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● *Technical Note*

REGISTRATION OF REAL-TIME 3-D ULTRASOUND TO TOMOGRAPHIC IMAGES OF THE ABDOMINAL AORTA

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Abstract—The purpose of this study was to develop an image-based method for registration of real-time 3-D ultrasound to computed tomography (CT) of the abdominal aorta, targeting future use in ultrasound-guided endovascular intervention. We proposed a method in which a surface model of the aortic wall was segmented from CT, and the approximate initial location of this model relative to the ultrasound volume was manually indicated. The model was iteratively transformed to automatically optimize correspondence to the ultrasound data. Feasibility was studied using data from a silicon phantom and *in vivo* data from a volunteer with previously acquired CT. Through visual evaluation, the ultrasound and CT data were seen to correspond well after registration. Both aortic lumen and branching arteries were well aligned. The processing was done offline, and the registration took approximately 0.2 s per ultrasound volume. The results encourage further patient studies to investigate accuracy, robustness and clinical value of the approach. (E-mail: reidar.brekken@sintef.no) © 2016 World Federation for Ultrasound in Medicine and Biology. All rights reserved.

Key Words: **Ultrasound, Multimodal, Registration, Image-guidance, Aorta, Aneurysm, Intervention, Endovascular.**

INTRODUCTION

Abdominal aortic aneurysm (AAA) is a vascular disease resulting in a local dilation of the abdominal aorta. To prevent rupture of AAA, a possible treatment is to place a stentgraft inside the aneurysm using endovascular aortic repair (EVAR).

EVAR is routinely guided by fluoroscopy. For more advanced procedures, cone-beam computed tomography (CBCT) can be used, providing intra-operative 3-D computed tomography (CT) images. Commercial systems incorporate methods for 2-D and 3-D image registration as well as fusion with pre-acquired CT (Carrell et al. 2010; Manstad-Hulaas et al. 2011; McNally et al. 2015; Törnqvist et al. 2015).

Ultrasound is a well-established modality for detection and follow-up of AAA (Brekken et al. 2011), but despite the potential for reducing exposure to radiation and radiopaque contrast material, it has not become recognized as a modality for guidance of EVAR. Compared to ultrasound, CT provides a better anatomic overview and is less user-dependent and less influenced by bowel gas and obesity. Because CT is routinely acquired for planning of the intervention, it would be beneficial to combine the real-time, dynamic information from ultrasound with anatomic overview from pre-acquired CT to strengthen the potential for using ultrasound for guidance of EVAR.

Various approaches for registration of ultrasound to CT or magnetic resonance imaging (MRI) have been suggested. Reinertsen et al. (2007) described a vessel-to-vessel registration method using an iterative closest point algorithm for registering MRI to reconstructed 3-D ultrasound in neurosurgery. Wein et al. (2008) presented a method for registering CT to reconstructed 3-D ultrasound using ultrasound simulation to make the CT data more ultrasound-like, followed by an algorithm optimizing the registration based on a similarity measure between the simulated and real ultrasound images. Another approach is to use a navigation system for tracking the

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ultrasound probe and to apply manual, landmark-based registration for fusion of MRI or CT and ultrasound. Kaspersen et al. (2003) presented a study using a landmark-based approach for registering CT to ultrasound volumes reconstructed from navigated 2-D ultrasound for guidance of EVAR.

More recently, several systems for real-time 3-D ultrasound, sometimes referred to as 4-D ultrasound, have become available in clinics. Automatic, real-time registration of these volumetric ultrasound data to pre-acquired CT has a promising potential for constituting a next step toward using ultrasound for guidance of EVAR. In this paper, we describe a feasibility study aimed at developing a new method for image-based registration of real-time 3-D ultrasound to pre-acquired tomographic images of the abdominal aorta.

METHODS

Data acquisition

Data were acquired from a volunteer with a normal aorta (male, 27 y old), who had CT images previously acquired for other reasons, and from a soft-silicon flow phantom replicating an abdominal aorta (A-S-N-002-B; Elastrat, Geneva, Switzerland). CT was acquired before the phantom was submerged in water and attached to a pump to circulate a blood-mimicking fluid (BMF-US; Shelly Medical Imaging Technologies, London, Ontario, Canada) with steady flow. The volunteer gave informed consent to participate in the study, and only anonymized data were stored and used for further analysis. According to the guidelines from the regional ethics committee, technical and methodological development work using only anonymized data does not have to be submitted for approval, and the study was therefore not assessed by the committee. The study involved an additional ultrasound examination only, and was therefore not considered to be of any undue risk to the volunteer.

Real-time 3-D ultrasound was acquired using a matrix probe (4 V) connected to a GE Vivid E9 ultrasound scanner (GE Vingmed Ultrasound, Horten, Norway). B-mode and Power Doppler ultrasound data were streamed through a digital research interface from the scanner to the open-source navigation system CustusX (www.custusx.org) (Askeland et al. 2015), and stored for offline processing.

Registration method

The open-source segmentation tool ITK-snap (version 2.4.0, www.itksnap.org) was used for segmenting the lumen from the pre-acquired CT data (Yushkevich et al. 2006). The segmentation was exported as a binary volume, which was subsequently imported into MATLAB (Version R2014 b; The MathWorks, Inc., Natick, MA, USA) for generation of a triangulated surface with unit normal vectors for each surface point (vertex).

The purpose of the registration was to find a rigid-body transformation matrix T_{CT}^{us} moving the model from the original (CT) coordinates p_{CT} to ultrasound coordinates p_{us} (eqn [1])

$$p_{us} = T_{CT}^{us} \cdot p_{CT} \tag{1}$$

The position of the model relative to the ultrasound data was initialized by manually indicating a point in the model approximately corresponding to the center of the ultrasound data. Both Power Doppler (US_{PD}) and B-mode (US_{Bmode}) data were evaluated at the surface of the model in the current position to determine if each surface point was inside or outside the lumen, thus imposing either an outward- or inward-directed virtual force at each point, respectively. In detail, a lumen probability map (LPM) was generated, with positive values likely to be inside lumen and negative values likely to be outside lumen (eqn [2]):

$$\mathrm{LPM} = w \cdot \frac{US_{PD} - A_{PD}}{B_{PD}} - (1-w) \cdot \frac{US_{Bmode} - A_{Bmode}}{B_{Bmode}} \tag{2}$$

where A_{PD} and A_{Bmode} define threshold values for the lumen in Power Doppler and B-mode, respectively. B_{PD} and B_{Bmode} are parameters used for normalizing the LPM to range approximately from -1 to 1, and w is a weight parameter ranging from 0–1, where 0 includes only B-mode and 1 includes only Power Doppler. In our study, we used $A_{PD} = 10$, $A_{Bmode} = 120$, $B_{PD} = 100$ and $B_{Bmode} = 120$. For the human case, w was set to 0.5, and 0.95 for the phantom. Because the signal outside the aorta was low for the phantom (submerged in a water bath) compared to the *in vivo* data, w needed to be changed. For pulsating flow, the Doppler intensities vary during the heart cycle. Because no electrocardiogram data was recorded, the systolic frame was manually chosen and used for US_{PD}.

The contributions from each point (i) were summed up to a net force (F) and torque (τ) (eqn [3])

$$\begin{aligned} \vec{f_i} &= \mathrm{LPM}_i \cdot \overrightarrow{en_i} \\ \vec{F} &= \frac{1}{K} \cdot \sum_i \vec{f_i} \\ \vec{\tau_i} &= \frac{1}{K} \cdot \sum_i \vec{r_i} \times \vec{f_i} \end{aligned} \tag{3}$$

where en is the outwards unit normal vector of each point, and r is the distance from the center of the model to each point on the model surface. K is the number of