Computers in Industry 69 (2015) 61–71

Contents lists available at ScienceDirect

Computers in Industry

journal homepage: www.elsevier.com/locate/compind

A real time displacement estimation algorithm for ultrasound elastography

Minhua Lu [a], Yanan Tang [a], Ruichao Sun [a], Tianfu Wang [a], Siping Chen [a], Rui Mao [b,*]

[a] National-Regional Key Technology Engineering Laboratory for Medical Ultrasound, Guangdong Key Laboratory for Biomedical Measurements and Ultrasound Imaging, Department of Biomedical Engineering, Shenzhen University, Shenzhen 518060, China

[b] Guangdong Province Key Laboratory of Popular High Performance Computers, College of Computer Science and Software Engineering, Shenzhen University, Shenzhen 518060, China

ARTICLE INFO

Article history:
Received 9 February 2014
Received in revised form 16 July 2014
Accepted 26 September 2014
Available online 24 October 2014

Keywords:
Real-time
Ultrasound elastography
Displacement estimation

ABSTRACT

Motion tracking algorithms to derive the local displacement distribution inside soft tissue from ultrasonic radio frequency (RF) signals are critical for ultrasound-based techniques, especially for ultrasound elastography. Normally, there is a trade-off between precision and computational cost. In this study, we present a framework combined with block matching algorithm (BMA) and phase shift method with prior estimation (PSPE). BMA is first applied to the RF signals, the axial and lateral displacements obtained are then used as a prior estimates of the PS method to calculate a more precision axial displacements. The performance of the algorithm is evaluated with synthetic ultrasound RF data, and it is found that both SNRe and CNRe of our method are significantly higher than those of phase-shift as a prior estimation (PSPE) method. Elasticity phantom and clinical data are also used to verify the usefulness of our algorithm, respectively. The results show that our method is robust to image the complex tissue motions.

1. Introduction

The field of medical imaging is advancing at a rapid pace. Imaging modalities like X-ray radiography, X-ray computed tomography (CT), ultrasound, nuclear imaging, magnetic resonance imaging (MRI), and optical imaging have been used in biology and medicine to visualize anatomical structures as large as the lung and liver and as small as molecules. Ultrasound is considered the most cost-effective among them. It is used routinely in hospitals and clinics for diagnosing a variety of diseases. It is the tool of choice in obstetrics and cardiology because it is safe and capable of providing images in real time. However, until the late 1980s, biomechanics of soft tissue is not any part of acoustics, although qualitative assessment of tissue elasticity by manual palpation has been widely used since ancient times and is still in use today. The principle behind clinical palpation is the significant difference in elastic properties between normal and diseased tissues [1,2], and even normal tissues in different biological status. However, manual palpation is considered to be subjective, inaccurate, and highly operator-dependent, whose efficacy is limited to the abnormities located relatively closed to the skin surface, while it is difficult to detect small and/or deeply located pathological lesions and organs.

Ultrasound has been widely used to differentiate cysts from solid tumors in tumor imaging because liquid-containing cysts are typically echo poor. However, certain solid tumors that are harder than surrounding tissues are missed sometimes by ultrasound because their echogenicity is similar to that of surrounding tissues. These tumors or harder tissues are identifiable if their elastic properties can be imaged. During the past 2 decades several elasticity imaging techniques have been developed for measuring the elasticity of tissues quantitatively, using ultrasound [3–10], magnetic resonance imaging [11–13], and other imaging modalities [14,15]. Compared with conventional morphological images, images of elasticity are able to display the distribution of stiffness/elastic properties of tissue, and thereby provide more valuable diagnostic information. Elasticity images have been shown to be able to provide new opportunities for the detection and diagnosis of cancers in the breast [7,16,17], prostate [18], and liver [19,9], and in other clinical applications [20,21] associated with assessments of the elastic properties of soft tissue.

Ultrasound elastography has been evolving into a useful and promising technique due to its real-time capability and ease of implementation. Among the various elastographic techniques

* Corresponding author. Tel.: +86 755 2653 4207.
E-mail address: mao@szu.edu.cn (R. Mao).

http://dx.doi.org/10.1016/j.compind.2014.09.006
0166-3615/

[22–24], quasi-static ultrasound elastography is particularly popular, whose basic steps are as follows: (1) a set of radio-frequency (RF) signals is collected from the specimen in its undeformed state; (2) the specimen is compressed by external loading, which can be assumed to be quasistatic, and another set of RF signals is recorded; (3) motion-tracking techniques, such as widely used cross-correlation techniques, are applied to estimate the displacement field between the two sets of RF signals recorded in the previous two steps; and (4) so-called elastograms are reconstructed/computed from the displacement field.

Elastography can be broadly classified into two groups based on the mechanisms underlying the generation of the elastograms. In the first group, the relative mechanical attributes are calculated directly from tissue displacements, and axial, lateral, and/or shear strains are calculated from the tissue displacement field and then inverted to produce elastograms [25,26]. Although this interpretation of the strain images may be affected by certain artifacts such as target hardening and bidirectional shadows [27], these images have received considerable attention over the past 20 years because it is feasible to obtain them in real time using a commercially available ultrasound system. In the second group, intrinsic elastic parameters are reconstructed quantitatively from the measurement data (also tissue displacements) [28–31]. The reconstruction process involves finding an optimal solution to an inverse problem with constraints, such as the assumption of plane-strain situation, knowledge of the boundary conditions, and several other assumptions. Modulus elastograms, such as the distribution of Young's modulus, can be thus obtained with the optimal solution. Modulus elastograms can greatly suppress the artifacts in strain images, but their quality is highly dependent on the precision of displacement measurements, which can be easily destroyed by many uncertainties owing to the ill-posedness nature of the inverse problem, while the robustness of many reconstruction methods can be readily affected by displacement measurements with a poor signal-to-noise ratio (SNR) [24,28,32].

For both the calculation of strain images and the reconstruction of elastic parameters, accurate estimation of tissue displacements is the first important step that will critically affect the image quality. This has prompted the development of different motion-tracking techniques for recovering tissue displacements during the past 2 decades [3–5,9,33]. A widely used displacement estimation technique is time-delay estimation (TDE) [3,26]. TDE methods generally involve finding the best-matching segment in the delayed RF signal for a specific segment in the reference RF signal by computing the maximum or minimum of a pattern-matching function. Cross-correlation is the most commonly used pattern-matching function in TDE, but several other matching techniques have also been employed, such as those based on correlation coefficients [34], hybrid-sign correlations [35], the sum of absolute differences [36], and the sum of squared differences [37]. The performances of these matching techniques have been comprehensively surveyed [38]. TDE provides accurate estimation of axial displacement, but it is normally time-consuming. An alternative way is to use phase-shift estimation (PSE) originating from Doppler techniques, which has the advantage of efficient calculation of displacements [5,39,40] and hence is more feasible to implement in commercial ultrasound systems.

These 1D real-time tracking algorithms may perform well in phantom, however, it is difficult to obtain a high quality strain map *in vivo* because of the complexity of tissue motion under compression. 2D elastography is a better choice for estimation of both axial and lateral displacements. In recent years, some new algorithms, such as dynamic programming (DP) [19] approach, and a fast normalized cross-correlation calculation method [41], have been developed to realize real-time 2D elastography. In these methods, an interpolate algorithm is needed to get the subsample displacements, which greatly influences the precision of the displacement estimation. A motion estimation refinement framework has recently presented by Zhou and Zheng [10]. It can obtain the subsample displacements directly with optical flow (OF) method, but the OF method may cost heavy computation, so that it will have a relatively lower frame rate.

In this paper, we proposed a new fast framework of displacement estimation which combined block matching algorithm (BMA) and phase shift with prior estimation (PSPE) methods. The framework of the algorithm is first briefly described, followed by the descriptions of both the BMA and PSPE algorithms used in our method. Then synthetic data are generated to evaluate the robustness and accuracy of our strategy. Moreover, the promising results obtained from real ultrasound phantom data and clinical data show a great potential of our approach in elastography. Finally, the discussion on the experimental results and the conclusion of this study are given in the Section 5.

2. Related works

Motion tracking algorithms have been widely used in ultrasound field, especially in ultrasound elastography or elasticity imaging. Many methods have been proposed to for motion estimation, including phase-domain methods [42,43], time-domain or space-domain methods [44,45], and spline-based methods [46,47]. Time-domain (1-D) or space-domain (2-D) methods have been widely and frequently used because of their high accuracy, precision, and resolution, and relative simplicity in implementation. Typically, an observation reference signal/image window (i.e. kernel) is defined in one frame of ultrasound data (either B-mode image, RF, or envelope signal) and subsequently compared with different candidate windows (i.e. comparison windows) from another frame of ultrasound data, within a pre-defined search range. A cost function is calculated to quantify the similarity, or matching, between the reference and comparison windows. The motion is estimated from the temporal or spatial shift between the reference window and the best-match comparison window, which gives the maximum (or minimum) cost function value.

Numerous cost functions have been developed for different motion estimation methods. These methods include, but are not limited to, non-normalized cross-correlation, normalized cross-correlation, normalized covariance, sum of absolute differences (SAD), and sum of squared differences (SSD) [44,45]. Research efforts have been concentrated on the performance comparison of different methods [36,44,45,48–50].

Although the time-domain methods exhibit high computational cost (in terms of point operations), normalized cross correlation and normalized covariance often are said to yield optimal estimates. The main advantage of these two algorithms is that they consider the energy of the two signals and are thus able to generate more precise estimates. The normalized covariance differs from the normalized correlation because it takes into account the mean of the reference and delayed signals over the observation windows. The sum absolute differences (SAD) and sum squared differences (SSD) algorithms have been shown to perform as precisely as the normalized cross correlation, at least in two-dimensional (2-D), and have a reduced computational complexity [36]. This reduced complexity arises from two main factors, the elimination of multiplications and particularly the elimination of the normalization. Non-normalized correlation also has been used extensively for motion tracking in ultrasound field, yielding reasonable results in term of precision and speed of execution. The main difference between normalized and non-normalized correlation is that the energy of the signals is not taken into account in the latter algorithm, thus reducing the number of