Ultrasound Viscoelastic Imaging of Breast Lesions: A Practical Hybrid Freehand Technique for Data Acquisition

Andrew Di Battista, J. Alison Noble
Institute of Biomedical Engineering, Oxford University, Oxford, United Kingdom
andrew.di-battista@wolfson.ox.ac.uk

Ruth English
Oxford Radcliffe Hospitals NHS Trust
Oxford, United Kingdom

ABSTRACT
Ultrasound imaging of the breast is a standard method in breast cancer screening, along with mammography. The viscoelastic properties of soft tissue can provide supplementary information for radiologists to consider in their assessment of pathology and tissue characterization. Measuring these properties generally entails acquiring a time sequence of ultrasound images and calculating parametric data from them. As images are necessarily accumulated over time, acquisition is limited by the frame rate and memory capacity of the ultrasound machine, and practical considerations such as movement from the clinicians' hands and patient breathing.

This paper describes a technique for hybrid-freehand imaging of viscoelasticity (HYFIVE). It involves acquiring a time sequence of images making use of a simple purpose-built canister enclosure for the ultrasound probe which allows for a stable and accurate manipulation of applied forces, without the need of motors, sensors or other sophisticated and costly parts. A sequence of ultrasound strain images was computed and a first order Kelvin-Voigt tissue model fit to the resulting strain vs. time curves to obtain parametric data related to tissue stiffness and viscosity. Experiments were conducted on both gelatin phantoms and clinical patient data.

INTRODUCTION
A distinctive feature of ultrasound (US) imaging is a high frame rate compared to other medical imaging modalities. Various techniques in ultrasound elastography have been developed to exploit this feature in order to determine parameters which may help differentiate soft tissue types and characterize breast lesions. In [1], cyclical compression was used to map out a spatially varying hysteresis loop; its normalized area was used to generate a parametric image. Another method [2] measures the creep resistance of tissue due to a sudden applied step force. It is this technique which forms the starting point of this paper's work.

MATERIALS AND METHODS
The probe canister device design is depicted in Fig. 1. The US probe is secured firmly in the canister by specially designed caps at the top and bottom (not shown). A weighted ring or hoop (500g) is free to slide up and down the outside canister wall which has tracks cut into it. Track bolts keep the motion of the ring vertical but still allow the clinician holding the apparatus via the ring to manipulate the probe side to side as well as rotate it about the vertical axis. Once the probe is in position, the clinician simply lifts/slides the ring up to produce what can be approximated as a unit step-off force. The probe remains set above the region of interest to free the creep response of the underlying tissue (see Fig. 2).

To represent the tissue response, a simple first order Kelvin-Voigt model was assumed. Conventional elastography models tissue as an ideal spring, with spring constant $k$ and strain $\varepsilon$. The addition of the dashpot introduces viscosity $\eta$ into the equation. It has
been shown [2] that the time constant \( \tau = \eta / k \) can provide a useful tissue characterization parameter.

\[ F = \eta \varepsilon + k \varepsilon \]  

Figure 3-The Kelvin-Voigt Model and governing equation

Strictly speaking, the applied force should be treated as a ramp function. Given a ramp interval \( \Delta \) defined in Fig.2 and the tissue model of Eq.1, it can be shown using Laplace Transform analysis that for \( T >> 0 \), \( t' = t - T \), and taking the overall change in force to be unitary, then

\[ \varepsilon(t) = \begin{cases} \frac{1}{k} \left[ 1 - \frac{t'}{\Delta} + \frac{t'}{\Delta} (1 - e^{-t'/\Delta}) \right], & 0 < t' < \Delta \\ \frac{1}{\eta} \frac{t'}{\Delta} e^{-t'/\Delta} (e^{t'/\Delta} - 1), & \quad \text{for } t' > \Delta \end{cases} \]  

By the method presented in [3], strain was calculated over sequential pairs of images acquired at 13 fps and accumulated to produce the data points shown in Fig.4. The accumulation ran from the last image to the first and was then time reversed giving all the curves a common final baseline of 0% strain.

Curve fitting involved optimizing the \( \eta \) and \( \Delta \) parameters which would be expected to vary spatially; \( 1/k \) could be found empirically from the maximum accumulated strains. As Eq.2 is piecewise continuous, an exhaustive, coarse to fine search over a pre determined range of \( \eta \) and \( \Delta \) values was used to find the fit with the minimum mean squared error (MMSE).

![Curve fitting](Figure 4)

RESULTS

Figure 5 depicts the results from the clinical data of a patient with an invasive ductal carcinoma (IDC). The maximum accumulated strain was approximately 16%. Contrast in the images was defined as the ratio of average strain outside the lesion over the average strain inside the lesion.

DISCUSSION

A time sequence of < 2sec was shown to be suitable for generating parametric data and the resulting high contrast colour mapped images that clearly distinguish between lesion and surrounding healthy tissue. Moreover, the \( \eta \) image in this case displays a 20-30% increase in contrast compared to the standard elastogram.

A dedicated optimization scheme for the curve fitting would be one obvious goal for future work. In addition, making a set of easily adjustable weighted rings would allow for compensation for the variation in tissue density across different patients.

Preliminary results would suggest that HYFIVE may have a practical role in the clinical environment.

ACKNOLEDGEMENTS

The authors would like to acknowledge the cooperation and enthusiasm of the breast screening clinical team at Oxfords Churchill Hospital in assisting in this research.

REFERENCES

