Numerical modeling of ultrasonic data for acoustic elastography

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Abstract

Development of new and improvement of existing tissue characterization methods is in the research focus now due to non-invasiveness and highly informative diagnostic value of procedure. In most techniques mechanical elastic properties of soft biological tissues are investigated from analysis of corresponding ultrasonic data. The latter can be obtained either experimentally or simulated numerically using specific programming tools. The brief overview of ultrasonic data modeling for purposes of elastography is presented in the following report. The document is based on a presentation conducted during the Moscow-Bavarian Joint Advanced Student School in March 2011 and serves more as an extention to the slides rather than an independent scientific paper.

Contents

1 Introduction

1.1 Elastography: definition

Over the last two decades an imaging modality for measurement and visualization of tissue stiffness called elastography has evolved as a quantitative alternative to manual palpation [1, Ophir]. Today this complementary procedure is an important non-invasive tool for the study of soft tissues and diagnosis of pathologies [2, Rouger]. This approach involves imaging tissue (e.g. with ultrasound (US)) under an applied load while measuring the internal motion. The goal is to determine mechanical properties of tissue from which a pathological reference can be proven. Such mechanical attributes of soft tissue may include the shear or Youngs moduli, Poisson's ratio or any of longitudinal or shear strains that occur in tissues as a response to an applied load [3, Ophir].

1.2 Ultrasonic signals analysis

A common approach to determine a resultant deformation is by correlating two subsequent sets of echo-lines obtained prior to and immediately following a small compression. Even a small compression (strain) of the tissue results in a small compression of the signal (similar to frequency modulation). Echo lines are then subdivided into small temporal windows which are compared pairwise by using cross-correlation techniques, from which the change in arrival time of the echoes before and after compression can be estimated. Then local strains are estimated as the gradient of the time delay (or displacement) and visualized in a gray or color scale.

1.3 Block diagram of the elastography method

In order to obtain the information about changes going on inside the object exposed to compression and to establish viability of proposed computing algorithms it is crucial to have a reliable simulation model. The block diagram illustrates the major parts of the computer simulation model used to synthesize ultrasound images from the object in the pre- and post-deformed condition and to calculate displacements.

- The outputs of the mechanical block are longitudinal or shear displacements for each point of a simulated phantom after the external load is applied. The deformation is modeled using Finite Element Method.
- In the second acoustic block the corresponding pre-/post-deformation echo - signals and B-mode images are generated using Field II simulation programme[4, Jensen].
- The purpose of the last block is to analyze these newly obtained simulated ultrasonic data and to estimate tissue motion. To assess the reliability of the algorithm the estimated displacements are then compared to the known displacements computed on the first stage using FEM.

Prevalent use of elastography in the process of ultrasound investigation explains the necessity and practical benefit of central acoustic block modeling which allows one to get realistic images of internal phantom structures and can then be implemented in the general simulation scheme. In modeling the stepwise approach is preferable and the task in general can be divided into following stages:

- transducer simulation and optimization of array parameters (considering the effects of every parameter on ultrasound field);
- setting different geometry and scattering properties of phantom and simulation of corresponding ultrasound images;
- computing new positions of scatterers representing phantom using FEM.

2 Transducer simulation

2.1 Modeling of linear array

To make the process of transducer parameter optimization easier for user and more visual a special graphical user interface was developed. Although it is applicable only for linear array, all necessary settings are provided: option for a complete description of transducer characteristics (including frequency, number, size and shape of elements), manipulating of ultrasound beam such as focusing, apodization, steering angle and general settings which define the type of excitation and medium properties.

2.2 Optimization of parameters

Varying one parameters and keeping constant the rest we can observe the resultant beam patterns and assess the effect these parameters have on the field structure.

Defining criteria of rational choice of transducer parameters are minimization of main lobe width, reduction of the side lobes angles with the main lobe which all results in a more directional beam pattern. From our observations it can be seen that by increasing the size of individual element or the total number of elements as well by increasing the center frequency the desired field structure can be achieved.

2.3 Comparison with experiment

Predicted pressure fields can be compared with measured pressure response[5, Guenther]. It is easy to realize if the model of experimental ultrasound scanner is known. Then all physical dimensions and technical characteristics of commercial linear probe can be found in published specification and equivalent settings are defined for simulation. A satisfactory agreement between simulation and experiment is achievable in the focal plane and then in the far-field, which is clearly not the case in the near-field. In the close proximity to transducer surface mechanical and electrical coupling between elements, cross talk, element nonuniformity and nonlinearities preclude high correlation between real and theorized fields.

3 Simulation of ultrasonic data

3.1 Algorithm of image simulation

After transducer parameters are defined and optimized the next move is to define targets that will be imaged. The imaging of artificial phantom is done by simulating and summing the field from a collection of point scatterers. Homogeneous tissue is, thus, made from a collection of randomly placed scatterers that have a scattering strength with a Gaussian distribution. Multiplying the amplitude or by setting it to zero it is possible to simulate high scattering regions or cysts. After series of processing steps original (raw) RF-signals are transformed into more typical B-mode ultrasound images. Although brightness images look realistic unprocessed echo-signals contain potentially more information on the investigated tissue which is later used for tissue motion analysis.

3.2 Data storage

Both rectangular and polar cross-sectional images can be simulated and saved in DICOM format, which is known to be a standard for handling, storing and transmitting information in medical imaging. By using special tags and fields associated with US it is possible to fill in all necessary information about image, transducer and phantom parameters.

4 Deformation modeling

To estimate tissue motion we need two series of echo-lines pre- and postcompression. Computation of deformed condition of phantom after external load is applied using

Finite Element Method is perfomed in FEMLAB package[6, Zahiri]. The model with circular inclusion being 5 times harder that of the substrate (to imitate nearly uncompressible lesion) is meshed and then is being compressed from the bottom side. It is the same as if the us probe was compressing from top side but ensures that we keep stay in the probes frame of reference in which images are typically processed. As here we simulate non-rigid motion then the interpolation procedure must be used to find the resultant displacement for arbitrary point in the region. On the final stage the second series of post-compression ultrasound echo-lines is to be simulated thus the interlink between mechanical and acoustical blocks of modeling scheme is realized.

5 Conclusion

In conclusion the results which have been achieved are summarized. First the developed GUI allows one set all basic array parameters and observe the field pattern. After weve chosen optimal parameters weve achieved a good correspondence between simulated and measured fields especially in the focal plane and far-field. Varying the spatial positions of scatterers and their scattering amplitude it is possible to simulate realistically looking images with hyper or anechoic regions and then store them in DICOM database. The algorithm of interlinking mechanical and acoustical blocks using Finite Element Modeling of post-deformed scatter positions was also analyzed and potentially the output data can be used for the tissue motion analysis in the last reconstruction block.

References

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