REGISTRATION-BASED PROPAGATION FOR WHOLE HEART SEGMENTATION FROM COMPOUNDED 3D ECHOCARDIOGRAPHY

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ABSTRACT

Whole heart segmentation of 3D ultrasound (US), also referred to as echocardiography or simply echo, is useful in cardiac functional analysis to achieve quantitative diagnostic information of the heart. However, characteristics of US imaging such as limited field-of-view, artifacts and inconsistent intensity distribution makes automated approaches a challenge. In this paper, we present a framework for automatic whole heart segmentation from 3D echo. This work is motivated by the new technology of compounding 3D echo from 2D matrix array transducers. We propose to use the registration-based segmentation framework and adopt a new similarity measure combining local phase, intensity information and local geometry for registration. The experimental results demonstrated the proposed method had achieved an accuracy of 6.4% volume difference against the gold standard for the left ventricle segmentation and an average accuracy of 14% for segmentation of all four chambers and myocardium.

***Index Terms*—** Whole heart segmentation, ultrasound, echocardiography, image registration

1. INTRODUCTION

Ultrasound (US) is a safe, low-cost and real-time interactive imaging technique, providing anatomical and diagnostic information for functional analysis. Three dimensional (3D) echo has the potential of providing more accurate quantitative analysis. However, most methods require some form of segmentation process, which can be difficult to automate due to the typically low signal-to-noise and small field-of-view in US imaging.

Several papers have reported successful attempts in US segmentation, see the survey in [1]. The proposed segmentation methods used deformable models or active contours to search for optimal boundaries with prior constraints to maintain a realistic shape. By contrast, no work has been published, to the best of our knowledge, on employing image registration techniques to perform segmentation propagation, probably due to the difficulty of achieving a successful nonrigid registration using echo images. Furthermore, the reported works mainly focus on the left ventricle or myocardium due to the limited field-of-view of the acoustic window. Whole heart segmentation however can provide more useful information for computational modeling or guidance of interventional procedures.

Recently, compounding techniques, which align and then combine intensity information from multiple image volumes, can produce 3D echo images with a much wider field-of-view and better image quality than a single US scan, in particular when a large number of volumes are used [2]. These compounding techniques can significantly reduce the intensity variations of tissues presented in single-scan US image, as shown in Fig. 1. This results in an increased consistency of image intensities between the template and image to be segmented, and so has the potential to increase registration accuracy and robustness using intensity information.

Furthermore, the local phase analysis of US images provides a continuous scalar value to describe the feature presented in the image. Hence, intensity-based registration can also be applied to US images using the local phase as feature intensity. Also, Haber and Modersitzki argued that a similarity measure computed from local geometrical information such as normalized gradient was applicable to images with different intensity distributions [3]. The local geometrical information is based on image structures. Therefore, we propose to use more information from the compounded echo images for the computation of the similarity measure, including intensity information, local phase information and local geometrical information.

Finally, it has been shown that the registration-based segmentation propagation framework can achieve a successful application in cardiac magnetic resonance (MR) images [4]. One advantage of this segmentation framework is that it does not require a training phase from a large dataset for different pathologies to perform the segmentation of an unseen case, unlike the segmentation using statistical prior knowledge. Instead, it only needs the prior segmentation of the template image, the atlas, in order to perform the propagation. By using the locally affine registration method (LARM) to further initialize the anatomical substructures, this framework has been shown to be applicable to a wide range of patholo-

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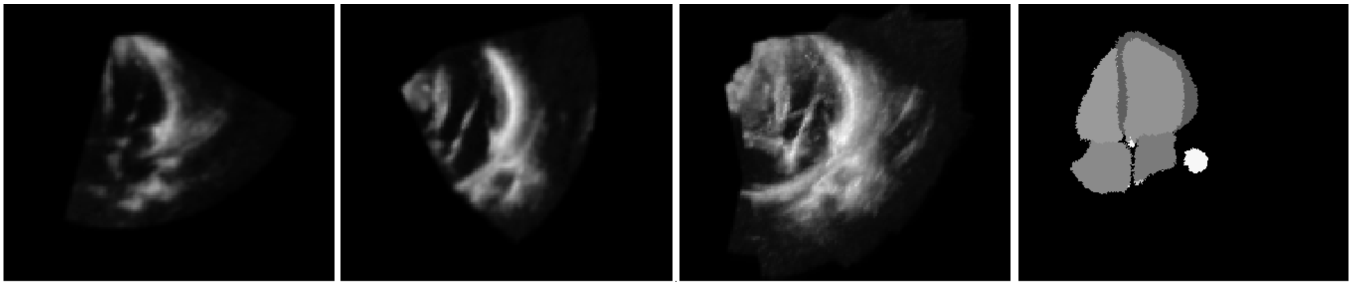

(a) (b) (c) (d)

Fig. 1. Example images and outputs from algorithms used during segmentation process, (a) view from apical scan; (b) view from parasternal view; (c) compounded image from 12 scans; (d) corresponding segmentation labels of (c).

gies [9]. This segmentation framework also has the potential of being applicable to other modalities, including US imaging given a similarity measure for echo registration is achievable. Therefore, we propose a new similarity measure for registering compounded 3D echo and apply it to the whole heart segmentation framework.

This paper is organized as follows: section 2 presents the methodology; section 3 demonstrates the segmentation results of twelve propagation cases; and our conclusions and discussions are given in section 4.

2. METHOD

2.1. Image compounding

Compounding large numbers of US images not only can improve the image quality by reducing noise and artifacts and increasing signal-to-noise ratio, but more importantly it can extend the field-of-view and reduce intensity inconsistencies caused by the different incidence angles of the US beams. The wider coverage provides the potential to carry out whole heart functional analysis, while the reduction in intensity inconsistencies increases the usefulness of the intensity values in differentiating tissues and boundaries.

The following techniques are adopted to produce the compounded images. Firstly, we acquire over ten US images for each subject using a variety of acoustic windows and probe positions. The aim is to achieve a maximal coverage of the heart from different angles. Triggered by electrocardiography (ECG) gating, wide sector acquisitions are taken with volunteers positioned in the left lateral decubitus position, breath-hold at end-exhale. At the same time, the echo transducer is attached with an array of light emitting diodes to enable its tracking during acquisitions. A probe calibration matrix is calculated as described in [6]. This matrix along with the probe tracking matrix is used to provide a starting estimate for image alignment. The alignment of all echo images is achieved by performing the semi-simultaneously (groupwise) registration scheme [7] and using the similarity measure of local phase and orientation information proposed in [5]. Fig. 1 (c) shows the four-chamber view of a compounded image from 12 scans using the maximum method [2].

2.2. Registration for segmentation propagation

We use three different types of information extracted from the compounded 3D echo: image intensity I, local phase ϕ and local geometric information $\vec{v}$.

The local phase ϕ, providing a quantitative, continuous and contrast invariant description of local features in images, can be derived using the monogenic signal [5]:

$$\phi(\mathrm{x}) = \mathrm{atan2}\Big(\sqrt{\sum_{i=1}^{3}(g(\mathrm{x}) * h_i(\mathrm{x}) * I(\mathrm{x}))^2},\ g(\mathrm{x}) * I(\mathrm{x})\Big), \tag{1}$$

where, g is a zero mean bandpass filter such as the log-Gabor filter [5], convoluted with I to constitute the even component of the signal; $\{h_i\}$ are the odd anti-symmetric filters in the spatial domain, $i = 1, 2, 3$, whose expression in frequency domain H_i is:

$$H_i(u_1, u_2, u_3) = \frac{u_i}{\sqrt{u_1^2 + u_2^2 + u_3^2}}. \tag{2}$$

The local geometric information, which denotes the perpendicular direction to the boundaries, can be computed from the normalized gradient [3]:

$$\vec{v}(\mathrm{x}) = \frac{\nabla I(\mathrm{x})}{||\nabla I(\mathrm{x})||}\ . \tag{3}$$

Combining intensity, local phase and local geometric information, we propose the following metric for the similarity measure of target image I_t and source image I_s:

$$\begin{aligned} \mathcal{C} = w_c \mathtt{NCC}(I_t, I_s) + &w_f \mathtt{NCC}(\phi_t, \phi_s) \\ &+ w_v \tfrac{1}{|\Omega|} \textstyle\int_{\mathrm{x}\in\Omega} [\vec{v}_t(\mathrm{x}) \cdot \vec{v}_s(\mathrm{x})]\ , \end{aligned} \tag{4}$$

where, Ω is the image volume, $\mathtt{NCC}(I_t, I_s)$ and $\mathtt{NCC}(\phi_t, \phi_s)$ are the normalized cross correlation (NCC) of the image intensity and phase; w_c, w_f, and w_v control the weightings given to each term.

For registration of heart images, Zhuang *et al.* [4] proposed to use a locally affine registration, LARM, to initialize the anatomical substructures between the atlas and the unseen image. LARM defines the local affine transformations based on the anatomical substructures of the heart, and it can preserve the local shape of substructures and provide a robust registration for the following nonrigid registration.

Therefore, we also adopt the locally affine transformation model for the substructure initialization registration. The segmentation propagation includes the following three steps:

- A rigid registration to localize the heart structure.
- LARM to further initialize the anatomical substructures, using four local affine transformations associated with the blood cavities of the four chambers.
- A free-form deformation (FFD) registration [8] to refine the registration of the local details.

Gradient ascent optimization is used for the three image registration processes. In the LARM and FFD registrations the local phase is not transformation invariant, and so updating local phase is computationally expensive. Therefore, we only re-estimate the local phase after every few iteration steps, typically 10 steps to ensure the errors do not accumulate.

3. EXPERIMENT

To construct gold standard segmentation for validation, we acquired a 3D isotropic MR image at the end-diastolic cardiac phase (same phase as used for the compounded echo images) for each subject before acquiring the echo images. An automatic whole heart segmentation of the MR image was then achieved using the segmentation method proposed in [9] where a mean accuracy of 9.1% volume difference was reported for the segmentation of the four chambers. This segmentation of the MR image was then assumed to be the gold standard segmentation of the corresponding 3D echo with a rigid transformation from the MR image to the echo image. The rigid transformation was obtained using a feature-based registration algorithm [6]. The registrations were also visually inspected, and small manual corrections were carried out if errors were observed. It should be noted that independent segmentations from MR and echo images may not perfectly match [10].

Twelve propagation cases were evaluated using four sets of echo and MR data and a validation strategy as follows: A dataset was chosen to represent the atlas, which was comprised of the compounded 3D echo and the label image produced by segmenting the MR volume, an example is shown in Fig. 1 (c) and (d). This atlas was then propagated to segment the other three datasets. The above was repeated four times, taking a different dataset to represent the atlas each time. Segmentation results were assessed by comparison with the MR derived gold standard using the volume difference measure,

$$\mathrm{VD} = \frac{2(|V_{seg} - V_{gd}|)}{V_{seg} + V_{gd}} \times 100\%, \tag{5}$$

where V_{seg} is the resultant segmentation, while V_{gd} is the gold standard. The volume measure was performed on the blood pools of the four chambers and the left ventricle myocardium. The average volume of the five regions was also computed, referred to as Whole Heart.

Table 1 gives the mean segmentation accuracy of the twelve cases, while Fig. 2 illustrates the visual results of four segmentation cases. The performance of the segmentation algorithm is promising, in particular, the algorithm achieved success in heart regions which were not fully covered in the compounded image. This was probably owing to the good initialization of LARM which registered the four chambers with locally affine transformations and hence maintained the local shapes of the four chambers between the atlas and the unseen image.

The segmentation achieved the best result in the left ventricle, the main interest of many echo segmentation works. This accuracy ($6.4 \pm 4.7\%$) is also comparable to the result ($6.5 \pm 4.9\%$) reported in the MR segmentation in [9]. The segmentation of the two atria was worse. This may be caused by heavy artifacts present in the region of the left atrium and because of incomplete coverage of the right atrium, as seen in the example images Fig. 2.

Table 1. Segmentation accuracy presented as volume differences. Column *before* shows the result before segmentation propagation i.e. the difference between the atlas and unseen image, column *after* shows the result after the propagation.

(%)	Left Ventricle	Left Atrium	Right Ventricle
Before	23 ± 14	29 ± 15	16 ± 9.0
After	6.4 ± 4.7	16 ± 12	9.4 ± 9.5
	Right Atrium	Myocardium	Whole Heart
Before	22 ± 11	22 ± 12	22 ± 13
After	16 ± 12	15 ± 13	14 ± 12

4. CONCLUSION AND DISCUSSION

In this work, we have proposed a framework for automatic whole heart segmentation from 3D compounded echo using image registration. A new similarity measure has been proposed to combine three types of information, local phase, intensity value, and local orientation to improve the registration performance. The experiments demonstrated the proposed method achieved an accuracy of 6.4% volume difference against the gold standard for the left ventricle segmentation, comparable to an MR segmentation [9] and an average accuracy of 14% over all four chambers and the myocardium.

The segmentation performance is particularly affected if there are large variations in heart coverage between the atlas and unseen image. Therefore, in the future we plan to investigate the use of specific acquistion protocols, based on standard echo views which we believe should improve the field-of-view of the compounded volume, and increase consistency

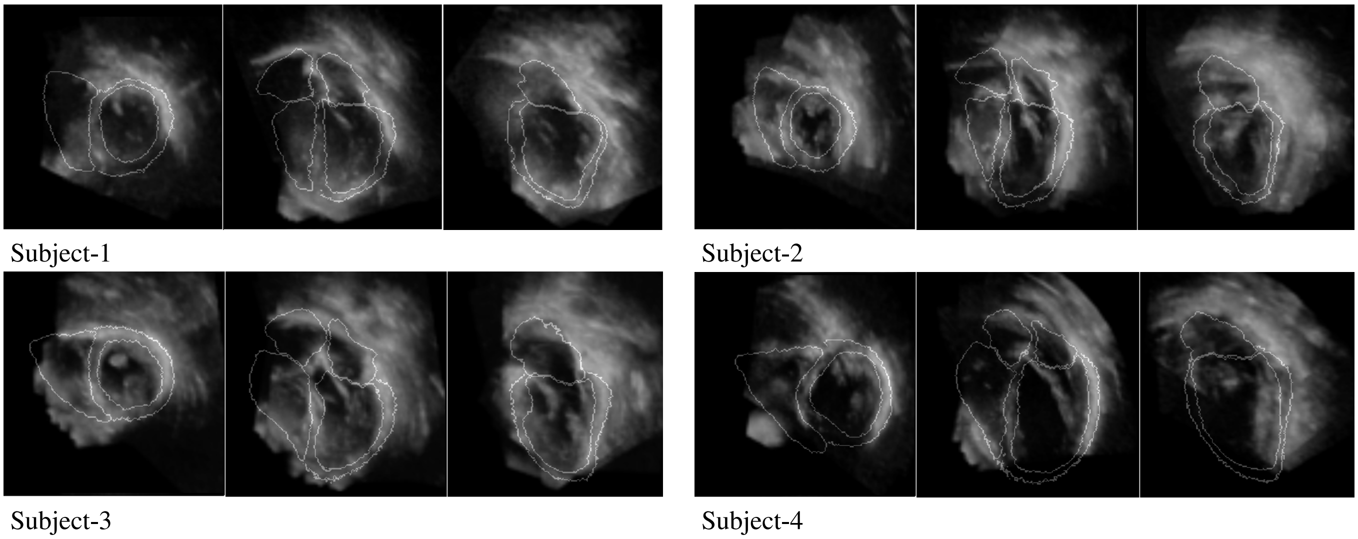

Fig. 2. The segmentation propagation results of four cases. Results are displayed in the short-axis view, the four-chamber view and the two-chamber view.

between datasets.

Echo is by far the most commonly used imaging modality for cardiac diagnosis. The ability to automatically and accurately extract information from echo has huge clinical potential. Our segmentation framework, which makes use of compounded 3D echo volumes, has shown the potential of using an atlas-based approach to carry out fully automatic whole heart anatomical segmentation. The transformation model used in this approach has shown to be robust against large shape variability of the heart from a wide range of pathologies in MR segmentation [9]. The same applicability of this framework for compounded echo is expected and the validation will be our future work.

5. REFERENCES

[1] J. A. Noble and D. Boukerroui, "Ultrasound image segmentation: A survey," *IEEE Transaction on Medical Imaging*, vol. 25, no. 8, pp. 987–1010, 2006.

[2] Cheng Yao, John Simpson, Christain Jansen, Andrew King, and Graeme Penney, "Spatial compounding of large sets of 3D echocardiography images," in *SPIE, Medical Imaging*, 2009.

[3] Eldad Haber and Jan Modersitzki, "Intensity gradient based registration and fusion of multi-modal images," in *MICCAI 2006*, vol. 4191 of *LNCS*, pp. 726–733, 2006.

[4] Xiahai Zhuang, Kawal Rhode, Simon Arridge, Reza Razavi, Derek Hill, David Hawkes, and Sebastien Ourselin, "An atlas-based segmentation propagation framework using locally affine registration – application to automatic whole heart segmentation," in *MICCAI 2008*, vol. 5242 of *LNCS*, pp. 425–433, 2008.

[5] V. Grau, H. Becher, and J. A. Noble, "Registration of multiview real-time 3-D echocardiographic sequences," *IEEE Transactions on Medical Imaging*, vol. 26, no. 9, pp. 1154–1165, 2007.

[6] Y. L. Ma, G. Penney, C. Rinaldi, M. Cooklin, R. Razavi, and K. Rhode, "Echocardiography to magnetic resonance image registration for use in image-guided cardiac catheterization procedures," in *Physics in Medicine and Biology*, vol. 54, no. 16, pp. 5039-5055, 2009.

[7] Christian Wachinger, Wolfgang Wein, and Nassir Navab, "Three-dimensional ultrasound mosaicing," in *MICCAI 2007*, vol. 4792 of *LNCS*, pp. 327–335, 2007.

[8] D Rueckert, P Aljabar, R A Heckemann, J V Hajnal, and A Hammers, "Diffeomorphic registration using b-splines," in *MICCAI 2006*, vol. 4191 of *LNCS*, pp. 702–709, 2006.

[9] Xiahai Zhuang, Kawal Rhode, Reza Razavi, David J Hawkes, and Sebastien Ourselin, "Free-Form Deformations Using Adaptive Control Point Status for Whole Heart MR Segmentation," in *FIMH 2009*, vol. 5528 of *LNCS*, 303-311. 2009.

[10] S. Takuma, T. Ota, T. Muro, T. Hozumi, R. Sciacca, M. R. Di Tullio, D. K. Blood, J Yoshikawa, and S. Homma, "Assessment of left ventricular function by real-time 3-dimensional echocardiography compared with conventional noninvasive methods," *Journal of the American Society of Echocardiography*, vol. 14, pp. 275–284, 2001.