Cardiac Three-Dimensional Magnetic Resonance Imaging and Fluoroscopy Merging
A New Approach for Electroanatomic Mapping to Assist Catheter Ablation

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Background—Modern nonfluoroscopic mapping systems construct 3D electroanatomic maps by tracking intracardiac catheters. They require specialized catheters and/or dedicated hardware. We developed a new method for electroanatomic mapping by merging detailed 3D models of the endocardial cavities with fluoroscopic images without the need for specialized hardware. This developmental work focused on the right atrium because of the difficulties in visualizing its anatomic landmarks in 3D with current approaches.

Methods and Results—Cardiac MRI images were acquired in 39 patients referred for radiofrequency catheter ablation using balanced steady state free-precession sequences. We optimized acquisition and developed software for construction of detailed 3D models, after contouring of endocardial cavities with cross-checking of different imaging planes. 3D models were then merged with biplane fluoroscopic images by methods for image calibration and registration implemented in a custom software application. The feasibility and accuracy of this merging process were determined in heart-cast experiments and electroanatomic mapping in patients. Right atrial dimensions and relevant anatomic landmarks could be identified and measured in all 3D models. Cephalocaudal, posteroanterior, and lateroseptal diameters were, respectively, 65±11, 54±11, and 57±9 mm; posterior isthmus length was 26±6 mm; Eustachian valve height was 5±5 mm; and coronary sinus ostium height and width were 16±3 and 12±3 mm, respectively (n=39). The average alignment error was 0.2±0.3 mm in heart casts (n=40) and 1.9 to 2.5 mm in patient experiments (n=9), ie, acceptable for clinical use. In 11 patients, reliable catheter positioning and projection of activation times resulted in 3D electroanatomic maps with an unprecedented level of anatomic detail, which assisted ablation.

Conclusions—This new approach allows activation visualization in a highly detailed 3D anatomic environment without the need for a specialized nonfluoroscopic mapping system. (Circulation. 2005;112:3769-3776.)

Key Words: catheter ablation ▪ atrium ▪ mapping ▪ magnetic resonance imaging ▪ imaging

Radiofrequency catheter ablation techniques have had a dramatic impact on the treatment of various forms of arrhythmias. Fluoroscopy-guided catheter navigation and ablation of accessory pathways, AV nodal slow pathways, and atrial flutter are accepted as first-line therapy with excellent outcomes.1-3 In the late 1990s, catheter ablation of more complex arrhythmias, such as postsurgical intra-atrial reentrant tachycardia, ventricular tachycardia, and atrial fibrillation, encouraged the development of new nonfluoroscopic mapping techniques in which electrical data and anatomic information are combined into 3D electroanatomic models.4-6 These systems continuously record the location of a roving mapping catheter and create a map of electrical activity in 3D space. However, they require a dedicated system and often special catheters, which makes these mapping systems expensive. Moreover, by limiting the anatomic information to points collected by a roving catheter, the resulting model is only a rough geometric approximation of the endocardial cavity, because points that are difficult to reach are not included. The latest technological developments allow merging of these electroanatomic maps with MRI- or CT-based 3D models.7 However, this approach requires registration of two 3D models with often very differing appearances to project the electrical activation map on the anatomic model. Most research work has focused on 3D modeling of the left atrium.8-11 The right atrium is less suitable for 3D modeling because of its anatomic complexity and inhomogeneous filling with gadolinium and iodine con-
The endocardial contours of the heart chambers were manually delineated on the MRI images with in-house developed software that allowed for cross-checking of the contours with those in other imaging planes (Figure 2). The software also compensated for positional differences in orthogonal planes that were acquired during different breath holds in the bFFE sequence, with automated 3D image registration by maximization of mutual information. All manually delineated contours in different 2D imaging planes were then combined into a single 3D surface model with a scattered data interpolation algorithm based on radial basis functions. Right atrial dimensions were measured directly on the 3D model along the cephalocaudal, posteroanterior and lateroseptal axes. In addition, posterior isthmus length, eustachian valve height, and coronary sinus ostium height and width were measured on the 3D models.

Integration of 3D Model in Biplane Fluoroscopic Framework
All radiofrequency catheter ablation procedures were performed with a conventional biplane fluoroscopy system (Bicor, Siemens) and standard electrophysiological mapping and ablation catheters. A custom software application was developed to allow integration of the 3D anatomic models into the biplane fluoroscopy framework. This integration process consisted of 2 steps:

1. Fluoroscopy geometry calibration: To render the 3D model in the biplane fluoroscopic framework in its proper aspect ratio, a calibration procedure was performed based on the fluoroscopic view angles and a calibration object. The fluoroscopic view angles were determined automatically from the DICOM (Digital Imaging and Communications in Medicine) header information of the fluoroscopic images. As a calibration object, we used intracardiac catheters with known interelectrode distances, which were measured on fluoroscopic images in both viewing planes (a minimum of 7 electrode positions were marked in each plane).

2. Image registration: After the calibration procedure, the 3D model could be introduced in the fluoroscopic framework in its proper aspect ratio at a standard offset. Initial model rotation was calculated from the fluoroscopic view angles. Subsequently, the transparent model was carefully manually translated and rotated by means of a 3D mouse (SpaceMouse, 3Dconnexion Inc) to properly align it with angiographic images of the heart in 2 projections (at end systole) by a modified version of the visual matching technique (Figure 3). The acquisition of angiographic images was performed with the same end-systolic triggering interval as for the MRI acquisition to avoid synchronicity mismatch.

The feasibility and accuracy of this integration process were assessed in 2 ways:

1. Phantom experiments: Registration by the proposed visual matching technique was tested on a set of simulated fluoroscopy images of a plastic endocardial negative heart cast. These fluoroscopic
images were rendered in a 3D environment together with the 3D model of the cast, which was positioned at a random translational and rotational offset. Subsequently, the 3D model was visually aligned with the simulated fluoroscopic images from 20 different offsets by 2 independent operators (1 cardiologist familiar with cardiac anatomy and 1 clinical engineer). Alignment error in the $x$, $y$, and $z$ axes, angular deviation, and the time to complete the task were recorded.

2. Patient validation: After registration of right atrial 3D models with right atrial angiography in 9 patients, we measured the distance between heart boundaries visible on angiography and projected boundaries of the 3D model rendered in the fluoroscopy framework. Distance between boundaries was defined as the area between the 2 manually outlined boundaries divided by boundary length (Figure 4). This alignment error was evaluated separately for posterior, inferior, and anterior borders in the right anterior oblique view and for lateral and septal borders in the left anterior oblique view. If no clear border could be visualized on angiography (eg, due to incomplete filling with contrast agent), this region was not used for measurement.
Projection of Catheter Positions From Fluoroscopic Images to 3D Model

After proper image calibration and registration, the position of intracardiac catheters could be projected from the fluoroscopy images to the 3D model. For this purpose, 3 anchor points of the catheter were manually marked on the fluoroscopy images and automatically projected onto the 3D model as a virtual catheter. Further modifications of the catheter curve were possible in the 3D environment using 3D crosshairs. The position of the virtual coronary sinus catheter within the coronary sinus ostium of the 3D model was used for additional evaluation of correct fluoroscopy model alignment in 11 patients (Figure 5).

Electroanatomic Mapping on 3D Model

During conventional electrophysiological mapping of arrhythmias, multiple positions from different catheters were projected onto the 3D model by the method described above. Their electrograms were recorded, and bipolar activation times on different electrode pairs (2 mm spacing) were graphically represented on the 3D surface, which resulted in a color-coded electrical activation map of the arrhythmia (Figure 6). During this developmental work, activation times were manually marked on the catheter electrograms. Fractionated and double potentials could also be represented on the 3D model, as well as scar tissue and orifices. Multielectrode mapping catheters were used for rapid electrical data acquisition at different catheter positions to accelerate the mapping process.

The investigational review board approved this study, and informed consent was obtained from all patients.

Statistical Analysis

Data are presented as mean±SD unless otherwise stated.

Results

Image Acquisition and Processing to 3D Model

Imaging time required for acquisition of the 3D bTFE sequence exceeded imaging time for the bFFE sequence by more than 10-fold (15±2.5 versus 1±0.2 minutes). Time required for manual contouring of the right atrium was not different between the 2 sequences (bFFE 1.5±0.4 hours, bTFE 1.6±0.4 hours, P=NS), even though more slices had to be contoured in the bTFE sequence (bTFE 120±5 axial slices, bFFE 20 slices in the axial, sagittal, and coronal planes). This was explained by increased complexity of contouring and cross-checking contours in different imaging planes in the 3 orthogonal bFFE acquisitions.

The accuracy of distance measurements was evaluated on an MRI-based 3D model of a static plastic endocardial negative heart cast. Distance measurements of structures on the 3D model corresponded exactly to distances on the 3D model (differences <0.5 mm). The endocavitary volume...
coronary sinus ostium height and width were 16 ± 3 and 12 ± 3 mm, respectively, and eustachian valve height was 8 ± 5 mm for both sequences, with a mean measurement difference of 0 ± 1 mm. The coronary sinus ostium height and width were 16 ± 3 and 12 ± 3 mm, respectively, in the bFFE sequence versus 15 ± 4 and 10 ± 3 mm in the 3D bFFE sequence, with a mean measurement difference of 0 ± 1 mm. The mean eustachian valve height in its middle portion and coronary sinus ostium height and width on 3D models of the same patients based on the bFFE and 3D bFFE sequences (n = 11). The mean eustachian valve height was 8 ± 5 mm for both sequences, with a mean measurement difference of 0 ± 1 mm. The coronary sinus ostium height and width were 16 ± 3 and 12 ± 3 mm, respectively, in the bFFE sequence versus 15 ± 4 and 10 ± 3 mm in the 3D bFFE sequence, with a mean measurement difference of 0 ± 1 mm. The mean eustachian valve height in its middle portion and coronary sinus ostium height and width on 3D models of the same patients based on the bFFE and 3D bFFE sequences (n = 11). The mean eustachian valve height was 8 ± 5 mm for both sequences, with a mean measurement difference of 0 ± 1 mm. The coronary sinus ostium height and width were 16 ± 3 and 12 ± 3 mm, respectively, in the bFFE sequence versus 15 ± 4 and 10 ± 3 mm in the 3D bFFE sequence, with a mean measurement difference of 0 ± 1 mm. The mean eustachian valve height in its middle portion and coronary sinus ostium height and width on 3D models of the same patients based on the bFFE and 3D bFFE sequences (n = 11). The mean eustachian valve height was 8 ± 5 mm for both sequences, with a mean measurement difference of 0 ± 1 mm. The coronary sinus ostium height and width were 16 ± 3 and 12 ± 3 mm, respectively, in the bFFE sequence versus 15 ± 4 and 10 ± 3 mm in the 3D bFFE sequence, with a mean measurement difference of 0 ± 1 mm.

The registration procedure in the heart-cast experiment could be completed within 5.0 minutes in 95% of cases by the cardiologist and in 70% of cases by the engineer. The initial random alignment errors were 0 ± 15 mm translational and 0 ± 5° rotational for the x, y, and z axes separately. After the registration procedure, the mean translational alignment error in the x, y, and z axes (ie, horizontal and vertical axes in the left anterior oblique view, respectively) for 20 registration procedures of the heart-cast model was 0.0 ± 0.2, 0.4 ± 0.3, and 0.2 ± 0.3 mm, respectively, for the cardiologist (overall mean 0.2 ± 0.3 mm) and 0.3 ± 0.3, 0.2 ± 0.3, and 0.4 ± 0.3 mm, respectively, for the engineer (overall mean 0.2 ± 0.4, difference = NS; Figure 7A). The mean angular deviations around the same x, y, and z axes were −1.0 ± 1.6°, −0.7 ± 0.6°, and 0.1 ± 1.3°, respectively, for the cardiologist and −0.3 ± 1.2°, −0.4 ± 0.6°, and −0.3 ± 0.8° for the engineer (difference = NS; Figure 7B).

The median distances between angiographic and 3D model boundaries in 9 patients were 1.9 mm (interquartile range 1.0 to 2.7 mm) for posterior, 2.4 mm (1.9 to 3.3 mm) for inferior, and 2.5 mm (1.5 to 9.9 mm) for anterior borders in the right anterior oblique view. In the left anterior oblique view, there was a median distance of 2.1 mm (interquartile range 1.7 to 3.1 mm) between lateral boundaries and 2.1 mm (2 to 5.1 mm) between septal boundaries (Figure 7C). When this methodology was used to calculate alignment errors between the boundaries of the MRI-based 3D model of the heart cast and its static fluoroscopy projection, alignment errors for all borders were negligible (0.37 to 0.6 mm). This suggests that most of the observed alignment errors in patients are due to cardiac motion or changes of the endocardial cavity dimension between imaging and the ablation procedure.

Coronary sinus catheters correctly followed the course of the coronary sinus as visualized on the 3D models in all 11 patients. Their most proximal segment projected correctly within the coronary sinus ostium in 7 of 11 patients and within 5 mm of the ostium in all 11 patients. This small error at the level of the coronary sinus ostium is most probably due to catheter-related pressure exerted in this region (see below).

**Projection of Catheter Positions and Electroanatomic Mapping**

The construction of a color-coded activation map during tachycardia or localized pacing was possible in all patients in whom it was attempted (n = 11). They included patients with clockwise or counterclockwise atrial flutter (n = 4), accessory pathways (n = 2), AV nodal reentrant tachycardia (n = 2), and complex arrhythmogenic substrates such as periatriotomy or congenital reentrant tachycardia (n = 3) (Figure 6). The average time needed for completion of these activation maps (based on current manual annotation of activation times) was 30 ± 6 minutes for an average of 62 ± 25 mapped points. There were no significant differences in total procedural times, fluoroscopy times, and number of radiofrequency applications compared with patients who were treated conventionally in the same time period for the same type of arrhythmias (Table). The mean number of radiofrequency applications in the group with complex arrhythmogenic substrates was higher (although not statistically significant) in the electroanatomic mapping group owing to 1 patient requiring a total of 17 radiofrequency applications.

**Discussion**

The main results of our research are the development of a mapping system that (1) is primarily based on highly detailed anatomic information, (2) does not require special mapping...
catheters, and (3) can be applied in a regular biplane fluoroscopy setup. Moreover, by using cardiac MRI for image acquisition, we did not invoke extra radiation for the patients (although our methodology could also be applied to cardiac CT).

Although MRI- or CT-based 3D volume rendering of the left atrium and pulmonary veins is frequently performed to assess left atrial anatomy (such as for catheter ablation of atrial fibrillation⁹,¹⁰), a similar approach for the right atrium has not yet been developed. Intravenously administered contrast agents are not homogeneously distributed in the right atrium (in contrast to the left atrium) and therefore do not allow easy segmentation and 3D visualization of the right atrium. Nevertheless, the majority of radiofrequency catheter ablation procedures are right-sided, and exact knowledge of right atrial anatomy could certainly benefit mapping procedures in patients with complex arrhythmogenic substrates, such as those with congenital heart disease or previous cardiac surgery. Therefore, we developed a methodology to construct right atrial 3D models without contrast agents by manually delineating and cross-checking endocardial contours in different imaging planes. This resulted in surface-rendered 3D models of the right atrium with an unprecedented level of detail, which also represented fine anatomic structures relevant to electrophysiology procedures. The time required for developing such a detailed 3D model of the right atrium is still substantial (1.5 hours), but it may be reduced by automating certain steps in the delineation process with model-based segmentation approaches.¹⁶ The preprocedural availability of a detailed and anatomically correct model may also allow preprocedural planning and selection of specialized guiding sheaths and ablation catheters. This is an important advantage compared with existing nonfluoroscopic mapping systems.

The methodology of manual contouring in differently oriented imaging planes allowed the use of the bFFE MRI sequence with 6-mm slice thickness to construct 3D models

Figure 7. A and B, Translational and rotational error as measured in 40 visual matching registration procedures of the plastic heart cast and simulated fluoroscopy images. X represents horizontal axis in right anterior oblique view; Y, vertical axis in right anterior oblique view; Z, horizontal axis in left anterior oblique view. C, Distance between borders visualized on angiography and 3D model in 9 patients.
The integration of detailed cardiac 3D models in the biplane fluoroscopic framework is a new concept. Its main advantage is the wide availability and general use of biplane fluoroscopy in electrophysiology laboratories around the world. The registration of the 3D models with angiography images was performed by a visual matching technique. Theoretically, automated registration of magnetic resonance images with fluoroscopy is possible with intensity-based registration measures, but we found these methods not to be reliable enough for use in clinical practice. The visual matching procedure could be performed with high accuracy and reproducibility by 2 independent operators. The maximal translational alignment errors in the heart-cast experiments were <1 mm, and the median distance between angiographic and model borders in patients ranged from 1.9 to 2.5 mm. The large error for the anterior border (median 2.5 mm, interquartile range 1.5 to 9.9 mm) is explained by the fact that this border represents the tricuspid valve in the right anterior oblique view. Very small differences in timing of the angiography image acquisition can have a profound impact on valve position and are the most likely explanation for the observed difference. The observed alignment error is certainly acceptable for use in clinical practice. This was confirmed by the finding of correct catheter projection with respect to anatomical landmarks (coronary sinus ostium experiments) and consistent electroanatomic maps of complex arrhythmias.

To further diminish alignment errors of different borders and enhance the anatomic accuracy of the mapping process, we are investigating a periprocedural correction of the 3D model to angiographic images. Such an approach for periprocedural fine-tuning of 3D models is desirable because changes in atrial dimensions might occur owing to (1) differences in heart rhythm and volume-loading conditions between MRI imaging and ablation procedure and (2) pressure exerted from sometimes highly curved catheters on the endocardial wall. The projection of the coronary sinus catheter (<5 mm) outside the coronary sinus ostium of the 3D model in 4 of 11 patients might have been caused in part by the latter effect. The quantification of these changes and the need for periprocedural correction in different clinical settings will be the subject of future investigations. Acquisition of MRI images and angiographic images was triggered to the same end-systolic interval to avoid synchronicity mismatch. This not only allows imaging during atrial diastole but also causes less error when changes in heart rate occur between MRI acquisition and the ablation procedure, because systolic duration is less influenced by heart rate than diastolic duration.

The current delay in completion of the activation map (30 ± 6 minutes) did not result in a significant increase in total procedure times, although more patients are needed for further evaluation of procedure and fluoroscopy times, number of radiofrequency applications required, and clinical outcome within different patient groups. Moreover, by automating the marking of catheter positions on the fluoroscopic images and the determination of activation times, the time required to complete the activation map will be shortened so that this electroanatomic approach can be used efficiently in clinical practice without the need for specialized catheters or hardware. The lower cost of this software and fluoroscopy-based approach may bring the benefits of electroanatomic mapping to a broader array of electrophysiology laboratories and to more patients.

**Limitations**

In its present form, our new concept for electroanatomic mapping has some shortcomings compared with available electroanatomic mapping systems. Currently, real-time tracking of the position of the ablation/mapping catheter is not possible. However, the ability to indicate an electrode in biplane fluoroscopic images in both the right and left anterior oblique views by simply clicking on it allows for its localization on the 3D model (in an epicardial or endocardial...
view), a maneuver that only requires a few seconds. This is a limitation compared with other merge technologies, but we think that it might suffice in most procedures. In the future, we intend to combine real-time catheter tracking within the 3D angiography framework. MRI acquisition was performed at least 1 day before the ablation procedure, with possibly different heart rate, rhythm, and volume-loading conditions. The effect of these factors on model accuracy must be determined in future work. The accuracy of the MRI-based 3D models was assessed by calculating the alignment error between heart boundaries of the 3D model and angiography images. A more direct evaluation of model accuracy is not possible. In our opinion, this is the most realistic method to evaluate model accuracy in vivo, despite the absence of a “gold standard.” Measurements on a phantom, although not subject to cardiac motion, showed the high accuracy of our 3D angiography merge process.

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References
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