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Measurement of Speed of Sound in Skull Bone and Its Thickness Using a Focused Ultrasonic Wave

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To quantitatively improve brain SPECT (single photon emission computed tomography), we propose a noninvasive method for determining the speed of sound in skull bone and its thickness. We assume a spherical shell structure for the skull, and the speed of sound to be constant in the bone layer. Small ultrasonic transducers are arranged on the outer surface of the bone layer to form a transducer array. The array is excited to produce an ultrasonic pulse wave focused at a point on the inner surface by assigning an appropriate time delay to each of the transducers. The pulse is propagated to the focal point, and a reflected spherical pulse wave is produced at that point. The reflected pulse wave is detected by the same transducers, and the output from each transducer is added to give a summed signal. When the time delays are optimized, we obtain the maximum envelope amplitude for the signal. A simulation study is conducted to confirm the method, and the potential error is examined for ideal conditions. From the simulation results, we conclude that the proposed method is suitable for SPECT if the transducers are arrayed with a sufficiently large aperture. [DOI: 10.1143/JJAP.41.3327]

KEYWORDS: thickness, speed of sound, skull bone, focused ultrasonic wave, transducer array

1. Introduction

In SPECT (single photon emission computed tomography), the quantitative accuracy is affected by factors such as measurement noise, attenuation and scattering in the sample, and detector response. In particular, compensation for attenuation and scattering in skull bone is the most important issue to be considered for brain SPECT.[1,2] In order to achieve accurate compensation, knowledge of the thickness and density of skull bone is required. These quantities are, in general, measured by X-ray CT or transmission-dependent convolution subtraction (TDCS);[3] however, ultrasonic methods are preferable because they are less hazardous to the human body. By measuring the acoustic impedance Z and the speed of sound v in the bone, the density ρ can be determined from the relation $Z = \rho v$.

We have previously proposed an in vivo method for measuring the acoustic impedance of bone with unknown thickness.[4] Such an in vivo method is important because most conventional ultrasonic methods can only be used under in vitro conditions.[5,6] We also developed a measurement method that simultaneously determines both the speed of sound in skull bone and its thickness through in vivo measurements.[7] However, in the previous method, the skull bone is assumed to be a flat slab, i.e., a layer with parallel planar surfaces. Therefore, when we apply the method to actual skull bone, accurate measurements cannot always be achieved because real skull bone has a locally spherical shell structure.

In this paper, we propose an in vivo method to simultaneously measure the speed of sound in skull bone and its thickness, even when the skull bone is assumed to have a spherical shell structure. We assume that the speed of sound is constant in the shell and that the shape of the outer surface of the skull bone is known. An array of small transducers is arranged on the surface, and the transducers are excited to produce an ultrasonic pulse wave focused at a point on the inner surface of the bone layer by assigning an appropriate time delay to each transducer. The pulse is propagated to the focal point, and a reflected spherical pulse wave is produced at that point. The reflected pulse wave is detected by the same transducers, and the output from each transducer is added to give a summed signal. When the time delays are optimized, we obtain the maximum envelope amplitude for the signal. The proposed method, called the focusing method, is also applicable to the case where the shape of the skull bone is a flat slab, for which it gives more accurate results than the previous method.[7] In the case of a flat slab, the transmitted wave can be focused on the closest point to the center transducer on the inner surface of the bone. In contrast to the previous method, however, the focusing method requires a long measurement time to achieve high accuracy because the processing involves a search strategy. To shorten the measurement time, we also propose a nonfocusing method to provide a rough estimate that can be used as an initial value for the focusing method. We conduct simulations for a flat slab structure under ideal conditions in order to examine the degree of possible measurement error. From the results obtained through these simulations, we confirm the feasibility of the proposed method.

2. Measurement Method

2.1 Principle: focusing method

As shown in Fig. 1, we assume that the shape of the skull bone has a locally spherical shell structure, and arrange an array of ultrasonic transducers on the surface of the skull bone. If each elemental transducer is excited with the appropriate lead time in advance of the center element, the ultrasonic wave generated by the transducer array is focused on a point P as indicated in Fig. 1. The focal point is located on the inner surface of the bone and is the point closest to the center element. To simplify explanation, the transducer array is assumed to be one-dimensional, i.e., the transducers are arranged as a curved line on the outer surface of the cross section of the bone as shown in Fig. 1. With the assumptions that the speed of sound in the bone v is constant, the num-

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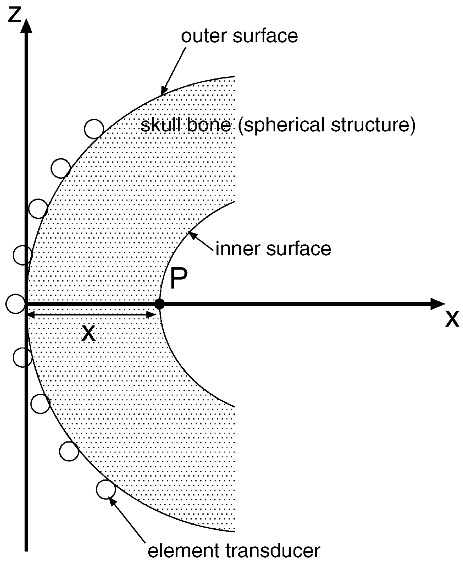

Fig. 1. Schematic diagram of shape of skull bone and arrangement of transducers with coordinate system.

ber i and coordinates $\{(x_i, z_i)\}$ of each element are known, and that each element is a point sound source, the appropriate lead time for each transducer τ_i can be calculated using the following equation

$$\tau_i = \frac{\sqrt{z_i^2 + (x - x_i)^2} - x}{v}. \tag{1}$$

The objective of the measurement, here, is to determine x and v using the above equation. If the aperture size and the number of elemental transducers in the array are both infinite, the amplitude pattern of the focused wave can be modeled as a delta function with the above appropriate lead times. However, the lead times are initially unknown, and if inappropriate values are employed, complete focusing cannot be achieved. In this case, the amplitude of the envelope of the echo signal is smaller than that with appropriate lead times satisfying eq. (1). Therefore, in the proposed measurement procedure, $\{\tau_i\}$ is varied sequentially and the amplitude of the envelope of each signal obtained from the focused pulse waves formed with each $\{\tau_i\}$ is measured. τ_i is varied by changing x and v in eq. (1). The amplitude of the envelope of the echo signals can then be plotted with respect to these x and v values. The values of x and v that give the echo signal with maximum amplitude among all of the measurements are considered to be the true values. Note that the focusing method can be applied to skull bone with arbitrary shape as long as the transmitted wave can be focused on the point closest to the center transducer on the inner surface of the bone.

2.2 *Initial value measurement: nonfocusing method*

The focusing method described in the previous section is essentially based on a search strategy in the parameter space of x and v. This search strategy involves an inherent quantization error which is proportional to the quantization size of the parameter space, in addition to the unavoidable errors caused by the finite aperture size and the finite number of elements in the array. In order to reduce the quantization error and the time required to measure the amplitude of the envelope of the echo signals, both the quantization size and the total definition size of the parameter space should be small. This requires that the values of x and v be estimated prior to conducting measurements using the focusing procedure. In this section, we propose a nonfocusing measurement method that provides a rough estimate of x and v for use in the focusing method. Using these rough values, we can set the total definition size of the parameter space to be much smaller that without premeasurement. The search for the true values of x and v in this small parameter space around such rough estimates then becomes much more efficient.

The round-trip time t_0 of the ultrasonic pulse from the center elemental transducer to the point P in Fig. 1 is given by

$$t_0 = \frac{2x}{v}. \tag{2}$$

We additionally assume that the ultrasonic pulse transmitted from the element at position (x', z') is reflected at P and received by the element at $(x', -z')$ with time delay t, i.e., the two elements are arranged symmetrically with respect to the x-axis. Then, t can be expressed as follows.

$$t = \frac{2\sqrt{z'^2 + (x - x')^2}}{v}. \tag{3}$$

From eqs. (2) and (3), the equation including only one unknown variable x is derived as

$$\frac{t}{t_0} = \frac{\sqrt{z'^2 + (x - x')^2}}{x}. \tag{4}$$

Consequently, the following quadratic equation is obtained.

$$\left\{ \left(\frac{t}{t_0} \right)^2 - 1 \right\} x^2 + 2x'x - (x'^2 + z'^2) = 0. \tag{5}$$

Since x must be positive, the solution of eq. (5) is given as

$$x = \frac{-x' + \sqrt{x'^2 + \left\{ \left(\frac{t}{t_0} \right)^2 - 1 \right\} (x'^2 + z'^2)}}{\left(\frac{t}{t_0} \right)^2 - 1}. \tag{6}$$

Repeating the measurement of t in eq. (3) for the pair of elements at (x', z') and at $(x', -z')$, we obtain the data set $\{t\}$. We can then obtain x with respect to each t using eq. (6), and calculate the average $\bar{x}$, for example, the sample mean of the above set $\{x\}$. Using this average $\bar{x}$, $\bar{v}$ can be obtained from eq. (2).

If we can obtain many measurements for t and model the exact statistical characteristics of the measured values of $\{t\}$ and t_0, we can calculate the minimum variance unbiased estimators of x and v, which are more accurate than the above $\bar{x}$ and $\bar{v}$. However, such optimal estimators for the nonfocusing method cannot be calculated analytically. The focusing method is superior to the nonfocusing method from the point of view that the statistical characteristics of the measurements are not required in the former. Additionally, the accuracy of the nonfocusing method when applied to skull bone having an asymmetric shape with respect to the x-axis is degraded, whereas the accuracy of the focusing method is not. On the basis of these considerations, we use the simple estimates $\bar{x}$ and $\bar{v}$ obtained by the nonfocusing method as initial estimates for the focusing method, i.e., in the focusing method, a small search region in the parameter space can be selected around $\bar{x}$ and $\bar{v}$.

3. Numerical Evaluation of Focusing Method

An ideal transducer array would have an infinite number of elements and an infinite aperture size. The potential accuracy of x and v obtained by the focusing method may therefore be affected by the finite number of elements and finite aperture size of a practical transducer array. To evaluate the accuracy of the focusing method in the absence of other complicating factors, such as the structure of the bone and the uniformity of the speed of sound, we conducted simulations of the focusing method using a flat slab structure under ideal conditions. Ideal conditions, here, mean that reflection occurs only at the center point P, v is uniform in the bone, each transducer element is an ideal point source, and that there is no measurement noise. The thickness x is set at 5 mm, the speed of sound v is 3000 m/s, and the other parameters are set as shown in Table I. The elements of the transducer array are arranged along a one-dimensional straight line.

Varying the predicted x from 4.0 mm to 6.0 mm and the predicted v from 2600 m/s to 3400 m/s, we calculated the waveform of the transmitted wave for the lead time formulated in eq. (1), where x and v in eq. (1) are varied within the predicted parameter space. We then calculated the waveform of the echo signal received by the transducer array with the same lead time as that set for transmission. The quantization sizes of x and v were set at 0.01 mm and 4.0 m/s, respectively. The echo signal for values of x and v equal to the set values, i.e., $x = 5$ mm and $v = 3000$ m/s, is shown in Fig. 2. The transmitted pulse was modeled to have a Gaussian envelope, as used in practice.

Table I. Values of parameters used in simulations.

Center frequency [MHz]	Pulse width [μs]	Sampling frequency [MHz]	Number of transducers
2	2	100	33

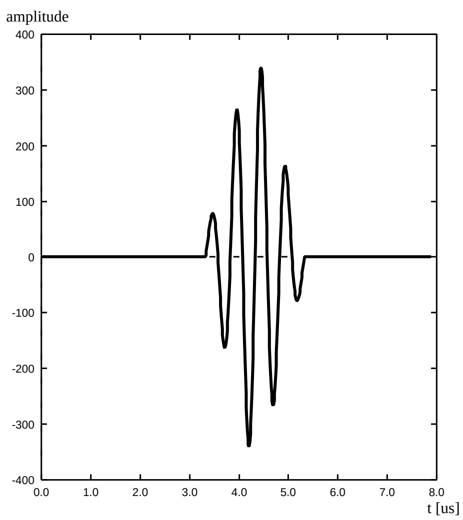

Fig. 2. Echo signal corresponding to true values of x and v.

In the actual reflection process, the echo signal consists of the waves reflected at both the center point P and in the neighborhood of P. In the present simulation scheme, it is not possible to determine the effect of sidelobes of the focused wave because only the amplitude of the envelope of the ultrasonic wave reflected at P is calculated in these simulations, and the amplitude of the wave reflected in the neighboring region mainly depends on the number of elements in the transducer array. The aperture size of the transducer array mainly determines the sharpness of the focusing, which is also predicted to affect the performance of the focusing method. These calculations were then performed for aperture sizes of 5 mm, 10 mm and 20 mm. The resulting map of amplitude is shown in Fig. 3 for these three aperture sizes. In these maps, the set values of

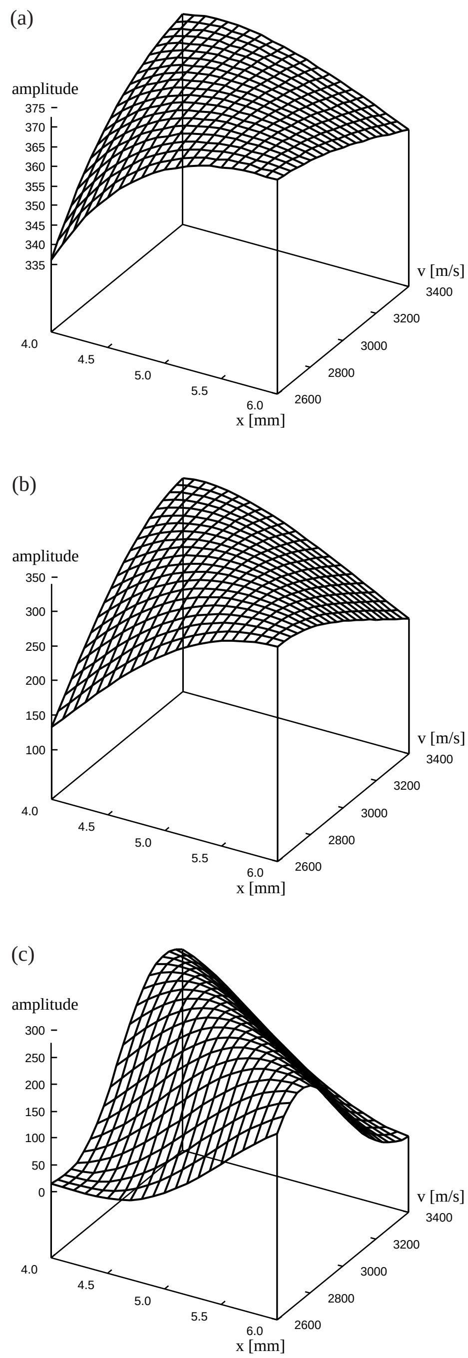

Fig. 3. Map of amplitude of envelope of echo signal for numerical simulations with aperture sizes of (a) 5 mm, (b) 10 mm, and (c) 20 mm.

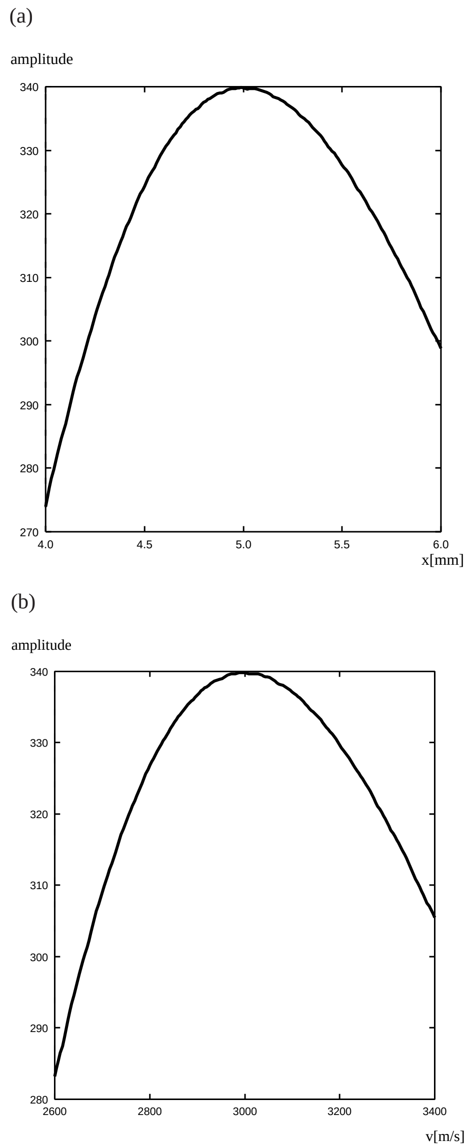

Fig. 4. Cross sections of (a) x and (b) v through center of map shown in Fig. 3(b).

Table II. Evaluated relative measurement errors of thickness and speed of sound for 5 mm layer thickness.

Aperture size [mm]	Error of x [%]	Error of v [%]
5.0	13.4	13.2
10.0	1.8	1.2
20.0	0.6	0.3

Table III. Evaluated relative measurement errors of thickness and speed of sound for 3 mm layer thickness.

Aperture size [mm]	Error of x [%]	Error of v [%]
3.0	14.3	13.3
6.0	4.7	3.1
12.0	1.0	0.4

x and v are positioned at the center of the parameter space. Cross sections of x and v through the center of the map in Fig. 3(b) are shown in Fig. 4.

The error in these calculated results can then be determined by comparing the point (x, v) corresponding to the echo signal with maximum amplitude in these maps to the set values $x = 5$ mm and $v = 3000$ m/s. This error will be indicative of the degree of error involved in the practical measurements of x and v. The errors for each aperture size are listed in Table II. Similar simulations were also conducted for a different layer thickness of $x = 3$ mm, with aperture sizes of 3 mm, 6 mm and 12 mm. The errors are shown in Table III. The results indicate that the measurement accuracy of the speed of sound in skull bone and its thickness depends significantly on the aperture size. In particular, whereas the errors for the aperture size of 5 mm in Table II and 3 mm in Table III are larger than that of the previous method,[7] the errors for the other aperture sizes are comparatively very small. This method is therefore suitable for practical measurement if a transducer array with sufficiently large aperture is used.

4. Conclusion

In this paper, we proposed a noninvasive method to simultaneously determine the speed of sound in skull bone and its thickness using an array of transducers. The method consists of a focusing method and a nonfocusing method, providing good accuracy and short measurement time. The potential measurement error of the focusing method, representing the major component of error in the measurement system, was evaluated through simulations under ideal conditions. In the future, we intend to conduct detailed simulations under realistic conditions, and examine the effects of sidelobes, spatial variations in the speed of sound in the bone, spherical-shaped bone, and measurement noise. We also plan to confirm the effectiveness of the measurement method in practical experiments.

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