AN OPTICAL REGISTRATION METHOD FOR 3D ULTRASOUND FREEHAND SCANNING

Carsten Poulsen, Dr. Peder C. Pedersen
Worcester Polytechnic Institute
Electrical and Computer Engineering
Worcester, MA, USA
pedersen@wpi.edu

Dr. Thomas L. Szabo
Boston University
Biomedical Engineering
Boston, MA, USA
tlszabo@bu.edu

***Abstract*—Three-dimensional ultrasound is emerging as an important adjunct to conventional 2D scanning, in particular in obstetrics and cardiology. Three-dimensional ultrasound makes it easier to visualize objects, pathologies and injuries. Three-dimensional ultrasound may be implemented in form of free hand scanning, which can be augmented with a registration system that records the position and orientation of each scan plane. The registration system may be in the form of an optical or EM source, mounted on the transducer, and with corresponding sensors placed in the room. In terms of portable ultrasound, free hand scanning is the most practical way to acquire 3D images.**

In this paper we present a new and very compact type of optical registration system, which tracks the position of the transducer on the skin surface. This is done continuously by acquiring images of the skin at the transducer location with a CCD-array, attached on the side of the transducer. With this position information, a sequence of 2D ultrasound frames, obtained with freehand scanning and thus likely to have unequal spacing and varying lateral position, is interpolated into a sequence of ultrasound frames with equal spacing and fixed lateral position. The implementation uses the Terason ultrasound scanner [1] and Sonocubic 3D software, operating in Windows XP.

To validate the registration method, a phantom with a rod-shaped inclusion was scanned in an uneven scan path. Five scans were made of a known volume. The volume estimations, based on the interpolated scan planes, fell within 96-106% of the actual volume, with a mean around 101%. Thus, we have demonstrated that the compensation algorithm is working as expected.

Keywords : Ultrasound, 3D freehand scanning, optical positioning, surface tracking.

I. INTRODUCTION

In classic freehand scanning for 3D imaging, the ideal scanning goal is to move the transducer in a straight line perpendicular to the scan planes at a constant rate of speed. In other words, the idea is to move the transducer in a way close to what a mechanical arm with constant speed would do. Naturally this is very difficult to do in reality.

Two major sources of distortion occur when the transducer is moved across the skin. For small distances often used for 3D imaging, the movement can be considered to be approximately planar. The intended movement of the transducer is along the x axis if the lateral axis is y and depth axis is z. The first error is the nonuniform rate of speed with which the transducer is moved. The second error is that the transducer deviated from the y axis by an offset at each scanplane acquisition.

The usual assumption for 3D interpolation made in this form of 3D scanning is that the scan planes are parallel, equispaced, and co-axial (aligned along the center of the x axis). In freehand imaging, both nonuniform movement and offsets occur (as shown in Fig. 7) that are not accounted for in conventional cartesian interpolation algorithms, and consequently, distortion occurs.

In order to produce undistorted surface renderings and volume measurements with freehand 3D ultrasound scanning, we have implemented and tested a new method for acquiring the transducer position. The motivation for the system is to augment the existing free hand scanning techniques, to make it a quantitative 3D imaging tool. In this way the size of volumes, for example, that of internal bleedings, can be estimated.

The 3D ultrasound imaging system makes use of two Windows applications, Terason and Sonocubic, which have been developed for freehand scanning without a positioning system. The added positioning system is in the form of an optical sensor with a DSP-processor (Digital Signal Processing) that is interfaced to a computer via a USB-interface. A DLL (Dynamic Link Library) makes it possible to interface Sonocubic to the driver for the optical sensor and to provide Sonocubic with the information necessary to position the scan planes correctly.

II. SYSTEM OVERVIEW

An overview of the system is shown in Fig. 1. Terason is a PC-based ultrasound scanner [1] developed by Terason Corporation, a division of Teratech. The Terason 2000 imaging system functions as a regular 2D ultrasound scanner, but it uses a PC as the platform instead of custom-built hardware. In the optional 3D version, 2D images can be transferred from Terason to Sonocubic continuously, with help from a shared memory interface. Along with each image is resolution information that describes how large a pixel is in a "real world" measure (pixels/meter).

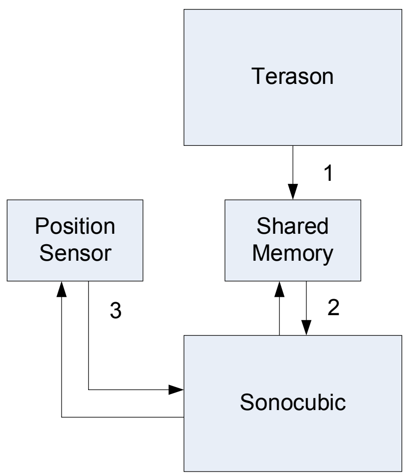


Figure 1. Block diagram of the system

In Sonocubic, a 3D ultrasound rendering software, the scan planes are collected and stored for 3D visualization. Using the position system, each scan plane is associated with a position.

This makes it possible for Sonocubic to first position the scan planes at their true physical location (laterally and axially) in a fine sub grid. An interpolation algorithm is then used for projecting the collected scan planes onto a uniformly spaced and laterally aligned, coarser main grid for visualization and distance measurements. The interpolated data contained in the main grid can be written to a file using the standard Windows AVI-file format. In this way data can be extracted from Sonocubic and used for further processing. A series of low-pass filters in Matlab is used to clean and enhance the collected data. By counting pixels in each scan plane a simple volume estimation can be done.

III. Optical Sensor

To get the positioning information of each acquired scan plane an optical sensor with a DSP-processor is being used, in the form of the Agilent ADNS-2610. This sensor is found in many optical computer mice. It basically consists of a CCD (Charge Coupled Device) camera that acquires images of the surface at a very high rate (1500 fps) and a DSP algorithm that makes a cross-correlation between consecutive images. By using the cross-correlation algorithm the distance the optical sensor has moved can be determined.

The camera consists of 18x18 pixels and has a resolution of 400 dpi in each dimension. It is thus covering an area of approx. 1.31 mm^2 when a lens with a 1:1 magnification is used, as is the case for this work. The cross-correlation concept is shown graphically in Fig. 2, where the optical sensor has moved -2 pixels in the x-direction and -3 in the y-direction. Note that it is the optical sensor, not the surface, that is moving, hence the negative movement.

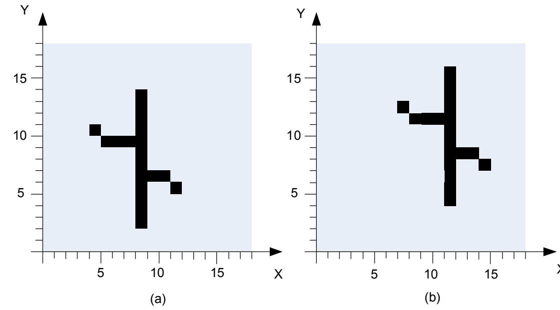


Figure 2. Concept of image captured before and after a movement

To be able to track the movement of the optical sensor requires surfaces to have an optical texture and/or to be rough. If the surface is optically rough, light projected at an angle can create shadows that makes it convenient to track on it.

For ergonomical purposes a design where an optical image conduit (an optical fiber bundle) is placed between the skin and the optical sensor was evaluated. Two different approaches were tried; in one approach the LED illuminated the surface through the optical fiber bundle (see Fig. 3(a)), and in the other approach the surface was illuminated by an LED mounted near the surface and with a lens in front of the optical fiber (see Fig. 3(b)).

For both approaches, we found that it was possible to track on surfaces with an optically highly irregular pattern, like a USAF 1951 test pattern [3] or a wooden table. For the first case Fig. 3(a) we found two problems. One was that the focal depth of the optical fiber bundle is very short, making it necessary for the fiber to be in complete contact with the skin at all time. Quantitatively this is within 0.5 to 1 mm. Another and bigger problem is that there is no shadowing to make surface discontinuities stand out and make tracking possible. This is very important for tracking on the skin.

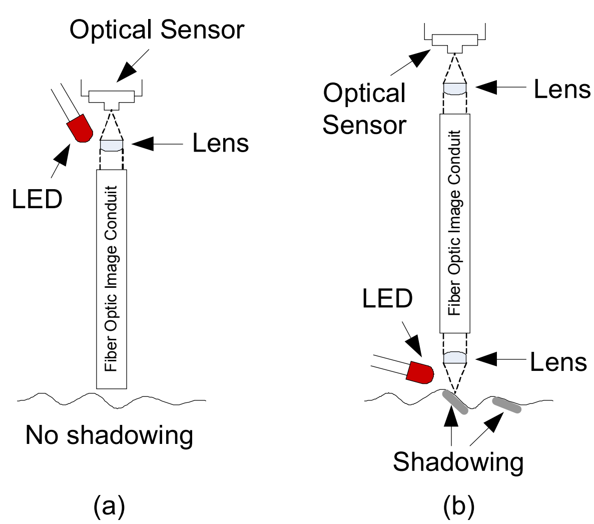


Figure 3. Two methods of mounting the optical image conduit and illuminating the surface

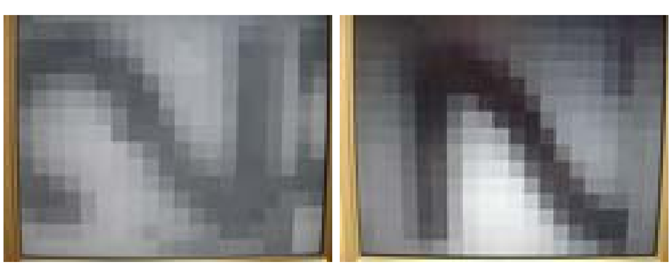
In the second case Fig. 3(b) we tried to overcome the problems with a design that provided greater focal depth and greater shadowing. Adding an extra lens in front of the fiber bundle helped with the focal depth, but also blurred the image more. We think that this blurring is produced by interference patterns since each optical fiber in the fiber bundle transmits light entering from an angle as well as the light entering straight in. There may also be a small coupling between each fiber element, which could contribute to the blurring.

Based on these results it was therefore decided not to use the optical fiber bundle. Instead a small custom housing was made for mounting a single lens in front of the optical sensor. This is shown in Fig. 4.

Figure 4. Lens holder with sensor mounted on transducer

The optical sensor has a serial interface, through which the position information is sent to the computer. Other types of data can be extracted, such as the image which the sensor has captured, or a surface quality (SQUAL) value. The SQUAL-value is a representation of how many features that are available for the optical sensor to see, and it can be used to determine whether it is possible to track on a surface or not.

In order to extract the image and SQUAL values from the optical sensor, an FPGA was programmed and interfaced to provide a serial interface on one side and a VGA (Video Graphics Array) output on the other. This tool proved to be essential for evaluating whether the optical sensor was in focus or not and to compare the different setup of fiber and lenses that we had. In Fig. 5 two images (described in caption) are shown of the number 2 in group 0 in a USAF 1951 chart.



(a) (b)

Figure 5. Image from USAF 1951 chart taken through optical fiber with two lenses (a) and through a single lens in the customized housing (b)

As can be seen the image is much clearer when a single lens is used in the customized housing Fig. 5(b), as opposed to the optical fiber bundle with two lenses Fig. 5(a). It is expected that the image is mirrored in Fig. 5(b) since a single lens inverts the image. Two lenses make it non-inverted.

IV. Positioning Software

In order to correct for the distortions, the position information for each scan plane must be acquired and transferred to the modified Sonocubic software. The Sonocubic software was modified to utilize this information, and to alter its interpolation algorithm we used the position information from an Agilent ADNS-2610 optical sensor. The only information we are interested in is the position, and that makes it possible to use the hardware found in a commercial computer mouse. The mouse hardware interfaces to the USB interface on a PC and by using a mouse filter driver we can extract the position data from a specific optical sensor. The mouse filter driver we use is CPN mouse [2].

A DLL (Dynamic Link Library) called 'Pos3D.dll' was developed by us for interfacing Sonocubic to the mouse filter driver. The change in position is continuously updated inside the optical sensor and the driver stack. Using the driver stack in a polled mode therefore made it possible to access the mouse filter driver and acquire the change in position since the previous time Sonocubic asked for it. Finally, the modified 3D software corrects scan plane positions for scan speed and offsets.

V. Computer Phantom

A test case was made where a computer generated phantom (a pyramid) was created by a test program called Teratest. This is done without having the optical position system attached to the ultrasound system. The setup can be seen in Fig. 6. Teratest calls the 'Pos3D.dll' to obtain a position based on the computer generated movement shown in Fig. 7.

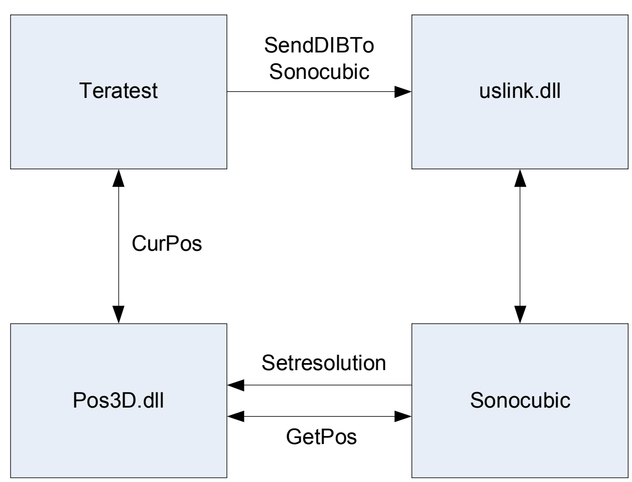


Figure 6. System setup for testing compensation algorithm with computer phantom (DIB = Device Independent Bitmap)

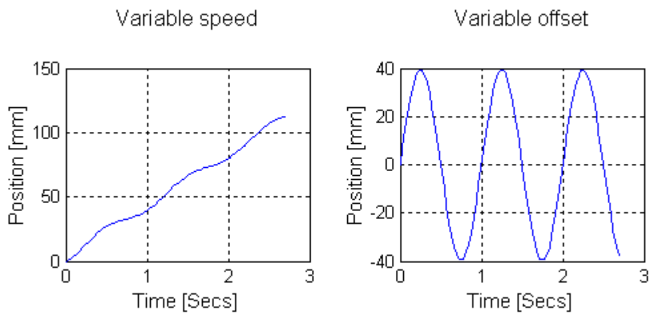
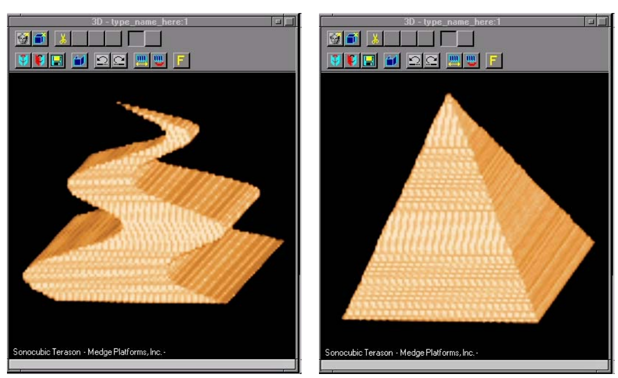


Figure 7. Computer generated movement of transducer with an uneven scan speed and offset

Based on the current position, found by calling 'CurPos', a scan plane is created that has an offset and a size that corresponds to how an ultrasound image would look if the scanner were moved over a real phantom. The scan plane is sent to Sonocubic via a shared memory interface, defined in 'uslink.dll'. Sonocubic reads from this shared memory interface and finds the resolution of the first scan plane from the header information. Based on this resolution a conversion factor is stored in 'Pos3D.dll'. The conversion factor relates the pixel movements with a real world movement, which is necessary in order to make a correct compensation. The 'GetPos' function in 'Pos3D.dll' is then called to obtain the current position. This position is the same as found with the 'CurPos' function, and if the compensation is working a compensated 3D ultrasound phantom should be the result.

In Fig. 8(a) the result from a "scan" on the computer generated phantom is seen without compensation. In Fig. 8(b) the same phantom is seen but with compensation. As it can be seen the compensated scan matches the shape of the computer phantom, so the compensation algorithm is working.



(a) (b)

Figure 8. Computer generated ultrasound phantom without (a) and with (b) compensation algorithm applied.

VI. 3D PHANTOM

For testing the performance of the system in real life we needed a 3D phantom that can be used for 3D freehand scanning. A suitable 3D phantom was not found available on the market, and we therefore decided to develop our own custom phantom. A recipe for making agar-based tissue-mimicking material was used, in which graphite powder is used to produce the backscatter characteristics [4]. A cylindrical agar piece, containing no graphite powder and with the dimensions shown in Fig. 9, was made and embedded in the tissue-mimicking material. The ultrasound image of the phantom resembled a cyst, as seen in Fig. 10 (a). One problem we encountered was that the surface of the phantom material was black and smooth (i.e. no optical texture or roughness), making it impossible for the optical sensor to track correctly on it. A solution to this was found by adding methylparaben (an ingredient already being used) near the surface of the phantom. The methylparaben surface degraded the image quality slightly.

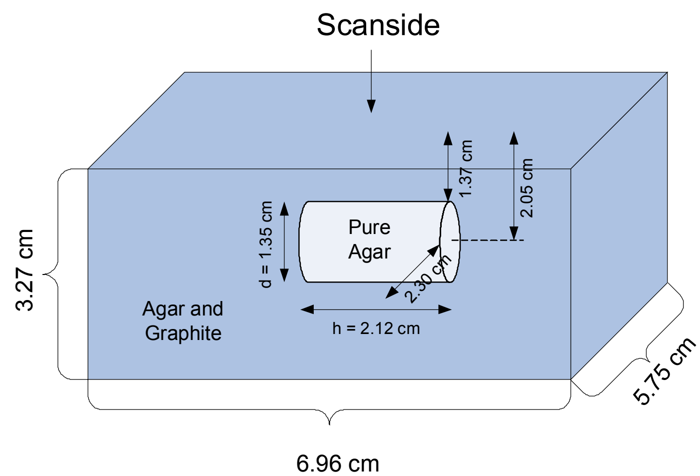
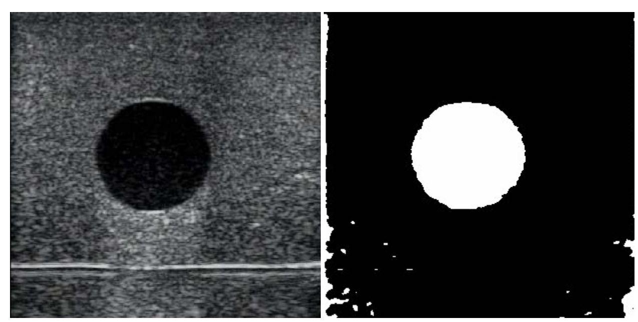


Figure 9. Dimensions of phantom with pure agar rod placed inside



(a) (b)

Figure 10. 2D ultrasound image from custommade phantom. Before (a) and after (b) image enhancement algorithm

VII. RESULTS

Five different scans were made of the phantom using the positioning system. This was carried out along a non-linear scan path, with an offset of approximately 1 cm from center. The scan planes were collected in the modified Sonocubic software. After the acquisition the modified interpolation algorithm calculated the data values for the voxels in the maingrid. The series of scan planes in the maingrid was then saved to an AVI-file for image enhancement in MATLAB.

In MATLAB, a Wiener filtering using a 5,5 neighborhood, followed by a nearest neighbor algorithm with a 3,3 neighborhood and a threshold was used to remove noise from the image. An inversion of the data was subsequently made to make visualization easier, so black parts appeared white and vice versa. The results can be seen in Fig. 10 (b) and Fig. 11 (a).

In each scan plane the sum of the pixel values were calculated and summed up. This sum of pixel values is equal to the sum of the voxels in the 3D volume and is therefore a representation of the size of the pure agar rod expressed in voxels. Since the voxel size is known the real volume of the pure agar rod can be found.

(a) (b)

Figure 11. 3D ultrasound image of pure agar rod (a) Actual image of pure agar rod after it was taken out of the phantom (b)

The volumes found were compared with the actual volume of the pure agar rod (3.03 cm^3), based on the physical dimensions. We obtained the results shown in Table 1

TABLE I.

RESULTS FROM VOLUME ESTIMATIONS

Scan no.	1	2	3	4	5
Volume[cm^3]	3.10	2.91	3.22	3.11	3.01
Norm. vol.	1.02	0.96	1.06	1.03	0.99

As it can be seen the system is able to determine the volume correctly with a highest deviation of 6% from the actual volume. The mean is at 3.07 (101%) and the standard deviation is 0.11 (3.72%).

VIII. CONCLUSION

In this paper we have presented a new positioning method for 3D ultrasound scanning. The physical size of the positioning system is small and well suited for portable applications. The positioning system overcomes some of the limitations that exist with freehand scanning and, when used together with a filtering algorithm, makes it possible to do very accurate volume estimations.

The system we have presented only has two degrees of freedom. We are in the process of implementing a system with angle information. This will give up to 5 degrees of freedom (only missing the depth). A patent is pending for the method described in this paper.

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CPN mouse filter driver : http://cpnmouse.sourceforge.net/
CPN = Colored Petri Nets

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