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Motion compensation for brain PET imaging using wireless MR active markers in simultaneous PET–MR: Phantom and non-human primate studies

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ABSTRACT

Brain PET scanning plays an important role in the diagnosis, prognostication and monitoring of many brain diseases. Motion artifacts from head motion are one of the major hurdles in brain PET. In this work, we propose to use wireless MR active markers to track head motion in real time during a simultaneous PET–MR brain scan and incorporate the motion measured by the markers in the listmode PET reconstruction.

Several wireless MR active markers and a dedicated fast MR tracking pulse sequence module were built. Data were acquired on an ACR Flangeless PET phantom with multiple spheres and a non-human primate with and without motion. Motions of the phantom and monkey's head were measured with the wireless markers using a dedicated MR tracking sequence module. The motion PET data were reconstructed using list-mode reconstruction with and without motion correction. Static reference was used as gold standard for quantitative analysis.

The motion artifacts, which were prominent on the images without motion correction, were eliminated by the wireless marker based motion correction in both the phantom and monkey experiments. Quantitative analysis was performed on the phantom motion data from 24 independent noise realizations. The reduction of bias of sphere-to-background PET contrast by active marker based motion correction ranges from 26% to 64% and 17% to 25% for hot (i.e., radioactive) and cold (i.e., non-radioactive) spheres, respectively. The motion correction improved the channelized Hotelling observer signal-to-noise ratio of the spheres by 1.2 to 6.9 depending on their locations and sizes.

The proposed wireless MR active marker based motion correction technique removes the motion artifacts in the reconstructed PET images and yields accurate quantitative values.

Introduction

Simultaneous Positron Emission Tomography (PET) and Magnetic Resonance (MR) imaging (PET–MR) is a novel and promising hybrid modality generating substantial interest in recent years. Individually, these two imaging modalities have complementary advantages and disadvantages in terms of sensitivity, molecular specificity, and spatial and temporal resolution. Their strengths may be integrated, providing new insights in neuroimaging through the concurrent acquisition of synergistic anatomical, functional and biochemical information (Mandeville et al., 2013; Sander et al., 2013).

Brain PET scanning plays an important role in the diagnosis, prognostication (Silverman et al., 2001), and monitoring of many brain diseases including dementia, which brings overwhelming burden to patients, families, and society (Mehta and Thomas, 2012). PET also plays an important role in neuroimaging of animals including non-human primates for radiotracer development (Gunn et al., 2011; Parker et al., 2012) and in understanding of brain diseases (Eberling et al., 2003; Howell and Murnane, 2011).

Motion artifacts from head motion are one of the major hurdles in brain PET. Dynamic brain PET studies can last more than 1 h; voluntary and involuntary head motions are almost inevitable. This is particularly true in the elderly or patients with dementia or movement disorders. Head restraints are often used to reduce head motion during the acquisition, but they cause discomfort for patients, and head movements still cannot be completely eliminated (Bloomfield et al., 2003). Furthermore,

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perturbation of neurochemical processes can be introduced by the stress caused by restraint in animal studies (Arnsten, 2000; Fueger et al., 2006). Anesthesia is also often used to keep animals still during acquisition; studies showed that anesthesia can also perturb the neurological process under study (Angel et al., 1999; Hendrich et al., 2001; Lindauer et al., 1993; Nakao et al., 2001; Patel et al., 2008; Tsukada et al., 2000; Tsukada et al., 1999).

Many approaches have been explored in the effort to correct motion artifacts. Depending on whether the motion is estimated from the acquired PET data or by other instrumentation, the approaches can be divided into two groups: auto-correction and assisted-correction. For the auto-correction techniques, the measured PET data are divided into temporal frames, and the motion is then estimated between temporal frames from the PET data. The estimated motion field can then be used to transform the reconstructed images (Friston et al., 1995; Tellmann et al., 2006; Woods et al., 1992) or the sinograms (Hutton et al., 2002; Kyme et al., 2003) of each temporal frame to a reference frame. The accuracy of motion estimation using this approach is limited by the noise of PET images, which increases as the data set is divided into temporal frames for a dynamic image sequence. Moreover, the fact that the motion estimation relies on the generation of images or sinograms limits its temporal resolution; this method is not suitable when the activity distribution is fast changing or the object is fast moving. The reconstruction algorithms of the assisted-correction approaches are similar to auto-correction techniques except that the motion information is instead measured using an instrument other than the PET camera, such as video/infrared cameras (Bloomfield et al., 2003; Goldstein et al., 1997; Picard and Thompson, 1997), and approaches with structured light (Olesen et al., 2012, 2013). Similar approaches have also been applied to motion correction in MRI (Schulz et al., 2012; Zaitsev et al., 2006). One advantage of these optical motion tracking approaches is that they are independent of the PET–MR acquisition, so that no changes to MR pulse sequence are required, in contrast to MR navigator-based methods. Another advantage is that the optical methods, in principle, are capable of achieving high frame rate. Some of these approaches monitor the motions of the reflectors attached to the subject's head; some observe a portion of the subject's face. But they all require an unobstructed view from the cameras to the reflectors or the subject's face. This is challenging for PET–MR head scan because the view from outside of the scanner is blocked by the head coil, especially for the modern head coils with large number of channels. There are RF contamination and MR compatibility issues associated with installing cameras inside of the scanner. Moreover, these optical systems require complicated calibrations.

Conventional MR navigator methods (Ehman and Felmlee, 1989; Wang et al., 1996) can be used to track motion with temporal resolution less than 20 ms. However, such methods cannot be used to track head rotation. Catana et al. (2011) used the cloverleaf navigator method (van der Kouwe et al., 2006) to track head motion for PET motion compensation. Although this method can track both translation and rotation, its motion tracking accuracy suffers from off-resonance effects, gradient instabilities, as well as signal contamination from non-rigid motion of the neck. Moreover, the cloverleaf navigator method requires approximately 20 s of motionless data to calibrate. Petibon et al. (2013) used image-based MR motion tracking for non-rigid motion compensation in cardiac PET. However, it generally lacks temporal resolution because of the long scanning time needed for acquiring the entire image volume.

Motion monitoring using wired MR active markers dates back to 1986 (Ackerman et al., 1986). It has been used for prospective motion correction and device tracking in MR imaging (Derbyshire et al., 1998; Dumoulin et al., 1993). Recently, wireless MR active markers gained interest in the MR community for prospective motion correction due to its simpler manufacturing, easier setup and better patient-friendliness (Garnov et al., 2011; Ooi et al., 2013; Sengupta et al., 2013). In this work, we propose to use wireless MR active markers to track head motion in real time during a brain PET–MR scan and incorporate the motion measured by the markers in a modified listmode PET reconstruction algorithm. This makes PET images, which are reconstructed from data acquired on a simultaneous PET–MR scanner, free of motion artifacts. Phantom and non-human primate experiments were conducted to evaluate the proposed methods. Our method is dedicated to simultaneous PET–MR scanners.

Material and methods

Simultaneous PET–MR scanner

All acquisitions in this study were performed on a commercially available whole-body simultaneous PET–MR scanner (Siemens Biograph mMR, Siemens Healthcare, Erlangen, Germany). This scanner consists of an MR-compatible lutetium oxyorthosilicate (LSO) crystal based PET camera inside a 3 Tesla MR scanner, which provides 258 mm axial field of view and 4.4 mm full width at half maximum (FWHM) transverse spatial resolution at 1 cm off the center. The simultaneity and spatial alignment between PET and MR scanners are essential for performing MR-assisted motion correction in brain PET.

Accurate temporal alignment of the PET and MR data was ensured by using simulated electrocardiography (ECG) signals which are recorded in the PET listmode and physiological signal logging on the MR scanner with a modification to the MR pulse sequence [code by Michael Erb, obtained from Siemens Integrated Development Environment for Applications (IDEA) discussion board www.mr-idea.com].

Motion tracking with wireless MR active markers

Active marker hardware

The wireless MR coils work by inductive coupling with the body RF coil of the scanner (and with the separate receiver coil if used). The RF magnetic field of the transmit pulse excites an RF current in the wireless coil, which in turn excites the spins in the water sample. Although the body coil-wireless coil mutual inductance is quite small, the RF field (B_1) produced in the water sample is quite large because of the large filling factor and the high quality factor (Q-factor) of the wireless coil, such that an RF flip angle of only a fraction of a degree (measured with respect to the body coil for the imaged subject, not the tracking sample) excites an intense signal from the wireless coil. Again, mutual inductance between the wireless coil and the receive coil (or body coil) couples the MR signal of the water sample into the receiver, yielding a readily detectable signal.

As shown in Fig. 1, the wireless MR active markers were built by installing a solenoidal wireless MR miniature coil around a spherical microsample cell with a volume of 18 μL (model 529-A, Wilmad-LabGlass, Vineland, NJ, USA). The sphere was filled with a degassed solution of deionized water doped with 1.25 g/L $NiSO_4 \cdot 6H_2O$ and 5 g/L NaCl. The sphere was immersed in liquid nitrogen while the air was pumped out, and the neck was flame-sealed under vacuum. The MR coils were built by winding 4 turns of 14 AWG (American wire gauge; 1.63 mm diameter) bare copper wire into a solenoid about 9.1 mm long by 9.8 mm outside diameter, resulting in an inductance of about 82 nH. The capacitance required to resonate the coils at 123.14 MHz (the exact Larmor frequency of the Siemens "3 T" scanner) is about 20 pF, which was made up with a combination of a fixed ATC (American Technical Ceramics, Huntington Station, NY, USA) nonmagnetic chip capacitor and an adjustable Johanson type 9341 nonmagnetic SMD (Surface Mount Device) trimmer capacitor (Johanson Manufacturing Corporation, Boonton, NJ, USA). Because the coils are not wired to an external circuit, no impedance matching circuitry is needed. The adjustable capacitor allowed the coil to be fine-tuned with the water sample in place by observing the resonance frequency while the coil is probed with a coupling loop connected to a Hewlett–Packard 8753C vector network analyzer displaying reflected power. The typical