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Measurement of brain function in pre-school children using a custom sized whole-head MEG sensor array

Blake W. Johnson[1*], Stephen Crain[1,2], Rosalind Thornton[1,2], Graciela Tesan[1], Melanie Reid[1]

[1]Macquarie Centre for Cognitive Science
Macquarie University
Sydney, Australia

[2]Department of Linguistics
Macquarie University
Sydney, Australia

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*Corresponding author:
Dr Blake W Johnson
Macquarie Centre for Cognitive Science
Macquarie University
Sydney, New South Wales 2109
Australia
Tel +61 2 9850 6879
Fax +61 2 9850 6059
Email bjohnson@maccs.mq.edu.au

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Abstract

Objective: Conventional whole-head MEG systems have fixed sensor arrays designed to accommodate most adult heads. However arrays optimised for adult brain measurements are suboptimal for research with the significantly smaller heads of young children. We wished to measure brain activity in children using a novel whole-head MEG system custom sized to fit the heads of pre-school aged children.

Methods: Auditory evoked fields were measured from seven 4-year-old children in a 64-channel KIT whole-head gradiometer MEG system.

Results: The fit of heads in the MEG helmet dewars, defined as the mean of sensor-to-head center distances, were substantially better for children in the child helmet dewar than in the adult helmet dewar, and were similar to head fits obtained for adults in a conventional adult MEG system. Auditory evoked fields were successfully measured from all seven children and dipole source locations were computed.

Conclusions: These results demonstrate the feasibility of routinely measuring neuromagnetic brain function in healthy, awake pre-school-aged children.

Significance: The advent of child-sized whole-head MEG systems opens new opportunities for the study of cognitive brain development in young children.

Introduction

Imaging of cognitive function in the developing human brain remains an important challenge for cognitive neuroscience in the 21st century. Measurements of brain function in young children are needed to further our understanding of the neural mechanisms for acquisition of language and other important cognitive functions, for the developmental pathways involved in devastating disorders of cognition including autism, and to allow surgeons to safely navigate in the vicinity of cortical regions that support crucial cognitive functions when extricating pathological brain tissues. However, neuroimaging of toddlers and young pre-school aged children poses special challenges for brain research. Ethical and safety considerations may prevent or discourage routine experimentation in healthy children with techniques such as PET neuroimaging, which involves the use of radiopharmaceuticals, or fMRI neuroimaging, which requires very strong magnetic fields.

In contrast, magnetoencephalographic (MEG) neuroimaging is an attractive technique since it measures brain activity in an entirely passive manner; hence the procedure poses no conceivable risks to developing tissues. A further advantage is that the high temporal resolution intrinsic to MEG neuroimaging is well-matched to the dynamics of human cognitive processes, and the spatial resolution is generally considered superior to EEG. Consequently, there is emerging interest in using MEG to study human brain function in normally developing (Fujioka et al., 2006; Parviainen et al., 2006; Rojas et al., 2006; Simos et al., 2005) and clinical pediatric (Bast et al., 2007; Gaetz et al., 2009; Gaetz et al., 2008; Oram Cardy et al., 2008; Widjaja et al., 2008) populations.

A significant barrier to the use of MEG with children derives from the fixed spatial configurations of MEG sensors. The superconducting sensors in modern whole-head systems are immersed in liquid helium in an insulated dewar helmet whose size and shape must satisfy two opposing constraints. The first consideration is that all sensors in the helmet must be positioned as close to the head as possible. Since neuromagnetic fields decay rapidly with distance, sensors that are positioned too far away from cortical tissue will be ineffective in detecting brain signals. The second constraint is the natural variability in the size and shape of human heads. Commercial MEG dewar helmets are designed to fit the majority of heads but are not optimised for heads at the extremes of the size distribution in a population. The majority of toddlers and pre-schoolers fall in one of the problematic tails of this distribution since they have heads that are as much as several cm smaller in diameter than adults. Consequently many or most of the sensors in commercial MEG systems will not be effective in detecting neuromagnetic signals from young children's brains.

There are also formidable practical difficulties in measuring MEG from pre-school aged children. MEG measurements demand that head and body movements must be minimized during sessions which can last several to many minutes, and young children typically have a very limited capacity to curtail their waking movements and to attend to experimental instructions and stimuli.

For these reasons, previous MEG studies of children have almost exclusively used non-helmeted MEG systems that do not sample the entire cortex (e.g. Rojas et al., 2006), or have studied kindergarten and school-aged children (e.g. Bast et al., 2007; Gaetz et al., 2009; Oram Cardy et al., 2008; Parviainen et al., 2006; Simos et al., 2005), pediatric patients measured under anesthesia (e.g. Gaetz et al., 2009) or children in the upper end of the pre-school age range (e.g. Fujioka et al., 2006).

In the present article we report on our experiences in measuring brain responses from a group of healthy and alert four-year-old children, using a novel whole-head MEG system with a helmet dewar custom-sized to fit the heads of preschool-aged children. Our objective was to demonstrate the capabilities of this custom MEG system using a simple auditory evoked field (AEF) paradigm. In particular, we wished to demonstrate that brain responses that have been quite difficult to measure from young children with conventional adult sized MEG systems, can be readily measured using the custom MEG, because it ensures that sensors are easily and effectively positioned in range of the brain and that head movement is well-constrained. These advantages open new opportunities to measure brain function in a developmentally crucial age-range for which there is currently a paucity of data.

Method

Subjects

Seven children (four females) aged between four and five years (mean 4.6 years; min 4.4 years, max 4.8 years) participated in the auditory evoked field experiment. All children were developmentally normal and were fluent in English.

Data on sensor-to-head center distances were obtained from an existing database of thirty-three healthy adults (mean age 29.3) who had participated in several unrelated MEG experiments from 2008-2009. Twelve of these subjects had participated in a separate study (unpublished data) employing acoustic stimuli identical to those presented to children in the present study. AEFs from one representative adult in that study are shown in the present article to provide a basis of comparison between AEFs measured from adults in a conventional adult system and AEFs measured from children in a child sized system.

MEG Systems

Brain responses were measured in two whole-head systems at the KIT-Macquarie Brain Research Laboratory. The adult system (Model PQ1160R-N2, KIT, Kanazawa, Japan) consisted of 160 coaxial first-order gradiometers with a 50 mm baseline (Kado et al., 1999; Uehara et al., 2003). The child system (Model PQ1064R-N2m, KIT, Kanazawa, Japan) system consisted of 64 first-order axial gradiometers with a 50 mm baseline. Sensor labels and positions for the two systems are shown in Figure 1. The sensors in each system are divided into five broad regions: left temporal, right temporal, frontal, central, and occipital.

Figure 1 about here

Child MEG Description and Specifications

The gradiometers of the child MEG had 15.5-mm coil diameter positioned in a glass fiber reinforced plastic (GFRP) cryostat for measurement of the normal component of the magnetic field from human brain (i.e. the direction of the baseline coincides with the direction normal to the human skull). The mean distance between neighboring channels was 38 mm and the distance between sensors and the outer dewar surface was 20 mm. Inner dimensions of the helmet dewar are shown in Figure 2 (top). The dewar helmet was designed to fit a maximum head circumference of 53.4 cm, accommodating >90% of heads of five year old US Caucasian boys and > 95% of five year old US Caucasian girls, (Roche et al., 1987). The helmet fit drops systematically with increases in head circumference at older ages, dropping by 10 years of age to < 50 % of heads for boys and < 75% of heads for girls. For adults, the helmet fits < 5% of males and < 15% of females.

Figure 2 (bottom) shows the relationship between the digitised head surface and MEG sensors for a representative subject (subject C3). The distance between MEG sensors and head surface for this subject was 36.4 mm (mean of central sensors), 38.2 mm (frontal sensors), 30.5 mm (occipital sensors), 38.9 mm (left temporal sensors) and 36.6 mm (right temporal sensors). Occipital sensors are closest to the head surface due to supine positioning. The 10.5 mm gap between the outer surface of the dewar and the occipital head surface is due to the marker coil cap and hair.

The cool-to-warm length of the cryostat was 20 mm. Gradiometer operation in this system is done by a direct offset integration technique, allowing linearization of the superconducting quantum interference device (SQUID) sensor characteristics. The position and direction and sensitivity of each gradiometer is calibrated after the liquid helium is added to the cryostat using a set of room temperature, precisely machined coils placed in a concave part of the cryostat feeding a standard electrical current. The typical intrinsic noise of the gradiometer is about 5 fT/√Hz at white noise frequency region. The output of each sensor was digitized with 16-bit resolution.

Figure 2 about here

Magnetically shielded room

Both MEG systems were contained within a single large magnetically-shielded room (Fujihara Co. Ltd., Tokyo, Japan). The room was constructed of two layers of aluminium frames covered with three layers of permalloy plates and one layer of copper.

Stimuli

The acoustic stimuli were broadband noise stimuli chosen because the frequency content is more naturalistic than pure tones, and because they are known to elicit consistent and measurable differences in hemispheric lateralization of the AEFs (Chait et al., 2004). The stimuli were derived from a psychophysical paradigm described in detail in previous publications (Hautus & Johnson, 2005; Johnson et al., 2003; Johnson et al., 2007). The stimuli were designed digitally using LabView software (Version 8.6, National Instruments, Austin, TX). Two identical broadband Gaussian noise processes of 500 ms duration were generated at a sampling rate of 44.1 kHz. The bandwidth of the two spectrally-flat noise processes was determined by a low-pass Butterworth filter with a corner frequency of 1200 Hz. Stimuli were windowed with a $\cos^2$ function yielding 4 ms rise and fall times.

The stimuli were generated on two channels of a 16-bit converter (Model NI USB 6251, National Instruments, Austin TX). The level of the sounds was adjusted using programmable attenuators (model PA4, Tucker Davis Technologies, Alachua, FL) to yield 70 dB SPL at the eardrum. Stimuli were delivered to listeners using insert earphones (Etymotic Research Inc. Model ER-30, Elk Grove Village, IL). Stimuli were presented binaurally with a random interstimulus interval between 800 and 1200 ms.

Since young children do not have the capacity to perform the demanding tasks that are commonly employed in experimentation with healthy adults, acoustic stimuli were delivered to child participants in this study while they viewed a video program projected onto a screen in the magnetically shielded room. The video aided the continued engagement of the child in the MEG environment and also helped to minimize movement artifacts during the recording session. The volume of the video soundtrack was left on and experimental sounds were superimposed on the soundtrack. Since the focus of the present study was on time-locked evoked responses to repetitive stimuli, the non-timelocked soundtrack has little effect on the evoked responses (McArthur et al., 2003). Child-friendly data acquisition techniques were employed to convey instructions, facilitate engagement with the experiment, and minimize movement artefacts during acquisition.

MEG acquisition

Prior to MEG measurements, five marker coils were placed on an elasticised cap on each subject's head and their positions and the subject's head shape were measured with a pen digitiser (Polhemus Fastrack, Cochester, VT). Head position was measured by energising the marker coils in the MEG dewar before and after AEF measurements, and the amount of head movement during the recording session was calculated by subtracting the pre-recording position of each coil from the post-recording positions. Movement tolerance was set at a threshold maximum of 5 mm for any individual coil. Shim pads were placed in the adult helmet to reduce children's head movement.

All measurements were carried out with subjects in a supine position in the MEG environment. An experimenter remained in the shielded room with the children during the recording session, seated on the bed of the unused MEG system. MEG data were acquired with a sampling rate of 1000 Hz and bandpass filtered from 0.03-200 Hz. No "synthetic gradiometer" noise processing was performed on the data.

MEG analysis

Sensor-head centre distances were calculated as the Euclidean distances between the nearest coil of each gradiometer and the centre of a sphere fitted to the digitised head shape using the least-squares minimization procedure. Since an optimal positioning of a participant's head in the MEG dewar helmet should minimize the sensor-to-head-centre distances summed over sensors, the average of the sensor-to-head centre distances in a given region provides a reasonable index of the "goodness-of-fit" of a subject's head in a MEG dewar helmet that provide a rough guide of comparison between subjects, between different regions of the head, and between different MEG instruments. Formally, we defined an overall "goodness of fit index" as:

$$GOF = \frac{\sum_{i=1}^{n}(S_i - HC)}{n}, \text{ where}$$

n = number of MEG sensors;
S_i = sensor position in Cartesian coordinates;
HC = head centre position in Cartesian coordinates; and
S_i-HC = the Euclidian distance (norm) between a given sensor and the head centre.

Regional goodness of fit indices were also calculated for the five sensor regions delineated in Figure 1.

MEG data were analysed off-line using BESA version 5.2.4 (MEGIS Software GMbH, Grafelfing, Germany) and MEG-MRI coregistration was performed using BrainVoyager version 1.10 (BrainInnovation BV, Maastricht, The Netherlands). Auditory evoked magnetic fields (AEFs) were averaged over a 600 ms epoch, including a 100 ms pre-stimulus interval. For each subject, two separate AEFs were averaged over 100 trials to show reproducibility of the responses. Source analyses were computed using AEFs averaged over all 200 trials. AEFs were baselined and digitally filtered (0.16 – 30 Hz). For analysis of channel waveforms, the 10 largest amplitude AEF channels (5 in each hemisphere) were identified using an automated peak detection algorithm in BESA 5.2. For each hemisphere, the root mean square (RMS) of the magnetic field amplitude was calculated across the five channels. Source analysis of AEFs were computed using a spherical model of the volume conductor, fitted to the digitised head shape, and two symmetric dipole sources. For adult AEFs, dipoles were fitted over an interval from the onset to the peak of the M100 component. For child AEFs, dipoles were fitted over an interval from the onset to the peak of the P100m component. Statistical comparisons of right and left hemispheric peaks in the child AEFs were computed with paired comparisons t-tests for mean magnetic field amplitudes and peak latencies in the intervals of 80-150 ms (P100m component) and 180-250 ms.

Results

Good AEFs were obtained from all seven children in the child MEG system. Each measurement session in the child system took approximately 20 minutes including positioning of the child in the MEG instrument. Positioning of children in the adult helmet took about 10 minutes longer for placement of shim padding. One child (C1) completed a subsequent recording session in the adult system; however no AEF was discernable in the recorded data. Two other children (C2 and C3) were placed in the adult MEG but measurements were terminated before completion due to gross movements (e.g. sitting up) or a request by the child. As it became clear that the length of the adult system measurement sessions were straining the capacities of the children, we did not attempt to obtain data in the adult system for the remaining four participants.

Head movement

Table 1 shows that all seven children maintained their head positions below the movement threshold during recordings in the child MEG helmet. The child who did complete the measurement session in the adult system was able to maintain head position below the movement threshold.

Table 1 about here

Adult heads in adult MEG

Mean overall GOF for 33 adults was 146.3 mm (sd = 6.4 mm). Figure 3A shows that regional GOFs varied systematically over the five regions in the adult MEG system and ANOVA confirmed that the regional differences were significant (($F(4,128) = 206.3$, $p < .001$). Smallest distances were at bilateral temporal areas (left temporal GOF = 139.0 mm, right temporal GOF = 137.3 mm). The 1.7 (+/- 5.4) mm mean difference between right and left temporal regions indicates that heads were positioned reasonably symmetrically in the helmet. A paired-differences t-test (2-tailed) confirmed no significant difference in fit between the temporal regions ($t(32) = 1.8$, $p > .05$). Poorest fits were for frontal (GOF = 155.0 mm) and central regions (GOF = 169.1 mm).

Figure 3 about here

Child heads in adult MEG

Overall GOF for three children's head positions measured in the adult MEG (Figure 3A) were 162.7, 163.6, and 164.7 mm. All GOF for these children were greater than 2.6 sd's above the mean for the adult GOF and all were worse than any of the individual adult participants (max GOF = 158.8 mm). Left-right temporal GOF differences for the three children were 12.2, 22.5 and 9.0 mm, corresponding to 2.3, 4.2, and 1.7 standard deviations for adults in the adult system, indicating that left-right symmetry was not well-accomplished for the children.

The main problems associated with the poorer fit of children's heads in the adult system are illustrated in Figure 4. Near optimal positioning of an adult head can be achieved and maintained with little effort: the experimenter need only determine that the head is fully inserted and symmetrically positioned in the dewar helmet. In contrast, the head of a four-year old child cannot be fully inserted due to the smaller crown-neck distance, and symmetrical lateral positioning cannot be achieved and maintained without the insertion of shim pads in the temporal regions.

Figure 4 about here

Child heads in child MEG

Mean overall GOF for the seven children in the child MEG was 136.1 mm (min 124.9; max 149.7), an improvement of some 20-30 mm over the GOFs for the three children that were measured in the adult MEG and 1.6 sd above the mean for the adult GOF. Figure 3B shows the regional GOFs obtained for the seven children in the child system, compared to the mean regional GOFs for adults in the adult system. Poorest fits were obtained for frontal (mean GOF = 162.0 mm) and central (mean GOF = 154.3 mm) regions, followed by temporal (left = 130.4 mm; right = 130.0 mm) and occipital (124.6) regions. ANOVA confirmed that the regional differences were statistically significant ($F(4,24) = 25.7$, $p < .001$). A paired samples t-test showed no significant difference between right and left temporal GOFs ($t(6) = .13$, $p > .05$), confirming that the heads were positioned symmetrically in the helmet.

Figure 3C illustrates the improved positioning of a child's head in the child MEG compared to the same child's head in the adult MEG (Figure 4B).

AEFs

Channel waveforms

Figures 5 and 6 compare AEF waveforms for an adult participant measured with the adult MEG and a child participant measured in the child MEG system. The adult responses show a typical AEF response composed of M50 (mean latency 63 ms, corrected for tubephone transit time), M100 (127 ms), M150 (183 ms) complex of peaks, with the largest response being the M100 peak. The M100 response shows anterior to posterior flux reversals in both hemispheres (see also Figure 4A) and is well-modelled with two (downward-directed) equivalent current dipoles in bilateral auditory cortices

(Figure 4D). The M50 and M150 responses exhibit anterior to posterior flux reversals that are oppositely directed to the M100 response.

Figures 5 and 6 about here

The child AEF showed a maximal amplitude response, termed P100m (Fujioka et al., 2006; Heim et al., 2003; Paetau et al., 1995), at a latency of about 120 ms (corrected for tubephone transit time). The anterior-posterior flux reversal of the P110m response was oppositely directed to that of the adult M100, and was followed by a circa 200 ms peak with flux reversal spatially opposite to the P100m.

Considering the children as a group, Figure 7A shows grand mean of the channel RMS waveforms for the seven children. The mean channel RMS waveforms showed a larger amplitude P100m response over the left hemisphere (mean 101.0 fT) compared to the right hemisphere (mean 74.08 fT) and this difference was significant ($p = .007$). The left lateralization of the P100m was visibly apparent in the individual waveforms of six of the seven children (see Figure 5). The right hemisphere P100m had a later mean latency (133 ms) than the left hemisphere (121 ms), a difference that approached but did not reach significance ($p = .08$). The circa 200 ms response had a later mean latency in the right hemisphere (215 ms) than in the left (195 ms) but this difference was not significant ($p > .05$).

Figure 7 about here

Source Analysis

Table 2 shows the results of dipole modelling for the 7 children. All data were well- modelled with two symmetric dipole sources and the grand mean dipole were located in bilateral superior temporal gyri (Figure 4E). All dipoles were located within 25 mm of the grand mean dipole location.

Table 2 about here

The grand mean source waveforms (Figure 7B) were characterised by the P100m peak. As for the channel waveforms, the left hemisphere P100m source peak (mean over the 90-150 ms time window = 44.0 nAm) was larger than the right hemisphere (22.1 nAm, $p = .02$). No other amplitude or latency differences reached significance.

Discussion

The present results show measurements of brain function from a sample of seven healthy and alert four-year-old children, an age for which there is currently a dearth of MEG and functional neuroimaging data. These data demonstrate that it is feasible to measure brain responses, using a child customized sensor configuration, from children who are difficult or impossible to study with conventional MEG or functional neuroimaging techniques.

The AEFs measured in the present study concur with child AEFs and auditory evoked potentials (AEPs) described by other investigators. AEFs in our study were characterised by a large amplitude peak at about 120 ms after stimulus onset. In contrast to the adult M100 response, the topography of the child 120 ms response exhibited a "positive up" dipole pattern best modelled by upward pointing dipolar sources. The 120 ms response corresponds to the circa 100 ms latency "P100m" response described previously in children (Fujioka et al., 2006; Heim et al., 2003; Paetau et al., 1995). All three of these studies have also reported a left lateralization of this response. The EEG version of the P100m, the P100 auditory event-related potential, has been measured in children as young as 1-3 years of age (Barnet, 1975; Courchesne, 1990; Kushnerenko et al., 2002). Although they have similar latencies, the P100m which dominates the child AEF is not comparable to the M100 of the adult AEF since they have opposite polarities. It remains unclear if the P100m matures into one of the adult AEP/AEF responses, but a likely candidate is the M50 response which also exhibits a prominent left hemispheric lateralization in the adult AEF (Chait et al., 2004).

In the present study we did not attempt a formal comparison between the adult and child MEG systems for measuring AEFs from children and we do not suggest that it is impossible to measure brain responses from children using the adult system. However, our experiences indicate that the adult sized system poses some important problems for measurements in children as young as four years. First, since the smaller head volume of children results in a poorer fit in the adult helmet, many of the sensors will be at a suboptimal or entirely ineffective distance from the brain. Second, it is more difficult to position children's heads symmetrically in the adult helmet, likely resulting in different signal-to-noise ratios for responses measured from the two hemispheres. Third, the necessity for shim padding substantially increased the set-up time in the adult system, a crucial consideration in the context of children who have a limited capacity to tolerate the boredom and confinement of the MEG system. These three factors have played major roles in the relative lack of MEG data available from pre-school children in the current literature. We note that P100m response described in the three previous reports described above were measured using either a single hemisphere MEG system, or were measured primarily from older children using adult whole-head systems. The 1995 report by Paetua et al. measured AEFs from 3-15 year olds using a 22 channel, single hemisphere MEG array, while the whole-head (adult system) MEG study of Heim et al. (2003) measured responses from 8-13 year olds. The whole-head (adult system) MEG study of Fujioka et al. (2006) studied 4-6 year olds over multiple sessions; however the mean age of children at the first recording session was 5 ½ years, at the upper end of the pre-school age-range.

The main advantages of the customized helmet dewar are threefold. First, the smaller radius of the sensor configuration is optimised to bring the sensors into range of the neuromagnetic signals. Second, the smaller helmet allows full insertion of a child's head into the dewar. Full insertion is prevented in adult dewar helmets because of the smaller crown to shoulder **distance** in children. These two factors are **crucial in recording brain activity using** MEG because of the rapid attenuation of neuromagnetic signals with distance. Finally, the customized child helmet aids in the symmetric positioning of the head and limits the freedom of movement of the child's head within the dewar. In comparison to the adult MEG system, these features of the child MEG system strongly facilitate the setup time, appropriate positioning, and measurement time for acquiring MEG signals from children.

Those familiar with healthy and active four-year old children will appreciate that head size is only one of the challenges to **conducting studies with children using** functional neuroimaging. It is also essential to customize the experimental environment, instructions, methodology and procedure to accommodate the cognitive and attentional capacities and motivational requirements of children. In the present study we set up a child-friendly laboratory environment with colourful drawings and cartoon characters on the walls. Children and parents were familiarized with the laboratory on one or more visits prior to the actual experiment. Parents were also permitted to stay in the MSR with their children during the recording sessions, providing ongoing reassurance for the child and helping the experimenter to detect and minimize movements. Further child-customized procedures are necessary if the child is required to actively engage in an experimental task; the present experiment was considerably simplified by the fact that experimental stimuli could be ignored while the children viewed a cartoon.

The comparison with adult MEG responses measured in a conventional adult system shows that the design of the child system has several limitations. First, the child system has substantially fewer sensors than the adult system, predicated by the smaller helmet volume. Second, the helmet has relatively poor coverage of the anterior frontal and temporal regions, predicated by a design requirement to leave children's faces fairly unconfined. This is necessary in a cognitive neuroscience research context to prevent claustrophobia and to facilitate communication and eye contact between the children and the experimenters and/or caregivers who remain in the magnetically shielded room. In the current study, this limitation resulted in relatively sparse coverage of the anterior flux maxima/minima of the AEFs; however the anterior flux reversals were covered by at least one sensor channel in each participant and we were able to calculate reasonable dipole locations for all participants.

Conclusions

These are the first MEG data measured from preschool-aged children using a helmet dewar custom-sized to accommodate children's smaller-sized heads and necks. Measurements of GOF show that the child helmet achieves a fit comparable to or better than that typically achieved for adult heads with a conventional MEG system, and substantially better than achieved for child heads in the adult system. The improved fit allows full insertion of the head into the dewar, concentrates the sensor array within a radius that is effective for measuring signals from children's brains, and limits movements of the head during recordings. The advent of customize-sized MEG systems will facilitate routine neuroimaging studies of cognitive development in young children and help to fill a large gap in the current neuroscientific literature on development of brain function.

Figure legends

Figure 1. Top. Schematic diagram of sensor layout for 160 channel adult system showing distribution of sensors in each of five regions. Bottom. Sensor layout for 64 channel child system.

Figure 2. Top: Inner dimensions of child helmet dewar in mm. R = radius. Bottom: Relationship between digitised head surface (grey) and MEG sensors (large yellow circles) for subject C3. Arrowheads indicate fiducial points (nasion, left perauricular, right preauricular).

Figure 3. A. Regional GOFs for mean of 33 adults, and three children measured in the adult MEG system. B. Regional GOFs shown for seven children measured in the child MEG system, compared to mean of 33 adults. Error bars for adult data indicate 1 standard deviation.

Figure 4. A-C. Sensor positions shown with respect to head for: A. An adult in adult MEG. B. A child in adult MEG. C. Child in child MEG. Adult head is reconstructed from an individual subject's MRI scan. "Child" head is from BESA standard (mean) MRI. M100 topography is shown in A, M120 topography in C. Magnetic flux measured by MEG sensors is projected onto the surface of the head. Blue indicates negative flux (entering head), red indicates positive flux (leaving head).

D. Mean dipole source locations for adult M100 AEF. E. Mean dipole source location for child M120 AEF. Adult sources are superimposed on an individual adult's MRI scan. Children's sources are shown on BESA standard MRI.

Figure 5. Channel AEF waveforms for an adult measured in the 160-channel adult MEG (top), and for a child measured in the 64-channel child MEG (bottom). Note the left

hemisphere lateralization of the M120(P100) response in the child AEFs. AEFs from circled positions are enlarged in Figure 5. AEFs are averages of 200 trials. Latencies in the figures are not corrected for 17 ms sound transit lag from the Etymotic tubephones.

Figure 6. Representative AEFs from an adult (A) and child (B) from sensor positions circled in Figure 5. Two averages each of 100 trials are shown superimposed. Latencies are not corrected for sound transit lag from the Etymotic tubephones.

Figure 7. A. Mean root mean square (RMS) channel waveforms for the seven children. B. Mean source waveforms for the seven children. Latencies in this figure are corrected for a 17 ms lag between activation of the earphone transducers and arrival of sounds in the ear canal.

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