An Imaging Method Using the Interaction between Ultrasound and Magnetic Field

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Abstract A new imaging method using the classical Hall effect has been developed in the context of diagnostic applications. “Hall effect imaging” (HEI) relies on ultrasonic signal generated by a pulsed current through the sample in a strong magnetic field. Hall effect images reflect the dielectric distribution of the sample. Phantom images have been collected with a single crystal sensor. Since dielectric parameters vary greatly among soft tissues and between normal and pathological states, HEI holds promises for human imaging.

Introduction

The classical Hall effect describes the charge separation phenomenon in a conductive object moving in a magnetic field [1]. This charge separation is the result of the opposing Lorentz forces on the positive and negative charges, and leads to an externally detectable voltage, the Hall voltage. The Hall voltage amplitude is determined by the strength of the Lorentz force and the charge density and mobility. Denote the magnetic field $B_0$, the velocity of motion $v$, and the apparent conductivity of the object $\sigma$, the Hall voltage $V_h = \sigma v B_0$.

This is a mechanism that relates conductivity to a voltage signal via motion. If spatially localized motion is created with ultrasonic pulses, images reflecting the electrical properties of an object can be formed in the same manner as echo ultrasound.

To illustrate the dependence of the Hall voltage on the electrical and acoustic properties of the sample, consider a one-dimensional example:

\[ \frac{\partial}{\partial z}(\sigma / \rho) \]

Fig.1 Relation between the Hall voltage and the $\sigma / \rho$ gradient in the sample.

An ultrasound transducer generates a longitudinal wave packet along the "Z" axis perpendicular to the magnetic field $B_0$ (Fig.1A). A step change in conductivity $\sigma$ and mass density $\rho$ occurs between positions $z_1$ and $z_2$ (Fig.1B).
Using the equation of wave propagation, the Hall voltage can be expressed in terms of the ultrasound momentum \( M(z, t) \) and the spatial gradient of \( \sigma/\rho \):

\[
V_h(t) \propto B_0 \int_{\text{soundpath}} M(z, t) \frac{\partial}{\partial z} \left[ \frac{\sigma(z)}{\rho(z)} \right] dz
\]  

(1)

In applying Eq. 1 to the example above, the integrand becomes non-zero at the interfaces \( z_1 \) and \( z_2 \). Thus the time course of the Hall voltage contains two peaks (Fig. 1D). The \( \sigma/\rho \) gradient at \( z_1 \) and \( z_2 \) are in opposite directions, so are the polarities of the Hall peaks. In this fashion HEI converts spatial information into the time domain much like echo ultrasound.

The method described above is the voltage detection method of HEI. Based on the reciprocity relation of linear electro-mechanical systems [2], HEI can also be carried out in reverse. In the reverse mode, an electric field impulse is applied to the sample. Any location in the sample responds to the local electric field with a current density proportional to the local apparent conductivity. At interfaces of changing conductivity the current density becomes discontinuous, and so are the Lorentz forces on the currents. The discontinuities of the Lorentz forces result in ultrasonic pulses emanating from these interfaces. These pulses are then received by ultrasonic sensors on the surface. The times of arrival of the pulses mark the depth of the interfaces, while lateral resolution can be achieved by scanning successive lines with a single crystal sensor, or with phased array sensors.

The forward and reverse modes are theoretically a reciprocal pair, therefore they produce identical images. However, the reverse mode, or ultrasound detection mode, have some practical advantages. It is less susceptible to electromagnetic interference in the environment, since the signal is acoustic, and the level of external acoustic noise in the MHz range is usually low. Ultrasound detection also enables the use of phased array transducers and fast 2D or 3D image formation.

**Experimental Methods**

To demonstrate the feasibility of HEI, a simple device was constructed to image objects suspended in a bucket of saline solution, placed in a 2 tesla field:

![Diagram of the experiment to image with HEI a sample immersed in a saline bucket.](image)

The bucket was filled with 0.4% NaCl solution. A rectangular polycarbonate block (cross section 6.0cm \( \times \) 1.2cm) was immersed in the saline. In the ultrasound detection mode, a planar piezoelectric transducer (Panametric V314) was the ultrasound sensor. Pulsed electric field across the bucket was driven with a 3.5 kV biphasic pulse of 1\( \mu \)s nominal pulse width. Pre-amplification was realized with a 60dB broadband (10 kHz to 10 MHz) low noise preamplifier (Miteq). After a passive bandpass filter and another 20dB gain, the signal was recorded with a PC based digital oscilloscope (Gage Model 1012).

Two dimensional images were formed with the line scan method by moving the transducer in 0.5 cm increments across the bucket, while recording the ultrasound signal at each position. These traces were displayed side by side in gray scale after a magnitude calculation, to form a 2-dimensional image. To eliminate direct
electromagnetic coupling between the pulser and the ultrasonic transducer, careful shielding of the transducer was necessary.

The resulting image is shown below:

![Image](image.png)

Fig.3 An HEI image of a polycarbonate block in a saline bucket.

**Discussion**

In the ultrasound detection mode of HEI, the limit on sensitivity is the maximum current allowed in the object. In biological subjects the threshold is set by nerve stimulation. The duration of the electrical impulse determines the length of the ultrasound pulse it produces, and therefore the spatial resolution of the image. To achieve millimeter or higher resolution, the pulse duration must be on the order of a microsecond, or one hundredth the strength-duration time constant of human sensory and muscular nerves [3]. In this short time limit the nerve stimulation threshold is established as the product (electric field)×(pulse duration) ~ 2×10^3 Vs/m [4]. Based on this index, the peak pressure of the ultrasound signal from a fat-muscle interface can be estimated for a range of magnetic field strength:

![Graph](graph.png)

Fig.4 Peak ultrasonic signal vs. spatial resolution from a fat-muscle interface in the ultrasound detection mode, for a variety of field strengths, based on sensory nerve stimulation thresholds.

In water like samples at room temperature, the rms noise level for 1MHz bandwidth centered at 3MHz is 0.16 pascal/(radian)½, thus real-time imaging is possible at high field strength. Given the magnetic field strength, the S/N ratio can be greatly increased without causing sensory nerve stimulation by using biphasic excitation pulses, e.g., a complete sine wave cycle. The threshold for biphasic pulses is much higher than
monophasic pulses [5]. For in vivo applications with compact magnets, signal averaging may be necessary to achieve sufficient sensitivity.

**Conclusion**

Although this imaging technique is still at an early stage of development, it holds promises for good soft tissue contrast. This is because, compared to the parameters that give image contrast in echo ultrasound and magnetic resonance imaging (MRI), conductivity and dielectric constant vary widely among tissues[6]-[8]:

<table>
<thead>
<tr>
<th></th>
<th>sound V (meter/sec)</th>
<th>MRI T1/T2 (millisecond)</th>
<th>Conductivity (S/m)</th>
<th>Dielectric constant</th>
</tr>
</thead>
<tbody>
<tr>
<td>muscle</td>
<td>1560</td>
<td>450, 65</td>
<td>0.71, 2200</td>
<td></td>
</tr>
<tr>
<td>fat</td>
<td>1470</td>
<td>150, 150</td>
<td>0.04</td>
<td></td>
</tr>
<tr>
<td>whole blood</td>
<td>1560</td>
<td>525, 260</td>
<td>0.71, 2040</td>
<td></td>
</tr>
<tr>
<td>liver</td>
<td>1540</td>
<td>250, 45</td>
<td>0.20, 1970</td>
<td></td>
</tr>
<tr>
<td>kidney</td>
<td>1560</td>
<td>400, 70</td>
<td>0.38, 2540</td>
<td></td>
</tr>
<tr>
<td>spleen</td>
<td>1560</td>
<td>400, 110</td>
<td>0.63, 1460</td>
<td></td>
</tr>
</tbody>
</table>

Therefore Hall effect imaging can potentially give comparable contrast level to MRI, while retaining the speed of ultrasonic imaging. In correlating electrical constants to tissue pathology, measurements in excised human and animal tissues showed that in the ultrasonic frequency range, conductivity is elevated 2 to 10 times in malignant tumor in several organs. For this reason, if phased array ultrasonic sensors can be adapted to HEI to obtain high quality images, it may be a useful diagnostic tool, and provide information that are not visible with the existing array of non-invasive imaging techniques.

**References**