum\textsuperscript{31} has demonstrated the attractiveness of an electrostatic based (capacitive) microfabricated ultrasonic transducers (MUTs). MUTs have already demonstrated superior performance compared to piezoelectric based transducers in air-coupled applications, and it is believed that they have the potential to reach the same standards in liquid applications. MUTs’ inherent advantages include micromachinability, low cost fabrication, diversity of geometries, and the possibility to integrate electronics on chip.
APPENDIX A: TRANSDUCER MANUFACTURE

The fabrication of a 2D array transducer can be broken into five ‘big picture’ steps:

1. Bond the piezoelectric ceramic to the printed circuit board using conductive epoxy.
2. Dice and sub-dice the ceramic into the desired element pattern.
4. Apply a common conductive layer across the top of the elements.
5. Seal the device for operation in fluid.

Each of these steps is described in detail below, including a listing of all necessary materials and equipment for each step.

1. **Bond the piezoelectric ceramic to the printed circuit board using conductive epoxy.**

Materials:

- piezoelectric ceramic (PZT-5H??)
- printed circuit board (PCB)
- conductive silver epoxy with hardener (1 gm dispensers, ChoBond 584-29)
- primer (Chemlock AP-131)
- 100% ethanol
- solvent for cleaning (acetone)
a) Remove conductive silver epoxy from the freezer and warm to room temperature for 1 hour. Heat the oven to 50°C.

b) Clean the surfaces of the PZT and PCB to be bonded with a solvent and allow the surfaces to dry for a few minutes.

c) Mix 3 parts 100% ethanol : 1 part primer by volume. For safety, mix the ethanol and primer under a hood and wear gloves. Using a clean brush, apply the primer lightly to the surfaces of the PZT and PCB that will be bonded. Allow the primer to dry for 30 minutes taking care to shield the PZT and PCB from dust and particles.

d) Thoroughly mix the silver epoxy and hardener within the dispenser. Apply a small drop of silver epoxy to the PZT and using a razor, spread the silver epoxy to make a thin coat on the surface, removing any excess epoxy.

e) Place the PZT/epoxy surface onto the PCB using the fiduciary markers on the top of the PCB for alignment. Sandwich the PZT/PCB between the acrylic and rubber, with the rubber against the PZT. Use the vice to hold everything in place and apply even pressure (Figure 6.1).

f) Place the vice into the oven for 4 hours at 50°C.
Figure 6.1 Picture of vice holding the PZT/PCB between acrylic and rubber.

2. Dice and sub-dice the ceramic into the desired element pattern.

Materials: black wax (optional)

4 inch silicon wafer (optional)

Equipment: dicing saw (Disco)

diamond blade (Disco, NBC-ZH2040)

Note 1: If using a new diamond blade, be sure to break in the blade by making 10 cuts each at 5 mm/s, 7 mm/s, 10 mm/s, and 13 mm/s.

Note 2: If the PCB substrate is not large enough to be held tightly by the vacuum in the dicing saw, you may choose to adhere it to a 4 inch silicon wafer using black wax.
a) Determine the dicing and sub-dicing pattern necessary to isolate each element based on the pattern of electrical connections in the PCB. Figure 6.2 illustrates the dimensions used for a 32x32 element array with 0.5mm pitch.

b) If sub-dicing is required, perform the sub-dicing of each element cutting approximately 70% into the PZT.

c) Dice into the PCB so as to electrically isolate each element, taking care not to dice through any electrical tracks on the inner layers of the PCB.


Materials: resin epoxy (Hysol, RE2039)
            hardener (Hysol, HD3561)
            modeling clay

Equipment: hot-plate
           stirrer
           oven
a) Mix 10 resin epoxy : 3 hardener by weight. Continue mixing over a warm hot-plate at 40°C to reduce viscosity.

b) Create a small barrier (anti-moat) around the circumference of the PZT elements using the modeling clay.

c) Place small drops of epoxy on one side of the PZT elements and the epoxy will wick under capillary forces through the kerfs of all the elements.

d) Cure for 3 hours at 60°C.

4. Apply a common conductive layer across the top of the elements.

Materials: gold-plated polyester
          conductive silver epoxy with hardener(1 gm dispensers, ChoBond 584-29)

Equipment: oven
           straight razor
           2 pieces of acrylic
           vice

a) Remove conductive silver epoxy from the freezer and warm to room temperature for 1 hour. Heat the oven to 50°C.

b) Thoroughly mix the silver epoxy and hardener within the dispenser. Apply a small drop of silver epoxy to the PZT and using a razor, spread the silver epoxy to
make a thin coat on the surface, removing any excess epoxy. The backfilling of
the kerfs will prevent the conductive epoxy from seeping into the kerfs and
shorting out the elements.

c) Use the silver epoxy to bond a piece of gold-plated polyester across the top of all
the elements. The polyester should be just large enough to cover all the elements.
d) Use the vice to apply even pressure to the PZT/PCB and polyester between the
two pieces of acrylic.
e) Cure for 4 hours at 50°C.

5. Seal the device for operation in fluid.

Materials: Sylgard 184 silicone elastomer, base and curing agent (Dow Corning)

Equipment: stirrer
disposable container for mixing

a) Mix the 10:1 by weight base to curing agent and apply over the entire surface
of the transducer to fully insulate the transducer from fluids.
b) Cure for 48 hours at room temperature or 2 hours at 50°C.
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