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HIGH FREQUENCY ULTRASOUND FOR HIGH RESOLUTION IMAGING: TECHNICAL CONCEPTS AND APPLICATIONS IN DERMATOLOGY

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Abstract - Ultrasound is widely applied in medical diagnosis for the non invasive imaging of tissues and organs. Conventionally, ultrasound in the range from 1 MHz to about 10 MHz is utilized. With increased frequencies the resolution is improved, and, at the same time, penetration is decreased due to increased tissue attenuation. For the imaging of tissues on a microscopic level, larger frequencies have to be applied. For skin imaging applications, systems working with high frequency ultrasound in the 20 MHz range were successfully established. Besides applications of high frequency ultrasound (HFUS) in dermatology, imaging techniques in terms of ultrasound biomicroscopy (UBM) became of interest in intravascular ultrasound (IVUS) imaging of coronary arteries and ultrasound microscopy for the characterization of tissue and bone specimens. Recently, 'small animal imaging', i.e. the imaging of small laboratory animals, especially of mice, has produced large interest as well. In this paper, technical concepts of HFUS based systems for high resolution imaging are discussed. Next to morphological B-mode imaging, applications of HFUS for high resolution blood flow imaging and HFUS elastography for the assessment of mechanical tissue properties are addressed.
Keywords: High frequency ultrasound, skin imaging, dermatology, ultrasound biomicroscopy, small animal imaging

1. Introduction

Ultrasonic imaging in medical applications is usually based on pulse echo measurements, i.e. an ultrasound transducer is excited with a pulsed signal, and ultrasound waves are emitted and backscattered or reflected at acoustical inhomogeneities. Ultrasound waves traveling back to the transducer are converted into an electrical signal. Based on the assumption of a constant speed of sound in the tissue, backscattered and reflected signals can be assigned to axial distances in sound propagation direction proportional to the time of flight. The ability to distinguish between axially adjacent structures and, thus, the axial resolution, depends on the pulse width of the system's response and therefore on the bandwidth of the system [1-3]. The resolution in directions orthogonal to sound propagation depends on the sound beam width, which is determined by the transducer aperture dimension and the system's center frequency. Because of these conditions imaging is improved with increasing ultrasound frequencies and bandwidth. UBM takes advantage of these conditions, utilizing frequencies in the 20 MHz range and above. On the other hand, the attenuation of tissue and water, which is very often used as coupling medium between the transducer and the tissue, increases significantly with increased frequencies, limiting the penetration depth in a system with a given limited dynamic. Thus, a compromise between a good resolution and maximum penetration depth has to be found. HFUS in the range up to 100 MHz range is very well suited for high resolution imaging of the uppermost skin layers, which is of high interest in dermatology as well as cosmetics. We have developed a HFUS skin imaging system, working in this frequency range. HFUS specific problems in the design of this system are addressed in this paper.

While the aim of B-mode ('brightness') ultrasound is to image tissue morphology, Doppler techniques are utilized to assess blood flow velocities as functional parameter. Functional imaging can also take advantage of HFUS, enabling high resolution measurements. We have developed a 50 MHz pulsed wave Doppler (PWD) system for flow imaging in small vessels [4-8].

Ultrasound elastography, first introduced in 1991 by Ophir et al., is a method for imaging elastic properties of tissues. In this approach mechanical strains inside the tissue are calculating estimating displacements in frames of consecutively acquired frames of radio frequency (rf) echo signals, while the tissue is mechanically loaded applying an external force. Recently, we have developed a HFUS based elastography system, utilizing 20 MHz ultrasound, for skin elasticity imaging [9-10].

2. Design of a High Frequency Ultrasound (HFUS) Skin Imaging System

Due to the lack of HFUS arrays, ultrasonic skin imaging is based on single element transducers up to now, performing mechanical scans. Spherically focused transducers with a converging sound beam characteristics in the near field and a diverging beam in the far field are utilized. Imaging is performed in the focus region, resulting in an optimized transversal resolution and measurement sensitivity. Water is used as coupling medium between the transducer aperture and the skin, see Fig. 1 a). In Fig. 1. b) the axial and transversal resolution depending on the spectral characteristics and dimensions of the transducers are given.

It becomes obviously that imaging is improved with increased bandwidth and center frequency, i.e. utilizing high frequency, broadband ultrasound. A disadvantage of the fixed focus single element transducer is that a focused imaging is only possible inside the limited and fixed focus region. Imaging can be improved acquiring echo signals with the focus positioned at different depths performing mechanical lateral and depth scans. B-mode images over a large depth range can then be composed joining the acquired echo signals, which results in a more depth invariant and in a homogeneous resolution.

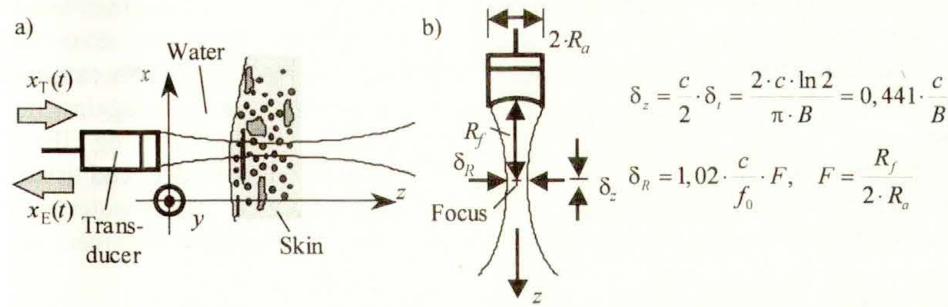

Fig. 1. a) Skin imaging with strongly focused single element transducer: Transmit signal $x_T(t)$, echo signal $x_E(t)$; b) Imaging properties: Axial resolution δ_z, transversal resolution δ_R, speed of sound c, bandwidth B, center frequency f_0, aperture radius R_a, focus length R_f

We have previously termed this concept B/D-scan technique ('brightness / depth') [1-3].

Applying HFUS, specific problems have to be addressed. Electrical signals, which are reflected at the terminations of transmission lines inside the system can cause significant distortions of the system's point spread function (PSF) because delays easily reach the echo signal pulse width. We have proposed to utilize network analysis (NWA) and time domain reflectometry (TDR) to assess mismatching in connection with nonlinear expander / limiter networks for transmit / receive switching [11].

3. HFUS Pulsed Wave Doppler System

Blood flow velocities are assessable by analyzing the Doppler frequency shift or time shifts in frames of echo signals, which result from the backscattering at blood particles. During the measurement trains of pulsed signals with a predefined pulse repetition frequency (PRF), which determines the range of unambiguously measurable flow velocities, are transmitted at discrete transducer positions. We have developed a pulsed wave Doppler system with an arbitrary function generator with subsequent power amplifier for transmit signal synthesis. Utilizing HFUS in the 50 MHz range, echo signals are directly sampled with an analog to digital (A/D) converter. Color flow maps and power mode images are processed offline [4-8].

4. HFUS based Elastography of the Skin

Ultrasound elastography relies on the spatially resolved estimation of displacements inside the tissue during the application of an external deformation analyzing acquired echo signal frames. Mechanical strains are calculated as the derivative of estimated displays. The major problem is to apply the external deformation in an appropriate way, i.e. displacements should be small enough to avoid a decorrelation of consecutively acquired echo signals. On the other hand, deformation must be large enough to enable a reliable displacement estimation. In conventional elastographic applications like breast or prostate imaging, an external compression can easily be performed pressing the ultrasound transducer against the organ. As was shown above, the conditions are more complicated applying HFUS for skin imaging, because of the need of a water path between transducer and tissue. Thus, a feasible technique is to apply a vacuum inside a chamber with a water path at the skin surface and to cause suction, drawing the skin into the chamber, see Fig. 2.

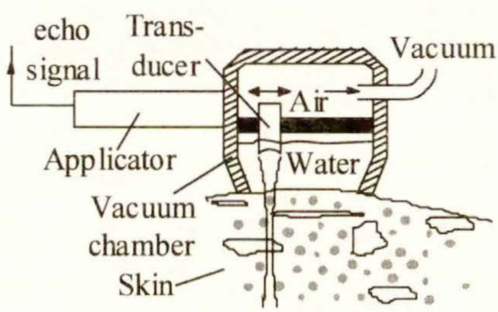

Fig. 2. Skin elastography system: Suction applied to the skin surface with a low pressure inside a vacuum chamber

Rf echo signal frames are acquired while the pressure inside the chamber is stepwise decreased. Axial displacements in sound propagation direction are estimated by cross correlating the analytical rf echo signals. The slope of a linear regression fit of estimated axial displacements delivers estimates of the mechanical axial strain [9,10].

5. Results

HFUS Skin Imaging

We have implemented a HFUS skin imaging system with a single element PVDF-transducer with a focus length $R_f = 4.3\,\text{mm}$ and aperture radius $R_a = 1\,\text{mm}$, resulting in a minimum axial resolution $\delta_z = 9.2\,\mu\text{m}$ and a minimum transversal resolution $\delta_R = 50\,\mu\text{m}$. Ultrasound in this frequency band enables the high resolution imaging of the uppermost skin layers. In Fig. 3, a B-mode image, acquired at the transition at the wrist with the implemented system is shown.

The skin surface appears as hyperechoic band because of the strong difference in the acoustic impedances of the water path and the skin. The uppermost layer of the epidermis, the stratum corneum, is visible as hypoechoic band. Beneath, a second hypoechoic region, the stratum malpighii, can be seen, whereas the subjacent dermis appears hyperechoic.

Moving from the arm to the hand at the wrist, the epidermis thickness increases. Echo poor structures inside the dermis are hair follicles. Underneath, the subcutaneous fat is visible.

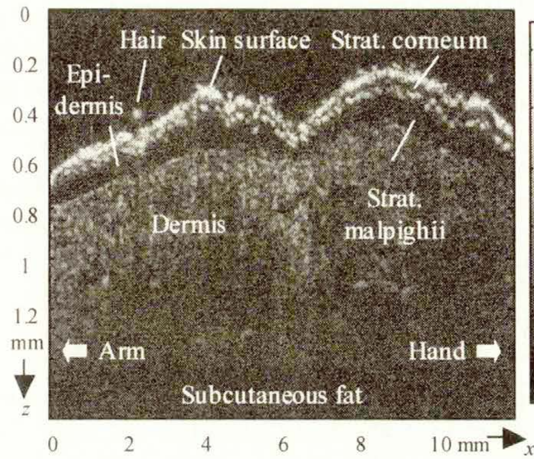

Fig. 3. HFUS B-mode image of skin at wrist area [11]

HFUS Pulsed Wave Doppler Technique

Utilizing HFUS in the 50 MHz range, blood flow in small veins with diameters down to 300 μm and small blood flow velocities in the range up to 4 mm/s at the back of a hand was successfully detected and imaged.

HFUS Skin Elastography

Frames of rf echo signals were acquired utilizing 20 MHz ultrasound during the stepwise decreased pressure at the skin surface. Strain images show that the subcutaneous fat is strongly elongated under these conditions, whereas the mechanical strain inside the dermis is very small in healthy skin. Results from measurements at burned skin show a high elongation in the subcutaneous fat dermis as well as in the dermis, which indicates the change of elasticity due to the destructed collagen fibers. The elastic properties of nevi were found to be significantly different than the properties of the surrounding dermis.

6. Discussion and Conclusions

In the implemented systems, HFUS is applied for skin morphology and functional imaging. Thus, HFUS supports the non invasive and high resolution diagnostics of living skin on a microscopic level. However, frequency bands have to be selected properly and matched to the applications to account for the frequency dependent attenuation. HFUS in the range up to 100 MHz was found to be well suited for the imaging of the uppermost skin layers, whereas the already well established 20 MHz ultrasound technique is mainly aimed at the imaging of the whole skin and the underlying subcutaneous fat. For blood flow imaging applications, HFUS in the 50 MHz range was successfully utilized.

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