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● *Original Contribution*

A CLOSED-FORM DIFFERENTIAL FORMULATION FOR ULTRASOUND SPATIAL CALIBRATION: MULTI-WEDGE PHANTOM

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Abstract—Calibration is essential in freehand 3-D ultrasound to find the spatial transformation from the image coordinates to the sensor coordinate system. Calibration accuracy has significant impact on image-guided interventions. We introduce a new mathematical framework that uses differential measurements to achieve high calibration accuracy. Accurate measurements of axial differences in ultrasound images of a multi-wedge phantom are used to calculate the calibration matrix with a closed-form solution. The multi-wedge phantom has been designed based on the proposed differential framework and can be mass-produced inexpensively using a 3-D printer. The proposed method enables easy, fast and highly accurate ultrasound calibration, which is essential for most current ultrasound-guided applications and also widens the range of new applications. The precision of the method using only a single image of the phantom is comparable to that of the standard N-wire method using 50 images. The method can also directly take advantage of the fine sampling rate of radiofrequency ultrasound data to achieve very high calibration accuracy. With 100 radiofrequency ultrasound images, the method achieves a point reconstruction error of 0.09 ± 0.39 mm. (E-mail: rohling@ece.ubc.ca) © 2014 World Federation for Ultrasound in Medicine & Biology.

Key Words: **Ultrasound, Calibration, Closed-form, Multi-wedge, Differential.**

INTRODUCTION

Ultrasound is a tolerable, portable, inexpensive and real-time modality that can produce 2-D and 3-D images. Therefore it is a valuable intra-operative imaging modality to guide surgeons aiming to achieve higher accuracy of the intervention and improve patient outcomes.

In all the clinical applications that use freehand tracked ultrasound, the challenge is to precisely locate the ultrasound image pixels with respect to a tracking sensor on the transducer. This process is called *spatial calibration*, and the objective is to determine the spatial transformation between the ultrasound image coordinates and a coordinate system defined by the tracking sensor on the transducer housing. Occasionally, it is also required to perform the calibration intra-operatively, which poses other important challenges. Intra-operative calibration requires a fast, robust and easy procedure using a calibration device that is portable, small and compatible with the sterile environment (Peterhans et al. 2010). Even an error of about a degree can cause points to be located several millimeters from their true positions (Gee et al. 2005). One example is ultrasound guided out-of-plane needle insertion (Najafi and Rohling 2011), where 0.5° error in the needle trajectory's angle could result in more than 5 mm error in the intersection point. Even higher accuracy is required in an in-plane needle insertion procedure.

The ultrasound calibration problem has been investigated over the last two decades. Most of the calibration techniques are based on imaging an artificial object with known geometric parameters. Prior knowledge of the phantom, along with ultrasound images of the phantom, is required to calculate the calibration parameters. The phantom can be as simple as a fixed point target which can be constructed as a bead (Amin et al. 2001; Detmer et al. 1994), crossed-wires (Melvær et al. 2011; Trobaugh et al. 1994; Yaniv et al. 2011) or the center of a sphere (Brendel et al. 2004). Alternatively, the phantom can be a fixed plane as in the single-wall method (Najafi et al. 2012b; Prager et al. 1998; Yaniv et al. 2011) or in its variant called the Cambridge phantom (Prager et al.

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1998). Another class of calibration phantoms consists of thin wires, usually with N- or Z-shaped patterns (Boctor et al. 2004; Chen et al. 2009; Pagoulatos et al. 2001; Peterhans et al. 2010). Another approach is based on registration of 2-D ultrasound images with the 3-D model of the phantom (Bergmeir et al. 2009; Blackall et al. 2000; Lange et al. 2011). Some calibration methods are based on precise manual alignment of the ultrasound image with a thin planar phantom (Gee et al. 2005; Lindseth et al. 2003). There are also some methods that do not require a phantom and use a calibrated stylus (Hsu et al. 2008; Khamene and Sauer 2005; Muratore and Galloway 2001) or use changes in speckle from transducer movements (Boctor et al. 2006b). Detailed reviews, comparison of different calibration methods and a summary of various validation techniques can be found in survey papers by Mercier et al. (2005) and Hsu et al. (2009).

Calibration methods can also be categorized as either iterative optimization techniques (*e.g.*, Detmer et al. 1994; Melvær et al. 2011; Prager et al. 1998) or closed-form solutions (*e.g.*, (Boctor et al. 2004; Chen et al. 2009; Najafi et al. 2012b). Iterative methods are subject to sub-optimal local minima, are sensitive to initial estimates and, therefore, are less robust in general than closed-form solutions (Eggert et al. 1997). They also typically require more input data to achieve the same level of accuracy.

Despite all these efforts, it is not easy to find a method for ultrasound calibration that is fast and easy to perform while at the same time being highly accurate. This may adversely affect the result of ultrasound-guided interventions and possibly limit the scope of new applications. Accurate, absolute localization of phantom features in ultrasound images is the biggest challenge in the calibration process (Lange et al. 2011) and is the limiting factor in increasing the accuracy of calibration.

One reason is the blurry appearance of features caused by the finite resolution of the ultrasound images and the presence of noise. There are also image formation errors resulting from speed of sound variations, refraction and a finite beam width, all of which contribute to distortions in the shape of the depicted features.

Finite resolution of the ultrasound and imaging artifacts, for example, causes small point-shaped objects to appear as short lines in the ultrasound images (Chen et al. 2009); therefore, a cross section of a wire in an ultrasound image does not appear in the shape of a circle, but rather as an asymmetric cloud with a width of several pixels. This makes it very challenging to accurately localize the actual intersection point. The finite beam width also affects the accuracy of feature localization of planar objects (Prager et al. 1998). A plane appears in the ultrasound image as a line with a thickness related to the beam width and plane orientation. Unlike point features, the shape of the line remains unchanged along the line as long as the plane's inclination relative to the beam direction is not high.

To improve accuracy, the need for absolute localization of the calibration phantom features should be eliminated. We propose to achieve this by using a differential measurement of the relative distance between two image features. Advancements in differential measurements for ultrasound motion tracking in recent years have enabled accurate measurements of the relative locations of phantom features. This accuracy can be as high as a few microns when radiofrequency (RF) ultrasound data are used (Walker and Trahey 1995).

The differential measurement can be especially accurate when the shapes of the features are very similar. Because the appearance of a wire varies with depth in the image (Chen et al. 2009), it is more difficult to perform accurate measurements of the distance difference of point features unless only pairs of features with similar shapes are selected. Planar phantoms, on the other hand, are ideal because the plane appears as a line with uniform thickness. In fact, echo RF pulses in all RF scan lines exhibit a similar pattern because they all experience the same physical situation except for a variation in the ultrasound focus (and, therefore, beam width), which slightly varies according to the depth. The relative axial distance of line features can be measured as the relative shift between echo RF pulses and is, therefore, a differential measurement. This measurement can be performed very accurately, especially with RF data. Redundancy of measurements, because of the presence of many scan lines, is another advantage of planar objects over wires. Redundancy allows averaging of measurements, which can also reduce error from measurement noise.

Closed-form calibration based on measurements of the relative pose of two different images using hand-eye coordination technique has been proposed before (Bergmeir et al. 2009; Boctor et al. 2004). Although relative measurements have an inherent advantage, accurate estimation of the relative 3-D poses of 2-D ultrasound images remains a challenge. This work is based on in-plane differential measurements which can also directly take advantage of the fine sampling rate of RF ultrasound data to achieve high measurement accuracy. Moreover, wedges have been used for alignment and calibration before (Afsham et al. 2011; Boctor et al. 2006a). The proposed multi-wedge phantom has been specifically designed and optimized for closed-form differential calibration.

Our aim was to make calibration more reliable for day-to-day clinical use by reducing the calibration error while keeping the calibration procedure as fast and simple as possible and suitable for intra-operative use.