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Passive twin-layer spatial-temporal phase-interference compensator for improved ultrasound propagation: A computer-simulation and experimental study in acrylic step-wedge samples

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ABSTRACT

Transcranial ultrasound wave degradation created by variations in both thickness and tissue composition is a significant impediment to diagnostic and therapeutic interventions of the brain. The current '*active*' solution is to vary the transmission delay of ultrasound pulses, inherently necessitating electronic control of each individual transducer element. This paper advances the sonic-ray concept of ultrasound wave propagation by hypothesising that wave degradation can be minimised if both the transit-time and propagation path-length for all sonic-rays are made constant.

A computer simulation was performed to investigate the dependence of an ultrasound signal upon spatial and temporal matching of all sonic-rays propagating through a sample exhibiting significant variation in transit time. Ultrasound propagation through a cylindrical acrylic 20-step wedge sample was considered, showing that phase-interference is removed only if there is complete spatial and temporal matching. A '*passive*' concept for achieving this is introduced within this paper, consisting of a twin-layer ultrasound phase-interference compensator (TL-UPIC). An experimental study was performed in transmission-mode on six cylindrical acrylic step-wedge samples, ranging from 2 to 20 steps, each creating a corresponding number of sonic-rays. A TL-UPIC model corresponding to each step-wedge sample was designed, such that the propagation path-length (spatial) and transit-time (temporal) was constant for all sonic-rays, and replicated using 3D-printing. Time- and frequency-domain analysis demonstrated that incorporation of the TL-UPIC successfully removed phase-interference in all cases.

It is hypothesised that the TL-UPIC concept may be applied to both pulse-echo mode diagnostic imaging and transmission mode therapeutic applications; and that it is also applicable to either single-element or multi-element array transducers, of any practicable frequency and dimension.

1. Introduction

The human skull may be considered to consist of a sandwich of marrow-infused cancellous bone between two layers of essentially solid cortical bone. The transit-time of an ultrasound wave through the skull can vary significantly due to anatomical inconsistencies in both the composition and overall thickness of the skull [1]. This has the potential to create phase-interference and hence significant wave degradation, which may impede both diagnostic imaging and therapeutic interventions of the skull. A number of '*active*' approaches have previously been considered to compensate for variation in transit-time, adjusting the transmission delay of ultrasound pulses from a large array of individual transducer elements; essentially a phased-array approach. By first analysing the variation in transit time, time-reversal compensation may be applied [2,3,1]. Analytical [4] and simulation [5] approaches have also been considered based upon X-ray CT data. A passive micro-bubble-derived aberration correction technique has recently been described [6], comparing actual and predicted transit times at each element of a hemispherical array. Although these techniques have successfully demonstrated a reduction in transcranial ultrasound wave degradation, they are inherently complex, necessitating electronic control of each of the individual transducer elements, which typically numbers 1000 in total; further, they are inherently restricted to the ultrasound system being utilised.

The author has previously hypothesised that the primary ultrasound attenuation mechanism in complex porous media such as cancellous bone is phase-interference caused by the variation in

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transit time as detected by the phase-sensitive ultrasound receive transducer [7]. He also proposed a concept describing the propagation of ultrasound through such a complex structure as an array of parallel sonic-rays. The transit time of each sonic-ray is determined by the relative proportion of the two constituents of differing propagation velocity, for example bone tissue and marrow, regardless of the sample's structure. Hence, transit time minima (t_{min}) and maxima (t_{max}) correspond to propagation through entire bone and marrow respectively. A transit time spectrum (TTS) describes the proportion $P(t_i)$ of sonic-rays having a particular transit time (t_i) between t_{min} and t_{min}. Reflection and refraction are not incorporated, being considered to be secondary phenomena. The sonic-ray concept has been initially validated through a comparison of experimental measurement and computer simulation in acrylic step-wedge samples of varying transit time heterogeneity immersed in water [8].

The author hypothesises that the sonic-ray concept may be applied to create a '*passive*' twin-layer ultrasound phase-interference compensator (TL-UPIC) that would significantly reduce wave degradation in the skull. The first aim of this paper was to investigate through computer simulation, the relationships between partial and full spatial-temporal matching upon phase-interference induced ultrasound wave degradation. The second aim was to create and experimentally validate a passive TL-UPIC.

2. Methods

2.1. Replication of Phase-Interference

Since the ultrasound velocity of calcified bone tissue is approximately twice that of marrow, typically 3000 m/s and 1450 m/s respectively, we may replicate these using acrylic and water, having ultrasound velocities of 2670 m/s (v_a) and 1483 m/s (v_w at 21.1 °C) respectively. Further, we may replicate the variation in ultrasound transit-time through the skull by incorporating a cylindrical acrylic step-wedge test sample immersed in water. If two circular single-element ultrasound transducers are co-axially positioned, an ultrasound wave may be propagated in transmission mode, one transducer serving as transmitter, the other as receiver; as shown schematically in Fig. 1. We may consider this propagating ultrasound wave as an array of parallel sonic-rays emanating from spatial coordinates (x,y) on the transmit transducer surface. The validity of this simplistic concept has been supported by successful estimation of both liquid [9] and solid [10] volume fractions, as well as discriminating overlapping ultrasound signals [11].

The propagation path of each sonic-ray is described by $z_o = (z_a)_{xy} + (z_w)_{xy}$, where z_o refers to the constant transducer separation, with $z_{a(xy)}$ and $z_{w(xy)}$ describing a variable thickness of acrylic and water respectively, illustrated schematically in Fig. 1 and referred to as Case 1. The corresponding transit time (t_{xy}) is given by $t_{xy} = (z_a)_{xy}/v_a + (z_w)_{xy}/v_w$. Minimum ($t_{min}$) and maximum ($t_{max}$) transit time values will correspond to entire acrylic and water, described by $t_{min} = z_o/v_a$ and $t_{max} = z_o/v_w$ respectively. Sonic-rays of different transit time will therefore be detected by the receive ultrasound transducer. Phase-interference between two, or more, sonic-rays will occur if their transit time difference (dt) is less than the pulse length (PL) of the propagating ultrasound wave, inherently resulting in degradation of the received ultrasound wave [12].

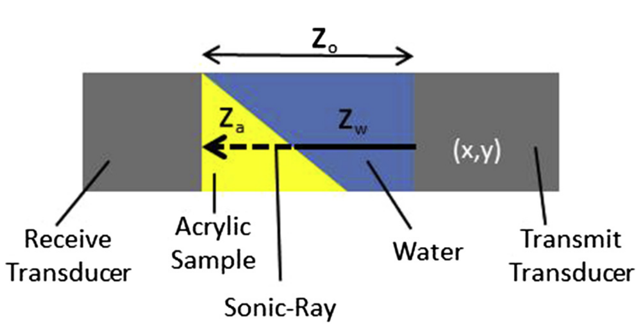

Fig. 1. Schematic diagram representing two co-axially aligned ultrasound transducers immersed in a water bath along with an acrylic step-wedge test sample (Case 1).

2.2. Mathematical modelling of a single-layer UPIC of solely spatial, solely temporal, and spatial-temporal matching

If we now incorporate a single-layer UPIC into the experimental system, as illustrated in Fig. 2, we may investigate the implications of solely spatial, solely temporal, and spatial-temporal matching.

For a sonic-ray travelling through the nth wedge-step, we may define the total transit time as $t_n = [t_w + t_a + t_u]_n = [(z_w/v_w) + (z_a/v_a) + (z_u/v_u)]_n)$, and the total dimension as $z_o = z_w + z_a + z_u$.

2.2.1. Spatial-temporal matching

If the velocity of the solid sample material (say acrylic, v_a) and UPIC (v_u, say 3D-print material) is equal, hence $v_a = v_u$, the combination of solid and corresponding UPIC will inherently be both spatially- and temporally-matched, illustrated in Case 2 of Fig. 3a. The minimum thickness (transit time) of UPIC equals the maximum thickness (transit time) of acrylic; and vice versa. The thickness of UPIC for the nth wedge-step is given by $(z_u)_n = z_o - (z_a)_n$, and the total transit time for nth wedge-step (t_n), being constant, is $t_n = (z_a)_n/v_a + (z_o - z_a)_n/v_u$.

If their velocities are different however, we may design the UPIC to be spatially-matched (Fig.3a) or temporally-matched (Fig.3b). In both scenarios, the velocity of the UPIC material may be less than ($v_a > v_u$) or greater than ($v_a < v_u$) the solid sample material.

2.2.2. Spatial- but not temporal-matching

The equations for UPIC thickness and corresponding total transit time for the nth wedge-step are retained; the velocity of the UPIC material may be less than ($v_a > v_u$), or greater than ($v_a < v_u$), the solid sample material; described by Case 3 and Case 4 respectively.

2.2.3. Temporal- but not spatial-matching

The transit time for the sample and UPIC for the nth wedge-step is constant, given as $t_{a\text{-}max} = (z_{a\text{-}max}/v_a)$. The thickness of UPIC for the nth wedge-step is $(z_u)_n = (v_u \cdot [t_{a\text{-}max} - (z_a/v_a)_n])$, and the corresponding thickness of water is $(z_w)_n = z_o - (z_a+z_u)_n$. The transit time through water for nth wedge-step is $(t_w)_n = (z_w/v_w)_n$; from which the total transit time for the nth sonic-ray is $(t_n) = t_n = t_{a\text{-}max} + (t_w)_n$. The two conditions of UPIC material velocity being less than ($v_a > v_u$), and greater than ($v_a < v_u$), the solid sample material are described by Case 5 and Case 6 respectively.

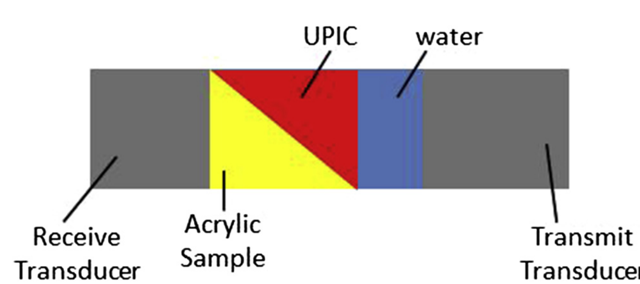

Fig. 2. Schematic diagram of an acrylic step-wedge sample and UPIC immersed in a water-bath between two ultrasound transducers.