Dynamics of Lateral Ventricle and Cerebrospinal Fluid in Normal and Hydrocephalic Brains

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Purpose: To develop quantitative MRI techniques to measure, model, and visualize cerebrospinal fluid (CSF) hydrodynamics in normal subjects and hydrocephalic patients.

Materials and Methods: Velocity information was obtained using time-resolved (CINE) phase-contrast imaging of different brain regions. A technique was developed to measure the change of lateral ventricle (LV) size. The temporal relationships between the LV size change, CSF movement, and blood flow could then be established. The data were incorporated into a first-principle CSF hydrodynamic model. The model was then used to generate specific predictions about CSF pressure relationships. To better-visualize the CSF flow, a color-coding technique based on linear transformations was developed that represents the magnitude and direction of the velocity in a single cinematic view.

Results: The LV volume change of the eight normal subjects was $0.901 \pm 0.406\%$. Counterintuitively, the LV decreases as the choroid plexus expands, so that they act together to produce the CSF oscillatory flow. The amount of oscillatory flow volume is $21.7 \pm 10.6\%$ of the volume change of the LV from its maximum to its minimum.

Conclusion: The quantification and visualization techniques, together with the mathematical model, provide a unique approach to understanding CSF flow dynamics.

Key Words: brain ventricle movement; CSF dynamics; visualization; CINE phase-contrast; CSF modeling

BACKGROUND

The LV volumetric change is one of the driving forces for CSF movement. The magnitude and timing of these movements needs to be measured to understand quantitatively how much the LV motion contributes to the CSF movement. A periodic 10% to 20% volume change of the LV was measured by Lee et al (7) based on the change of MR image signal intensity. This approach likely contains a serious overestimation. The brain tissue movement within a cardiac cycle is only a small fraction of a pixel, as found by Enzmann et al (4) and our own measurements using the time-resolved (CINE) phase-contrast technique. Estimating the ventricle size change in the resolution of pixel size is therefore highly inaccurate. Furthermore, flow artifacts and partial volume effects can contribute to the change of signal intensity at the edge of the ventricle. A technique similar to the approach of Oyre et al (8) is instead used in our work. The edge positions of the ventricle throughout the

DISTURBANCES OF THE CEREBROSPINAL FLUID (CSF) flow in the brain can lead to hydrocephalus, a condition affecting thousands of people annually in the United States. Considerable controversy exists about fluid and pressure dynamics, and about how the brain responds to changes in flow patterns and compression in hydrocephalus. Some information to help understand CSF flow dynamics is currently available from MRI, including measurements of CSF flow pattern and velocity at various locations along the CSF pathways (1–3) and of brain motion (4,5). However, integration of these measurements to explain CSF flow dynamics is incomplete. We have used MRI techniques to measure the lateral ventricle (LV) size change and its temporal relationship with intracranial blood flow and CSF movement along the CSF pathways. The ventricular size changes and CSF flow patterns that we have found are consistent with the dynamics of intracranial phenomena predicted by a first-principles model introduced by Linninger et al (6). This model, in turn, can be used to predict intracranial pressure (ICP) dynamics in normal and hydrocephalic brains. A new quantitative color-coding technique is introduced to better visualize the CSF flow patterns.

Contract grant sponsors: Medtronic Inc.; Gilbert Asher; Max Cooper. Most of the research reported here was conducted while D.C.Z. was working in the Brain Research Imaging Center at the University of Chicago.

Received August 22, 2005; Accepted May 24, 2006.

DOI 10.1002/jmri.20679

Published online 6 September 2006 in Wiley InterScience (www.interscience.wiley.com).
cardiac cycle are estimated based on the velocity values of the ventricle edge points measured by the CINE phase-contrast technique. But, unlike Oyre et al (8), the directions in which the edge positions move are not assumed in our technique. Along with ventricular movement, CSF flow rates were measured at the junction of the aqueduct of Sylvius (AS) and the fourth ventricle (V4) and at the midcoronal section of the third ventricle (V3), and blood flow rate was measured in the basilar artery. Since all velocity measurements could be referenced to the cardiac pulse, their temporal relationship could be established. Because of the choroid plexus’ complex shape, its motion (which also drives CSF movement) could not be measured.

Figure 1. The five approximate locations where the two-dimensional CINE phase-contrast images were collected. A midsagittal slice is shown. The other four locations are as follows: an axial slice across the middle of the LV (a), an axial slice across the junction between the AS and V4 (b), a midcoronal slice at V3 (c), and an axial slice nearly perpendicular to the basilar artery in the prepontine region (d).

Figure 2. Demonstration of the color-coding technique. a: The center of the color circle represents zero velocity. The edge of the color circle represents velocity of 5 mm/second or higher. The 0° or 360° line is indicated. The direction of the color circle is indicated by the S (superior), I (inferior), A (anterior), and P (posterior), with the positive directions of S–I and A–P. The velocities at a time frame of the cardiac cycle at two locations (one at the middle of V4 and the other near the entry of the foramen of Magendie) are equivalently represented by the locations indicated at the color circle. \( V_{SI} = -1.34 \text{ mm/sec} \) or \( V_{mag} = 4.07 \text{ mm/sec} \). \( V_{AP} = -3.84 \text{ mm/sec} \). Angle = 199°.

b: The velocity view contrast is enhanced by setting the edge of the color circle to represent 2 mm/second or higher. \( V_{SI} = -5.15 \text{ mm/sec} \) or \( V_{mag} = 5.50 \text{ mm/sec} \). Angle = 291°.

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directly. The timing of its decrease and increase was assumed to be synchronous with the blood flow at the basilar artery.

A first-principles model for pulsatile CSF flow, whose mathematical formulation has been presented by Linninger et al (6), relates three dynamically interacting systems: the cerebral vascular system, the CSF-filled ventricular and subarachnoid spaces (SASs), and the brain parenchyma. With the inputs from MR measurements, the CSF pressure and velocity fields throughout the brain can be derived, as well as the dynamics of parenchyma stresses, strains, and displacements using the laws of elastodynamics. The direct MRI measurements and the calculated results from the model provide not only an understanding of normal CSF flow dynamics but also important predictions about the pressure and flow rate changes in hydrocephalus.

To provide a better visualization of CSF movement, a new color-coding technique for cinematic flow visualization has been developed. Traditional cinematic flow visualization in CINE phase-contrast MRI has been limited to either the magnitude or one vector component of the velocity at a time. As a consequence the complex nature of the flow pattern is not fully represented. The new color-coding technique combines both the magnitude and direction of the CSF flow velocity in one cinematic view. Using color maps to represent direction is not new in imaging. For example, color mapping has often been applied to the directional visualization of white-matter fiber tracks in diffusion tensor imaging (DTI) (9). However, the direct translation of the DTI color mapping technique to flow is not appropriate because fibers do not require the differentiation of two opposite directions, as is necessary to depict flow patterns. Our new color-coding technique represents flow in all directions, and expands color mapping to the time frame, while maintaining the quantitative nature of the CSF flow dynamics.

MATERIALS AND METHODS

Data Acquisition and Velocity Calculation

The twodimensional CINE phase-contrast technique (10,11) was applied to collect CSF flow data from 11 subjects (eight normal subjects from 23 to 52 years old, and three with hydrocephalus) on a 3T GE Signa system (GE Medical Systems, Milwaukee, WI, USA) equipped with a standard quadrature birdcage head coil. All volunteers signed the consent forms approved by the Institutional Review Board at the University of Chicago.

Of the three subjects with hydrocephalus, one subject has mildly enlarged ventricles but was neurologically normal. The second subject has the signs and symptoms of adult communicating hydrocephalus, and large ventricles. The third subject has the signs and symptoms of adult obstructive hydrocephalus, and moderately enlarged ventricles.

The two-dimensional CINE phase-contrast images were collected at five different locations (Fig. 1): 1) the midsagittal slice to view the major CSF pathways; 2) an axial slice across the middle of the LV to investigate the LV volumetric change; 3) an axial slice across the junction between the AS and V4 to measure the CSF flow rate; 4) a midcoronal slice at V3 to measure the CSF flow rate; and 5) an axial slice nearly perpendicular to the basilar artery in the prepontine region to measure the blood flow rate. For the first two locations, velocities in all three directions, velocities in all three directions, were collected at five different locations (Fig. 1): 1) the basilar artery in the prepontine region to measure the blood flow rate. For the first two locations, velocities in all three directions were measured to investigate the flow dynamics based on the simple four-point method (11). Images at 16 equidistant time frames were reconstructed per cardiac cycle. For the latter three locations, only the velocity perpendicular to the slice of interest was measured so that data could be collected with a higher temporal resolution. The simple two-point method was used to calculate the velocity (11). Images at 32 equidistant time frames were reconstructed per cardiac cycle. For all studies, flow compensation and peripheral gating were applied. For CSF flow measurement, a low maximum measurable velocity (VENC) of 5 cm/second was chosen as the limit so that a reasonable velocity resolution could be achieved. For the blood flow measurement of the basilar artery, a VENC of 100 cm/second was chosen. Other acquisition parameters were: TE = 8.4 msec, TR = 18 msec, flip angle = 20°, field of view (FOV) = 24 cm, slice thickness = 5 mm, matrix size = 256 × 128 for the midsagittal acquisition and 256 × 192 for the other acquisitions, number of excitations = 2, and full phase FOV for the midsagittal acquisition, but 75% phase FOV for the other acquisitions to achieve an effective matrix resolution of 256 × 256.

The CSF pathway was segmented for analysis based on the T2-weighted fast spin echo (FSE) image (TE = 100 msec, TR = 4200 msec, echo train length = 16, FOV = 24 cm, slice thickness = 5 mm, interslice spacing = 1 mm, number of slices = 16, matrix size = 256 × 256) in which CSF was enhanced. The velocity at every pixel within the regions of CSF was calculated. To reduce the possibility of a spatially-dependent offset velocity due to eddy currents or head motion, the velocity at each pixel location was corrected by basic subtraction of the time-averaged “velocity” of a nearby solid brain tissue “background” within a 29 × 29 mm² region having this pixel at its center (4,5,12). In calculating the velocity of the solid brain tissue, the velocity at each pixel location was corrected by basic subtraction of the time-averaged “velocity” of this pixel itself. These approaches are based on the fact that solid brain tissue does not accumulate net displacement over a complete cardiac cycle (4,5).

The flow rate at the midcoronal slice across V3, at the junction of the AS and V4, or in the basilar artery, is estimated by the multiplication of the average velocity at the cross-section of the CSF/blood pathway and the corresponding area. The cross-section of the fluid pathway is segmented based on an image that showed the best cross-section from the T2-weighted and T1-weighted images. The mean oscillatory flow rates of CSF at the two cross-sections were also calculated based on the average of the forward and backward flow rate magnitudes through a full cardiac cycle. The mean flow volume per cycle at the junction of the AS and V4 was calculated based on the average of the forward and backward flow volumes through a full cardiac cycle.