Error Analysis of Registering of Anatomical and Functional Cardiac Data Using External Markers

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Abstract

In this work the relative strengths of the different error sources in a skin marker based registration method for functional magnetocardiography (MCG) data and anatomical magnetic resonance (MR) images of the heart were evaluated. The registration method is needed a) to transform individual torso model, obtained from MR thorax images for inverse problem computations made in the coordinate systems of the bioelectromagnetic measurement device and b) to transform MCG solution of functional information of the cardiac electric excitation to anatomy of MR images. The objective was to analyze the most severe error sources in the registration method, and to reduce their magnitude if possible. Measurements were made with a phantom and on a volunteer. The sum of all registration error components was 6 mm. No specific error sources dominated and their contribution to the total error was approximately equal.

1 Introduction

Interpretation and comparison of anatomical and functional data from different medical imaging modalities can be accomplished by registration. Registration methods based on skin markers are widely applied because they allow match any imaging modalities in which the positions of markers can be accurately defined. The main objective of this study was to define the relative strengths of the different error sources in a skin marker based registration method for functional magnetocardiography (MCG) data and anatomical magnetic resonance (MR) images of the heart.

2 Materials and methods

2.1 Registration protocol

2.1.1 MCG recordings

MCG studies were performed in the BioMag laboratory of Helsinki University Central Hospital (HUCH). Multichannel MCG signals were recorded in a magnetically shielded room using a 99-channel cardiomagnetometer (4-D NeuroImaging, Helsinki, Finland) [1]. The position of the MCG recording system with respect to the patient was determined by attaching three marker coils (magnetic dipoles) to the skin. The magnetic fields produced by the coils were then used to calculate the sensor locations relative to the marker coils.

2.1.2 MCG markers

A set of nine external marker positions, here referred as MCG markers, was selected to registrate the MCG sensor system to MR images. The locations of the MCG markers were defined by attaching a cross-shaped object consisting of two silicone strips of rubber on the skin (see **Fig 1**). The three marker coils were used to define the MCG sensor coordinates in respect to the MCG markers.

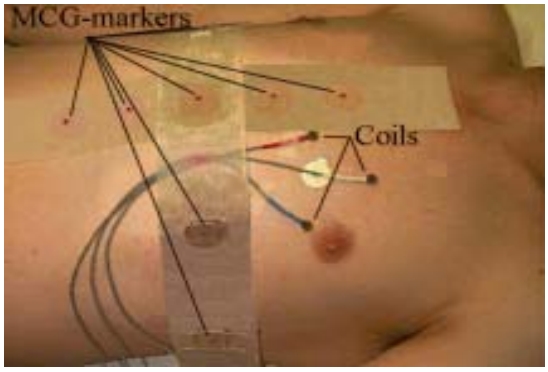

Fig. 1 Placements of the nine MCG markers and three marker coils on the chest in a typical patient study.

The separation between neighboring MCG markers was 5 cm in the head-feet direction and 10 cm in the left-right direction. The locations of the MCG markers and the marker coils were defined with a 3-D digitization system (3SPACE ISOTRAK II, Polhemus Inc., Colchester, VT, USA). The digitized MCG marker positions were stamped with non-toxic ink, visible only in ultraviolet light.

2.1.3 MRI markers

The nine MRI markers were constructed from two perpendicular tubes filled with 1 mmol/l $MnCl_2$ fluid, inserted inside a piece of plastic of 4.0 x 4.0 x 0.7 cm. Prior to MR imaging, the nine MRI markers were placed on the stamped positions on the skin. The cross-shaped figure of a marker was well visible in the MR images.

2.1.4 Registration

The MRI markers were first located manually from MR images, using a dedicated software. The nine marker coordinate sets (x, y, z) in the MCG and MRI coordinate systems, respectively, were registered using a non-iterative least-squares method [2]. Only rigid transformations including global rotations and translations were considered.

Thus far, our registration protocol has been applied to more than 50 patient studies [3]. The root mean square (RMS) error of the nine registered markers was about 6 mm; ranging from three to more than ten millimeters. The registration error was relatively high and should be reduced, because MCG localization accuracy of < 1 cm are desired for clinical purposes.

2.2 Separation of various error sources

A comprehensive study of all error sources in thorax registration would be laborious. In this work, the error sources were divided into five studies: A) the reproducibility of the 3-D localization using the digitization pen, B) the error in alignment of the patient, C) the error arising from repositioning of the MRI markers, D) the effect of different shapes in the measurement beds, and E) the localization error of the MRI markers from the images. Also F) target registration error (TRE) was defined in phantom studies by measuring error of twelve markers in thorax area and four markers in cardiac area which were not used to define registration parameters. Several other error sources can be found, especially from MR imaging, but the selected five studies cover the most significant registration errors in our MCG studies.

An independent measurement of separate error sources is very difficult. For example, the error of the reproducibility of the 3-D localization (study A) can not be avoided during the other studies. Therefore, we chose to quantify the difference between the RMS errors computed for various point sets and not to evaluate the absolute value of each error type individually. Consequently the effects of error sources are not cumulative in a scalar sense, because the errors are vector quantities in 3-D. The main objective of this study is to define the relative strengths of the error sources, instead of absolute magnitudes, in order to reduce the impact of the most severe ones. Estimating the differences of RMS errors gives a measure of their relative strengths.

2.3 Protocol for measuring the error sources

Various error sources were studied using a phantom and a volunteer. The phantom was constructed of wood. Its shape and size were defined from thorax boundaries extracted from MR images of a volunteer. Digitizations were carried out by two persons at the Biomag-laboratory. The MR images were acquired at the Department of Radiology (HUCH). The phantom and the volunteer were positioned supine into a 1.5 T Siemens Magnetom Vision imager (Siemens, Erlangen, Germany). Imaging was performed with the surface coil as a receiver and using a T1 weighted gradient echo sequence. A series of 32 contiguous transaxial images was acquired. Each slice was 10 mm thick. The matrix was 256 x 256 pixels with a 1.76 mm^2 pixel size.

2.4 Error studies

2.4.1 Study A

The reproducibility of the 3-D localization using the digitization pen was first studied. The MCG markers, a cross-shaped rubber strips including nine markers, were placed on the top of the supine positioned phantom or the volunteer. The MCG markers were digitized four times and the average set was calculated. The RMS error was calculated as an mean value of the RMS errors between digitized sets and the average set. Next the marker positions were stamped to the surface of the phantom or the volunteers skin. Also the stamped positions were digitized four times and the RMS error was calculated similarly than for the MCG marker digitizations. Global measurement procedure was repeated three times and the mean value of the three computed RMS errors were calculated.

2.4.2 Study B

The error caused by the alignment of the volunteer was studied next. The stamped positions from the skin of the supine positioned volunteer were digitized four times and the average set was calculated. Thereafter the volunteer arose from bed and get back into supine position for repeating the digitization procedure. Average sets were registered and the RMS error between them calculated. The global measurement procedure was repeated three times and the mean RMS error was calculated. This mean RMS error also included the error of reproducibility of the 3-D localization (study A) and it was subtracted from the computed error to get an estimate of the RMS error caused by the alignment of the volunteer.

2.4.3 Study C

The error of repositioning the MRI markers to the stamped positions of the surface of the phantom or the skin of the volunteer was studied next. The MRI markers were attached to the surface of the phantom or the skin of the volunteer and markers were digitized two times and the average set was calculated. Then the MRI markers were detached and reattached, and the measurement was repeated. The average sets were registered and the RMS error between them calculated. The global measurement procedure was repeated three times and the mean RMS error was calculated. The computed RMS error contains also the error of reproducibility of the 3-D localization using the digitization pen (study A) and it was subtracted from the computed error to get an estimate of the inaccuracy in repositioning of the MRI markers.

2.4.4 Study D

The shapes of the measurement beds were different; the MRI bed was concave while the MCG bed was flat. Two wooden wedge objects with the shape corresponding to the MRI bed were placed under the sides of the volunteer lying on a flat bed. The stamped positions from the skin of the volunteer were digitized two times and and the average set was calculated. Thereafter the volunteer arose from bed, wooden wedge objects were removed and volunteer get back into supine position for repeating the digitization procedure. The average sets were registered and the RMS error between them calculated. The global measurement procedure was repeated four times and the mean RMS error was calculated. The error of reproducibility of the 3-D localization (study A) and the error caused by the alignment (study B) was subtracted from the computed error to get an estimate of the error caused by wooden wedge objects.

2.4.5 Study E

The inaccuracy of the 3-D localization of the MRI markers from the MR images was studied in phantom experiment. The center of the cross from MRI marker was well visible in the MR images. The detection was done using two versions of an interactive software allowing a) only orthogonal slices or b) orthogonal and oblique slices. The point set located from MR images was registered with the point set digitized from MCG markers and the RMS error between them calculated. The errors due to the reproducibility of the 3-D localization using the digitization pen (study A) and the error arising from repositioning of the MRI markers (study C) were subtracted from the computed error. This measure gave an estimate of the increase of the error due to the 3-D localization of the markers from the images. In addition, this error contained also other error sources due to MR imaging itself.

2.4.6 Study F

Target registration error (TRE) was defined with phantom experiment in thorax and in heart area. A set of nine MCG markers and eight other markers, six markers in thorax area and two in heart area, were digitized from the phantom by using digitization pen. MR markers were attached to the same placements and then imaged in MR scanner and located from the images. The ditized nine MCG markers and corresponding set of MR markers were registered and same registration parameters were used for the other eight markers. The TRE was then calculated as mean RMS error of six thorax area markers and respectively for two heart area markers. The global measurement procedure was repeated two times.

3 Results

3.1 Study A

With *phantom* the mean RMS error of the 3-D localization from MCG markers was 0.5 mm and from the surface of phantom 0.6 mm.
With *volunteer* the mean RMS error of the 3-D localization from MCG markers was 0.9 mm and from the surface of skin 1.2 mm. The average value for the 3-D localization error for the volunteer was 1.1 mm.
The mean RMS error measure for the phantom demonstrated errors arising from different measurement events and from different users of the digitization pen. The error of the measurement device itself is also included in the error. The RMS error was higher with the volunteer than with the phantom because of the errors from the breathing and the elasticity of the skin.

3.2 Study B

The RMS error due to different alignments of the volunteer was 1.3 mm, ranging from 0.7 mm to 2.1 mm. The values were relatively low, because: 1) the markers in our protocol were attached in positions which not sensitive to alignment errors, and 2) the error was not cumulative in scalar sense, as mentioned above. The former reason was demonstrated by attaching 9 extra markers on other regions of the thorax, e.g. the shoulders. This increased the RMS error from 1.3 mm to 2.8 mm for the total of 18 markers.

3.3 Study C

For the volunteer, the RMS error of repositioning of the MRI markers was 0.2 mm, ranging from 0.1 mm to 0.5 mm. With the phantom, the error increased compared to volunteer result and was 1.9 mm. A probable reason for this was the large size of the markers (4 x 4 cm) which makes their attachment on the curved surface of the wooden phantom difficult.

3.4 Study D

The increase of the RMS error because of the wooden wedge objects was 1.9 mm, ranging from 0.8 mm to 2.9 mm. However, the increase of the RMS error was slightly overestimated because the difference in softness of mattress in measurement beds causes also part of the error but it was not considered separately in our measurements. If the locations of the markers were considered separately, lateral markers were clearly lifted up relative to the markers on the sternum.

3.5 Study E

The RMS error without oblique slices was 1.5 mm and with oblique slices, the error was 0.5 mm lower.
A sum of all error components for the volunteer was 6.0 mm as summarized in **Table 1**. The error values corresponded well to the average RMS error of about 6 mm in our patient studies. With oblique slices the sum of all error components was 5.5 mm. The static RMS accuracy of the digitization system, the error that could be expected from a specific point to be digitized, was specified by the manufacturer to be in normal mode 2.5 mm over a 10.16 cm - 71.12 cm motion box [4]. In our measurements we did use Polhemus equipment in quiet mode-state which utilizes data averaging to increase resolution by up to factor of three over normal mode.

Error source	**Human**	**Phantom**
Study A. The reproducibility of the 3-D localization	1.1 mm	0.6 mm
Study B. The error in alignment of the patient	1.3 mm	-
Study C. The error from repositioning of the MRI markers	0.2 mm	1.9 mm
Study D. The effect of different shapes in the measurement beds	1.9 mm	-
Study E. The localization of the MRI markers from the images	1.5 mm	1.5 mm
Sum of all errors	**6.0 mm**	**4.0 mm**

Table 1. Summary of the registration error sources.

3.6 Study F

The TRE error for totally twelve thorax area markers was 5.9 mm and 5.3 mm for four heart area markers. TRE was slightly smaller in heart than in whole thorax area. This was propably because heart area markers are in the central area of the nine MCG and MRI markers which were used to define registration parameters. TRE error of the phantom experiment was slightly bigger than defined RMS error of the markers used to define registration parameters (4.0 mm).

4 Discussion

A drastic reduction of the total RMS error was not easy to accomplish because any error sources appeared to dominate. The effect of two error components was, however, fairly easy to compensate: 1) the effect of the measurement bed shape (1.9 mm) was reduced by making a firm support, which copied the shape of the MRI bed. This support could be placed under a patient before digitization of the MCG markers. 2) The use of oblique slices appeared to be superior to orthogonal slices.
Breath holding was used during the MR imaging but not during the 3-D localization of the MCG markers. The breath holding also during the 3-D localizations would reduce the registration error between digitized point sets from MCG markers and reference point sets from MR images. The digitization system itself might have also error while defining digitization points. The effect of the digitization system error will be visible in the localization error of the MRI markers from the images (study E), since point set digitized from MR images does not include it.
TRE error in general is usually bigger than RMS error of the markers that are useded to define registration parameters. Still, about 1 mm difference of the heart area TRE error and RMS error of the markers used to define registration parameters (phantom experiment), gives an estimate that heart area registration error of our system is not very big.
In the registration technique that is used in our patient studies there exists several error sources. The five error sources mentioned above explain well the total RMS error of these studies. The results of our analysis are also applicable to other studies requiring registration between two or more imaging facilities on the basis of external markers.

5 Literature

[1] Montonen, J.; Ahonen, A.; Hämäläinen, M; Ilmoniemi, R.; Laine, P.; Nenonen, J.; Paavola, M.; Simelius, K.; Simola, J.; Katila, T.: Magnetocardiographic functional imaging studies in BioMag Laboratory, in Biomag96, Proc. Tenth Internat. Conf. on Biomagnetism, Aine, C. et al., Eds. New York, Springer, 1996, pp. 494 -497

[2] Arun, K. S; Huang, T. S; Blostein, S. D.: Least-squares fitting of two 3-D point sets, IEEE Trans. Pattern Anal. Machine Intell. 9(1987), H.5, S. 698-700

[3] Pesola, K.; Lötjönen, J.; Nenonen, J.; Magnin, I.E.; Lauerma, K.; Fenici, R.; Katila, T.: The effect of geometry and topology differences in boundary element models on magnetocardiographic localization accuracy, IEEE Trans. Biomedical Eng., 47 (2000), P. 9, S. 1237-1247

[4] 3SPACE ISOTRAK II User's Manual, Polhemus Inc., Colchester, Vermont, U.S.A., 1993