Estimation of wire phantom's position in ultrasound probe calibration based on polynomial equation

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ABSTRACT
The primary problem in the reconstruction of the 3D ultrasound image is the calibration of the probe. This paper presents a method of estimating wire phantom's position within the calibration system of the ultrasound's probe. Calibration process was done by probe scanning above the phantom, in this case using copper wires. Depends on the position and orientation probe to wires phantom, scanning process produced some types of images of phantom. The phantom will generate random distribution intensity in the B-mode ultrasound images. The intensity of the image often exceed a maximum value and produce image saturation. The saturated image, usually includes many saturated pixels. To find the accurate position of a phantom in the B-mode images, it requires modeling the distribution of saturated pixels. Data distribution in the ultrasound image was approximated using 5^{th} degree polynomial equation to compute the coordinate of image maximum value. This experiments show that one centimeter is equivalent to 122.72 pixels, and the average error is $(17x10^{-5})$ %. By calculating of the center of mass of wire image, obtained that one centimeter is 123.01 pixels, and an average error is $(20.9x10^{-5})$ %.

CCS Concepts
• **Computing Methodologies → Artificial Intelligence → Computer Vision Problem → Object detection.**

Keywords
Ultrasound image; image saturation; phantom position; polynomial equation.

1. INTRODUCTION
Ultrasound imaging is a non-invasive method in medicine for detecting and monitoring human's tissue properties. Some medical applications that use ultrasound imaging are monitoring of fetal development, monitoring bone density, breast imaging, etc.[1]. The quality and precision of ultrasound imaging are good, and the ultrasound imaging method has advantages in minimal risk of radiation in comparison with MRI method[2]. In addition, ultrasound imaging can be used for capturing the individual anatomy, visualizing the spread of local anesthetic, has minimal risk of complications, real time, and can be used to provide direct visualization of three-dimensional (3D) anatomy[3].

Ultrasound imaging is a method based on the use of acoustic waves. Echoes reflected by the tissues make possible to see what is invisible to the human eye. Reflecting echoes are received by ultrasonic sensors. Ultrasonic sensors are placed in a probe, as shown in Figure (1), and the probe is moved over the phantom in the scanning process. Two-dimensional (2D) B-mode images will be generated by the ultrasound probe scanning. To reconstruct the three-dimensional (3D) images, the 2D ultrasound images must be assembled and segmented.

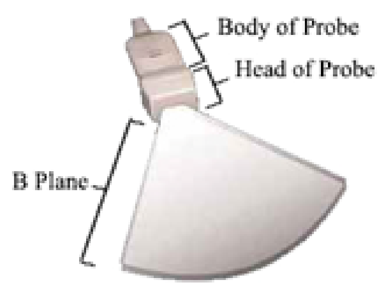

Figure 1. Ultrasound probe

The primary problem in the reconstruction of the 3D ultrasound image is the calibration of the probe. Significant variations in speed and position of the probe during the scanning process can make the produced 2D images have wide variations and will cause an error in its 3D image reconstruction. The easiest way to calibrate the probe is by scanning the object or phantom of known geometric form and then calculating the offset between the real object and the object produced in 2D images[4].

In this paper, we proposed a method for positioning the calibration phantom used in ultrasound's probe and estimating the position of objects in the produced 2D B-mode image. The proposed method used a polynomial equation to find the distribution of the generated ultrasound data. The maximum value of the polynomial function is used to determine the coordinate position of the phantom. Finally, the offset between the real phantom and the produced data can be measured. Hence, the

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variations in the 2D images that were produced using ultrasound imaging can be measured, and the error in its 3D reconstruction can be minimized.

2. RELATED WORKS

Awais, et al.(2015) used ultrasound probe with 6 degrees of freedom for calibration, tracked by optical marker and integrating this system with robotics biopsy[5]. The tracking result, then processed by mathematical model to calibrate the ultrasound probe. And this experiment result in a 1.67mm root mean square error.

Another research used the n- wire phantom in freehand ultrasound for probe calibration has been conducted by Bielmann, et al. (2015). This method makes a temporal and a spatial calibration to obtain different matrixes, which enable to know the setting, offset and transformation for getting a good 3D reconstruction[6].

J. Chaoui, et al. (2009) has proposed probe calibration using single point phantom[7]. This research used mathematical simulation to reduce calibration time and to eliminate user intervention. The simulation provides a wide accurate dataset using only one US image.

Other research has been conducted by Lee, et al. that proposed an efficient estimator for approximation of x-ray transmittance using low order polynomial [8]. The experimental result shows that the calibration process in photon counting detector (PCD) is only needed for bias correction, and the bias will converge into unbiased solution. The proposed low-order polynomial method gives comparable results with maximum likelihood (ML) method while the computational cost is reduced.

Pujiono, et al.(2016) also used a polynomial equation to restore the constancy of the basic colors of red, green and blue (RGB)[9]. This method was used to enhance underwater colorful coral reef in 5 m depth below the water surface. The proposed method has been evaluated using Peak Signal to Noise Ratio (PSNR) with the average value is 33.94.

Those researches show that polynomial equation can be used for calculating the offset or bias in numerical data. Since the 2D image is actually a sequence of numerical data in two-dimensional array, in case of a grayscale image, or three-dimensional array in case of the color image, then polynomial equation can be used for measuring the offset in the 2D grayscale images that was produced using ultrasonic imaging.

The 2D images produced using ultrasound imaging have high intensity that caused image saturation, hence image enhancement should be performed. Su-Ling, Lee (2015) on his research has presented about backlighting color image enhancement method using color saturation and image fusion [10]. However, only color saturation is used in the proposed enhancement method. Thus, it's useful to combine the information of intensity, color and depth to solve image enhancement problem include in the ultrasound image.

3. METHODOLOGY

In the scanning process, ultrasound probe driven manually above phantom, and produce several possibilities of positions and orientations. The probable position and orientation of ultrasound probe are perpendicular to the wire phantom or the probe position leaning to the right/left, that make an angle with the wire phantom, as shown in Figure (2). Significant variations of speed and probe position during the scanning process will make the produced 2D B-mode ultrasound images have wide variations and will cause an error in its 3D image reconstruction. It requires a calibration method to minimize errors. The easiest way to calibrate the probe is by scanning the object or phantom of known geometric form.

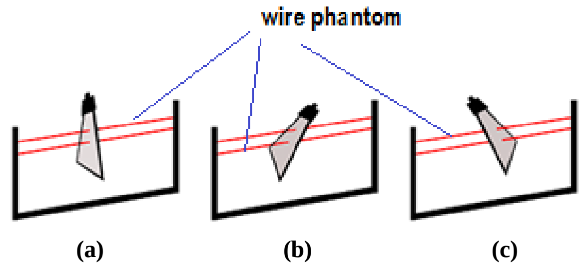


Figure 2. Probe's position in the scanning process

(a) Ultrasound probe is perpendicular to wire phantom,

(b)-(c) Ultrasound probe makes an angle with wire phantom

Depend on the position of the ultrasound probe, the phantom images will vary, although the ultrasound probe scanned the same of phantom.

The calibration phantom reflects echoes from ultrasound wave stronger than other media in these areas. Thus, the pixels in B-mode ultrasound image which has strongest intensity can be defined as representation of phantom's pixels. Figure (3a) shows a result of ultrasound scanning above two wires. These wires have the same size, but the scanning images have differences.

A 2D ultrasound image $I(i, j)$, is described as two dimensions function where I is the intensity of the image at the coordinates (i, j). A high intensity will cause image saturation. Saturation occurs when the measured intensity exceeds the maximum range capability of the ultrasound probe. Hence the saturated region of the image has the possibility as the phantom actual position. Based on the Figure (3b) the saturated area includes several pixels.

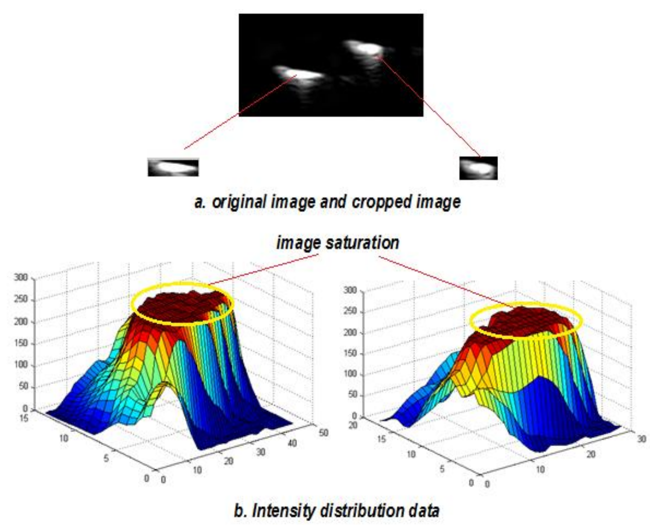


Figure 3. Image of wire phantom and saturated image

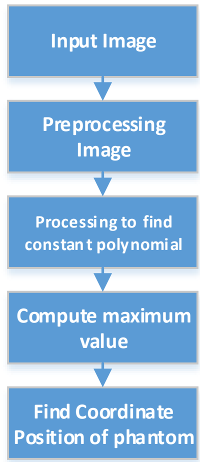


Figure 4. Processing ultrasound image

The steps for calculating phantom position follow the chart on Figure (4). To simplify the ultrasound image processing, the first step is to change the intensity in the grayscale mode. Image phantom has three RGB color components, and for the preprocessing step, these images are converted to grayscale follows the Equation (1). The images then normalized by dividing each pixel intensity by a maximum intensity of the dataset in the phantom images, according to the Equation (2).

$$I(i,j) = 0.333R(i,j) + 0.333G(i,j) + 0.333B(i,j). \quad (1)$$

$$Z(i,j) = \frac{I(i,j)}{\max(I(i,j))} \quad (2)$$

Where $Z(i,j)$ is the normalized value of a pixel in the phantom images, in the range of 0 – 1 (the maximum value for the saturated pixel is 1). Then, normalized data $Z(i,j)$ collected and plotted into the polynomial's fitting calculation to acquire the data distributions.

The intensity data in the ultrasound images have a random distribution and required a mathematical manipulation to put the random intensity data in a smooth curve. The classical two variables polynomial equation are formed based on this distribution. Due to the high difference in the intensity distribution data in each image, the low degree polynomial model can not be used in this case.

Ultrasound image is in two dimensions, hence the intensity of the image depends on i and j position, and the polynomial equation arranged in two variables. As shown in Figure (3b), the intensity distribution data have many local maxima or local minima, so the polynomial model must be in the high order.

The general form of n^{th} order polynomial equation in two variables is shown in Equation (3).

$$f(i,j) = \sum_{\substack{(k+m)\leq n \\ (k,m)\leq 0}} a_{km}\, i^k\, j^m$$

$$f(i,j) = a_{00} + a_{10}i + a_{01}j + \dots\dots + a_{n0}i^n + a_{(n-1)1}\, i^{n-1}j + \dots + a_{0n}j^n, \quad (3)$$

Where $f(i,j)$is polynomial function in (i,j), and a is polynomial constant. This polynomial fitting used to form the surface curves and fit the intensity data of ultrasound image to these surface curves. The distribution data and plot them in the polynomial surface fitting is shown on Figure (5) for image in Figure (3b). Figure (5) shows the intensity distribution data and surface fitting using the polynomial model. Then, the saturated pixels can be manipulated.

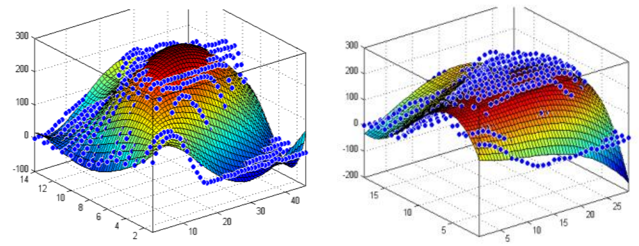


Figure 5. Polynomial Surface fitting

After the polynomial model found, the phantom position represented by the peak point of the polynomial curve fitting of each image. The maximum value of $f(i,j)$ calculated to acquire the position of the wire phantom in the ultrasound image. The peak point curve in Figure (5) is the maximum value that owned by the saturated pixel. And the coordinate of peak point determined from the polynomial equation.

4. EXPERIMENTAL RESULTS

In this research, the phantom was 1 mm copper wire mounted horizontally and connects the two sides of acrylic's water tanks. We used two parallel wires with a distance in 1 cm and 0.5 cm. Wire phantom was mounted 5 cm below the water surface. Sensor probe in the scanning process is an ultrasound type L15-7L40H US-5 with power gain 100%, -7 dB and frequency of 7.5 MHz. During the scanning process, the phantom placed in the water, and an ultrasound probe is moved above these wires, as shown in Figure (6).

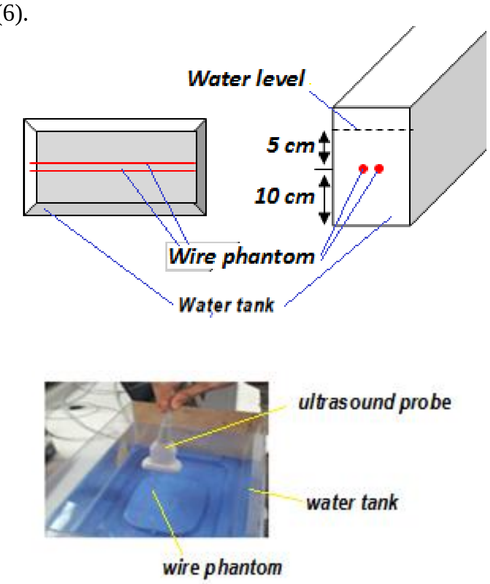
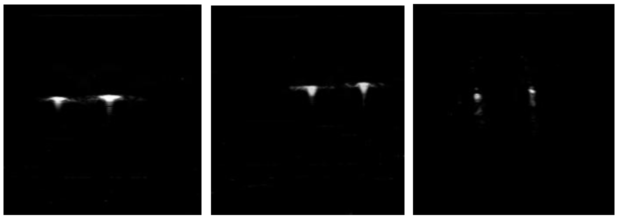


Figure 6. Wire phantom in the water tank (top view, side view and water tank photograph)

Figure (7) shows some B-mode ultrasound images as the result of ultrasound probe scanning on the wire phantoms. The size of each image is 450 x 500 pixels. Different images occur as the result of the differences in probe position and orientation in the scanning process. White node in this image assumed as representation of a wire phantom. The white nodes were processed to estimate the wire phantom accurate position.



a. B-mode image using 1 cm wire distance

b. B-mode image using 0.5 cm wire distance

Figure 7. Wire phantom scanning result in the water tank (B – mode ultrasound image)

First, the white node was cropped from B-mode ultrasound image for faster processing and to eliminate full-black images. The wire image coordinates in the B-mode image axis, (x, y) noted, as shown in Figure (8).

Figure (8) shows that upper left corner of wire image has coordinate (x, y) in the B-mode image axis. When weAs result of the process, it creates a new axis with the upper left corner of wire image in (0,0).

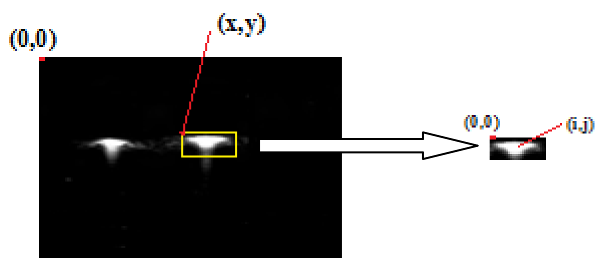
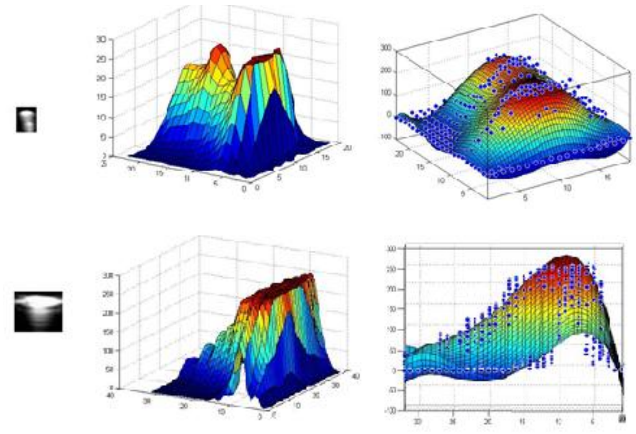
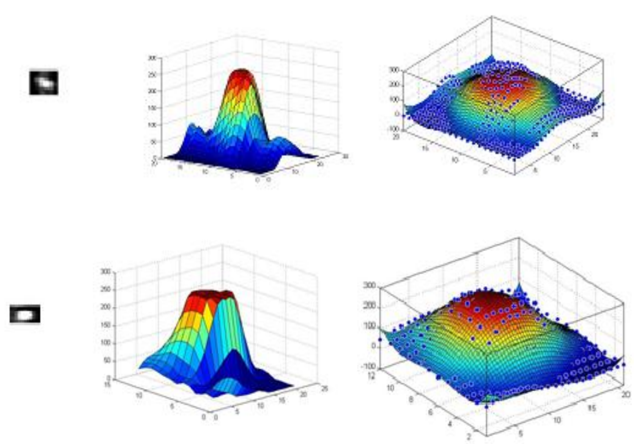


Figure 8. Coordinate of wire phantom in the two axis

Figure (9a) shows the wire images from various B-mode images. Figure (9b) visualized the saturated pixels in 3D visualization.





(a) (b) (c)

Figure 9. Image processing using fitting polynomial (wire image, 3D visualization and polynomial surface fitting)

Figure (9b) shows that most of the images have local maxima over 4 positions. So the next step is finding the polynomial equation that fit the data distribution of the wire image. This research used a 5th order polynomial as a model to get the peak point of the saturated image. The curve of polynomial fitting is shown in Figure (9c).

For example, 5th order polynomial equations for the first image in Figure (9a) are,

$$f(x,y) = 294.6 - 15.55x - 158.1y + 0.1161x^2 + 11.33xy + 43.61y^2 - 0.004628x^3 - 0.1944x^2y - 0.8975xy^2 - 6.093y^3 + 0.0002748x^4 - 0.002139x^3y + 0.0273x^2y^2 + 0.3833y^4 - 0.0001037x^3y^2 - 0.0004776x^2y^3 + 0.0006868xy^4 - 0.008607y^5 \quad (4)$$

The peak point calculated using polynomial equation of each images. The obtained peak point is the peak position coordinate of each wire images. Table (1) shows the estimated peak position wire phantom in the B-mode image axis. Based on Figure (8), the determined position of wire phantom in the B-mode ultrasound image (m, n), is calculated using equation (5).

$$(m, n) = (x + i, y + j) \quad (5)$$

Table 1. Estimation of position and distance wires in the B-mode image axis (1 cm distance)

No	Position (pixels)		Distance (pixels)
	Wire - 1	Wire - 2	
1	(236.2;193.9)	(358.6;187.4)	122.57
2	(64;132.9)	(185.8;132.7)	121.8
3	(128.1;220)	(246.9;216)	118.87
4	(135.1;244)	(252.7;235.7)	117.89
5	(148.1;213)	(273.9;202.3)	126.25
6	(87;204)	(212;212)	125.26
7	(135;150.6)	(253.4;115.2)	123.58
8	(110.2;118.8)	(230.8;128.4)	120.98
9	(136.5;136.7)	(260.3;144.8)	124.06
10	(145.9;120)	(269.1;132.5)	123.83

Table 2. Estimation of position and distance wires in the B-mode image axis (0.5 cm distance)

No	Position (pixels)	Distance

	Wire - 1	Wire - 2	(pixels)
1	(246;124.1)	(302;89.2)	65.98
2	147.7;111.4)	(209.8;104.7)	62.46
3	(249.6;106.5)	(310.3;104.1)	60.75
4	(125.7;108.2)	(180.9;135)	61.36
5	(168.2;102.5)	(226.2;110)	58.48
6	(301.2;136)	(359.8;154.4)	61.42
7	(189.2;136.7)	(248.1;144.5)	59.41
8	(209.5;155.1)	(269.9;164.4)	61.11
9	(326.7;98.9)	(387.6;109.7)	61.85
10	(267.5;135.6)	(325.1;158.1)	61.84

Table 3. Measurement Results

No.	Measurement (cm)		Error(%)	
	Polynom	CoM	Polynom	CoM
1	0.999	0.985	1×10^{-5}	15×10^{-5}
2	0.992	0.989	8×10^{-5}	11×10^{-5}
3	0.969	0.966	31×10^{-5}	34×10^{-5}
4	0.961	0.959	39×10^{-5}	39×10^{-5}
5	1.028	1.030	29×10^{-5}	30×10^{-5}
6	1.020	1.021	21×10^{-5}	21×10^{-5}
7	1.007	1.026	7×10^{-5}	26×10^{-5}
8	0.985	1.029	14×10^{-5}	29×10^{-5}
9	1.011	0.998	11×10^{-5}	2×10^{-5}
10	1.009	0.998	9×10^{-5}	2×10^{-5}

Table (1) and Table (2) shows the estimation of position and the measurement of distance between wires in B-mode image. The polynomial methods resulted that one centimeter is equivalent to 122.72 pixels in B-mode ultrasound image. As a benchmarks, the center of mass (CoM) methods applied on each wire images, and resulted that one centimeter is equivalent to 123.01 pixels.

5. CONCLUSION

This result proves that the estimated position of wire phantom in B-mode ultrasound image can be determined using polynomial fitting approximations, and that 1 centimeter is equivalent to 122.72 pixels. According to the measurement results and calculations in Table 3, the average error of probe calibration using polynomial equation is (17×10^{-5}) %. And as comparison, using the center of mass methods on the wire image, obtained 1 centimeter is equivalent to 123.01 pixels and has average error of (20.9×10^{-5}) %.

For future work, researches about the determination of the exact position of the B-mode image in the real world coordinates, and its three-dimensional image reconstruction can be conducted.

6. ACKNOWLEDGMENT

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