Published in final edited form as:

Ultrasonics. 2008 November ; 48(6-7): 568–573. doi:10.1016/j.ultras.2008.07.010.

Simulations of lesion detection using a combined phased array LHMI-technique

Janne Heikkilä¹ and Kullervo Hynynen²

¹University of Kuopio, Department of Physics, Yliopistonranta 1F, FIN-70211 Kuopio, Finland, Tel: +358 (0)17 162395, Fax: +358 (0)17 162585, E-mail: Janne.Heikkila@uku.fi

²Imaging Research, Sunnybrook Health Science Centre, 2075 Bayview Ave., Toronto, ON, Canada

Abstract

Ultrasound based elasticity imaging techniques have been developed during the past decades. Some of these techniques are based on an internal radiation force stimulation in which a transient or dynamic radiation force is produced by using a single or dual-frequency sonication. In addition, sonication and data acquisition can be implemented using combined or separate transducers. In this simulation study of lesion detection using localized harmonic motion imaging (LHMI), we used a combined phased array designed for simultaneous thermal ablation and lesion detection. In the sonication mode, a focused single-frequency amplitude-modulated sonication is used to induce harmonic motion and in the tracking mode, some of the array elements are used for pulse-echo tracking of the induced displacements. The results showed that the size of the lesion affected the induced displacement around the sonication point. The displacement tracking simulations demonstrated that these changes in the displacement distributions can be detected using only a few of the array elements in the tracking mode but the exact size of the lesion can not be detected accurately. The simulations also showed that two lesions having the radius of 2.5 mm can be distinguished if distance between these lesions is at least 2.5 mm.

Keywords

Localized harmonic motion imaging (LHMI); lesion detection; finite element method (FEM); pulse-echo imaging; simulations

Introduction

During the past two decades, extensive research has been carried out investigating different kinds of ultrasound based elasticity imaging techniques. In certain diseases the stiffness of tissue is increased dramatically, whereas the acoustic parameters may stay almost constant. This means that in some cases tumors might not be visible in traditional ultrasound imaging, but because of these changes in the mechanical properties, the tumors can be seen in elasticity imaging. [1]

In elasticity imaging techniques, an external or internal mechanical stimulation is used and the mechanical response of the target is measured. It has been shown in several previous ultrasound

elastography studies that the measured responses are dependent on the structural and material properties of a target and that the mechanical properties of a target can be estimated using the measured responses [1]. In techniques based on an internal time-varying stimulation, an ultrasound radiation force produced by one or two overlapping focused ultrasound beams are used [2]-[4]. Responses can be externally acquired by using a low-frequency hydrophone or pulse-echo imaging.

Localized harmonic motion imaging (LHMI) [2] is one of these ultrasound elastography techniques that are based on pulse-echo imaging of dynamic ultrasound exposure induced displacements or vibrations. In LHMI, the frequency of the ultrasound beam amplitude-modulation is typically between 50 Hz and 1 kHz and induced harmonic motion is tracked using a traditional pulse-echo imaging technique. Usually, a separate imaging transducer is used for motion tracking, but combined systems can be used in which both sonication and data acquisition is implemented using a single phased array. The pulse-echo imaging based elastography methods have been introduced to be usable in many different applications, e.g. in diagnostics [5], [6] and monitoring during focused ultrasound surgery [7], [8].

This study includes simulations of radiation force induced vibrations and displacement detection using pulse-echo imaging. We used a geometrically focused linear phased array that can be used for both the high power sonication to induce tissue motion and pulse echo detection of induced motion. This study expands our previous studies, in which different sonication configurations and displacement distribution detection using pulse-echo imaging have been simulated [9], [10]. In this study, we investigated the feasibility of this combined phased array LHMI technique to detect stiffness inhomogeneities in soft tissue which might be related to some diseases or they can be induced by elevated temperatures during focused ultrasound surgery exposures. The feasibility of this technique to detect the inconsistency and size of the lesion was simulated by using different tracking configurations.

Theory

LHMI is one of the elastography techniques in which a dynamic acoustic radiation force is used to excite a target. The easiest and most efficient way to produce this oscillating radiation force is an amplitude-modulated ultrasound beam [10]. The radiation force per volume element acting in an absorbing medium is [11]

$$F = \frac{2\alpha I}{c}, \quad (1)$$

where α is the absorption coefficient, $\mathbf{I}$ is the time average acoustic intensity vector and c is the speed of sound. The radiation force can be amplitude modulated by varying the intensity of the sonication beam.

Biological tissue structures exhibit very complex mechanical behavior during force excitation and many models to represent this behavior can be found from the literature. In this study, we used the equation of motion that can be written in the form [12]

$$\rho \frac{\partial^2 \mathbf{u}}{\partial t^2} = \nabla \cdot \boldsymbol{\sigma} + \mathbf{f}, \quad (2)$$

where ρ is the density, t is the time, $\mathbf{u}$ is the displacement, $\boldsymbol{\sigma}$ is the viscoelastic stress tensor and $\mathbf{f}$ is the ultrasound radiation force induced body force.

In LHMI, responses (i.e. radiation force induced displacements) are measured using pulse-echo imaging that is based on time delay estimation between the echoes measured during the

target movement. Many different time delay estimation techniques have been developed to detect small time lags between two RF-signals. If only the correlation function is used, very high sampling rate or curve fitting methods are needed to achieve good time resolution. Using so called zero phase techniques, this resolution can be increased also when a low sampling rate is used. The zero phase technique that we used in this study can be computed as follows [13]:

$$\phi = \arg(H(S_1) \times H(S_2)), \quad (3)$$

where ϕ is the phase of the correlation function, S_1 and S_2 are the acquired RF-signals, H is the Hilbert transform and $\times$ is used to denote the cross-correlation. The delay D ($D \in \mathbb{R}$) between S_1 and S_2 in units of sample point can now be obtained from the point where $\phi = 0$. Finally, the displacement d_x that caused the detected time shift into the echo signals can be computed as follows:

$$d_x = \frac{Dc}{2f_s}, \quad (4)$$

where f_s is the sampling frequency of the RF-signal.

Simulations

We used a combined 128-element focused linear phased array for both the sonications and pulse-echo imaging (Fig. 1). The width and height of the transducer was 92 mm and 28 mm, respectively, and the focal depth was 40 mm. The width of the element was 0.65 mm and the kerf was 0.05 mm. The base frequency of the amplitude-modulated sonication field was 1.1 MHz and the modulation frequency was 250 Hz. Three different central frequencies of the pulse-echo beam (3.3 MHz, 5.5 MHz and 7.7 MHz) and the sampling frequency of 150 MHz were used in the motion tracking simulations. Also the number of elements used in the tracking mode was varied from 4 to 16. The widths of the point-spread functions for all tracking configurations are shown in Table 1. When the focus of the tracking beam was scanned across the lesions the elements used in the tracking mode were chosen such that the location of the tracking focus (in the y direction) was always in the middle of the tracking elements. Both the sonication and tracking foci were electrically moved in the steps equal to the centre-to-centre distance between transducer elements. While the electrical steering the transducer was kept in a fixed position.

All simulations were computed in a heterogeneous soft tissue cylinder having the radius of 38 mm and the length of 100 mm in the y direction. The computation domain consisted of two components (Fig. 2): coagulated and normal tissue. Only the stiffness was different in these two components and all other parameters were equal. Young's modulus of the coagulated tissue was 140 kPa and 28 kPa was used for the normal tissue. In the resolution simulations, we used two spherical coagulated tissue volumes ($r = 2.5$ mm) separated by 1.5 mm or 2.5 mm. In the other simulations, we used the homogeneous domain and four concentric spherical lesions of different diameters. The radii of the lesions were 0.75 mm, 1.25 mm, 1.75 mm and 2.25 mm.

The induced displacement distributions were simulated using the finite element method (FEM) [12] and Field II-codes [14] were used in the pulse-echo imaging simulations. The displacement simulations were computed using all 128 elements in the sonication mode which means that we assumed that lack of few elements (elements used for the tracking beam) does not affect the sonication beam. Displacement distributions at the instant when the focal point of the sonication beam reached the maximum displacement were used in the displacement tracking simulations. In the displacement analysis, the window length was 8 times the wave length of the tracking beam.

Results

Fig. 3 shows results from the simulations in which two separated spherical lesions ($r = 2.5$ mm) were used to study the resolution and feasibility of LHMI to detect separate tumors. Two separation distances, 1.5 mm and 2.5 mm, were used in these simulations. Both the induced and tracked displacements were simulated and the profiles on the focal line of the transducer are shown. The vertical dashed lines show the locations of the boundaries of the lesions. The tracked displacements were averaged around the focal depth (± 0.5 mm) to achieve smoother profiles, whereas the induced profiles were not averaged. In these simulations, the maximum displacement on the sonication point was tracked while the joint sonication and tracking focus was scanned across the lesions.

It can be seen from these simulations that when these two lesions were separated by 2.5 mm they can be distinguished by using 8 or more tracking elements with all frequencies studied. When the separation distance was 1.5 mm, these two spheres cannot be distinguished if 3.3 MHz tracking beams were used. In the case of 8 or 16 tracking elements operating at 5.5 MHz or 7.7 MHz, the induced displacement profile can be found from the simulated tracked profiles.

Fig. 4 shows the results from the lesion detection simulations. Also in this figure, only the tracked displacements were averaged around the focal depth. The simulations show that LHMI is sensitive to changes in the size of the lesion. The most suitable parameters for the tracking beam that are used in this study are 16 elements operating at 5.5 MHz. When the tracking frequency is increased from 5.5 MHz to 7.7 MHz the induced profiles can be tracked better only when 8 elements are used. In the case of 16 tracking elements, 5.5 MHz yield better results than higher frequency. However, the exact size of the lesion is impossible to detect by using any of these configurations.

Discussions and Conclusions

The simulations showed that the proposed combined phased array technique can be used to detect regions that were stiffer than the surrounding soft tissue. Results also demonstrated that 8 or more elements have to be used in this transducer configuration to detect the separated lesions or get any estimate of size of the lesions. It can also be seen that better lateral resolution for estimating the size of the lesion was achieved if the number of elements was increased. Estimation of the size of the lesion might be improved if the joint sonication and tracking focus is moved in the smaller steps across the lesion.

When motion tracking is performed during radiation force application there is interference from the ultrasound used for the radiation force with the pulse-echo tracking. However, recent in vivo measurements [8] show that this interference is not an insuperable problem. In this model all interactions between the sonication and tracking beams have been neglected which may cause some errors into model's results. In addition, this model is sensitive to some of the parameters, such as the mesh density and the size of the focus. This means that variations in the computation mesh density at the focal region can cause variations to the computed radiation forces. That kind of errors can be seen in Fig. 4 in which the induced displacement profile without the lesion is not flat. In these simulations, there might be too large variations in the mesh density compared to the shape of the focus.

Further research is needed to optimize this technique so that both the mechanical properties and the size of the lesion could be imaged accurately. Possible improvement could be the use of smaller elements in the centre part of the transducer because it would produce better tracking beam at higher frequencies. Now, the element size has been optimized for 1.1 MHz sonication beam.

As a conclusion, results indicated that this combined phased array technique might be useable for example for tumor detection. In addition, this combined phased array technique could also be useful for simultaneous ultrasound surgery and its monitoring because the phased array transducer can be used to induce thermal coagulation of tissue and LHMI may be able to detect a growth in the coagulated tissue volume.

Acknowledgements

This study was supported by The Academy of Finland, NIH Grant R33 CA102884, and the CRC program.

References

1. Ophir J, Alam SK, Garra BA, Kallel F, Konofagou EE, Krouskop T, Merritt CRB, Righetti R, Souchon R, Srinivasan S, Varghese T. Elastography: Imaging the elastic properties of soft tissue with ultrasound. *J Med Ultrasonics* 2002;29:155–171.
2. Konofagou EE, Hynynen K. Localized harmonic motion imaging: theory, simulations and experiments. *Ultrasound Med Biol* 2003;29:1405–1413. [PubMed: 14597337]
3. Fatemi M, Greenleaf JF. Ultrasound-stimulated vibro-acoustic spectrography. *Science* 1998;280:82–85. [PubMed: 9525861]
4. Nightingale KR, Palmeri ML, Nightingale RW, Trahey GE. On the feasibility of remote palpation using acoustic radiation force. *J Acoust Soc Am* 2001;110:625–634. [PubMed: 11508987]
5. Trahey GE, Palmeri ML, Bentley RC, Nightingale KR. Acoustic radiation force impulse imaging of the mechanical properties of arteries: *In vivo* and *ex vivo* results. *Ultrasound Med Biol* 2004;30:1163–1171. [PubMed: 15550320]
6. Nightingale KR, Nightingale RW, Palmeri ML, Trahey GT. A finite element model of remote palpation of breast lesions using radiation force: Factors affecting tissue displacement. *Ultrason Imaging* 2000;22:35–54. [PubMed: 10823496]
7. Righetti R, Kallel K, Stafford RJ, Price RE, Krouskop TA, Hazle JD, Ophir J. Elastographic characterization of HIFU-induced lesions in canine livers. *Ultrasound Med Biol* 1999;1099–1113. [PubMed: 10574342]
8. Curiel L, Chopra R, Hynynen K. *In vivo* monitoring of focused ultrasound surgery using localized harmonic motion. 2007Submitted
9. Heikkilä J, Hynynen K. A combined array for localized harmonic motion imaging: A simulation study of feasibility of motion detection. *Proc IEEE Ultrason Symp* 2006:1148–1151.
10. Heikkilä J, Hynynen K. Investigation of optimal method for inducing harmonic motion in tissue using a linear ultrasound phased array – a simulation study. *Ultrason Imaging* 2006;28:97–113. [PubMed: 17094690]
11. Torr GR. The acoustic radiation force. *Am J Phys* 1984;52:402–408.
12. Hughes, TJR. Finite Element Method - Linear Static and Dynamic Finite Element Analysis. Prentice-Hall; Englewood Cliffs: 2000.
13. Pasavento A, Perrey C, Krueger M, Ermert H. A time-efficient and accurate strain estimation concept for ultrasonic elastography using iterative phase zero estimation, *IEEE Trans. Ultrason Ferroelectr Freq Control* 1999;46:1057–1067.
14. Jensen JA, Svendsen NB. Calculation of pressure fields from arbitrarily shaped, apodized, and excited ultrasound transducers, *IEEE Trans. Ultrason Ferroelec Freq Contr* 1992;39:262–267.

Fig. 1. Schematic illustrations of the focused linear phased array transducer (top) and combined sonication and tracking technique used in this study (bottom).

Fig. 2.

Two separated spheres used in the resolution simulations (top) and the spheres of different sizes used in the simulations in which the lesion size was changed (bottom).

Fig. 3.

Induced and pulse-echo tracked displacements on the focal line of the transducer when two different distances d (1.5 mm and 2.5 mm) between two spherical lesions and 4, 8 or 16 tracking elements operating at 3.3 MHz, 5.5 MHz and 7.7 MHz frequency were used. The vertical lines show the locations of the lesion boundaries. In these simulations, both the sonication and tracking focus was scanned across the lesions.

Fig. 4.

Induced and pulse-echo tracked displacement profiles on the focal line of the transducer when five different size of the lesion (no lesion, 0.75 mm, 1.25 mm, 1.75 mm and 2.25 mm) and 4, 8 and 16 tracking elements operating at 3.3 MHz, 5.5 MHz and 7.7 MHz frequency were used.

Table 1

The width of the point-spread function (PSF) for all sonication configurations

Frequency (MHz)	Number of elements	Width of the PSF (mm)
3.3	4	6.88
5.5	4	4.50
7.7	4	4.07
3.3	8	3.43
5.5	8	2.36
7.7	8	2.15
3.3	16	1.71
5.5	16	1.21
7.7	16	1.22