ISLAMIC UNIVERSITY OF TECHNOLOGY

Strain Estimation Techniques Using MATLAB Toolbox for Tissue Elasticity Imaging.

by

Sabbir Ahmed (092420)

Tehjib Alim Bhuiyan (092442) &

Waliul Islam Khan (092448)

A thesis submitted in partial fulfillment of the requirement for the degree of Bachelor of Science in Electrical & Electronic Engineering.

Academic Year: 2012-2013

Department of Electrical and Electronic Engineering

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Declaration of Authorship

We, Sabbir Ahmed(092420), Tehjib Alim Bhuiyan(092442), Waliul Islam Khan (092448) declare that this thesis titled **"Strain Estimation Techniques Using MATLAB Toolbox for Tissue Elasticity Imaging"** and the works presented in it are our own. We confirm that

- This work has been done for the partial fulfillment of BSc. in EEE.
- Any part of this thesis has not been submitted anywhere else for obtaining degree.

Submitted By:

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Sabbir Ahmed

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Tehjib Alim Bhuiyan

…………………………………

Waliul Islam Khan

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Approved By:

…………………………………

Prof. Dr. Kazi Khairul Islam,

Thesis Supervisor,

Professor,

Department of Electrical and Electronic Engineering,

Islamic University of Technology.

…………………………………

Prof. Dr. Md. Shahid Ullah

Head of the Department,

Department of Electrical and Electronic Engineering,

Islamic University of Technology.

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Abstract

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Strain Estimation Techniques using MATLAB toolbox for Tissue Elasticity Imaging.

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Ultrasound is a widely used modality for both therapy and diagnosis in medicine and Biology. Currently, in the field of medical diagnosis, ultrasound is responsible for about one in five of all diagnostic images. This research project examines the applicability of strain estimation algorithms to improve the clinical value of tissue elasticity images. Pathological changes can be frequently correlated to changes in soft tissue stiffness. Despite the fact that many cancers often manifest themselves as stiff focal lesions, there location and shape can make them difficult, if not impossible, to detect during a palpation-based physical examination. The mechanical properties of tissue cannot be measured

directly using any current imaging modality. However, an ultrasound-based technique known as elastography has evolved over the last fifteen years which is capable of estimating relative strain distributions in soft tissue. Under certain conditions, these strain images (elastograms) can give a clear depiction of the underlying tissue stiffness distributions, and thus, can be used as a clinical tool for the detection of pathological lesions.

Conventionally elastographic techniques estimate tissue strain by tracking spatial features found in congruent pairs of ultrasonic echo backscattered signals before and after a small, quasistatic compression is applied to the tissue surface via the ultrasound transducer. Although this technique has shown promise from a clinical perspective in detecting both benign and malignant lesions of the breast and prostate, it is sensitive to extraneous motions that ultimately compromise the strain estimation procedure. To alleviate this problem, the applicability of strain estimation algorithms and techniques were examined. Both simulation and experimental (in vitro) results obtained using an elastographic phantom indicate that this approach holds promise to improve the clinical value of the images produced. Also, initial results obtained using a novel elastographic animal model further support the efficacy of spectral-based strain imaging in vivo.

Summary

Medical imaging is vital to modern clinical practice, enabling clinicians to examine tissues inside the human body non-invasively. Its value depends on accuracy, resolution, and the imaged property (e.g., density). Various new scanning techniques are aimed at producing elasticity images related to mechanical properties (e.g., stiffness) to which conventional forms of ultrasound, X-ray and magnetic resonance imaging are insensitive. Elastography, palpography or strain imaging has been under development for almost two decades. Elasticity images are produced by estimating and analysing quasistatic deformations that occur between the acquisition of multiple ultrasound images. Likely applications include improved diagnosis of breast cancer (which often presents as a stiff lump), but the technique can be unreliable and difficult to perform. Practical imaging is based on freehand scanning, i.e., the ultrasound probe is moved manually over the surface of the tissue. This requires that elasticity images are calculated fast to provide a live display, and the images need to present meaningful elasticity data despite the poorly controlled properties of the deformations. This thesis presents technical developments towards clinically practical elasticity imaging. First, deformation estimation is examined to devise algorithms that are both computationally efficient and accurate. Second, the entire image formation process is considered, providing strain data accompanied by indications of accuracy, which are then appropriately scaled and displayed in elasticity images representing the value and reliability of elasticity data.

Displacements are estimated by matching windows of radio-frequency data between pre- and post-deformation ultrasound frames. Robust tracking ensures that displacement estimates can be found by searching over small ranges

without introducing large errors, location estimation corrects a well-known amplitude modulation artifact, and a “weighted phase separation” framework illuminates the scope for optimizing the speed and accuracy of deformation estimators.

Strain estimates derived from each estimated deformation provide a form of elasticity image. A method is devised for predicting the accuracy of each strain estimate, which is first applied for dynamic resolution selection: parameters are automatically modulated to produce images with fixed precision at variable resolution. This indicates the scope for using accuracy indicators, which are applied to greater practical advantage in an interface concept: Nonuniform normalization of strain data leads to “pseudo-strain” images. Values from multiple images are blended adaptively to produce a final display that is reliable, while indicating the level of uncertainty where data are less accurate.

This has made it possible to produce good 2D and 3D elasticity images by freehand scanning. Indeed, a clinical trial has recently been set up to evaluate the utility of this system in various clinical scenarios.

Keywords: medical imaging, 2D ultrasound, 3D ultrasound, elastography, elasticity imaging, strain imaging, deformation imaging, RF signal processing

CONTENTS

Introduction

Diagnostic ultrasound has become the most common medical-imaging modality, and it is far from stagnant as the number of clinical applications for ultrasound continues to grow. Ultrasound is routinely used in all clinical medicine specialties and in many areas of bioscience research. It is a popular modality because it is safe, noninvasive, portable, easy to use, relatively inexpensive, and displays images in real time (104). Even as a mature technology, advances are still being made, ranging from improvements in transducer design to new signal processing algorithms, including the introduction of new modes and applications, such as three-dimensional (3D) imaging. In this review, we focus on the current status of ultrasound imaging and the impact of new applications on the design of current and future ultrasound machines. We discuss the various algorithms, processing requirements, and challenges of the common diagnostic ultrasound modes, including B-mode and color flow imaging. We also discuss the processing challenges of several new algorithms for estimating strain and obtaining the strain image to detect the pictorial status more precisely.

The tactile properties of tissue continue to represent important information for modern medical practitioners. Huge investment has been directed at research and infrastructure for breast screening programmes in developed countries. Irrespective of the movement towards evidence-based medicine, with great emphasis being placed on measuring tangible changes in medical outcomes, manual palpation in the clinical breast examination is still widely regarded as an important procedure, contributing to a lowering ofthe breast cancer mortality rate . On the other hand, breast cancer also exemplifies the limitations of subjective examinations when the tools may be inadequate. For many years it was assumed that training women in self-examination of their own breasts would achieve better medical outcomes, owing to earlier cancer detection, but the accumulated evidence indicates that such training programmes have only one significant result: The rate of biopsies on benign lesions goes up, which may in fact be damaging to health.

DIAGNOSTIC ULTRASOUND IMAGING

Figure 1 schematically illustrates the processing stages of a typical diagnostic ultrasound system. The ultrasound acoustic signals are generated by converting pulses of a 2- to 10-MHz electrical signal (known as the carrier frequency xc) from the transmitter into an acoustic wave using a piezoelectrictransducer.As the acoustic wave pulse travels through the tissue, a portion of the pulse is

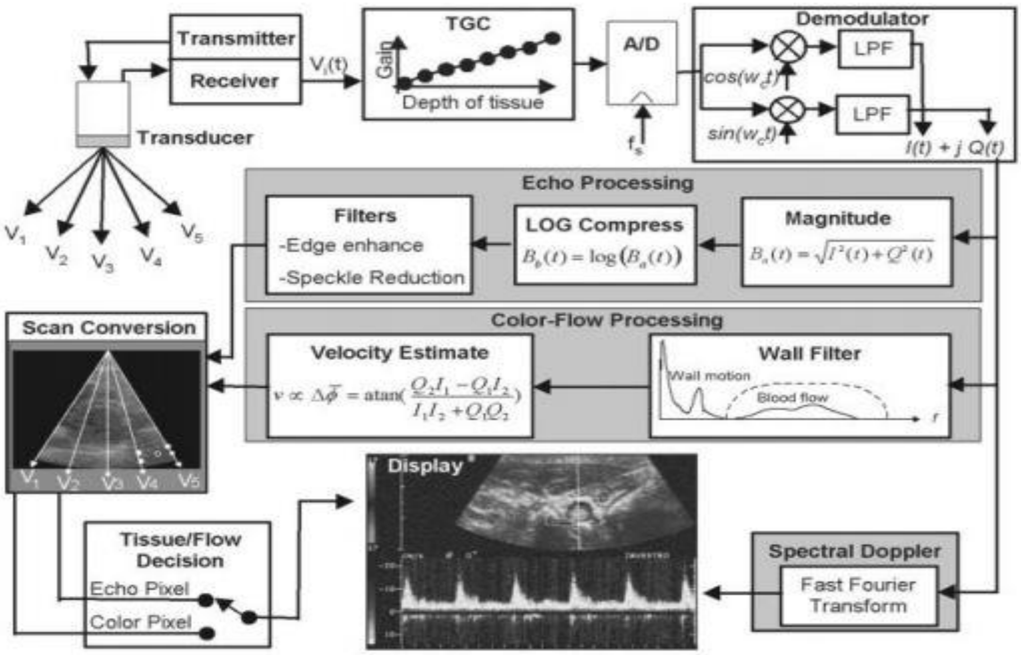


FIGURE 1: Block diagram of a typical diagnostic ultrasound machine.

reflected at the interface of materials with different acoustical impedance, creating a returned signal that highlights features, such as tissue boundaries long well defined beam line. The reflected pulses are sensed by the transducer, and converted into radio frequency (RF) electrical signals. The transducer emits the acoustic pulses at a pulse repetition frequency (PRF) typically ranging from 0.5

to 20 kHz, based on the time for the pulse to travel to the maximum target depth and return to the transducer.

As the acoustic wave travels through the tissue, its amplitude is attenuated. Therefore, the receiver first amplifies the returned signal in proportion to depth or the time required for the signal to return (i.e. time-gain compensation, TGC).The signal's attenuation also increases as xc is increased, limiting the typical ultrasound system to depths of 10–30 cm. Higher frequencies, such as 50 MHz, are limited to a,1-cm depth, but they offer the advantage of improved resolution, which could be useful for applications in eye, skin, and intravascular imaging.

After the RF analog signal is received and conditioned through time-gain compensation, it is typically sampled at a conservatively high rate (e.g. 36 MHz for a transducer with xc 4 7.5 MHz). The demodulator then removes the carrier frequency by using techniques such as quadrature demodulation to recover the return (echo) signal. In quadrature demodulation the received signal is multiplied with cos (xct) and sin (xct), which after low-pass filtering, results in the baseband signal of complex samples, I(t) ` jQ(t). The complex samples contain both the magnitude and phase information of the signal and are needed to detect moving objects, such as blood flow.

The samples of the signal obtained from one acoustic pulse (i.e. onebeam) are called a vector. Today's phased-array transducers can change the focal point of the beam as well as steer the beam by changing the timing of the firing of the piezoelectric elements that comprise the array. By steering these beams and obtaining multiple vectors in different directions along a plane (i.e. V1–V5 in Figure 1), a two-dimensional (2D) image can be formed. Depending on how the vectors are processed, the image can be simply a gray-scale image of the tissue boundaries (known as echo imaging or B-mode) or also have a pseudo color image overlaid. In addition, the spectrum of the blood velocity at a single location over time can be tracked (known as gated Doppler spectral estimation) and plotted in a spectrogram. By combining multiple slices of these 2D images, 3D imaging is also possible for these modes. Depending on the application, typical frame rates can range from 5 to 30 frames per second (fps) for 2D color-flow imaging to.50 fps for 2D B-mode imaging.

Strain Estimation

The first signal processing task for generating a strain image is estimation of the deformation throughout the scan region. The inputs are pre- and post-deformation frames of RF ultrasound data. Throughout this thesis, for simplicity, strain is only estimated in the axial direction. Errors in the fine component have previously been described as “jitter”, with errors in the bulk component described as “peak-hopping”; analysis characterising estimation performance has tended to focus on the size of fine errors. This chapter is about methods for correctly estimating the bulk component. Minimising fine errors is irrelevant unless bulk errors are rare.There is non-zero fine error in every displacement estimate, varying in size depending on data quality and algorithm performance, whereas the occurrence of bulk errors is more of a binary event: the local optimum selected as a displacement estimate either is or is not closer to the true displacement than every other local optimum that could have been chosen. The presence of bulk errors can be highly problematic, because they are amplified by later stages of signal processing, impacting disproportionately on strain estimates calculated using linear filters.

All methods for deformation estimation share common structural features. An individual estimate of displacement relative to the ultrasound probe is calculated at each point on a grid covering the scan region. A window of data centered on the point of interest in the pre-deformation frame is compared with windows of equal size at shifted positions in the post-deformation frame. A match is identified by calculating the similarity between the windows, noting the post-deformation window that registers the highest similarity. Finally, the displacement estimate is equal to the difference between the positions of the pre- and post-deformation windows. The displacement estimate can have both axial and lateral components if post-deformation windows are tested in multiple neighbouring columns. Otherwise, the lateral component of displacement is treated as zero.

Two features that distinguish between different displacement estimators are the similarity measure and the search range. The correlation coefficient between RF ultrasound data in pre and post-deformation windows is a popular choice of similarity measure used throughout this chapter. Alternatives are tested in later chapters. Regarding the search range, many authors favour exhaustive search, where the search range spans a large fraction or all of the data acquired over the dimension being searched.

Correlation coefficients are calculated for every integer shift along the sampled data to find the highest value.A single RF ultrasound frame spanning a square region of tissue typically consists of around 128 columns of RF ultrasound data, called A-lines, with several thousand RF samples along each A-line. With conventional beam forming the axial dimension in RF ultrasound frames is not only far more densely sampled than the lateral dimension, but it is also less likely to be affected by aliasing,has a higher centre-frequency, and higher bandwidth. Therefore, axial displacement estimation is far more accurate than lateral displacement estimation.

Figure 2 shows an example of exhaustive search along an A-line. The correlation coefficient is shown for a wide range of trial displacements. The highest peak is near the actual displacement, but wrong peaks occur at intervals roughly equal to the wavelength associated with the centre frequency of the ultrasound signal. The correct peak happens to be the highest in this example, but some of the wrong peaks are almost as high. Bulk errors occur in an exhaustive search whenever the highest peak is wrong.

The rate of bulk errors can be reduced by limiting the search range to include fewer wrong peaks. However, bulk errors can only be eliminated if the search range contains the correct peak, and no others sometimes a neighboring wrong peak is higher. Ensuring that a limited search range actually includes the correct peak is not trivial. “Tracking” refers to approaches in which displacement is estimated at different points in sequence; each search range is limited based on the values of displacement estimates that have already been calculated at other points nearby. When successful, this eliminates bulk errors.

Tracking in ultrasonic deformation estimation is usually motivated as a means to fast, efficient computation. Speed is an important advantage achieved by shortening the search range, which reduces the number of computations required. Fundamentally this is why exhaustive search is unsuitable for freehand elasticity imaging. However, increased accuracy is also potentially a significant benefit of tracking. In some situations it might even be the main advantage. Tracking does not necessarily reduce the rate of bulk errors. It causes bulk errors whenever the correct peak lies outside the search range. The first bulk error may skew the search ranges for subsequent displacement estimates, resulting in catastrophic failure due to error propagation.

On the other hand, strategies for robust tracking that avoid error propagation offer joint benefits of greater accuracy and greater speed.

This chapter describes and compares exhaustive search and various tracking strategies. Displacement estimates are calculated as integer values (number of samples axially and number of A-lines laterally). Results are presented in the form of displacement images, in which bulk errors are easy to identify by visual inspection. Accurate estimation of subsample displacements is considered in later chapters; this is separate to the issue of avoiding bulk errors. Novel Strategies introduced here include cross-seeding, multi-pass analysis, continuity checking, secure initialization, and coarse lateral tracking.

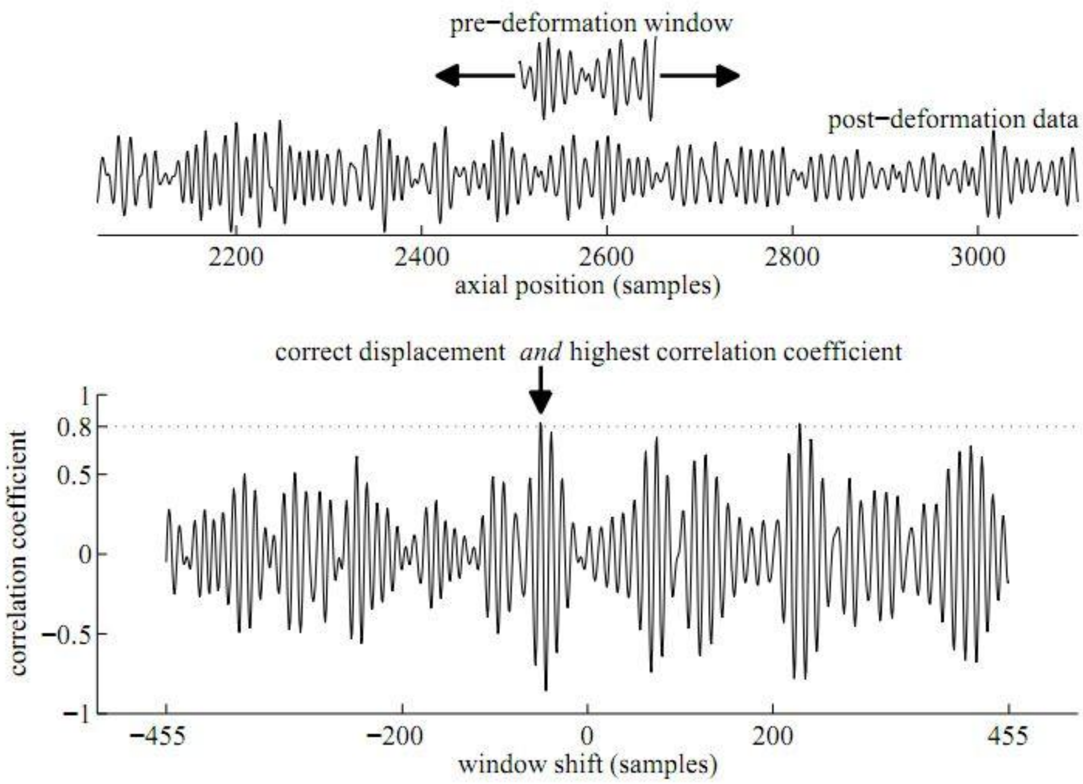


Figure 2: Exhaustive search in the axial direction: (a) A window of pre-deformation RF ultrasound data is shown above the corresponding post-deformation A-line. (b) Correlation coefficients for a large number of trial displacements.

Spectrum Analysis

Theory

An overwhelming number of formally defined mechanical properties could be considered in the analysis of human tissue. There is likely to be a subset of properties for which measurements or images would be clinically useful, and another subset of properties for which measurement or imaging is to some extent possible. The overlap is the type of property that may serve as the basis for successful elasticity imaging. An overview of relevant theory is presented framing the role of qualitative approaches to elasticity imaging.

When analyzing ultrasound images to estimate mechanical properties, the type of motion required is deformation, i.e., compression, expansion, or shear. This is most simply described by a field of displacement data, recording tissue motion as a function of spatial position. A more useful description is in terms of strain, i.e., quantities calculated by taking spatial derivatives of the displacement field to remove components associated with rigid body motion (bulk translation and rotation) which are unrelated to mechanical properties.

In 1D, strain is typically defined as the change in length divided by either the original or the final length. Definitions differ significantly when it comes to large strains (e.g., greater than 10%). Various definitions are suitable for 3D analysis, of which the most popular is also the simplest, and they converge when considering small deformations.

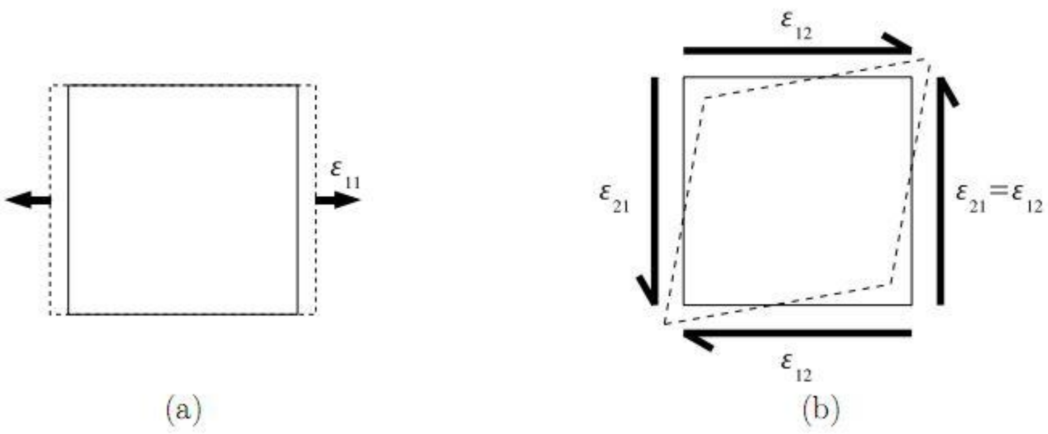


Figure 3: Components of 2D strain, εij , when a square (solid line) is deformed (dashed line): (a) Longitudinal strain (i = j) indicates shortening or lengthening in a particular direction. (b) Shear strain (i =/ j) indicates warping as shown.

Throughout this thesis, “strain” means elements from Cauchy’s strain tensor:

$$\boldsymbol{\varepsilon ij = ½(\partial ui/\partial xj + \partial uj/\partial xi)} \qquad i, j = 1, 2, 3, (1.1)$$

where ui is displacement in direction i, and xj is pre-deformation position in direction j. The different meanings of longitudinal and shear strain are indicated in Figure 3.Strain arises due to changes in the stress field (force per unit area) within the tissue, which in turn follows changes in forces acting internally or on boundaries, called “mechanical excitation”.

The tissue may be squashed by a compression plate, vibrated from its surface, struck on the surface by a small mass, or palpated internally by radiation force. The resulting stress field comprises

(1) an isotropic/hydrostatic component causing change in volume (change in pressure), and

(2) anisotropic components causing change of shape (shear stresses and anisotropic components of longitudinal stresses). Elements from the 3D stress tensor are labeled in Figure 4.

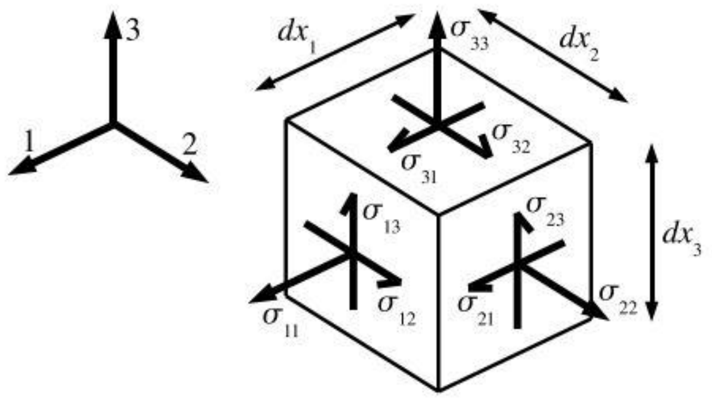


Fig 4: Elements from the 3D stress tensor labelled on an infinitesimally small cube.

In order to estimate mechanical properties, measurements or estimates of deformation must be combined with prior knowledge, estimates, or reasonable assumptions regarding mechanical excitation. In theory, deformation and excitation are linked via constitutive equations dependent on mechanical properties of the tissue. For example, the theory of linear elasticity based on Hooke’s law has been developed extensively by mathematicians neering, physicists and engineers.

It has proven useful for analysing the properties of man-made structures such as buildings and machines. Much of the behaviour of common engineering materials, such as steel and timber, is actually poorly described by linear elasticity, but the theory is valuable because engineers are familiar with situations in which it applies relatively accurately. The usual application for constitutive equations is to calculate deformation given knowledge of mechanical excitation and material properties: the “forward problem”.

Working in reverse, it is sometimes possible to calculate the values that must apply regarding material properties given that a known mechanical excitation causes a known (measured) deformation: the “inverse problem”. This can often be solved, but inverse problems are more difficult than forward problems. If a technique is required for producing quantitative images of a specific mechanical property by solving an inverse problem, one of the associated challenges is the need for accurate deformation measurements, which is considered at length throughout this thesis. Realistically, computational cost is also a significant issue, i.e., the time required to produce each image, and the amount of computing power that must be built into the ultrasound scanner. Both considerations apply similarly with regard to producing images that are only qualitatively related to mechanical properties.

Quantitative imaging presents three other fundamental challenges:

(1) Complexity: This is perhaps the greatest obstacle. The mechanical behavior of tissue is not accurately described by simple, convenient models.

(2) Computational stability: As the number of parameters in a model increases, the chance of a unique solution being available diminishes, and the quantity of data required for achieving acceptably low error increases dramatically.

(3) Unknown boundary conditions: Elasticity imaging is usually based on scanning a limited region of tissue, spanning a small volume (3D scans not encompassing the entire human body) or a slice (2D scans). This causes problems, because some approaches to quantitative analysis only produce correct solutions if the mechanical properties are known all over the boundary.

Elasticity imaging concepts

The following review is not exhaustive, but it should serve as sufficient context for judging the significance of the approach pursued in this thesis. The main ultrasonic concepts are summarized, and differences are highlighted.

Quasistatic:

The basic scanning procedure consists of

(1) recording a “pre-deformation” ultrasound image of unloaded tissue,

(2) applying a load, and

(3) recording a “post-deformation” ultrasound image.

The deformation between pre- and post-deformation ultrasound frames is estimated by a suitable signal processing technique, and analyzed to produce an elasticity image. That abstract covers numerous related techniques. Loading can be applied at a quasistatic rate by various means. For example, research into intravascular elasticity imaging exploits physiological excitation: artery walls deform because of the change in blood pressure over the cardiac cycle. The loading has a frequency of approximately 1 Hz, which seems suitable for quasistatic analysis. In other tissues, quasistatic deformation arises due to breathing. Many investigations have been performed using motion of the ultrasound probe as the source of quasistatic mechanical excitation. Tissue compresses when the probe is pushed firmly against the surface, and relaxes when the probe is held lightly (Figure 5). Suppose that motion of the ultrasound probe causes a uniform change in the axial component of longitudinal stress.

Strain at each point in the image is then inversely proportional to Young's modulus if the behavior is isotropic linear-elastic. The stress field is usually nonuniform, so strain data are ambiguous, but strain imaging is the simplest way of displaying quasistatic deformation data to provide a visual indication of variation in mechanical properties. Techniques exploiting probe movement for mechanical excitation divide into two subcategories. Automated scanning requires extra hardware, whereas freehand scanning is the usual technique employed in conventional sonography. The choice of scanning technique has significant consequences. Automated scanning entails mounting an ultrasound probe on a mechanical actuator to deform tissue by means of a carefully defined movement at the surface. For example, tissue can be compressed by translating the probe precisely 1 mm in the axial direction. The main advantage is that a certain probe movement is repeatable, especially if it is known to lead to good elasticity images in a particular application. Furthermore, the reliable acquisition of suitable pre- and post-deformation ultrasound data means that computations for producing an elasticity image can be performed off-line. This might buy time for elaborate computational methods to maximize the quality of the elasticity images. Equally, the need for correct prior knowledge of a suitable probe movement is potentially a disadvantage. The best movement for quasistatic elasticity imaging is likely to vary from one patient to the next. Automated scanning is cumbersome compared to the freehand approach that clinicians are accustomed to. Usually the sonographer holds an ultrasound probe in his or her hand and moves it manually over the patient's skin to produce images of accessible tissue from whichever angle proves to be the most informative.

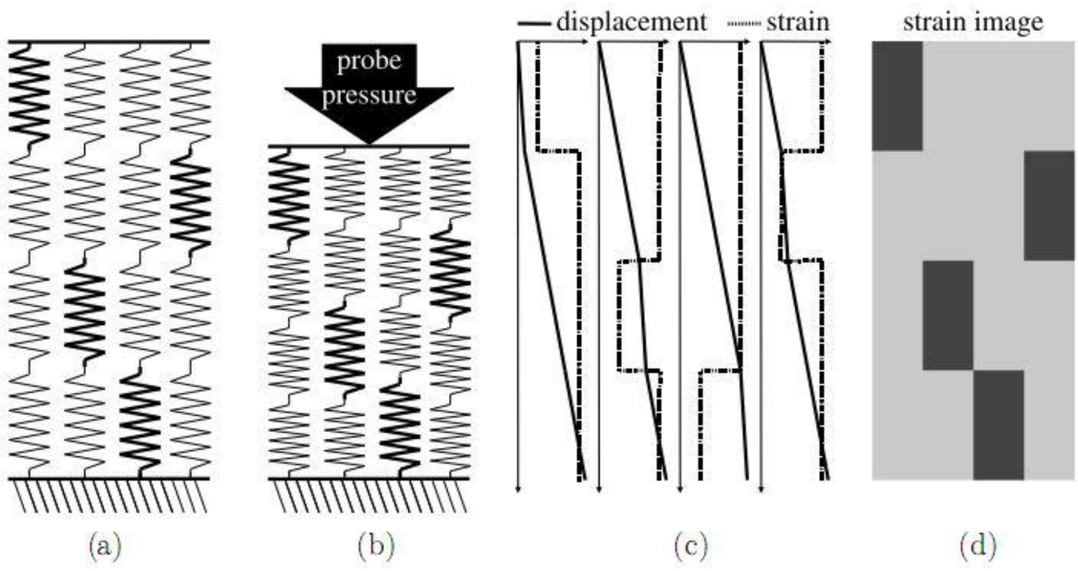


Figure 5: The principle behind quasistatic elasticity imaging.

(a) Inhomogeneous tissue is modelled as a set of springs, where different spring constants represent variation in Young's modulus. Bold springs are three times stiffer than the rest. An ultrasound image records the pre-deformation state of the tissue.

(b) Tissue deforms when the ultrasound probe is pressed firmly against the surface, and a post-deformation ultrasound image is recorded. The deformation here is exaggerated for illustrative purposes (25% compressive strain). It is likely that smaller deformations (strain on the order of 1%) are more appropriate when examining human tissue.

(c) Tissue displacement relative to the ultrasound probe is estimated by analyzing the ultrasound images, and strain is calculated by differentiation. If the stress throughout the tissue is uniform (as in this example), strain is inversely proportional to the spring constant, so the strain in bold springs is three times lower than elsewhere.

(d) Variation in the spring constant is revealed by setting pixel intensities according to the strain values.

There is inevitably some variation in the pressure applied through the probe between consecutive ultrasound frames in a freehand scan, which is a source of continuous mechanical excitation. The freehand scanning approach is hugely appealing, because it can potentially be implemented on conventional ultrasound machines with minimal modification to the hardware and scanning procedure. Elasticity imaging could be made available as a display option to switch on whenever required during routine sonography. Such a development might resemble the previous adoption of Doppler techniques for measuring blood flow, which has rapidly become an indispensable adjunct to conventional ultrasound imaging .

Compared to automated scanning, the deformation in a freehand scan is uncontrolled and unpredictable. Clinicians may be unable to perform prescribed probe movements accurately on a very fine scale. Therefore, freehand deformations may often be too small or too large for inferring mechanical properties, or the type of deformation may simply be unsuitable. Figure 5 shows a compression in which the probe is translated axially. Freehand scanning is bound to throw up more complicated movements, such as rotation about the elevational axis. Some freehand strain images must inevitably be less informative than automated strain images. On the other hand, a sequence of freehand strain images can test a range of deformations, so the best freehand images may be more informative than the best automated strain images. Furthermore, the variation within a sequence of freehand deformations could increase the reliability of inferences relating to mechanical properties.

The value of a strain image sequence must depend on the sonographer's scanning technique, which determines the deformation sequence. A live (real-time) strain display is likely to enhance the benefits of freehand scanning significantly by providing continuous feedback, so the sonographer can adjust his or her scanning technique towards whatever seems to work best. Live strain displays have been demonstrated in the past with frame rates similar to conventional ultrasound imaging (some tens of Hz). However, joint requirements for accuracy and speed place pressure on hardware and algorithm design.

Whether by automated or freehand scanning, quasistatic elasticity imaging may depict in fine detail the geometry of regions distinguished by their mechanical properties. The fundamental limit on resolution is the same as that of the ultrasound images from which the deformation estimates are derived. However, the usefulness of the images will depend on how accurately deformation can be estimated using ultrasound, and how closely inferences based on the deformation estimates can be made to correspond to quantitative mechanical properties of the tissue such as Young’s modulus.

The first point is critical, because raw displacement estimates are spatially differentiated when calculating strain or inferring mechanical properties, which tends to amplify noise. The signal strength can be increased by applying larger deformations, but very large deformations are unlikely to be desirable. Aside from the basic issue of patient comfort, the response is more nonlinear, and hence more difficult to interpret, as discussed earlier. Anyway, large deformations may not improve the signal-to-noise ratio, because an increase in the “deformation signal” causes greater de-correlation between successive ultrasound frames, which in turn increases the level of estimation noise. This topic is considered in detail throughout this thesis. The feasibility of accurately estimating tissue deformation using ultrasound data is a basic premise behind this work, although there may be no simple means of ensuring that all deformation estimates are sufficiently accurate.

On the second point, while nonuniformity in the stress field means that strain images along the lines of Figure 5 are not equivalent to Young’s modulus images, can strain data be converted into Young’s modulus images? The inverse problem is already greatly simplified by the assumption of isotropic, incompressible, linear-elastic behaviour introduced . Accepting the approximate nature of this analysis, it is possible to evaluate Young’s moduli throughout the scan region if the stresses or Young’s moduli are known at all points over the scan boundary. Several researchers have attempted to produce quantitative images on this basis, but unknown boundary conditions are a significant obstacle. It is (probably) impossible to measure all of the boundary stresses, so assumptions are needed regarding boundary values of Young’s modulus. It has been suggested that images of relative Young’s modulus can be produced by assuming uniformity over the boundary, but dramatic errors can occur if the assumption of uniformity is incorrect. The selection of appropriate prior assumptions is emphasised in model-based approaches, which may be more successful for specificwell-constrained tasks.

Challenges associated with the inverse problem should not impede simpler approaches to quasistatic elasticity imaging. Strain images represent useful information despite variation in stress. Boundaries in accurate strain images usually correspond to boundaries with respect to mechanical properties, although the size of differences in strain does not match the the size of differences in Young's modulus. Strain images cannot justify quantitative statements such as, "Young's modulus in the lesion is 2.5 times higher than in the background," but it may often be reasonable to make qualitative observations such as, "The round lesion (diameter of 3 mm) is much stiffer than the background."

To date, strain imaging has been the approach in almost every clinical trial of quasistatic elasticity imaging. In one instance, data were acquired by automated scanning, but freehand scanning has been tested more frequently with a focus on breast and prostate examinations. Successful strain images are sometimes better than conventional ultrasound for identifying tumours. Malignancy and different types of cancer may be distinguishable on the basis of either strain contrast or (perhaps more likely) differences in geometrical appearance between conventional ultrasound and strain images. Reports conclude unanimously that strain imaging could potentially be a useful clinical tool, but none has yet demonstrated a conclusive case for adoption into routine clinical practice. The issue of stress variation and boundary conditions seems not to be a major drawback, because the qualitative information is still useful. The question of how to produce accurate images is more of a problem, because strain images continue to be unreliable . The presence of many bad images makes interpretation difficult and laborious for sonographers, who are typically required to inspect sequences of stored strain images by eye to select those that are informative. This thesis contributes towards improving the reliability of strain images.

Concluding remarks and Future Works

Project summary

In this research project, novel elastographic algorithms and techniques were developed and evaluated with the objective of improving the clinical value of the resultant tissue elasticity images. Conventionally elastographic techniques estimate tissue strain by tracking spatial features found in congruent pairs of ultrasonic echo backscattered signals before and after a small, quasistatic compression is applied to the tissue surface via the ultrasound transducer. Although this technique has shown promise from a clinical perspective, it is sensitive to extraneous motions (i.e., decorrelation noise) that ultimately compromise the strain estimation procedure.

To alleviate degradation in the strain estimation procedure associated with decorrelation noise sources this research project examined the applicability of spectral elastographic techniques. Firstly, a 1-D spectral elastographic simulation was introduced and utilized to analyze the tradeoffs associated with spectral elastography. Specifically, it was demonstrated using simulated Strain Filters that increases in the transducer center frequency and/or bandwidth results in a corresponding increase in the SNRe, strain imaging sensitivity and dynamic range. It was also illustrated that by increasing the underlying spatial window length used to compute the spectral estimates that a corresponding improvement in the strain estimation process may be obtained. The effect of spectral density on strain estimation performance was also evaluated and it was determined that by increasing the spectral density,a corresponding improvement in the strain estimation procedure results.

For our thesis purpose we created the MATLAB toolbox(Fig:).And in this toolbox we use different algorithms which will help the physicians to diagnosis more accurately as these algorithms accumulated at the same place.So they can use it more quickly and effectively than before.

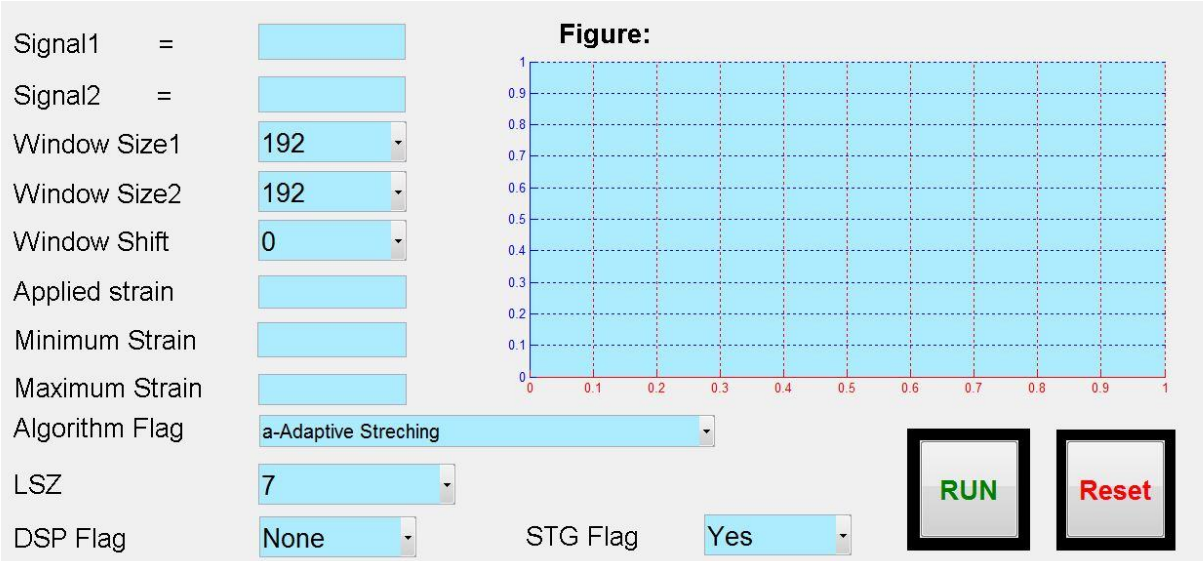


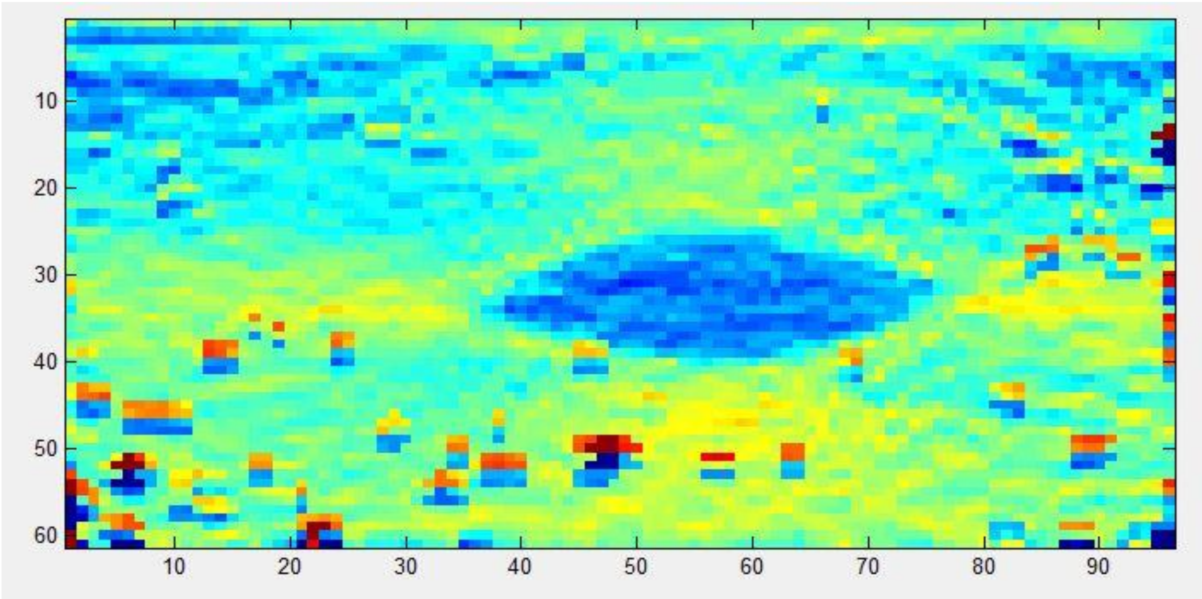
Figure 6: Graphical User Interface using MATLAB toolbox

Algorithms accumulated here are given below:

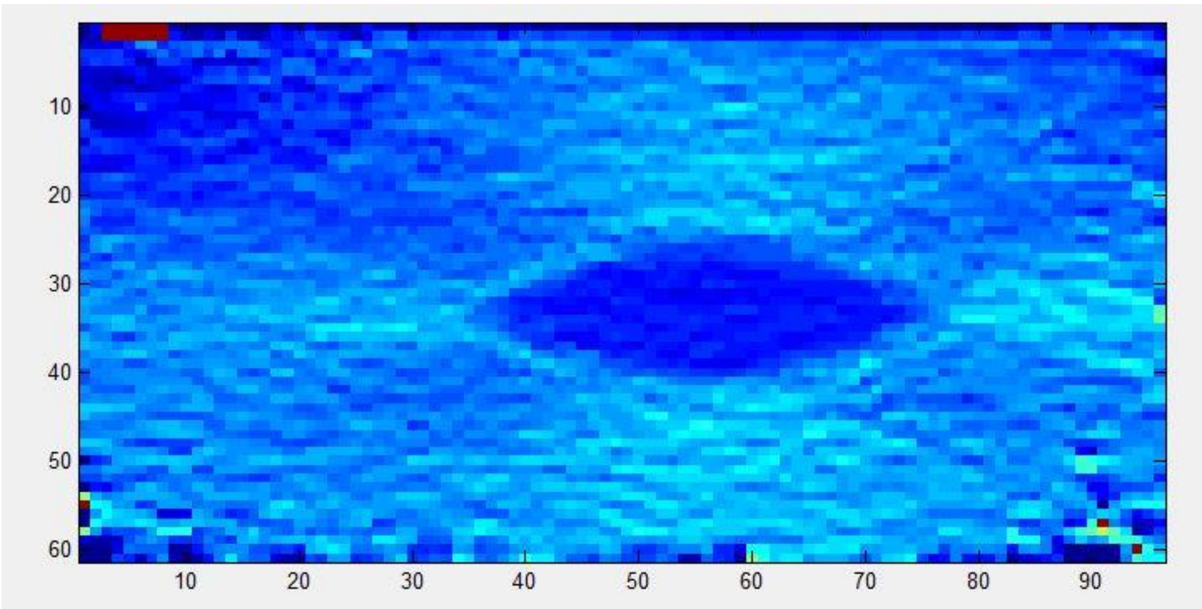
1. Adaptive Stretching
2. Uniform Stretching
3. Gradient(No Stretching)
4. Least Square Stretching
5. Least Square Uniform Stretching
6. Least Square Uniform Stretching-2
7. Least Square Correlation Maximum
8. Least Square Correlation Maximum Uniform Stretching
9. Variable Stretching
10. Least Square Variable Stretching
11. Smooth Spline(No Stretching)
12. Smooth Spline Uniform Stretching

Different results found using different algorithms:

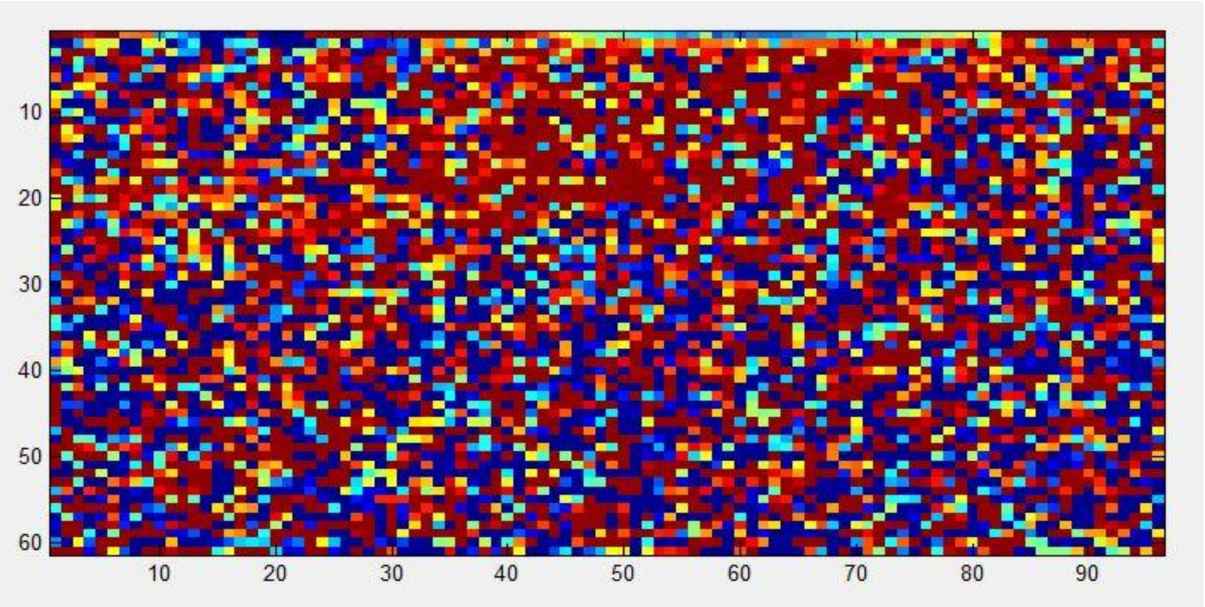
For Uniform Stretching:



For Gradient: It is the most primitive one.It is less time consuming.But for this case the quality of image is not satisfactory.



For Adaptive Stretching: It is time consuming but quality of the image is better than gradient algorithm.



Future Work:

1. Real time Ultra sound:

In this context we will work on real time ultra sound image. In case of emergency the signal can be received by e-mail also. So our toolbox should be designed so to extract the messages from the emailed signal.

2. To make some work about Ultrasound frequency:

When we are studying on ultrasound then we came to know that for deeper region we can't use the high frequency ultrasound. Due to this reason the resolution of the image is not satisfactory. So we want to work on this to improve the quality of image for deeper region.

2. Can we use RF signal instead of US signal?

To make sure, we have to compare the frequency of RF signal with the range of ultrasound used normally.

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