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A Method for Distribution Control of Aerial Ultrasound Radiation Pressure for Remote Vibrotactile Display

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Abstract: We propose a calculation method for phase and amplitude of ultrasound transducers in arrays, which are supposed to generate proper spatial distribution of radiation pressure on users' skin corresponding to vibrotactile sensation conveyed to users. The phased array technique has been applied for ultrasound transducer array and converged energy in a focus of magnified amplitude in the air has been successfully generated as well as vibrotactile stimulation in a point. In this paper we expand that phasing scheme for a single focus to arbitrary three-dimensional distribution amplitude with transducers located in three-dimensional disposition. We verify the validity of our method with numerical experiments.
Keywords: Tactile Display, Airborne Ultrasound, Amplitude Control, Inverse-problem

1. INTRODUCTION

It has been of a great interest of many researchers to reveal and establish a new application of human haptic information. A number of framework and devices have been investigated and proposed for display of tactile stimuli. They are often related to some specific sort of tactile stimuli which are related to certain virtual objects so that the user can tactually feel the existence of them. They can be categorized into mechanical and electrical methods. Most of current tactile displays are classified in former, on which mechanical actuator like wires, motors, vibrators are mounted. They basically are supposed to be attached on a part of human bodies and give tactile sensation on a region in contact with the devices. The same goes for the latter, which gives tactile stimuli via electrodes on the surface of the skins or connected to nerves transmitting sense signals.

Our approach has been different in terms of that we give vibrotactile sensation remotely. We have proposed a framework to generate a focal spot of airborne ultrasound, which enables us to feel localized tactile sensation in the air without wearing any devices on our bodies. Our newest prototype can generate a focus of static pressure of 7.4gf in force and switch the focal amplitude quantized into 320 levels with the sampling rate of 2 kHz (Figure 1). We fabricated multiple ultrasound transducer units which operate simultaneously in harmony. As a result, we realized enlarged aperture of an ultrasound phased array and generated a focus without blur in further region from the device than we did in the past research with a single transducer unit. With the framework we have developed several interfaces including the visual and tactile display which allows us to tactually feel virtual visible objects [1][2]

The concept of composing a larger aperture with multiple units can be expanded to three-dimensional aperture structures. Multiple units with different directions of their surface normal vectors can form more spatially complicated energy distribution in comparison with a single planar transducer array. With a proper spatial distribution of ultrasound pressure, it is possible to enable users to feel aerial tangible objects with their three-dimensional volumes such as a floating ball.

Figure 1: Our newest system of the 9-unit AUTD. The total area of the 3 by 3 units is 57×45cm^2. Vibrotactile sensation is produced on the user's palm by focused ultrasound emitted from the transducer array mounted above the user.

In order to create a given planer output amplitude distribution in a far field, it has been well known that the proper phases and amplitudes at arbitrary points on a lens as a function of two-dimensional coordinates are given as the spatial Fourier inverse transformation of the resultant image. However this solution is feasible only in generating a two-dimensional image which is parallel to the aperture [3]. When the resulting image should be three-dimensional, other solution will be necessary.

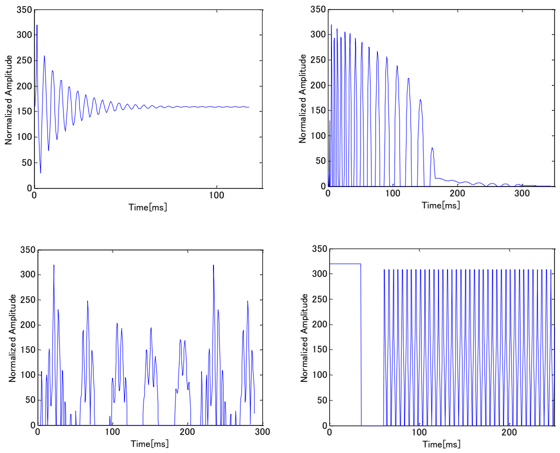

Figure 2: Examples of output radiation pressure waveforms. Each of them produces its own vibrotactile texture (impulsive, rapturous, trembling, buzzing.)

In this paper we propose the method to obtain appropriate output phase and amplitude of each transducer located in a three-dimensional way. The resulting image does not need to be planar. We formulate this issue as a simple inverse problem and solved with common least square error method. Several numerical examples to validate our method are given in the paper as well.

2. PRINCIPLES OF AIRBORNE ULTRASOUND TACTILE DISPLAY

Our research group has been researching and developing the airborne ultrasound tactile display (AUTD). The AUTD is a device which generates focused ultrasound on human skin by which remote tactile sensation is caused [3]. An array of ultrasound emitting transducers is mounted on it. The driving signal frequency of all transducers is set to be identical (40 kHz) and the phase delay on each transducer is set in order that the output ultrasound at the desirable focal point may be in phase. The focal ultrasound amplitude becomes very high and a phenomenon called radiation pressure arises [4]. It is a static pressure generated on the surface of an object blocking the sound propagation of very high amplitude. Its intensity P is proportional to the energy density of the sound field:

$$P = \frac{\alpha p^2}{\rho c^2} \tag{1}$$

where p[Pa] denotes the sound amplitude, c[m/s] denotes the sound velocity in the medium, ρ[kg/m^3] denotes the density of the medium and α denotes the coherent coefficient determined by the acoustic impedance of the medium and the object to block off the ultrasound propagation. When ultrasound propagates in the air and blocked off by liquid or solid, almost all of the ultrasound is reflected on the boundary and in this case the coefficient α becomes nearly 2. Here p is the instantaneous RMS of the sound amplitude. The direction of the pressure is normal to the object surface. By varying the sound amplitude temporally, the resultant radiation pressure intensity can be time-variant. Our current prototype can switch the focal intensity with a sampling rate of 2 kHz and 320-level quantization. The sampling rate 2 kHz is determined based on the knowledge that human beings can perceive only 1 kHz or lower vibrations on the skins [5]. This is an amplitude modulation of carrier ultrasound and its temporal envelope is what a user feels as vibrotactile sensation. To sum up, with this prototype we have successfully generated a rich variety of vibrotactile sensation on a single point on the skin surface (Figure 2) [6].

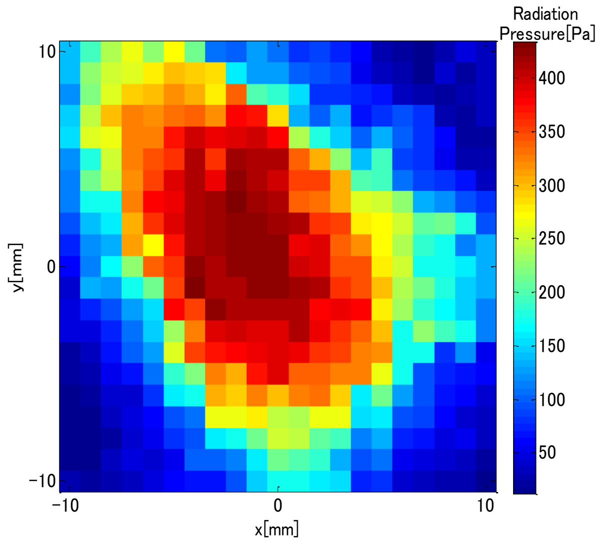

Figure 3: Spatial distribution of radiation pressure amplitude around the focal point 60 apart from the ultrasound emission surface.

There is another remarkable improvement of this version. A large aperture of ultrasound emitting surface is composed by integrating multiple ultrasound transducer array units [6][7]. With this implement it has become viable to make a non-blurred focus in a further region than with a previous version. Moreover, since the number of transducers was greatly increased, a radiation pressure focus whose intensity was dramatically increased was produced.

The reason why the spatial distribution produced by AUTD has been merely a sole point is that it is the optimum strategy to create the most intense energy density. Nevertheless, with a single array unit, the focal intensity was only 1.6 gf. Due to this limited output intensity, it has not been practical to creating other spatial distribution. The current version of AUTD comprises 9 units and produces a focus of 7.4gf intensity as a result (Figure 3) [6].

With the intensity mush increased, there would be no need for generated spatial amplitude distribution to be limited to a single point. It may be possible that users can feel tactile sensation with less localized amplitude distribution. If so, spatially enriched vibrotactile sensation that expends in the three-dimensional space can be realized. In the following, we describe how we create complicated spatial amplitude distributions with the AUTD.

3. SOUND PROPAGATION MODEL

In optics, it is known that when a two-dimensional planer image propagates and is projected on the screen parallel to the image plane, the spatial Fourier transformation of the image is obtained at the screen. Therefore in order to obtain the intensity distribution which creates a desired image on the screen, the Fourier inverse Transform of the image gives the answer. The same goes for ultrasound imaging [3].

However, when the resultant image is not parallel to the ultrasound array or it is defined in the three-dimensional space, the solution above is no longer available. Here we start with a simple and generalized model of sound propagation.

Let N be the number transducers and $\boldsymbol{r}_i$ be the position of the transducers with an index i (i = 1, … , N.) Each transducer transmits sinusoidal ultrasound wave with the amplitude p_i and phase shift θ_i. The common ultrasound wavenumber of all transducers is set to k. The complex expression of output sound amplitude of the ith transducer is given as

$$p_i e^{-j\theta_i}. \tag{2}$$

When all output sound interferes at the location $\boldsymbol{q}$, with their individual delays, the resultant sound pressure $p(\boldsymbol{q})$ becomes

$$p(\boldsymbol{q}) = \sum_{i=1}^{N} a_i p_i e^{-j(k\|\boldsymbol{q} - \boldsymbol{r}_i\| + \theta_i)}, \tag{3}$$

where a_i denotes the ith attenuation coefficient determined the distance between the transducer and the location of interest, usually the inverse proportional of powers of the distance $\| \boldsymbol{q} - \boldsymbol{r}_i \|$ or inverse exponential of it. By substituting the complex output amplitude of the ith transducer $g_i = p_i e^{-j\theta_i}$, we have

$$p(\boldsymbol{q}) = \sum_{i=1}^{N} a_i e^{-j(k\|\boldsymbol{q} - \boldsymbol{r}_i\|)} g_i. \tag{4}$$

Figure 4 depicts the positional relation between these parameters. The equation above indicates the relation between the series of output transducer signals and the resultant sound pressure distribution. In this equation, the parameter we want to obtain is simply expressed as g_i. In case of a single focus generating, g_i should be a multiple of $e^{j(k\|\boldsymbol{q} - \boldsymbol{r}_i\|)}$ so as to compensate the phase delay of each transducer.

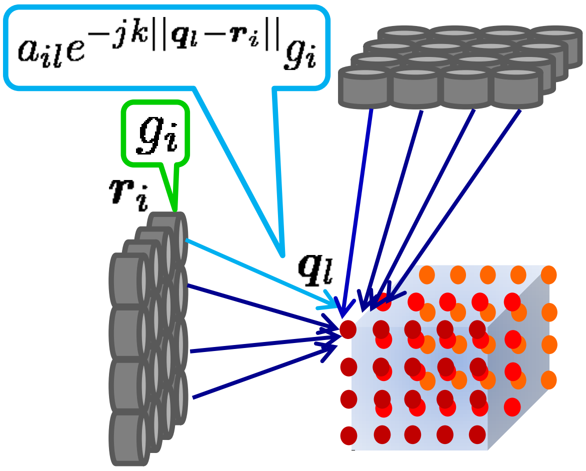

Figure 4: Schematic diagram showing the problem situation.

4. OBTAINING THE OUTPUT GAIN OF EACH TRANSDUCER

Here we consider the case of producing more complicated spatial amplitude distributions. We handle the problem with a discrete inverse-problem framework. We define the most proper g_i as the ones creating $p(\boldsymbol{q})$ with least error under given $p(\boldsymbol{q})$. $p(\boldsymbol{q})$ must be defined in advance at many discrete points, $\boldsymbol{q}_1$, … ,$\boldsymbol{q}_M$, namely. We apply a least-squared-error criterion to the problem. Finally, the problem is formulated as, "find g_i which minimizes the gross error

$$\sum_{l=1}^{M} \left\| p(\boldsymbol{q}_l) - \sum_{i=1}^{N} a_{il} e^{-j(k\|\boldsymbol{q}_l - \boldsymbol{r}_i\|)} g_i \right\|^2 ." \tag{5}$$

Here the attenuation coefficient a_i is replaced by a_{il}, a coefficient with two suffixes. This is because the attenuation ratio can be determined for all combinations of the transducers and points where an ultrasound image is produced (i.e. $\boldsymbol{q}_1$, … ,$\boldsymbol{q}_M$). In order to facilitate handling the equation above, we introduce linear-algebraic expressions. Let $\boldsymbol{b}_l$, $\boldsymbol{g}$ and $\boldsymbol{p}$ be defined as below:

$$\boldsymbol{b}_l = \left[a_{l1} e^{-j(k\|\boldsymbol{q}_l - \boldsymbol{r}_1\|)} \quad \ldots \quad a_{lN} e^{-j(k\|\boldsymbol{q}_l - \boldsymbol{r}_N\|)}\right],$$

$$\boldsymbol{g} = \begin{bmatrix} g_1 \\ \vdots \\ g_N \end{bmatrix}, \tag{6}$$

$$\boldsymbol{p} = \begin{bmatrix} p(\boldsymbol{q}_1) \\ \vdots \\ p(\boldsymbol{q}_M) \end{bmatrix}.$$

Components of $\boldsymbol{b}_l$ denote the phase and amplitude change via sound propagation. A series of the vector $\boldsymbol{b}_l$ can define a matrix $\boldsymbol{B}$:

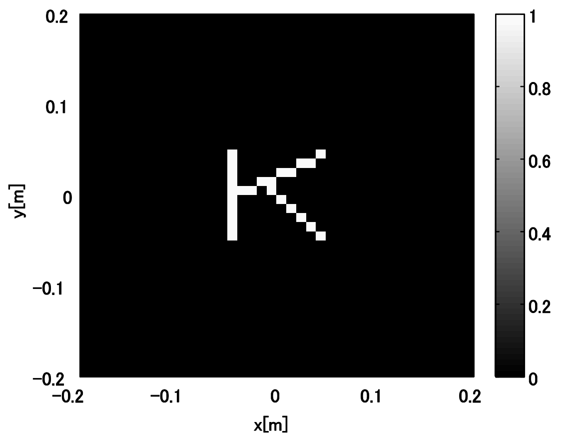

Figure 5: Desired spatial amplitude distribution to be produced under individual conditions.

$$\boldsymbol{B} = \begin{bmatrix} \boldsymbol{b_1} \\ \vdots \\ \boldsymbol{b_N} \end{bmatrix}. \tag{7}$$

With these vectors and matrixes, (5) can be rewritten as the following very simple form:

$$\|\boldsymbol{p} - \boldsymbol{Bg}\|^2. \tag{8}$$

For the minimization of a vector norm (8), in case the parameter is under no constraint, the solution is given by

$$\boldsymbol{g} = (\boldsymbol{B}^H\boldsymbol{B})^{-1}\boldsymbol{B}^H\boldsymbol{p}, \tag{9}$$

where $\boldsymbol{B}^H$ denotes the conjugate transpose matrix of $\boldsymbol{B}$. In general, this solution is determined identically only in the case the rank of $\boldsymbol{B}$ is greater than N. The rank of $\boldsymbol{B}$ highly depends on the number of ultrasound field points M and their spatial distributions. In case that the output gain vector should be obtained under some constraint, minimization method under constraints such as the Lagrange multiplier method can be applied.

5. NUMERICAL EXPERIMENT

We validated the effectiveness of our method through some numerical simulative experiments. Figure 5 shows the desired spatial distribution of sound pressure amplitude. Note that although in the experiments the produced pattern was defined only in two-dimensional plane, the proposed method can be applied to the patterns spreading in three-dimensional way. The pattern was defined at discrete points in lattice for every 10 mm. The simulation for generating this pattern was conducted for two different positions of pattern generating plane (1, 2) and two different arrangements of transducers (a, b). So we have results of simulations under four different combinations of each two experimental condition, (1a), (1b), (2a), (2b). These conditions are schematically depicted in Figure 6. The region (1) is a square *xy*-plane of 400 mm x 400mm and

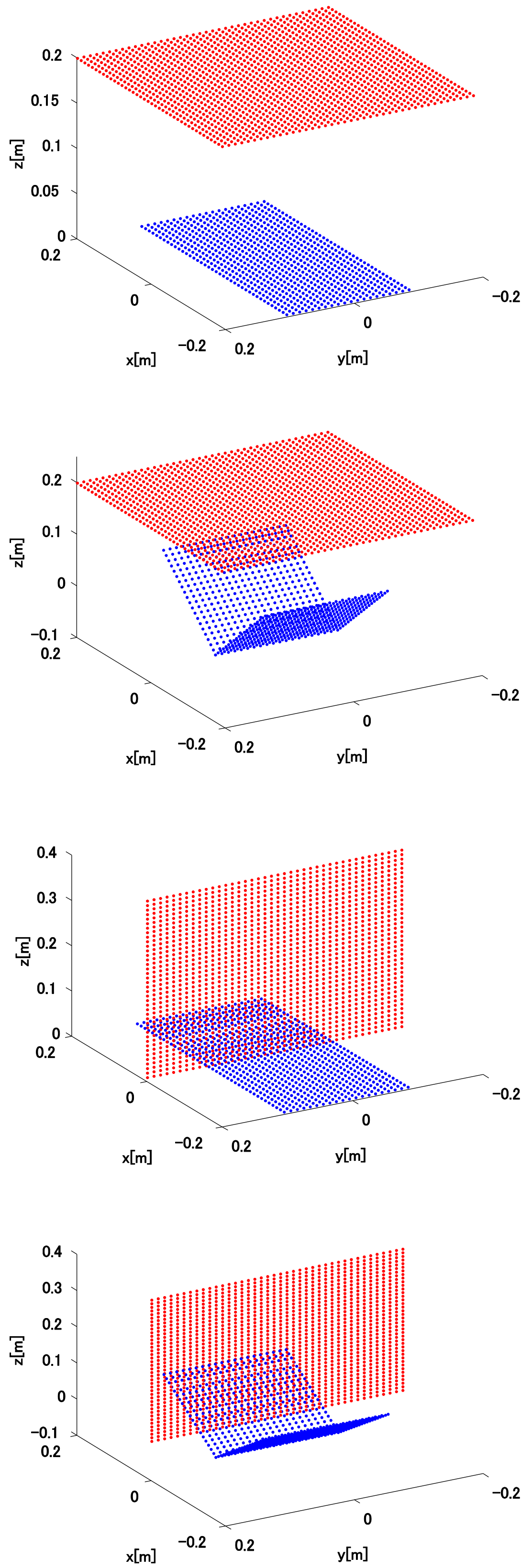

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Figure 6: Four location sets of transducers (blue points) and regions in which spatial patterns of sound pressure are generated (red points), 1a(top), 1b(second), 2a(third) and 2b(bottom) depicted respectively

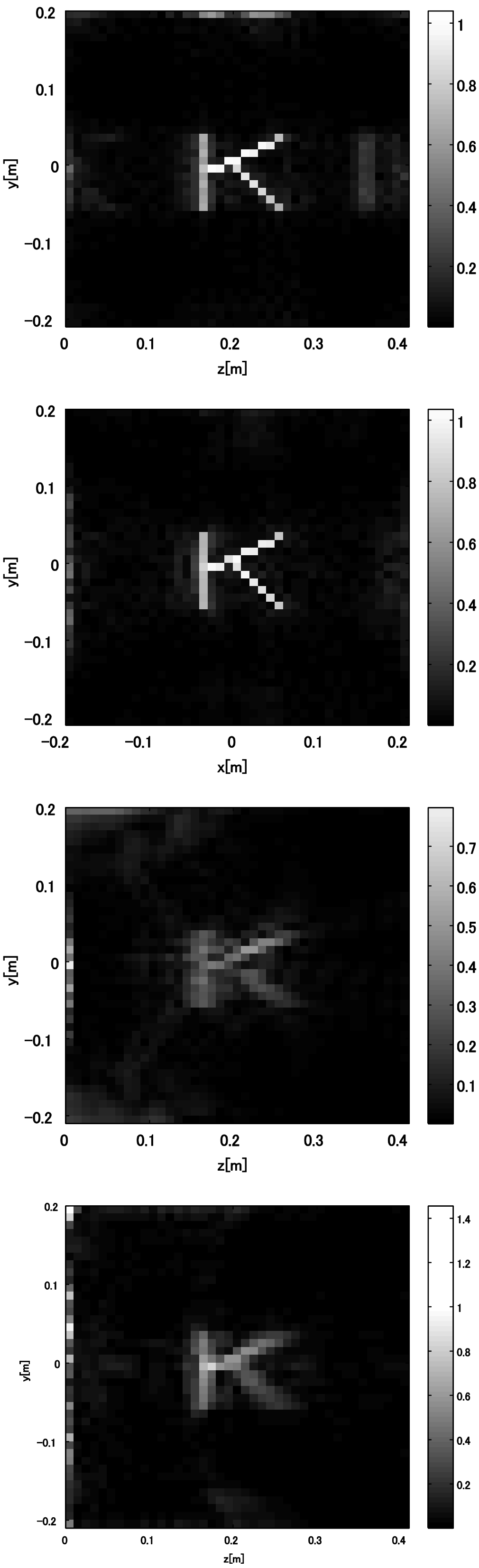

Figure 7: Normalized spatial amplitude distribution of sound pressure generated with the location 1a (top figure), 1b (second), 2a (third), 2b (bottom)

centered on the z axis at z = 200mm. The other region (2) is a square of same size which is parallel to the yz-plane and centered on the coordinate (x, y, z) = (0, 0, 200mm). The transducer arrangement (a) is a single planar lattice of every 10 mm. The lattice region is a part of the xy-plane centered on the coordinate origin and its size is 20mm in x axis and 40mm in y axis. The other arrangement (b) is created by folding the arrangement (a) in half toward the direction of the z-axis so that the ridge of the resultant transducer array may remain in contact with the xy-plane.

For each of these four arrangements, the optimum output gain of each transducer was calculated by the equation (9) and simulated the spatial distribution of amplitude produced by the obtained output sound waves. For simplicity, all attenuation coefficients are set to 1.

The results are shown in the Figure 7. Little difference in the resultant spatial amplitude between (1a) and (1b) is seen. On the contrary it is obvious that the imaging performance in (2a) is more degraded than in (2b). The spatial detail and maximum power are both lower in (2a). The reason is because it is difficult for transducers to make sharp focuses in the depth (z-axis) direction due to small extent of transducer locations in the xy-plane and no spatial flexibility of arrangement in z-axis. In the case of the arrangement (2b), the positional freedom of transducers is guaranteed in all of x, y and z axis.

The experiment has shown that the spatial freedom in transducer arrangement directly affects the imaging performance. The directive dependency of imaging performance can be unbiased by spatially uniform transducer arrangement.

6. SUMMERY AND THE FUTURE PERSPECTIVE

In this work we propose to construct a three-dimensional transducer array to create three-dimensional radiation pressure distribution for vibrotactile sensation with varying depth. The simple ultrasound propagation model is introduced, which is proved to be expressed in a linear-algebra manner. We obtained the optimum series of output amplitudes and phase delays of transducers by means of the minimization of a quadratic form. In the numerical experiment, we verified the validity of the obtained solution and discussed the relation between the transducer arrangements and the imaging performances with their dependency on the direction from the transducer array.

We are now implementing the actual device to create such three-dimensional vibrotactile sensation. To control the direction of the generated pressure is also included in our future work.

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