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A realistic volume conductor model of the neonatal head: methods, challenges and applications

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***Abstract*— Developing a realistic volume conductor head model is an important step towards a non-invasive investigation of neuro-electrical activity in the brain. For adults, different volume conductor head models have been designed and successfully used for electroencephalography (EEG) source analysis. However, creating appropriate neonatal volume conductor head model for EEG source analysis is a challenging task mainly due to insufficient knowledge of head tissue conductivities and complex anatomy of the developing newborn brain. In this work, we present a pipeline for modeling a realistic volume conductor model of the neonatal head, where we address the modeling challenges and propose our solutions. We also discuss the use of our realistic volume conductor head model for neonatal EEG source analysis.**

I. Introduction

The rapid development of non-invasive brain imaging techniques has opened new horizons in the study of brain structure and function. In patients with neurological disorders, magnetic resonance imaging (MRI) and electroencephalography (EEG) are the most important diagnostic tools, because they can capture features of brain anatomy and function, and map neuronal dynamics and degenerative processes. Since individual analysis of EEG and MRI have its strengths and weaknesses, there is an increasing demand for multi-modal EEG-MRI brain analysis (such as EEG source analysis) to synthesize the strengths inherent in each technique [1]. Nowadays, EEG source analysis has an important role in the management and diagnosis of various brain diseases such as epilepsy, birth asphyxia, strokes and tumors.

Developing a realistic volume conductor head model is a necessary and the most critical step towards EEG source analysis. Firstly, this is because the active anatomical zones of the brain are estimated using a volume conductor head model and EEG signals measured on the scalp, see Fig. 1. Secondly, the accuracy of source estimates (e.g. current dipoles) highly depends on the selection of a volume conductor head model. In EEG source analysis, both spherical and realistic volume conductor head models have been successfully used. A spherical head model is typically created as a set of concentric spheres representing different conductive layers like scalp, skull and brain, and is used to simplify and speed up source calculations. However, it had been showed that using a spherical head model instead of a realistic head model results in a dipole location error [2], [3]. This is particularly important when studied sources are located in the occipital or temporal lobes [4]. Therefore, it is necessary to develop a realistic volume conductor head model to minimize the dipole location error.

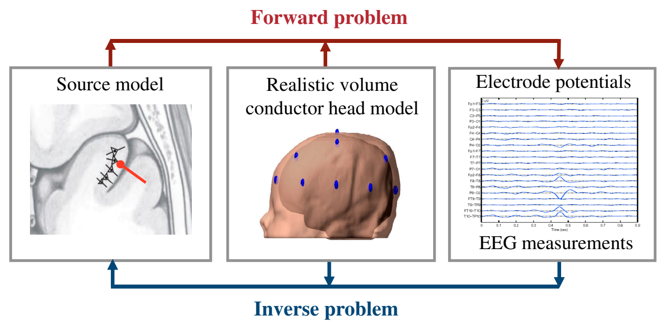

Fig. 1. An outline of the EEG source localization method. Using a volume conductor head model is an important step for solving both forward and inverse problems. Forward problem calculates the electrode potentials given the source and the head model. Inverse problem quantitatively estimates the source parameters in the head model for a given set of EEG measurements.

Nowadays EEG source analysis using a realistic volume conductor head model is routinely performed in adults (e.g. in patients with epilepsy) [5], but it is still in early experimental phase in neonates (e.g. in babies with asphyxia and hypoxic-ischemic encephalopathy (HIE)) [6], [7]. This is because EEG source analysis in neonates is a more complex task than in adults, where one of the technical difficulties is modeling a volume conductor head model. The most critical tasks in neonatal head modeling are: reconstructing a realistic conductive layers of the head (such as scalp, skull and brain) and assigning appropriate conductivities to the head layers. The first task is challenging due to the complex anatomy of the newborn brain (e.g. not developed skull, existence of fontaneles, brain myelination) and very often poor MRI quality. The second task is challenging due to insufficient knowledge of head tissue conductivities.

In this paper we present a pipeline for modeling a realistic volume conductor model of the neonatal head. We discuss the challenges in head modeling and propose our solutions. Also, we discuss the use of our realistic volume conductor head model for neonatal EEG source analysis. The paper is organized as follows: In Section II we present a modeling steps necessary for creating a realistic head model of the neonatal head. Modeling results are presented and discussed in Section III. Finally, we conclude this paper in Section IV.

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II. Methods

Creating a realistic volume conductor head model involves three equally important steps: (1) MRI segmentation of the head in different head structures such as scalp, skull and brain tissue; (2) selection of appropriate conductivities for each head structure; and (3) placement of the EEG electrodes on the scalp surface.

A. MRI segmentation

MRI is an excellent technique for imaging neonates because of the absence of ionizing radiation and the superior contrast of soft tissues and resolution compared with sonography. However, neonatal brain imaging presents set of technical and practical challenges in comparison to adult brain imaging, which make the head tissue segmentation in newborns more complex than in adults. The most obvious technical difficulty is in obtaining motion-artifact-free images, due to the presence of baby's motion during an MRI acquisition, which decreases the image quality. The two main practical difficulties are due to anatomical and developmental issues. Neonates are much smaller than adults and the differences between the appearance of neonatal and mature adult brains in MRI are significant (lower contrast-to-noise ratio, lower MRI resolution, inverted contrast of white matter (WM) and gray matter (GM)).

Due to significant anatomical differences between the adult and neonatal brains, MRI segmentation techniques developed for adults are not applicable for neonates. Therefore, the MRI segmentation of the neonatal brain has become a research focus in recent years [8], [9]. The existing methods are mainly focused on the problem of neonatal brain tissue segmentation (e.g. brain cortex, myelinated and non-myelinated WM), and are not developed to segment head structures such as skin, skull, and cranial cavity. Also, these methods rely on probabilistic atlases, where an atlas-based segmentation is often difficult and prone to errors, particularly in patients with brain lesions or with a brain-anatomy that significantly differs from the atlas template. For instance, in the EEG source localization studies where personalized volume conductor models are required (when direct comparison between localized seizures and lesions seen on MRI is needed) the atlas-based segmentation will give inaccurate results.

Thus, to segment different head structures we propose an atlas-free segmentation algorithm based on multi-modal T1-W and T2-W MRI. Before segmentation, several preprocessing steps are necessary. First, we use cubic spline interpolation to interpolate low-resolution MRI scans and obtain appropriate inter-slice resolution. Second, we perform bias field (intensity inhomogeneity) correction [10] and multimodal T1-W and T2-W MRI registration [11]. After preprocessing, we segment MR images into four head structures: scalp, skull CSF, and brain tissue. We use the brain extraction algorithm [12] to separate brain from non-brain tissue, multi-modal fuzzy c-means clustering [13] to segment scalp, brain tissue and CSF, and mathematical morphology to reconstruct skull.

The neonatal skull is one of the most challenging head structures to segment because it is not visible on MRI scans. Also, computer tomography (CT) scans, which are commonly used to reconstruct the skull in adults, are not available for neonates. Thus, to create the skull, we used the remaining voxels between the previously segmented brain and scalp, which resulted in a skull layer of thickness $1.4-2.8$ mm. Also, we reconstructed the anterior fontanelle as a thin skull layer (0.7 mm) at the top of the head. Finally, all segmented head structures are used to generate a cubic grid of the head model.

B. Head conductivities selection

Different head structures (brain, CSF, scalp and skull) have different conductivities of which the inappropriate setup may influence the lead fields of forward problems and the solutions of the inverse problems, see Fig. 1. Since neonatal head conductivities have never been directly measured, we estimate the head conductivity values based on existing studies for adults and small animals, see Table I.

TABLE I

Conductivity values of human head tissues

Component	Adult(S/m)	Neonate(S/m)
Scalp	0.43	0.43
Skull	$0.0067 - 0.015$	$0.033 - 0.2$
CSF	1.79	1.79
Brain	$0.2 - 0.48$	0.33

The conductivities of the neonatal scalp and CSF are assumed to be the same as in adults with the values 0.43 and 1.79 S/m, respectively [14], [15]. The conductivity of the neonatal brain is set to 0.33 [6]. The main challenge was to estimate the conductivity of the neonatal skull. Since the neonatal skull has incomplete bone development (contains the soft bone called cartilage), it is reasonable to assume that the conductivity of the neonatal skull is higher than in adults. In the study of Murray et al. [16] it was estimated that the neonatal skull conductivity should be between 0.2 and 0.033 S/m. This ranges from the conductivity of most soft tissues ($\sim$ 0.2 S/m), to five times the conductivity of the adult skull 0.0067 S/m. For our head model the skull conductivity is set to 0.033 S/m.

C. Electrode placement

To perform EEG source localization studies with a realistic volume conductor head model, it is necessary to project the exact electrode positions on the scalp surface of the head model. Typically this is solved by using a 3D digitizer, which transforms coordinates of electrode locations to the MRI coordinates. However, 3D digitizers are not available in most EEG laboratories. Thus, an alternative solution is to place the electrodes on the scalp surface using the internationally standardized 10-20 system and two anatomical landmarks: the nasion point and the inion point. In this work, both nasion and inion points are manually annotated on the segmented scalp surface and are used as anatomical

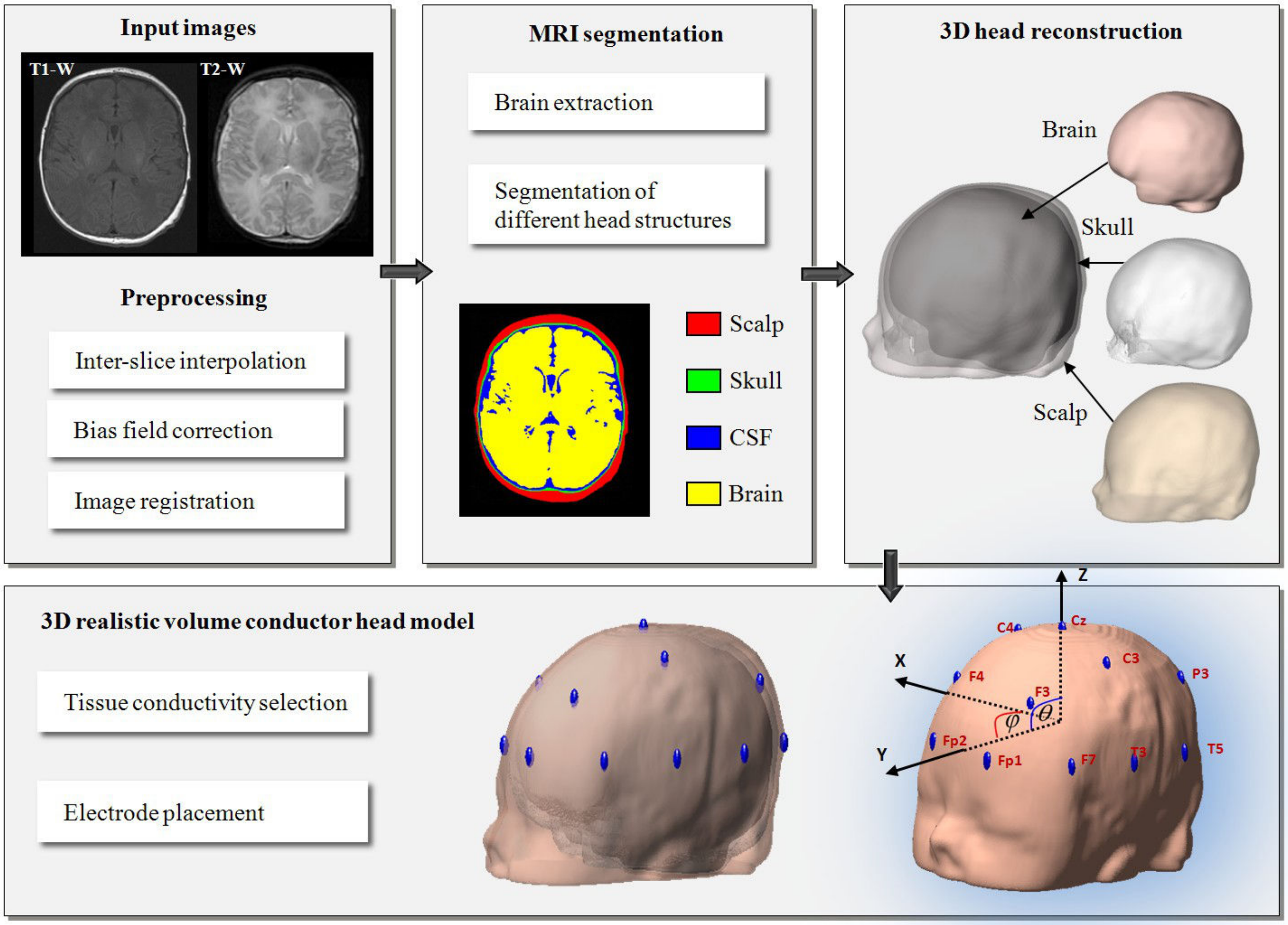

Fig. 2. An outline of the proposed realistic head modeling method. We also show the segmentation and 3D reconstruction results of a realistic volume conductor head model with 17 EEG electrodes.

markers for electrode placement. The electrode coordinates are described using two parameters: the latitude φ (contra-clockwise angle with the x-axis in the horizontal $x-y$ plane where $0^o \leq \varphi < 360^o$) and the azimuth θ (angle with the vertical z-axis where $0^o \leq \theta < 180^o$). All steps for the realistic volume conductor head modeling are summarized in Fig. 2.

III. Results and Discussion

The MRI data in this study were recorded at the Sophia Children's Hospital (Erasmus Medical Center, Rotterdam, the Netherlands), on a newborn preterm with gestational age of 40 weeks subject to asphyxia. Both T1-W and T2-W MRI were acquired on a Siemens 1,5T MRI scanner (256×256×20 voxel matrix, with a resolution of 0.7mm×0.7mm×4.2mm).

3D reconstruction results of a neonatal realistic volume conductor head model with 17 EEG electrodes are shown in Fig. 2. Our head model consists of four structures: scalp, skull, CSF, and brain tissue. The segmentation results were checked and approved by the expert physician. The proposed head modeling method is further successfully used in EEG source localization studies, where we explored the relationship of neonatal seizures to brain lesions visible on MRI [7], see Fig. 3. In the same study we also explored the influence of volume conductor head model errors on dipole locations.

In comparison to a boundary element method (BEM), which is commonly used to create the three-layer head model consisting of scalp, skull, and brain tissue [6], our method creates more realistic head model containing the ventricular system and CSF as an additional conductive layer. This is important because the electrode potentials are dependent on the total electrical field generated in the head caused by a current dipole. Also, the ventricles and CSF are used to better constrain the dipole location in EEG source analysis (e.g. current dipoles cannot be placed in the ventricles or CSF).

One of the biggest challenges in neonatal head modeling is to estimate the appropriate conductivity of the neonatal skull using available studies for adults and small animals. In the adult studies it has been reported that the conductivity of the adult skull ranges from 0.0067 to 0.015 S/m. Also, it has been suggested that the human brain-to-skull conductivity ratio is 15 [17] instead of the commonly used value of 80. Furthermore, in the recent animal study of Pant et al.

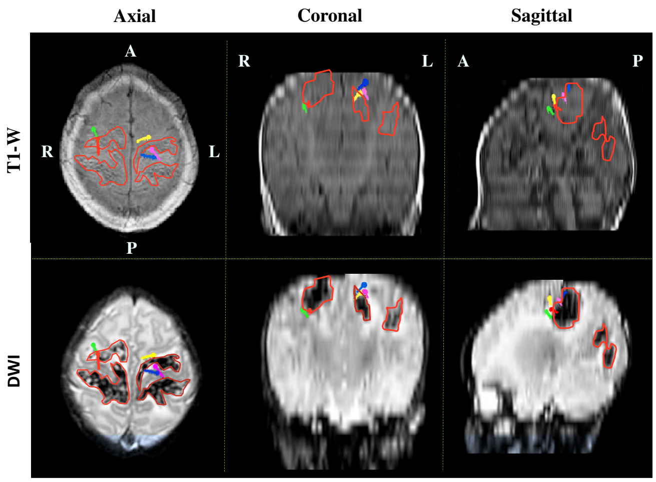

Fig. 3. Dipole localizations results of patient with bilateral white matter hemorrhages. 3D dipole fit results are plotted onto the patients T1-W and diffusion weighted images (DWI) using projections on three planes: axial, coronal and sagittal. Color of the dipole represents different EEG events (seizures) and brain lesions are delineated with a red color.

[18], it has been found that the average conductivity of the preterm/neonatal piglet skull (with the skull thickness 1.3 mm) is in the range 0.025-0.035 S/m at 1 kHz. They also found that the skull conductivity increased linearly with the skull thickness. For our neonatal head model we set the skull conductivity value to the similar value of 0.033 S/m, because the brain-to-skull ratio of 10 seems reasonable in comparison to adult studies [17]. Note that the piglet brain is a well-accepted preclinical model for neurodevelopmental research in humans because of its anatomic and physiologic similarities to humans, like similar patterns in brain development and growth. However, the similarity between the human neonatal skull and the piglet preterm/neonatal skull is still controversial.

For the electrode placement we used the standard 10-20 system to manually place the electrodes on the scalp in a relation to the anatomical markers. Although 3D digitized electrode positions describe the electrodes on the scalp more precisely than the standard 10-20 system, 3D digitizers are often not available in neonatal studies. In adult studies it has been shown that on average the digitized electrode locations deviate from the standard 10-20 positions by about 4^o. This electrode misallocation has not yet been measured in neonates, but it might be very similar if the fine adjustments based on the head shape and anatomical markers are made during electrode positioning. Depending on the application, this error can be tolerable or critical. For example, in studies where only relationship of seizures to brain lesions is studied [7] this error is tolerable, but for surgical planing in epilepsy, very high precision source localization is desired and digitization is required. Next to the electrode placement, the higher source localization accuracy can be obtained using higher electrode density on the scalp (e.g. 19, 32, 64 or 128 electrodes).

IV. Conclusion

In this paper we propose a method for creating a realistic volume conductor model of the neonatal head. Despite many technical difficulties and using only 17 electrodes, we successfully used our head modeling method in EEG source localization studies. The experimental results showed that EEG source imaging is feasible in newborn infants and with further improvements and validations it can be a useful diagnostic tool as it is now in adults.

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