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Volumetric Hall Effect Tomography – A Feasibility Study

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Abstract

Hall effect imaging is an ultrasound-based method of mapping spatial variations in the dielectric constants of an acoustically-uniform sample. This paper presents three-dimensional Hall effect images of phantoms obtained by scanning a single transducer across a two-dimensional grid, effectively simulating two-dimensional phased-array signal reception. The experiments demonstrate the feasibility of volumetric Hall effect tomography and show the advantage of volumetric scans over planar scans. The images reflect several limitations of the current scanning method and point to directions for further hardware development. The inherent limitations of Hall effect imaging are also discussed in light of these results.

Keywords

Biological; conductivity; dielectric constant; Hall effect; imaging; three-dimensional ultrasound; tomography

1. INTRODUCTION

Hall effect imaging¹ (HEI) is an ultrasonic method for mapping the heterogeneity in the electrical constants (conductivity and dielectric constant) of a sample that is acoustically uniform. The principle of the method is that, under a strong static magnetic field, a pulsed current through the medium produces ultrasonic signals from the Lorentz force. When these signals are received on the surface of the sample with an array transducer, an image can be reconstructed tomographically. This image reflects spatial variations in the electrical constants of the sample. This process can be reversed, where ultrasonic pulses are transmitted into the sample to produce rf voltage signals across electrodes attached to its surface. The forward and reverse modes of HEI give the same information. A detailed description of the method is given in reference¹.

Previous papers showed two-dimensional (2D) images collected with linear scans of single planar and focused transducers.^{1,2} Because the pulsed current in the sample is not localized to a specific area, the ultrasonic signal is generated in the entire volume, and spatial localization is only realized in the ultrasound signal reception. In other words, HEI is a tomographic technique. Whether using a 1D array transducer or a single transducer in a linear scan, the elevation profile is defined once by the elevation profile of the transducer. Therefore, the profile is significantly wider than in the case of echo ultrasound, where it is defined twice through both transmission and reception. In addition, HEI experiments are performed at low frequencies (1.0–2.0 MHz) to achieve sufficient signal-to-noise ratio (SNR)¹. The combination of the two means that the width of the elevation profile is 3 mm or more with the linear scans. As a result, it is difficult to depict irregular 3D structures with sufficient detail. To obtain higher lateral resolution without increasing the frequency, it is necessary to receive the ultrasound signal

over a significant portion of the sample surface area and use a full 2D tomographic reconstruction (see section 5 below).

Ideally, the shortest scan time is achieved with a 2D phased-array transducer that simultaneously receives and stores the signal from each element of the array. Practically, the technologies of large-scale 2D array transducers and receivers are still being developed.^{3–8} It is a formidable challenge to connect the thousands of elements in a full 2D array, so current 2D transducers have sparse elements.^{3–5} This is not suitable for HEI due to the SNR penalty. Full 2D hybrid array technology has been demonstrated,⁸ but is not yet available to the general users. Another difficulty with array transducers in HEI is that HEI signal reception is very susceptible to electromagnetic cross-talk noise, and the transducers have to be carefully shielded.² Current passive shielding techniques are effective for single element transducers,² but difficult to implement on array transducers.

A solution to these problems is to scan the sample surface with a well-shielded single transducer possessing a small contact point, thereby simulating a 2D array. In this approach, the single transducer serves as one element of a virtual 2D array, and it is scanned across a 2D rectilinear grid sequentially, collecting an ultrasound trace at each point of the grid. If the pulse excitation condition is identical for all the grid points, the resulting data set is the same as if it is collected simultaneously by an array of transducers that occupy all the grid points. While this approach lengthens the scan time many fold, it can be used on phantoms to study the resolution and contrast of 3D HEI images. The following sections describe the experimental details and present phantom images obtained in this fashion. These images clearly show the structures of the phantom, with superior resolution to the previous 2D images. The limitations of the current instrumentation and the inherent limitations of HEI are also revealed by these experiments and are discussed in the last section.

2. OVERALL EXPERIMENTAL SETUP

The general setup of the experiment is shown in figure 1. The base platform has an overhanging beam on which the point transducer-preamplifier assembly is attached. The axis of the transducer is always in the Z (vertical) direction. Under the transducer, a boat-shaped sample chamber rides on an X-Y stage assembly (Burleigh Instruments Inc., Fishers, New York), which moves the sample in the two orthogonal directions in the horizontal plane by up to 50.8 mm (2 inches). The piezoelectric drive motors of the X-Y stage have built-in position encoders and are capable of 1 μ m precision.

The sample chamber is 18.0 cm long, 7.5 cm deep and 11.5 cm wide at the midpoint. Two adhesive copper strips (3M, St. Paul, MN) are attached to the ends of the chamber as electrodes. The electrodes are connected via two coaxial cables to a single cycle rf pulse generator (Kentech Instruments Ltd., Oxon, UK). When triggered with a TTL signal, the pulse generator outputs a 2 MHz single cycle (500 ns long) current pulse of 40 ampere amplitude into the chamber. The current flow direction in the chamber is generally along the X axis.

The X-Y stage motors and encoders are connected to a 3-axis controller (Model 6000ULN, Burleigh Instruments Inc., Fishers, New York), which, in turn, is controlled by a PC via the Labview software (Version 5.1, National Instruments, Austin, TX). The ultrasound signal from the transducer is amplified by a 60 dB preamplifier (Model AU1291, Miteq, Hauppauge, NY) and a custom-made 40 dB second stage amplifier. It is then filtered with a 60 kHz high-pass filter (Model 3202, Krohn-Hite, Avon, MA) and a 5 MHz low-pass filter (Model BLP-5, Mini-Circuits, Brooklyn, NY). The signal is then digitized and stored with a PC digital oscilloscope (CompuScope 1012, Gage Applied Sciences Inc., Montreal, Canada) installed in the same PC that controls the X-Y stage.

The setup is placed inside a 2.5 tesla static magnetic field (Oxford Instruments, Oxon, UK) for the HEI scans. The field direction is along the Y axis. A Labview program written in-house controls all steps of the scan, including movement of the X-Y stage, pulse generator triggering and data acquisition.

In a typical scan sequence, the Labview program instructs the X-Y stage to start from the (0 mm, 0 mm) position and scan a 50 mm by 50 mm square grid at 1 mm increments in both directions. At each grid point, the Labview program sends a trigger pulse to the rf pulse generator and synchronously instructs the digital oscilloscope to collect the signal from the transducer at 20 M samples/second for 75 ms (1,500 sample points). This is repeated for 1,000 times and the average of the 1,000 traces is stored in the PC. The program then instructs the X-Y stage to move 1 mm along the X direction to the next grid point and repeat this process. When the X coordinate reaches 49 mm, the X-Y stage is incremented 1 mm in the Y direction and moved back to the X = 0 mm position. The above process is then repeated.

The entire system operates at 25 pulses/second, being limited by data storage speed. An entire scan lasts 28 hours. A complete data set contains 2,500 signal traces of 1,500 points each. In order to reconstruct a volumetric image of the sample from the raw data, one needs to know the angular response of the point transducer. This is detailed in the next section.

3. POINT TRANSDUCER DESIGN AND TESTING

The construction of the transducer-preamplifier assembly is based on a previous design for single element 1D scans.² As diagramed in figure 2, the core of the assembly is a 2.25 MHz/0.75 in broadband focused transducer (Krautkramer Branson, Lewistown, PA). The focal length of the transducer is specified as 50 mm. A conical hollow standoff is fitted to the transducer. The standoff is 40 mm long and made of delrin. It is filled with silicone oil as the acoustic medium. The tip of the cone has a small opening covered by a small cone of polyethylene sheet. The transducer is connected to the 60 dB preamplifier via an SMA connector, and a copper shield is fitted to the transducer-standoff assembly. The shield only comes into contact with the transducer at several small patches of velcro, thus preventing the vibration of the shield from entering the signal traces. This shield effectively suppresses the direct cross-talk noise as well as the Lorentz vibration noise in the transducer.² A cap consisting of a flat polyethylene sheet glued to a rigid foam collar is attached to the end of the conical portion of the shield, such that the flat sheet comes into contact with the tip of the transducer standoff via a small droplet of silicone oil. This scheme gives a constant acoustic contact area of about 0.5 mm diameter. The oil volume in the standoff also provides electrical insulation between the sample and the transducer. The length of the standoff is empirically determined for the best sensitivity.

Because the point transducer acts as individual elements of a 2D phased array, it is necessary to know its sensitivity profile with respect to incident waves at different angles from its axis, or the acceptance angle of the transducer. This is experimentally measured by scanning the transducer across a linear signal source. The linear source is a segment of 36 gauge (0.10 mm diameter) copper wire strung between the electrodes in the sample chamber along the X direction (Fig. 3). The chamber is filled with distilled water. When pulsed current flows through the wire in the presence of the static magnetic field in the Y direction, the Lorentz force on the current causes ultrasonic waves to emanate from the wire. The transducer is placed vertically above the chamber, with the tip cap touching the water surface. It is scanned at 1.0 mm intervals along the Y direction across the wire. At each position, an ultrasound signal trace is acquired. At the central position, where the tip of the transducer is directly above the wire, the distance between the two is 15 mm. Figure 4a shows the magnitude of the signal represented by a 2D gray scale image. From these data, it is determined that the angle span at which the signal

magnitude is halved is approximately 90° . Figure 4b is the real part of the complex signal. It shows that there is no significant phase shift over the 90° angle span. These data are used in the image reconstruction algorithm described below.

The standoff design of the point transducer allows effective shielding against cross-talk noises, but at the expense of sensitivity when compared to a small footprint transducer in direct contact with the sample. For this reason, it is necessary to have an estimate of the transducer sensitivity. For this measurement, a calibrated ultrasonic plane wave transmitter is used as the source and the transducer is pointed perpendicularly to the wavefront to maximize the received signal. The measurement gave a sensitivity of approximately 0.04 mV/Pascal at 2.0 MHz along the axis of the transducer. This is roughly one to two orders of magnitude lower than a conventional design and it becomes necessary to average over 1,000 acquisitions in the imaging experiments.

4. SAMPLE CHAMBER DESIGN

The sample chamber is made of 1.6 mm (1/16 in) thick fiber glass boards and 3.2 mm (1/8 in) thick plexiglass boards. The chamber is filled with agar gel containing 0.5% NaCl. The conductivity of the gel is approximately 0.4 S/m at 1MHz. This is comparable to those of muscle and blood at this frequency (0.71 S/m).^{9,10} Embedded in the agar are thin polyethylene tubes or rings of scotch tape to be detected by the 3D tomography. The axes of the tubes are along the Y direction parallel to the magnetic field, and perpendicular to the electric field (Fig. 1).

In another phantom construction, a rack is constructed in the form of two square end plates connected by four parallel thin fiberglass struts at the corners. The rack holds a block of agar gel with embedded objects and is immersed in 0.5% saline in the sample chamber (Fig. 3). The end plates are mounted with plastic bearings on slots in the sidewalls of the sample chamber. The axis of rotation is in the Y direction. The agar block has a square cross-section perpendicular to Y, and therefore four side surfaces that are parallel to Y. In 2D scans, each of the four sides can be rotated to face upward (perpendicular to the Z axis). The sides of the square end plates are 44.5 mm (1.75 in) long. The diameter of the supporting fiberglass struts is 3 mm.

In both phantom constructions, thin polyethylene tubes or scotch tape rings are embedded in the agar to act as electrical current barriers. They effectively render their internal volumes nonconductive but with much less effects on the acoustic wave propagation when compared with solid objects used in earlier HEI experiments^{1,2}.

5. 2D TOMOGRAPHIC IMAGE RECONSTRUCTION

The reconstruction algorithm to form the volumetric images is similar to full 2D phased-array image reconstruction in ultrasound imaging,^{11,12} but modified by the fact that in HEI, the signal is generated throughout the sample volume simultaneously. Referring to figure 5, the point transducer scans a grid pattern $\mathbf{R}(i, j)$ on the surface of the sample, with an even distribution of grid points. If the signal trace at point is (i, j) , is $S_{ij}(t)$ the rf pulse excitation occurs at $t = 0$, then the image amplitude of any point $\mathbf{r}$ in the sample volume is given by

$$A(\mathbf{r}) = \sum_{i,j} s_{ij}(t_d + |\mathbf{r} - \mathbf{R}(i, j)|/c) P(\theta) / |\mathbf{r} - \mathbf{R}(i, j)|. \quad (1)$$

Here t_d is the time delay introduced by the standoff of the point transducer (measured to be 29.5 μ s), c is the speed of sound in the medium (1,500 m/s), and $P(\theta)$ is the angular sensitivity

profile of the transducer (Fig. 4). The rationale behind this algorithm is as follows: if a point source at position $\mathbf{r}$ emits a pulse $A(\mathbf{r})$ that produces the signal S_{ij} at position (i, j) on the surface, with the delay and weight factor in Eq. (1), then the time reversal of this process, where a surface array emits S_{ij} with the same delay and weight factor, should produce a wave that converges to point $\mathbf{r}$, with intensity proportional to $A(\mathbf{r})$.

As shown in section 3 above, the half-height acceptance angle of the transducer is approximately 90° . The angular profile used in the image reconstruction is a Gaussian function $P(\theta) = \exp[-\theta^2/\theta_a^2]$, with $\theta_a = 54^\circ$.

The point response of this algorithm is tested with the data from the line source in figure 4. Figure 6 shows the point response function. In principle, the 1 mm grid density under samples the surface pressure field, which may give smearing artifacts at shallow depth. In practice, the point response function at 1 mm step size is satisfactory beyond 10 mm depth (Fig. 6). A grid of 0.5 mm pitch would take 110 hours to scan! For these reasons, the 1 mm step size was used for the phantom scans. As shown in the following section, the 3D images did not show significant artifacts from undersampling. This is partly due to the fact that the objects were embedded at depths below 10 mm and partly because the SNR of the data is not high enough to display low-level artifacts.

6. 2D TOMOGRAPHIC SCAN RESULTS

In the first scan, the sample chamber is filled with agar gel and two polyethylene tubes are placed end to end in the gel along the Y direction. The diameters of the tubes are 8.0 mm and 10.5 mm. The raw data set is a three-dimensional array of 50 by 50 traces of 1,500 points each. Figure 7 shows a cross-section of the set parallel to the X axis at $Y = 24$ mm. The initial high-intensity 'ping' at time zero comes from the residual cross-talk between the transducer and rf pulse generator. From its time of flight, the first horizontal band in the midrange is identified as the HEI signal from the contact interface between the transducer tip and the agar phantom. The time delay of this band is the delay time of the standoff ($29.5 \mu\text{s}$). The arc traces are from the embedded plastic tubes. The signals at the bottom edge corners come from the Lorentz force vibrations of the electrodes and are visible when the transducer is near the starting and end positions of the scan, where they are closest to the electrodes.

Figure 8 shows the tomographically-reconstructed image with surface rendering. Again, the magnetic field is in the Y direction, current flows in the X direction and the transducer axis is in the Z direction. The layer at the top of the image is the contact interface between the transducer and the phantom. The portions of the tube surface that are roughly parallel to the Z axis are not visible. Figure 9 shows two perpendicular cross-sections through the 3D image at $X = 24$ mm (slice perpendicular to the X axis) and $Y = 28$ mm (slice perpendicular to the Y axis), respectively. The signal dropout in areas parallel to the Z axis is attributed to two factors. The first is that the acoustic wavefront of the signal generally follows the contour of the tubes, radiating out in a roughly cylindrical fashion. The wavefront from the areas parallel to the YZ plane propagates laterally along the X direction and is thus not visible to the transducer above. The second factor is the characteristics of the HEI signal. The HEI signal originates from the Lorentz force on the pulsed current. Around the tubes, the current is diverted above and below them, thus creating high current densities at surface areas perpendicular to the Z axis and low current densities on the lateral sides. For these reasons, surfaces whose normal vectors are at greater than 45° angles from the Z axis cannot be seen in the scan.

The solution to this problem is to observe the phantom from all around. To realize this, the rotary rack is used as the phantom holder (Fig. 3). It holds a rectangular agar gel block and is fitted in the saline-filled sample chamber. The axis of rotation is in the Y direction. An

irregularly-shaped scotch tape ring is embedded in the center of the gel volume as the target object. The axis of the ring is oblique but is roughly along the Y direction. The gel block has a square cross-section perpendicular to the Y axis; therefore, four side surfaces are parallel to the Y axis. A 2D tomographic scan is carried out with each of the four sides rotated to face upward, perpendicular to the Z axis (the axis of the transducer). Figure 10 displays the individual surface rendered images from the four sides. The four posts visible in the images are the fiberglass struts of the rotating rack. One of the struts has a rectangular cross-section and produced secondary echoes visible in figure 9a. When these images are superimposed together by simple magnitude addition, the result is depicted with surface rendering in figure 11. Figure 12 shows a cross-section through the composite image. The scotch tape ring is mostly depicted in the composite image, except for a strip that remains roughly 45° from the transducer axis (Z axis) in all four scans.

7. DISCUSSION

The results above show that volumetric HEI tomography is capable of depicting electrical variations in a 3D structure. Currently, a single transducer is used to sequentially scan a 2D grid simulating a 2D array. This makes the total scan time several days for a complete four-sided scan. If a 2D array transducer coupled with fully parallel receivers are used, the scan speed is only limited by the repetition rate of the rf pulse generator, and is reduced to 20 seconds for a complete scan at 100 pulses/second. Therefore, with the development of array technologies it may be feasible to perform volumetric HEI scans *in vivo*, such as in biomedical applications.

In the above experiments, the scan surfaces are flat and form the four sides of a square. This is dictated by the movement of the X–Y stage and is less sensitive to signals from interfaces that are oblique to the transducer axis. This suggests that when designing a 2D array for volumetric HEI scans, the array elements should be arranged as part of a spherical surface to reduce angle dependent sensitivity variations. For human imaging, a large area spherical transducer limits the possible applications in that it can only be used in areas that allow large acoustic windows, such as the breast and the abdomen. It will be difficult if not impossible to image the heart noninvasively.

In Hall effect imaging, the ultrasound signal comes from the Lorentz force, which is always perpendicular to the magnetic field. For this reason, interfaces in the sample that are perpendicular to the magnetic field do not generate any signal and are therefore invisible. This is an inherent deficiency of HEI. Additionally, in biological samples, the conductivity or dielectric constant may vary several fold from one area to the next. This causes large variations in the electric field distribution, and ‘shadowing’ effects in regions surrounded by low conductivity tissue, such as fat. Even in visible areas, quantification is made difficult by the electric field distribution. Reliable quantitative results will have to come from correlation between HEI and other hybrid ultrasound-electromagnetic methods such as electro-acoustic imaging.¹³ This point is currently being investigated.

In the above experiments, the transducer is shielded with a waveguide standoff, and, as a result, much sensitivity is lost. This points to the need for better transducer designs without the large tradeoff of sensitivity. If such designs are developed, less signal averaging will be needed and the scan time can be shortened. Recent developments in fiber-optics based array transducers may provide a good solution.^{14,15} They are potentially immune to EM interference, and optical fiber connection to array elements avoids many of the problems of electrical connection.

The experiments described in this paper were done at 2.5 tesla. This is higher than most current high field clinical MRI scanners (1.5 T). Human imaging will need superconducting magnets

of similar field strengths. Fortunately, HEI magnets have much relaxed *uniformity* and stability requirements than MRI, which is often costly to achieve. Superconducting magnets often have the solenoid geometry to provide the highest possible field strength; therefore, a human HEI device may consist of a solenoid in which the subject is placed, similar to the common MRI scanners. However, without the stringent uniformity specifications, the length of the magnet can be as short as 50 cm for better access.

In conclusion, volumetric HEI tomography is demonstrated in static samples. Although the current single channel device takes hours to collect a 3D image, an imaging device incorporating a 2D array transducer with parallel receivers will be able to produce images in seconds. This technique may find biomedical applications such as mammography, where scan times on the order of seconds can be easily tolerated and the resulting information on the electrical constants improves the specificity of the exam.

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FIG. 1.

The setup of the volumetric scans. The magnetic field B_0 is in the Y direction. (1) The transducer-preamplifier assembly. The axis of the transducer is in the vertical Z direction. (2) The sample chamber. The current flow in the chamber is in the X direction. (3) The wires connecting to the electrodes in the chamber. (4) The X-Y stage and its drive motors. (5) The base platform.

FIG. 2.

A schematic diagram of the transducer-preamplifier assembly. The design of the tip is magnified in the inset. (1) The preamplifier. (2) The conical copper shield enclosing the transducer and the standoff. The shield is electrically connected to the outer casing of the preamplifier. (3) The 2.25 MHz/0.75 in broadband focused transducer. (4) The delrin hollow cone standoff casing. (5) The standoff is filled with silicone oil. (6) The circular foam collar of the tip cap. It is attached to the copper shield. (7) The polyethylene sheet of the tip cap. It is glued to the foam collar. (8) The small oil droplet maintains a 0.5 mm diameter contact area between the standoff tip and the polyethylene sheet of the tip cap.

FIG. 3.

The sample chamber with the rotary rack mounted in it. (1) The electrodes are two strips of adhesive copper tape. The chamber is positioned such that the current flow is in the X direction, perpendicular to the magnetic field. (2) The wires connecting to the electrodes. (3) The rotary rack that holds a block of agar gel, with target objects embedded inside. The axis of rotation is in the Y direction, parallel to the magnetic field.

FIG. 4.

The signal traces from the linear scan across a linear signal source. The horizontal dimension is the scan direction along Y axis. Each vertical line in the image represents the ultrasound time trace collected at that horizontal position. As the transducer scans across the source, the distance between the source and the transducer tip varies, leading to the varying delay times of the peak position, (a) The magnitude of the signal traces, showing the dependence of the signal intensity with respect to the angle of the line source relative to the axis of the transducer, (b) The real part of the signal traces. This shows that the phase of the signal is uniform over the measured range of angles.

FIG. 5.

A schematic representation of the 2D scan geometry, to illustrate the tomographic reconstruction algorithm. The transducer tip traverses the 2D grid in the XY plane and signal traces are collected at positions marked by the indices i and j . The electric field is generally in the X direction. The magnetic field is in the Y direction.

FIG. 6. Point response function reconstructed from the line source data shown in figure 4. The horizontal direction is the scan direction (Y axis) and the vertical dimension is the depth into the saline chamber.

FIG. 7.

Cross-section of the 3D data set collected with an agar gel phantom, at the position $Y = 24$ mm, parallel to the XZ plane. The current flow is in the X direction and the magnetic field is in the Y direction. Two plastic tubes are embedded in the gel, their axes along the Y direction. (1) The strong 'ping' from the cross-talk between the transducer and the pulse generator. (2) The HEI signal from the interface between the transducer tip and the phantom. (3) Signals from the Lorentz force vibration of the electrodes come into the depth of view when the transducer tip is close to them.

FIG. 8.

A 3D surface rendered version of the tomographically-reconstructed image of the agar gel phantom. The top layer is the surface of the phantom. The magnetic field is in the Y direction and the current flow is in the X direction. The axes of the plastic tubes are aligned in the Y direction. The axis of the transducer (not shown) is in the vertical (Z) direction. The visible portions of the tubes are generally perpendicular to the transducer axis (Z direction).

FIG. 9.

Cross-sections through the 3D image shown in figure 7. The current flow is in the X direction and the magnetic field is in the Y direction, (a) A cross-section perpendicular to the Y direction at Y = 24 mm, showing the larger tube, (b) A cross-section perpendicular to the X direction at X = 28 mm, showing both tubes.

FIG. 10.

Surface rendered 3D images reconstructed from four scans of the rotary phantom. The magnetic field is in the Y direction, the current flow is in the X direction and the axis of rotation is also in the Y direction. The transducer is placed vertically above the phantom along the Z axis. The rotary rack marked by the four struts holds a gel block with a square cross-section perpendicular to Y axis. A scotch tape ring is embedded in the gel block, its axis along the Y direction. Each of the four sides parallel to the Y axis is rotated to face up toward the transducer, i.e., perpendicular to the Z axis, and a scan is taken. The four scans show different facets of the tape ring. The four struts of the rack and the interface between the transducer tip and saline are visible in all four images.

FIG. 11.

Surface rendering of the composite image from the addition of the four images in figure 10. The magnetic field is in the Y direction, the electric field in the X direction. The upper right supporting strut has a rectangular cross-section and gave visible secondary echoes. The axis of the scotch tape ring is roughly in the Y direction. The portion of the ring at a 45° angle from the Z and X directions has low signal level, as it remains at 45° from the transducer axis (Z axis) in all four scans.

FIG. 12.
A cross-section through the composite image of figure 11 perpendicular to the Y axis. The axis of the tape ring is roughly in the Y direction, so the circumference of the ring is visible.