TRACKING OROMOTOR DYNAMICS USING WHOLE-CORTEX MAGNETOENCEPHALOGRAM SENSORS

Natasha Alves[1,2], Pascal van Lieshout[2], Douglas Cheyne[1,2]
The Hospital for Sick Children[1]; University of Toronto[2]

ABSTRACT

In this study, we propose a non line-of-sight, five-dimensional magnetoencephalogram (MEG)-compatible system for tracking the location and orientation of target points on the tongue, lips and jaw during articulation. The orofacial tracking system uses small, light-weight coils that are driven continuously with low-amplitude sinusoidal currents. The current carrying coils create magnetic dipoles that can be localized by least-square fits of the modeled magnetic field to the response of the whole-cortex MEG sensors. This paper discusses preliminary tests performed to evaluate the spatial and temporal resolution of the tracking system. Once implemented, the system will present a unique opportunity to study the neural mechanisms underlying speech and other oro-motor functions.

INTRODUCTION

An estimated 1% of Canadians, including 4% of pre-school children and 6-12% of seniors, have significant speech disorders [1]. Many speech disorders, such as stutter, aphasia and articulation disorders, have been linked to neurological damage or atypical interactions between regions of the brain [2].

In the production of speech, the brain orchestrates the coordination of over 100 muscles related to articulation, phonation and respiration at relatively fast rates [3]. The precise time course in which the brain coordinates these complex activities is relatively unknown, due to challenges in localizing activity to specific brain areas with sufficient temporal resolution. Conventional neuroimaging methods, such as functional magnetic resonance imaging (fMRI) and positron emission tomography (PET), are limited in their ability to capture the fast evolving changes in brain activity associated with the rapid movements and sensori-motor processing involved in speech production. In contrast, MEG offers the possibility of examining cortical activity with high temporal resolution (millisecond range) while maintaining good spatial accuracy (2-3mm) [4].

The understanding of the underlying neural mechanisms of speech requires the recording of MEG in parallel with articulatory kinematics, such as movements of the tongue, lips, jaw and vocal tract. Electromagnetic articulography (EMA) systems are typically used in speech studies to track multiple points on the articulators inside and outside the vocal tract [5]. EMA systems consist of alternating magnetic field transmitters positioned around the subject's head, and small receiver coils positioned on and in the subject's mouth. The motion of the articulators within the magnetic field induces current in the coils from which its position and orientation can be determined.

While EMA systems introduce magnetic interference and cannot be used in the MEG recording environment, the inductive measurement principle may still be exploited for measuring coil position and orientation in the oral cavity. Small coils carrying alternating currents create magnetic dipoles that can be localized within a whole-cortex MEG system. The objective of this study is to localize speech articulators by employing methods similar to those used for continuously monitoring the position of the subject's head within the whole-cortex MEG sensor [6]. We briefly discuss the methods used to locate and track the moving coils, and the preliminary tests performed to evaluate the system's ability to meet the temporal and spatial resolutions required for tracking speech kinematics.

METHODS

Instrumentation

Small (3mm diameter), light-weight (2g) coils, with inductances of approximately 550µH and resistances of about 26Ω were continuously excited with low-amperage sinusoidal currents in the 1400-1600Hz range. The coils similar to those used in EMA systems. They are much smaller than the coils used for head-localization within the MEG system and are driven with very small currents (micro-ampere range). Further, unlike head coils which are positioned close to the MEG helmet, the mouth coils may be positioned 15-20 cm below the helmet. The driving frequencies of the coils were chosen so that the coil's movement information is modulated outside the band-width of interest for neural activity, and such that each coil's signal is easily separated from other coils and the power-line harmonics [6]. A 151-channel whole cortex MEG system (CTF MEG^{TM}) was used to measure the magnetic fields induced by the current carrying coils. The MEG data were sampled at 12kHz.

Coil localization

The positions and orientations of the simultaneously activated coils can be evaluated from the forward solution of the magnetic fields induced by the coils. The magnetic induction by the k^{th} coil on the i^{th} MEG sensor is:

$$\beta(x_i, n_i; y_k, m_k) = \frac{3m_k(y_k - x_i)n_i \cdot (y_k - x_i)}{\left|y_k - x_i\right|^5} - \frac{m_k \cdot n_i}{\left|y_k - x_i\right|^3} \quad (1)$$

where x_i and n_i are the position and orientation of the i^{th} MEG sensor, m_k is the coil moment and y_k is the coil's position [7].

The optimum estimator of the coils positions and orientations is obtained by a least-squares fit of the magnetic field model (from equation 1) and the measured data (b_{ij}). The minimization function can be expressed as:

$$H = \sum_{i,j} \left(\sum_{k=0}^{K} s_{k,j} \beta(x_i, n_i; y_k, m_k) \right) - b_{ij} \quad (2)$$

where $k = 1, \ldots K$, $i = 1, \ldots I$, time $j = 1, \ldots J$ and $s_{k,j}$ is the source time function of coil k. When the source time functions of the coils are mutually orthogonal, the minimization problem for each coil can be solved independently. The detailed evaluation of function $H(\cdot)$ and its gradient is given by Munck et. al [7]. The Levenberg–Marquardt algorithm was used to find the values of y_k and m_k that minimizes $H(\cdot)$ [8].

Stationary dipole localization

The tracking system was validated in a phantom experiment where relative coil positions are known. Four mouth coils were equally spaced along the circumference of a cylinder of radius 7.5cm. The cylinder was placed approximately 7cm below the MEG helmet. Taking into consideration the diameters of the coils, it is expected that adjacent coils are separated by approximately 10.83cm, and non-adjacent coils are separated by 15.32cm. The four coils were energized with 2µA peak sinusoidal currents with frequencies of 1410Hz, 1470Hz, 1530Hz and 1590Hz, respectively. Five seconds of data were recorded and were spliced offline into non-overlapping 33.3 ms windows. The starting point of the least-squares position search was set at [x=0, y=0, z=-4cm] on the helmet co-ordinate system for each data window. Each data window was analyzed independently, i.e. *apriori* position information from previous windows was not used.

Moving dipole localization

Two mouth coils, separated by approximately 8.3cm, were secured to an inextensible styrofoam rod. The 5 foot rod was attached to an electrodynamic mini-shaker (2004E, Modal Shop) which moved the rod and the coils vertically with a displacement trajectory of $2.5\text{cm} \times \sin(2\pi 5t)$. The coil therefore moves at velocities of up to 50cm/sec. These speeds may be expected during speech movements. The MEG helmet was positioned vertically. The electrodynamic shaker was placed in the magnetically shielded room, about 5 feet away from the MEG helmet, and introduced a strong magnetic interference in the 0-100Hz range. However, magnetic interference in the 1kHz-3kHz range was minimal. The coils were energized with 20μA peak sinusoidal currents at frequencies of 1470Hz and 1530Hz, respectively. As in the stationary-dipole experiment, data were spliced into non-overlapping 33.3ms windows for locating the positions and orientations of the magnetic dipoles. For this test, the start point of the position search for the first data window was set at [x=0, y=0, z=-4cm] on the helmet co-ordinate system. Subsequent data windows used the least-squares position of the previous window as its search start position.

RESULTS

Stationary dipole tests

The relative distances between the coils computed from the dipole-fit were in the expected range of 10.83cm and 15.32cm. The position and relative distances between the four coils is shown in Figure 1. The variance of the distances was very small over the 5 second recording session.

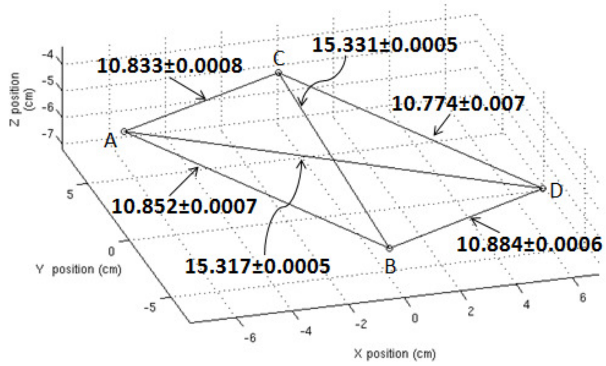

Figure 1. Mean position and relative distances of the four coils from the phantom experiment (mean ± standard deviation).

Moving dipoles tests

Figure 2 shows the time-domain signals and Figure 3 shows the frequency-domain signals of the MEG sensors for the moving-dipoles experiment. As seen in the figures, the current in the mouth coils act as carrier frequencies on which the coil's movement information is amplitude-modulated.

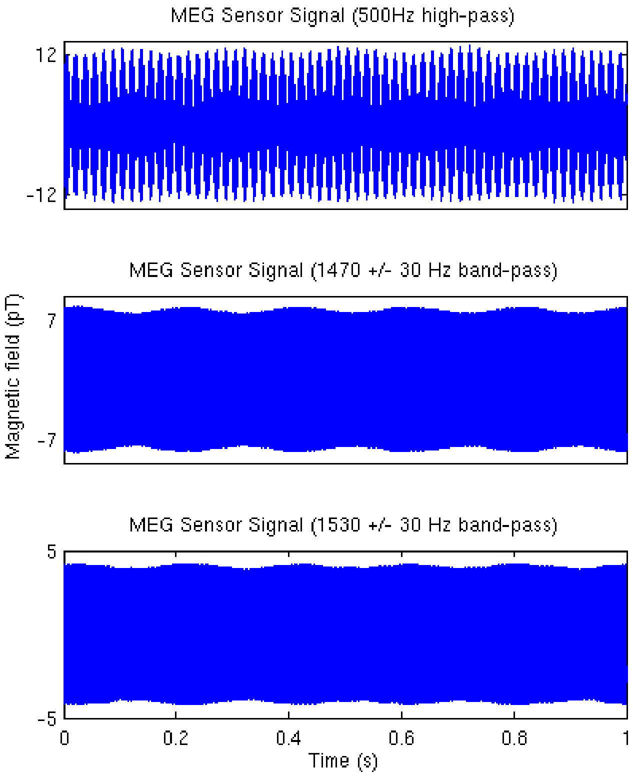

Figure 2. Signals recorded from an MEG sensor when 2 coils, driven at 1470Hz and 1530Hz, are moving at 5Hz. Band-pass filters at each coil's driving frequencies shows the coil's movement profile.

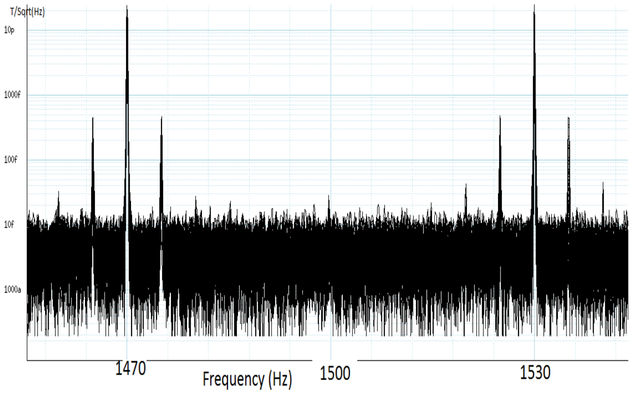

Figure 3. Power spectrum of signals recorded from MEG sensor, showing carrier coil frequencies and movement frequencies.

Figure 4 shows the x, y and z positions of the coils in the MEG helmet co-ordinate system. The 5Hz sinusoidal movement of the coils is clearly visible on the x and z axes. In spite of the high velocities of up to 50cm/sec, the distance between the two coils remained fairly constant at 8.256±0.0395 cm.

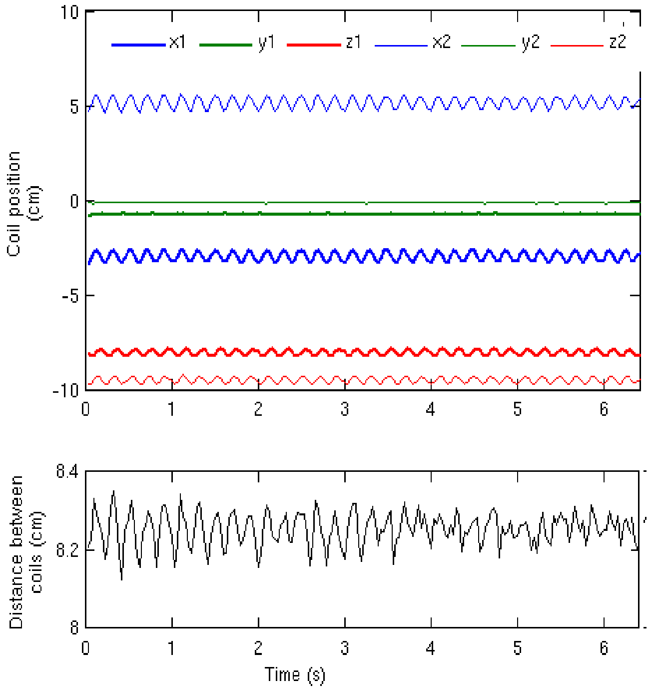

Figure 4. Dipole fit results from the moving coils. The least-squares dipole-fit x, y, and z positions of coil 1 (thick lines) and coil 2 (thin lines) are shown. The distance between the two coils, shown in the lower figure, remains fairly constant (±0.4mm).

DISCUSSION

While this work shows the potential of the MEG-compatible articulography system, there is a need to benchmark the position and orientation accuracies with existing ultrasound or EMA tracking systems. It is unclear how tracking accuracies would change with distance from the helmet or with current amplitudes. Further, unlike EMA systems whose coils have passively induced currents, the mouth coils used in the MEG system are excited with currents, necessitating the design of coil drivers that are optically isolated and pass safety standards before human testing.

ACKNOWLEDGEMENTS

The authors thank Dr. Harold Wilson, Dr. Paul Ferrari and Marc Lalancette. This work was supported by NSERC and the MITACS Elevate program.

REFERENCES

[1] Canadian Association of Speech-Language Pathologists and Audiologists, "Speech, language and hearing fact sheet (www.caslpa.ca)," 2005.

[2] M. Sommer, M.A. Koch, W. Paulus, C. Weiller and C. Büchel, "Disconnection of speech-relevant brain areas in persistent developmental stuttering," The Lancet, 360 (9330), pp. 380-383; 2002.

[3] R.D. Kent, "The uniqueness of speech among motor systems," Clinical Linguistics & Phonetics, 18 (6), pp. 495-505; 2004.

[4] M. Hämäläinen, R. Hari, R.J. Ilmoniemi, J. Knuutila and O.V. Lounasmaa, "Magnetoencephalography—theory, instrumentation, and applications to noninvasive studies of the working human brain," Reviews of Modern Physics, 65 (2), pp. 413-497; 1993.

[5] P.W. Schönle, K. Gräbe, P. Wenig, J. Höhne, J. Schrader and B. Conrad, "Electromagnetic articulography: Use of alternating magnetic fields for tracking movements of multiple points inside and outside the vocal tract* 1," Brain and Language, 31 (1), pp. 26-35; 1987.

[6] H. Wilson, "Continuous head-localization and data correction in a whole-cortex MEG sensor," Neurology and Clinical Neurophysiology, 2004 pp. 56; 2004.

[7] J. Munck, J. Verbunt, D. Ent and B. Dijk, "The use of an MEG device as 3D digitizer and motion monitoring system," Physics in Medicine and Biology, 46 pp. 2041; 2001.

[8] J. More, "The Levenberg-Marquardt algorithm: implementation and theory," Numerical Analysis, pp. 105-116; 1978.