Communication

MR Measurement of Cerebrospinal Fluid Velocity Wave Speed in the Spinal Canal

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***Abstract*—Noninvasive measurement of the speed with which the cerebrospinal fluid (CSF) velocity wave travels through the spinal canal is of interest as a potential indicator of CSF system pressure and compliance, both of which may play a role in the development of craniospinal diseases. However, measurement of CSF velocity wave speed (VWS) has eluded researchers primarily due to either a lack of access to CSF velocity measurements or poor temporal resolution. Here, we present a CSF VWS measurement methodology using a novel MR sequence that acquires unsteady velocity measurements during the cardiac cycle with a time interval <10 ms. Axial CSF velocity measurements were obtained in the sagittal plane of the cervical spinal region on three subjects referred for an MRI scan without craniospinal disorders. CSF VWS was estimated by using the time shift identified by the maximum velocity and maximum temporal velocity gradient during the cardiac cycle. Based on the maximum velocity gradient, the mean VWS in the three cases was calculated to be 4.6 m/s (standard deviation 1.7 m/s, $p < 0.005$) during systolic acceleration. VWS computed using maximum velocity alone was not statistically significant for any of the three cases. The measurements of VWS are close in magnitude to previously published values. The methodology represents a new technique that can be used to measure VWS in the spinal canal noninvasively. Further research is required to both validate the measurements and determine clinical significance.**

***Index Terms*—Cerebrospinal fluid (CSF) hydrodynamics, cine MR, pulse wave velocity (PWV), spinal cord pathology assessment, subarachnoid space (SAS).**

I. Introduction

The pulse wave velocity (PWV) in a compliant vessel increases as wall stiffness increases and has been of interest since arterial stiffness is thought to be a risk factor for arterial disease [1]. Craniospinal disorders, such as Chiari malformation, syringomyelia, and others, are thought to be linked with overall cerebrospinal fluid (CSF) system compliance, and hence, PWV measurements would also be of interest [2]–[5]. *In vivo* PWV has been quantified through invasive measurement of pressure in the CSF system [2]. However, researchers have found it difficult to obtain CSF PWV noninvasively due to a lack of accessibility to the skull and spinal vertebrae.

MRI has been employed to measure PWV in various parts of the circulatory system including the aorta and other major vessels by comparing the time to peak velocity between two locations along the vessel of interest [6]–[8]. Fielden *et al.* detailed a methodology for quantifying aortic PWV using cross correlation on 2-D probability of correct message receipt (PCMR) velocity data [9]. In addition, a "COMB" excitation can be used and time to onset of displacement can be measured to estimate PWV [10], [11]. These studies have assumed that the PWV and velocity wave speed (VWS) measured by MR are nearly equivalent, which is likely a reasonable assumption in cases where Womersley number is nearly constant along the fluid conduit and the radial measurement location of the velocity waveform in the vessel is consistent between measurement locations. Recent improvements in MR hardware and protocols have reduced the imaging time interval to values less than 10 ms. Furthermore, in-plane velocity encoding allows continuous sampling of the VWS propagation in both the temporal and spatial domains. This MR technique has been shown to produce a measurement of PWV in the aorta [9]. The present study employs a similar MR technique to determine the VWS in the cervical spinal canal.

A number of studies have reported a single value of VWS during the CSF flow cycle. VWS in the CSF was found to be 13.5 m/s by Williams [2], 2.2–4 m/s by Carpenter *et al.* [3], 4 m/s by Greitz *et al.* [4], and 12.4 m/s by Bertram *et al.* [12]. In an *in vitro* study of syringomyelia by Martin *et al.*, the VWS was found to vary during the CSF flow cycle from 2 to 26 m/s [5].

II. Methods

Two patients who were referred for MRI because of unspecified back pain, but without any previously diagnosed craniospinal disorders, and one patient who had fused vertebrae participated in the study. All three patients were undergoing a clinically ordered MRI exam of the spine. The protocol was approved by the University's Institutional Review Board (IRB). Patients were placed supine on a six-element spin array coil in a Philips Medical Systems 3.0 Tesla Intera MRI scanner. ECG leads or a peripheral pulse (PPU) fingertip gating was used to monitor heart rate. After a standard MRI spine exam, including T1- and T2-weighted sagittal and transverse images, a slice location was identified that passed through the center of the spinal cord and spinal canal in the sagittal plane. At this slice location, a cine phase contrast velocity scan was acquired with in-plane velocity encoding in the foot head direction at a value of 20 cm/s. Retrospective ECG or PPU gating was used to reconstruct three cases with 134, 154, and 141 frames, respectively, over the cardiac cycle. The transverse runaway (TR) for the sequence was 4.4 ms and the TE was 2.4 ms. Slice thickness was 8 mm and in-plane reconstructed pixel sizes were 1.48×1.48 and 1.37×1.37 mm in case 1 and cases 2 and 3, respectively. Overall scan time was 2–3 min, depending on the heart rate.

Image processing was conducted using MATLAB (Mathworks 13.0, Natick, MA). The distance traveled by the pressure wave in the spinal canal was determined by tracing the path along the midline of the anterior subarachnoid space (SAS), as shown in Fig. 1. Axial velocity

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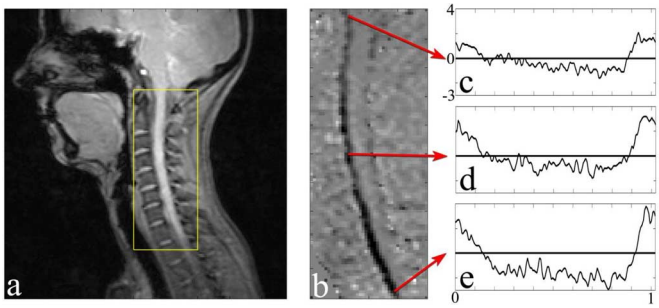

Fig. 1. (a) Sagittal geometry image with region of interest indicated (magnitude image). (b) Region of interest velocity image (phase image) as systole (T1 weighted). (c)–(e) CSF velocity traces at various levels within center of anterior spinal canal for patient 1.

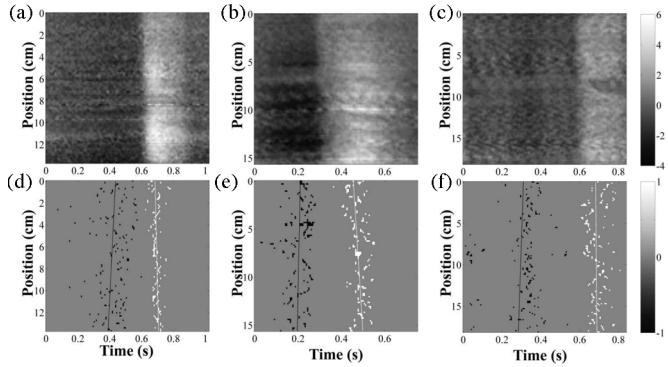

Fig. 2. (a)–(c) Raw velocity data for three patients along centerline of the anterior gap in the spinal canal as indicated in Fig. 1. Velocity scale indicated in centimeters per second. (d)–(f) Peak systolic acceleration (white) and deceleration (black) mappings with a linear fit.

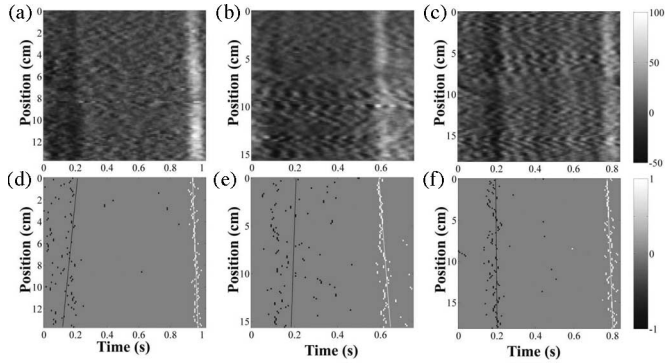

Fig. 3. (a)–(c) Velocity temporal gradient based on the velocity data for three patients in Fig. 2(a)–(c). Velocity gradient scale indicated in millimeters per square second. (d)–(f) Peak systolic acceleration (white) and deceleration (black) mappings with a linear fit and their scale.

(caudal and rostal) information was recorded for each pixel in the sagittal plane. The gap at the anterior side of the SAS ranged between two and five pixels. The pixel with the maximum systolic velocity was selected at each axial position to minimize wall boundary noise. This was generally near the center of the gap. It is important to select the pixels near the center of the gap since a phase shift can occur for velocity traces at different distances from the wall. This is a well-documented phenomenon for pulsatile (Womersley) flows and would add noise to the calculation of VWS.

The velocity waveform of each central pixel for all three cases was extracted for a total of 101, 106, and 124 vertical pixels, each having high temporal resolution with 6.7, 5.6, and 6.0 ms time intervals, respectively. Time points of maximum velocity (peak systole), minimum velocity (peak diastole), maximum gradient (systolic acceleration), and minimum gradient (systolic deceleration) were obtained. At each location, the velocity gradient throughout the cardiac cycle was based on slopes of lines fitted to an 11-time-point window obtained at each time point. The 11-point line was a fit obtained with linear least squares regression. The time of each computed gradient was considered to be that at the midpoint of the 11-time-point window. Time point of maximum acceleration and deceleration along the spinal canal was determined, and a linear regression was then performed on the time points versus position. The distance the velocity wave traveled divided by the time delay, represented by the slope of this linear fit, was assumed to be representative of the VWS at this time point in the cardiac cycle. Slope values that had a p-value less than 0.05 were considered to be statistically significant. A linear fit may not be the best technique to compute VWS since VWS may vary along the spinal SAS due to compliance changes. Several other curve fit techniques were examined. However, the linear fit produced the lowest overall p-values for the three cases.

III. Results

Fig. 2(a)–(c) shows the raw MR velocity data for each volunteer as a function of time and position. Axial position is indicated in Fig. 1 and begins at the base of the brain and extended caudally. Velocity is indicated by the grayscale, white and black representing velocity in the caudal and cranial directions, respectively. Gray represents zero velocity. An increasing time delay for the velocity front is observed for each case. Correspondingly, a decreasing time delay for the minimum velocity was observed for each case. VWS was computed as the slope of a linear fit for both the maximum and minimum velocity time points, which correspond to systole and diastole. The VWS was 12.7 and 7.2 m/s (standard deviation 13 and 4 m/s) during systole and diastole, respectively. Only two [Fig. 2(a) and (b)] of the three cases were statistically significant ($p < 0.05$) for the systolic VWS, and none was statistically significant for diastole. Thus, the maximum and minimum velocity time points did not provide a good measure of VWS using linear regression.

Fig. 3(a)–(c) shows the velocity temporal gradient for each patient as a function of time and position for the velocity data in Fig. 2(a)–(c). Similar to the velocity plots, the velocity gradient magnitude is indicated by the grayscale, white and black representing acceleration and deceleration during the cardiac cycle. A second series of plots identifies the time points at maximum acceleration and maximum deceleration during the systole [Fig. 3(d)–(f)]. VWS was computed as the slope of a linear fit for both the maximum and minimum velocity gradient time points, which correspond to systolic acceleration and deceleration. VWS was 4.6 and 10.6 m/s (standard deviation 1.7 and 11.1 m/s) during systole and diastole, respectively. Assuming a linear fit for the data, all acceleration VWS values were statistically significant (average $R^2 = 0.244$, standard deviation 0.126, $p < 0.005$), while none of the deceleration VWS values was shown to be statistically significant (Tables I and II). The coefficient of determination (R^2) and p-value were calculated based on a linear least squares regression fit for each dataset. These values were computed using MATLAB software with a function called mregress.m written by G. A. Reina (The Neurosciences Institute). The results for all cases are detailed in Tables I and II in which it is clear that only the maximum velocity gradient, corresponding to systole, produces statistically significant values of VWS.

TABLE I
CSF VWS IN SPINAL CANAL COMPUTED FROM PEAK VELOCITY METHOD AND PEAK GRADIENT OF VELOCITY

	CSF Velocity Wave Speed (m/s)			
	Peak Velocity		Peak Velocity Gradient	
Case	Systolic	Diastolic	Acceleration	Deceleration
1	6.9	3.2	4.4	1.6
2	3.7	11.2	3.0	7.2
3	27.7	7.2	6.3	23.0
Avg.	12.7	7.2	4.6	10.6
Stdev	13.0	4.0	1.7	11.1

TABLE II
LINEAR REGRESSION OF PEAK VELOCITY AND VELOCITY GRADIENT DATA

	Linear regression – R^2 [P-value]			
	Peak Velocity		Peak Velocity Gradient	
Case	Systolic	Diastolic	Acceleration	Deceleration
1	0.0628 [p<0.05]	0.0183 [N.S.]	0.3815 [p<0.005]	0.0307 [N.S.]
2	0.1345 [p<0.005]	0.0080 [N.S.]	0.2179 [p<0.005]	0.0039 [N.S.]
3	0.0025 [N.S.]	0.0049 [N.S.]	0.1329 [p<0.005]	0.0011 [N.S.]

TABLE III
SUMMARY OF WAVE SPEED MEASUREMENTS IN SPINAL SAS

Researcher	Year	Wave Speed	Technique
Williams	1976	13.5 m/s	pressure measurement in lumbar and cisternal CSF during cough
Greitz et al.	1999	4 m/s	MR velocity measurement
Carpenter et al.	2003	4 m/s	theoretical model
Bertram et al.	2005	12.2 m/s	computational model
Martin et al.	2005	2 - 26 m/s	compliant in vitro syringomyelia model
Present study	2008	4.6 m/s	in vivo cine MR using maximum velocity gradient

IV. DISCUSSION

VWS was computed for the pressure wave originating in the cranial compartment during the cardiac cycle. The slope of the peak velocity measurements during systole was positive, indicating that the wave was traveling caudally. This is the expected velocity wave direction since the CSF is moving caudally during systole, i.e., toward the tailbone. During diastole, the flow is reversed and the pressure in the CSF system of the spine is lower. Due to the nonlinear stiffening properties of the spinal tissues, the transient state of lower pressure results in a higher system compliance during diastole, which manifests itself in a lower wave speed through the system. However, the results during diastole in this study were not statistically significant, and thus, we cannot confirm VWS variation during cardiac cycle. Statistically significant data were obtained only for the wave traveling caudally using the maximum velocity gradient (during systole) as a time point marker.

Estimates of the VWS in the spinal SAS are few and the reported values vary significantly (Table III). The first measurement of wave speed was by Williams in 1976 via direct puncture into the lumbar and cervical spinal canals [2]. Phase lag was recorded after the subject was asked to cough, thereby obtaining an approximate wave speed of 13 m/s. Carpenter *et al.* noted that the pressure pulse likely originated between the lumbar and cisternal regions, which means that the assumption that the wave traveled from the lumbar to cervical region may not be correct [3]. Thus, the distance traveled was potentially shorter than assumed, which would mean that the wave speed was likely smaller than reported, and in Carpenter *et al.*'s assessment, ". . . the true value lay in the range of 4 to 5 m/s" [3]. This value is in agreement with Carpenter *et al.*'s computed value from numerical simulations. In 2005, Bertram *et al.* also developed a numerical model of the spinal fluid system and computed wave speed to be approximately 12 m/s [12]. However, Carpenter *et al.* and Bertram *et al.* assumed elastic modulus values and dimensions for the various tissues, which will likely have a significant impact on the computed CSF VWS magnitude.

Using an *in vitro* model representative of syringomyelia, Martin *et al.* demonstrated a variation in wave speed through the cardiac cycle (2–26 m/s) [5]. Because of the linear stiffening of the physical model's spinal cord, this VWS variation during the CSF flow cycle could be different *in vivo*. This study computed the wave speed within a model of the spinal SAS in a two-part experiment. First, MR was used to measure velocity and compute the flow waveform in the model. In a separate experiment, unsteady pressure measurements were obtained in the same model in the laboratory. These values were then used to compute the wave speed employing a modified form of the Moens–Korteweg equation. In 1999, Greitz *et al.* reported the CSF VWS of approximately 4 m/s [4]. This value matches the values from the numerical studies [3], [12]; however, details of how Greitz *et al.* obtained this value are not fully explained.

Thus, the paucity of *in vivo* VWS measurements in the spinal CSF space is indicative of the difficulty of this measurement. Numerical models can provide estimates of wave speed; however, the difficulty of determining tissue elasticity properties as well as accurate dimensions diminishes the reliability of these estimates. The *in vitro* study by Martin *et al.* was performed on a simplified flow model rather than a patient, which puts the VWS values in question.

This novel MR methodology with in-plane velocity encoding provides sufficient spatial and temporal resolution for CSF axial velocity such that the VWS can be detected by tracking the peak velocity gradient during systolic acceleration. The statistically significant values of VWS are similar in magnitude to previously published values. The MR method and image processing techniques could easily be incorporated into clinical protocols if found to have clinical relevance. For example, large values of VWS may provide an indication of elevated intracranial pressure. Chiari I malformation may produce an altered pressure environment that is detectable by this methodology as well. Further research is necessary to determine the accuracy of this method using *in vitro* models, possible accelerations of image acquisition, and the clinical importance of VWS in the spinal SAS and its relation to PWV.

It should be noted that one subject [Figs. 2(c) and 3(c)] had two disks that completely blocked CSF flow on the anterior side. In addition, there appeared to be artifacts due to subject motion during the MR scan. Despite these complications, the velocity gradient method obtained a value of VWS for this subject. This may indicate robustness of the VWS measurement. However, a large sample population and *in vitro* validation will be required to accurately determine the efficacy of this methodology. In addition, the data processing technique could be improved through optimization of correlation parameters.

V. CONCLUSION

We present a technique to determine *in vivo* CSF VWS, a potentially clinically useful parameter, to help understand the biomechanical environment of the spinal SAS. The methodology requires verification through phantom model studies and optimization of data processing.

Nonetheless, the technique shows good promise as it provides robustness, since it factors in many axial positions along the spinal canal.

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REFERENCES

[1] X. Zhang and J. F. Greenleaf, "Noninvasive generation and measurement of propagating waves in arterial walls," *J. Acoust. Soc. Amer.*, vol. 119, pp. 1238–1243, 2006.

[2] B. Williams, "Cerebrospinal fluid pressure changes in response to coughing," *Brain*, vol. 99, pp. 331–346, 1976.

[3] P. W. Carpenter, K. Berkouk, and A. D. Lucey, "Pressure wave propagation in fluid-filled co-axial elastic tubes. Part 2: Mechanisms for the pathogenesis of syringomyelia," *J. Biomech. Eng.*, vol. 125, pp. 857–863, 2003.

[4] D. Greitz, K. Ericson, and O. Flodmark, "Pathogenesis and mechanics of spinal cord cyst. A new hypothesis based on magnetic resonance studies of cerebrospinal fluid dynamics," *Int. J. Neuroradiol.*, vol. 5, pp. 61–78, 1999.

[5] B. A. Martin, W. Kalata, F. Loth, T. J. Royston, and J. N. Oshinski, "Syringomyelia hydrodynamics: An *in vitro* study based on *in vivo* measurements," *J. Biomech. Eng.*, vol. 127, pp. 1110–1120, 2005.

[6] E. Laffon, R. Marthan, M. Montaudon, V. Latrabe, F. Laurent, and D. Ducassou, "Feasibility of aortic pulse pressure and pressure wave velocity MRI measurement in young adults," *J. Magn. Reson. Imag.*, vol. 21, pp. 53–58, 2005.

[7] W. J. Rogers, Y. L. Hu, D. Coast, D. A. Vido, C. M. Kramer, R. E. Pyeritz, and N. Reichek, "Age-associated changes in regional aortic pulse wave velocity," *J. Amer. Coll. Cardiol.*, vol. 38, pp. 1123–1129, 2001.

[8] F. Wiesmann, S. E. Petersen, P. M. Leeson, J. M. Francis, M. D. Robson, Q. Wang, R. Choudhury, K. M. Channon, and S. Neubauer, "Global impairment of brachial, carotid, and aortic vascular function in young smokers: Direct quantification by high-resolution magnetic resonance imaging," *J. Amer. Coll. Cardiol.*, vol. 44, pp. 2056–2064, 2004.

[9] S. W. Fielden, B. K. Fornwalt, M. Jerosch-Herold, R. L. Eisner, A. E. Stillman, and J. N. Oshinski, "A new method for the determination of aortic pulse wave velocity using cross-correlation on 2D PCMR velocity data," *J. Magn. Reson. Imag.*, vol. 27, pp. 1382–1387, 2008.

[10] X. Shao, D. Y. Fei, and K. A. Kraft, "Rapid measurement of pulse wave velocity via multisite flow displacement," *Magn. Reson. Med.*, vol. 52, pp. 1351–1357, 2004.

[11] M. Tarnawski, G. Cybulski, D. Doorly, C. Dumoulin, R. Darrow, and C. Caro, "Noninvasive determination of local wave speed and distensibility of the femoral artery by comb-excited Fourier velocity-encoded magnetic resonance imaging: Measurements on athletic and nonathletic human subjects," *Heart Vessels*, vol. 9, pp. 194–201, 1994.

[12] C. D. Bertram, A. R. Brodbelt, and M. A. Stoodley, "The origins of syringomyelia: Numerical models of fluid/structure interactions in the spinal cord," *J. Biomech. Eng.*, vol. 127, pp. 1099–1109, 2005.