Ultrasonic focusing through the ribs using the DORT method

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Thermal ablation induced by high intensity focused ultrasound has produced promising clinical results to treat hepatocarcinoma and other liver tumors; however skin burns have been reported due to the high absorption of ultrasonic energy by the ribs. This study proposes a method to produce an acoustic field focusing on a chosen target while avoiding the ribs, using the decomposition of the time-reversal operator (DORT method). The idea is to apply an excitation weight vector to the transducers array which is orthogonal to the subspace of emissions focusing on the ribs. A linear array of transducers has been used to measure the set of singular vectors associated with a chest phantom made of three human ribs immersed in water, and to produce the desired acoustic fields. The resulting propagating fields have been measured both in the focal plane and in the plane of the ribs, using a needle-hydrophone. The ratio of the energies absorbed at the focal point and on the ribs has been enhanced up to 100 fold as demonstrated by the measured specific absorption rates.

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I. INTRODUCTION

High intensity focused ultrasounds (HIFU) allow non-invasive energy concentration onto soft tissues and can induce localized temperature elevation (30°C to 55°C) within a few seconds. The resulting high temperature can generate irreversible tissue necrosis while leaving surrounding normal tissues undamaged. This method is used to treat a variety of conditions: treatment benign prostate hyperplasia,[1] prostate cancer,[2,3,4,5] uterine fibroid,[6] kidney tumors,[7] liver tumors...[8]

However, in the case of the liver, the presence of the rib cage is a major problem: first, it acts as an aberrator that affects the focusing,[9] and second, due to the high value of the absorption coefficient of the bones,[10] overheating of the ribs can be quite important. Thanks to MR temperature monitoring on pigs in vivo, Daum *et al* reported temperature elevations during sonication 5 times higher on the ribs than in the intercostal space.[11] This overheating can cause up to third-degree skin-burns,[8,12,13] so that partial surgical removal of the ribs has been clinically performed.[14]

Several methods have been proposed to avoid ribs removal. They are all using a transducer array to produce a field that avoids sending energy on the ribs. Botros *et al.*[15] proposed a two-step method, based on the introduction of a virtual array along the intercostal spacings. Two propagation matrix are first defined, one between the target points and the virtual array and the other one between the virtual array and the focusing array. Thanks to the pseudo-inversion of the first matrix, the particle velocities on the virtual array are first computed, so that the resulting field focuses at the desired location. Finally, the particle velocities on the actual array are computed so that the resulting field produces the correct field on the virtual array. This numerical study demonstrated that it was possible to minimize the power on the ribs down to 1 to 3% of the total power. However, it required the pseudo-inversion of two propagation matrix and has not been tested on real ribs.

Another approach simply consists in switching off the array elements whose normal vectors cross the ribs (which means, basically, switching off the elements in front of the ribs): Liu *et al.*[16] showed through a numerical simulation that it reduces the temperature elevation at the ribs; this method could be very simple to implement but but may not be the optimal method since it does not take into account wave diffraction by the ribs nor shear mode conversion.

Time-reversal correction have been first introduced to correct phase aberrations. As such a technique is a matched filter, it maximizes the ratio between the energy deposition at focus and the total emitted energy, in a non absorbing media as well as in an absorbing media.[17] It thus naturally minimizes the energy deposition on the ribs: Aubry *et al.*[17] showed experimentally that a mean 83% decrease of the temperature elevation on the ribs could be achieved by time reversal compared to a non-corrected spherical wavefront. However, this robust and real-time adaptive focusing method requires a transducer being placed at the desired focal point thus making this technique invasive.

The present study proposes a novel non invasive solution. The DORT method (french acronym for Décomposition de l'Opérateur de Retournement Temporel) was derived from the analysis of acoustic time reversal mirrors.[18] It consists in finding the invariants of the time-reversal process thanks to the backscattered echoes recorded on the emission/reception array. These invariants contain informations about the medium and will be used to avoid sending energy on the reflectors (in the present case, the ribs). Basically, the desired emission signals (namely focusing on the region to treat) are projected orthogonally to the set of emissions that focuses on the ribs. This set can be determined by applying the DORT method to the ribs' echoes. As will be shown, this process allows to construct an emission that focuses at the chosen point while minimizing the energy deposition on the ribs. As similar technique has been used by H.C. Song *et al.*[19] in underwater acoustics: an oceanic waveguide was insonified while minimizing reverberations from the bottom of the guide.

In this study, a 128-elements linear phased array and a chest phantom made of three ribs immersed in water was used to demonstrate the efficiency of the method. The specific absorption rate (SAR) was chosen as the indicator of that efficiency. The SAR obtained after projection has been compared with those obtained without projection and those obtained by using the time reversal technique.

II. MATERIAL AND METHODS

II.A. Experimental setup

A linear echographic probe was used, made of 128 transducers with a 0,55 mm spacing. The central frequency was 1,5 MHz (1 mm wavelength in water). The transducers were driven by fully programmable parallel processed electronic channels with a 20 MHz sampling rate. This system allowed to insonify the medium and record the echoes in a flexible manner.

The pressure field propagating from the transducers was measured by a needle-hydrophone fixed to a three-axis motor with a 10 µm precision. The fields were recorded firstly in the focal plane and secondly in the plane of the ribs once the ribs had been removed.

Three degassed human ribs was placed in a geometry mimicking a human rib cage. The width of the ribs varied from 10 to 15 mm and their thickness from 5 to 10 mm. They were attached to a frame and the intercostal space was varied from 12 to 17 mm depending on the experiment. Their distance to the echographic probe was 30 mm+-5 mm. The layout of the array, the ribs and the focal spot are displayed in Fig. 1(a) and 1(b).

II.B. Specific absorption rate

To evaluate the efficiency of the method and compare the results from different sets of measurements, the specific absorption rate (SAR, in W.m^{-3}) at a given position x is defined as follows:

$$SAR(x)=\alpha(x)\frac{|P(x,f_c)|^2}{2\rho(x)c(x)}$$

where $\alpha(x)$ is the absorption coefficient at position x, P is the Fourier component of the pressure at point x (so that $P^2/2$ is the mean squared pressure for a monochromatic emission) and at the central frequency f_c, and ρc is the acoustic impedance. The SAR gain was calculated as the ratio between the measured SAR on the target and the spatial average of the SAR over the central rib surface.

As the focal point was chosen behind the central rib, the average was not calculated over the lateral ribs that were less insonified. Averaging over them would artificially increase the SAR gain for any focusing emission.

An SAR gain of at least unity indicates a higher energy absorption at the targeted focal spot than on the ribs.

We chose to calculate the SAR for a single frequency, because monochromatic signals are used for therapy (typically several seconds pulse length).

The values of α, ρ and c are taken from Liu *et al.*[16] except for the absorption coefficient of the bone (taken from Fry *et al.*[20]):

- $\alpha_{liver}=\mu_{focal}=13{,}5$ Np.m^{-1} and $\alpha_{bone}=214$ Np.m^{-1} where α is the absorption coefficient and μ is the attenuation coefficient.

- $\rho_{liver}=1055$ kg.m^{-3} and $\rho_{bone}=1450$ kg.m^{-3}

- $c_{liver}=1547$ m.s^{-1} and $c_{bone}=2300$ m.s^{-1}

The values at focal spot correspond to the values of the liver.

II.C. Time reversal

Time-reversal is a focusing method that relies on the invariance of the wave equation under time reversal.[21] It requires an array of transducers and a source at the desired focal point.[22]

The wave propagating from an impulse point source to the array of transducers is recorded and time reversed. If the medium is time invariant, the time reversed signal propagates as if going backward in time, and converges to the initial spot as illustrated in Fig. (2). This method is robust and efficient, even in complex propagating medium.

A scatterer embedded in the medium acts as a source by reflecting the energy it receives. By insonifying the medium, and recording the backscattered wave, one can learn to focus on a scatterer. But in the presence of multiple scatterers, time reversal does not teach how to focus on each of them separately.

II.D. Decomposition of the time reversal operator

The DORT method was derived from the theoretical study of iterative time-reversal mirrors. It consists essentially in the construction of the wave fronts that are invariants under a time-reversal process.[23] Those invariants appear as the eigenvectors of a matrix called the time-reversal operator which describes the time-reversal process.

Considering an array of N transducers, and assuming that the system is linear and time-invariant, one can build in the frequency domain a N*N matrix $\mathbf{K}(\omega)$, the elements of which are the inter-element transfer functions at the pulsation ω. The time reversal operator (TRO) is then defined as $\mathbf{K}(\omega)\mathbf{K}^*(\omega)$.[23]

This operator describes the iteration of the following sequence: transmission, reception, and time-reversal operation (equivalent to a phase conjugation in the frequency domain). This operator is hermitian with positive eigenvalues. Its eigenvectors are the invariants waveforms of the time reversal process [indeed, if $\mathbf{V}(\omega)$ is the transmitted signal and is an eigenvector of the time reversal operator, the signal received on the first array is $\lambda(\omega)\mathbf{V}(\omega)$].

From a mathematical point of view, the diagonalization of the TRO is equivalent to the Singular Value Decomposition (SVD) of the array response matrix $\boldsymbol{K}(\boldsymbol{\omega})$. Indeed, the SVD decomposition is $\boldsymbol{K}(\boldsymbol{\omega})=\boldsymbol{U}(\boldsymbol{\omega})\boldsymbol{\Lambda}(\boldsymbol{\omega})\boldsymbol{V}^{\dagger}(\boldsymbol{\omega})$, where $\boldsymbol{\Lambda}$ is a real diagonal matrix and $\boldsymbol{U}$ and $\boldsymbol{V}$ are unitary matrices. It means that the columns of $\boldsymbol{V}$ are the eigenvectors of the TRO and the singular values are the square-roots of the eigenvalues.[23]

It is now well known that if the medium is composed of several point-like, isotropic and well resolved scatterers, the number of non-zero eigenvalues of the TRO is equal to the number of scatterers and each eigenvector is associated to one scatterer.[23] This means that applying the phases and amplitudes of a chosen eigenvector to the elements of the array produces a wave front focusing on the corresponding scatterer. In general, a scatterer is not isotropic nor point-like and is associated to several invariants of the time-reversal operator. These invariants were studied by A. Aubry *et al.* in the case of a solid cylinder.[24] The number of invariants is equal to the rank of $\boldsymbol{K}(\boldsymbol{\omega})$ which is of the order of the number of resolution cells in the scatterer.[25] Each eigenvector associated to a significant eigenvalue focuses on one portion of the scatterers.

In practice, the array response matrix **K** is measured by emitting a pulse on each array element successively, and measuring the corresponding echoes on the N transducers. An appropriate time-window is then used to select the echoes from the ribs and a Fourier transform of the selected signals is performed at each frequency within the transducers bandwidth . Then, looking at the singular values, the singular vectors are separated into two categories: a first set focusing on the ribs (the eigenvectors associated with highest singular values) and a second set that do not send energy on the scatterers (the eigenvectors associated to the smallest singular values).

II.E. Projection

The first step of the method consists in defining a set of emission signals $\mathbf{A}(\omega)$ focusing on the the target point, independently of the effect of the ribs. To this end, a time-reversal experiment is performed in the absence of the ribs: an impulse is emitted by a hydrophone located at the target location and the corresponding B-scan is measured on the array and time-reversed. The Fourier transform of this time-reversed B-scan gives rise to the $\mathbf{A}(\omega)$ on demand. If this B-scan is used to focus through the ribs, part of the wavefront is not affected by the ribs and produces a focusing at the desired point but part of it is absorbed and scattered by the ribs. Instead of using a time-reversal experiment, a conventional non-corrected cylindrical wave front could have been used. Nevertheless, thanks to time reversal, the phase defects of the transducers are automatically corrected. Moreover, such signals will have the same bandwidth than the signals acquired by the K matrix, thus enabling a fair comparison between all the techniques presented in this article. In the whole paper, the "non-corrected" signals will correspond to **A**(ω).

A(ω) is then projected orthogonally to the first set of eigenvectors:

$$A_{projected}(\omega)=A(\omega)-\sum_{i=1}^{i_{max}}\left(V_i^{\dagger}(\omega)A(\omega)\right)V_i(\omega)$$

where $\mathbf{V}_i$ is the i-th eigenvector and i_{max} is the number of eigenvectors belonging to the first set of eigenvectors. An inverse Fourier transform is then performed to obtain a temporal signal.

II.F. Experimental protocol

The distance between the chosen focal point and the echographic array was equal to 77 mm.

In order to demonstrate the efficiency of the projection method, three types of focusing emissions have been compared:

1 - without the ribs (free space), the impulse response from the desired focal spot to the array was recorded and time-reversed, providing emission **A**. It is well-known that emission **A** focuses in free space at the desired focal spot.

2 - the same measurement was done through the ribs, providing emission **B**.

3 - finally, a DORT experiment was run, and the eigenvectors sets were discriminated as described in section IIE. To allocate each eigenvector to the correct subset, and find $\mathbf{i_{max}}$, the singular values distribution must be examined.

A typical distribution of singular values is displayed in Fig. (3) for each of the 5 sequences.

i_{max} was set to 40: beyond that value, the singular values can be neglected. The numerical back-propagations of the eigenvectors at central frequency showed that beyond this value, the eigenvectors did not focus on specific points of the ribs, but would rather spread energy on a large number of points [see Fig. (4)]. On the opposite, Fig. (5) shows that the first eigenvector focuses where the ribs are. The same value (40) was kept for all frequencies. Emission **A** was then projected orthogonally to the first set of eigenvectors, providing emission **C**.

Finally, the focal spot obtained for these 3 emissions (with the ribs) and the field obtained in the plane of the ribs (the ribs having been removed) were measured.

This sequence was repeated 5 times, for 5 different geometrical configurations.

III. RESULTS

Typical emissions **A** to **C** obtained for the third set of measures are shown in Fig. (6). The corresponding pressure fields measured in the focal plane along the x axis are displayed in Fig. (7) as a function of time. The square of the corresponding time peak amplitude is plotted, in a dB scale in Fig. (8). In order to estimate the efficiency of the focusing, each plot is normalized by the total emitted energy (sum over the transducers of the square of the time-peak of the emitted B-scan).

The beam patterns are very similar, with the same -3 dB width. One has to note that depending on the experiment, the -3dB width was ranging from 1 to 2 mm, which is small enough for therapeutic purpose.[15] Emission **A** is on average 5 dB below emission **B** and **C** because part of the energy is absorbed or scattered by the ribs.

On the opposite, in the plane of the ribs, the fields differ significantly, as can be seen in Fig. (9) (obtained with the third set of measures).

For both emissions **B** and **C** [Fig. (10a)], the amplitude of the field drops significantly between -3 mm and +10 mm. This corresponds to the position of the central rib.

A closer look to the same plot displayed in a dB-scale [Fig. (10b)] shows that the amplitude drop is not constant over the central rib: it decreases by 12 to 28 dB.

The SAR gain were calculated for each set of measures, for all three emissions. The results are gathered in table I. Experiments are ordered with an increasing intercostal distance.

For emission **A** , the SAR gain is smaller than 1: the ribs heat more than the region to treat, leading to unavoidable skin-burns.

For emission **B** and **C**, the SAR gain is always larger than 1, thanks to ribs sparing. Not surprinsigly, the SAR gain is always larger with the invasive approach (emission **B**).

IV. DISCUSSION

For transribs liver therapy the set of emission signals has to meet two properties: focusing at the chosen location and avoiding energy deposition on the highly absorbing ribs.

The first condition was seen to be easily fulfilled because the aberrating effect of the ribs is small: as a matter of fact, the reflection and absorption by the ribs are so high that the amplitude of the field that propagated through the ribs (and has been distorted by the ribs) is negligible compared to that of the field that propagated in the intercostal space. The location of the focal spot has been shown to be correct and its width acceptable for liver ablation.

The second condition is harder to fulfill, because even if very little energy is sent on the ribs the SAR gain may be small because the absorption coefficient of the bone is high. This is indeed what is observed: even though emissions **B** and **C** are very similar (and so are the resulting fields measured on the ribs' surface), their SAR gain differ by a sightly higher pressure field on the ribs with emission **C**. As a result, the SAR gain is smaller with the non invasive approach.

Nevertheless, the SAR gain is significantly enhanced by the projection technique: it is 20 to 106 fold higher compared to the non-corrected approach, as shown in Table 1. Variations of the SAR gain are observed from a set of measures to the other, for each type of emission. For emission **A**, the variations can be explained by a simple geometric factor: the larger the rib surface is on the way between the transducers array and the focal spot, the smaller the SAR gain will be. In case 1 and 2, the intercostal space is bigger than in case 3, which is bigger than in case 5. So, the SAR gain decreases because more energy is sent on the ribs. Case 4 is special: one of the gaps between two ribs was placed just in front of the middle of the transducers array, where the energy is concentrated: the surrounding ribs only saw the weaker lateral parts of the emission, thus increasing the SAR gain.

The same comment applies to emission **B**: if the intercostal space is smaller, less energy can be sent to the focal spot, so that one would expect a decrease of the SAR gain from one experiment to the other. Nevertheless, the time reversal approach is more complex and in some cases, depending on the geometry, some energy might be deposited on the ribs. In such a case, it means that some energy has been able to cross the ribs on its way to the array, in the first step of the time reversal process. As some ribs were thinner on their sides, ultrasound rays arriving on these points could be incompletely absorbed. Furthermore, guided modes at the surface of the ribs can also radiate toward the array in the first step of the time reversal process. Of course, the amount of transmitted energy varies, depending on the exact geometry (namely the angle between the rib and the transducers array). Even though no obvious trend can be related to intercostal spacing, the SAR gain is strongly enhanced by this time-reversal approach in all cases, confirming the results previously obtained.[17]

Similarly, the efficiency of the non invasive DORT projection approach is not constant: as for the time reversal approach, different amounts of information can be collected with the DORT method, depending on the exact geometry. In this case, SAR gain seems to depend less on the intercostal space than on the orientation of the ribs. Nevertheless, the efficiency of this novel noninvasive approach is clearly demonstrated as the SAR gain is at least 20 fold better than without correction.

V. CONCLUSION

In this study, the DORT method was applied to chest wall models composed of three ribs immerged in water. Contrary to most DORT studies, the information provided by the measured array response matrix was used to avoid sending energy on the scatterers instead of focusing on them. Experimental results showed that the projection of an emission orthogonally to the eigenvectors associated with the scatterers leads to an emission that avoids these scatterers. The projected emission was compared with the original one, and with an invasive time-reversal experiment. It was shown that the projected emission lead to a focusing similar to the time-reversed emission, and could enhance up to 100 fold the efficacy of the ribs sparing .

Such a noninvasive approach could be implemented clinically in order avoid skin-burns in transribs liver therapy. The use of multi-element therapeutic arrays would be mandatory, but such systems are becoming more and more widespread.[26-30]

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Tables

Table I. Comparison of the SAR gain obtained with three different emissions, in 5 different geometries. The last column gives the enhancement of the SAR gain for the emission C.

Index of the case	Intercostal spacing / mm	SAR gain for emission A (time reversal in water)	SAR gain for emission B (time reversal with the ribs)	SAR gain for emission C (projection of emission A)	$\frac{SAR\,gain(C)}{SAR\,gain(A)}$
1	30	0.12	13	3.0	25
2	29	0.12	23	7.5	62.5
3	24	0.085	11	9.0	106
4	24	0.14	7.0	2.8	20
5	20	0.017	1.4	1.3	76

Figures Captions

Fig. 1.a : *Schematic diagram of the experimental setup.*

Fig. 1.b: *Pictures of the ribs.*

Fig. 2 : *Principle of time reversal mirror.*

Figure 3 : *Typical singular values distribution at central frequency (1.5 MHz)*

Figure 4 : *Numerical repropagation of the eigenvectors from the echographic array to the plane of the ribs (f=1.5 MHz). The lateral position at 0 mm corresponds to the center of the echographic array. (a) first eigenvector, (b) 37th eigenvector, (c) 40th eigenvector*

Figure 5 : *Superposition of the numerical back-propagation of the 20 first eigenvectors from the echographic array to the plane of the ribs. The lateral position at 0 mm corresponds to the center of the echographic array. The ribs' position is shown in dotted line. (f=1.5 MHz).*

Figure 6 : *Typical emissions : (a) A (obtained with a time-reversal experiment in water), (b) (obtained with a time-reversal experiment through the ribs), (c) C (obtained by projection of emission A orthogonally to the first 40 eigenvectors)*

Figure 7 : *Pressure field measured in the focal plane after (a) emission A (b) emission B, (c) emission C.*

Figure 8 : *Normalized energy distribution (dB scale) in the focal plane obtained with emission A (dashed line), B (full line) and C (dotted line). The lateral position at 0 mm corresponds to the center of the focal spot.*

Figure 9 : *Pressure field measured in the plane of the ribs after (a) emission A (b) emission B (c) emission C. The lateral position at 0 mm corresponds to the center of the focal spot.*

Fig. 10 : *Normalized energy distribution (left dB scale and right linear scale) in the plane of the ribs obtained with emission A (dashed line) , B (solid line) and C (dotted line). The lateral position at 0 mm corresponds to the center of the focal spot.*