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Are intracranial pressure wave amplitudes measurable through lumbar puncture?

Behrens A, Lenfeldt N, Qvarlander S, Koskinen L-O, Malm J, Eklund A. Are intracranial pressure wave amplitudes measurable through lumbar puncture?
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Objective – The aim of this study was to investigate whether pulsations measured in the brain correspond to those measured in lumbar space, and subsequently whether lumbar punctures could replace invasive recordings. *Methods* – In ten patients with normal pressure hydrocephalus, simultaneous recordings of the intracranial pressure (ICP; intraparenchymal) and lumbar pressure (LP; cerebrospinal fluid pressure) were performed. During registration, pressure was altered between resting pressure and 45 mmHg using an infusion test. Data were analyzed regarding pulsations (i.e., amplitudes). Also, the pressure sensors were compared in a bench test. *Results* – The correlation between intracranial and lumbar amplitudes was 0.98. At resting pressure, and moderately elevated ICP, intracranial pulse amplitudes exceeded that of lumbar space with about 0.9 mmHg. At the highest ICP, the difference changed to −0.2 mmHg. The bench test showed that the agreement of sensor readings was good at resting pressure, but reduced at higher amplitudes. *Conclusions* – Compared to intracranial registrations, amplitudes measured through lumbar puncture were slightly attenuated. The bench test showed that differences were not attributable to dissimilarities of the sensor systems. A lumbar pressure amplitude measurement is an alternative to ICP recording, but the thresholds for what should be interpreted as elevated amplitudes need to be adjusted.

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Key words: intracranial pressure; spinal puncture; cerebrospinal fluid pressure; hydrocephalus; normal pressure; pulse pressure waves

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Introduction

Arterial blood pressure pulsations with high amplitudes are an important risk factor in cardiovascular disease (1–3). Most likely, large cardiac-related intracranial pressure pulsations are also involved in many neurological diseases, for example stroke, vascular dementia, and hydrocephalus. There is also an emerging concept of 'pulse wave encephalopathy', describing how altered pulse waves damage the brain (4–6). However, their impact on the brain is difficult to study for a neurologist, as it involves invasive procedures, that is, implanting a pressure sensor into the brain parenchyma. This is a well-developed technique although associated with risks (7, 8). It would be of clinical advantage if the arterial pulsations of the brain, reflected by intracranial pressure (ICP) pulsations, could be studied using a standard lumbar puncture.

There is an increasing interest in the pulsatile nature of ICP in the field of Normal Pressure Hydrocephalus (NPH). NPH is a syndrome comprised of the symptom triad of impaired gait, dementia, and urinary incontinence, in addition to the findings of enlarged ventricles and disturbed cerebrospinal fluid dynamics. The treatment for the condition is implantation of a ventricular shunt. In the search of sensitive and specific shunting criteria, there is a development of indications based on the amplitude of the intracranial pulse pressure (9–11).

Intracranial pressure measured via lumbar space agrees with ICP in brain tissue (12). However, reports on agreement between lumbar and

intracranial pulse pressure amplitude are scarce (13–15). Accessibility of this parameter via the lumbar route would give the neurologist a diagnostic tool to investigate CSF disturbances beyond the level of the mean pressure. A prerequisite is, however, that the lumbar readings reflect true ICP pulsations. The aim of the study is to determine whether ICP waves are correctly measured through lumbar puncture. To account for hardware differences, the sensor devices were compared in a bench test.

Material and methods

Clinical material

The study was based on 10 patients with idiopathic NPH. Their mean age was 72.4 years, and they had symptoms of gait disturbance, memory deficiency, and urinary incontinence. MRI showed communicating hydrocephalus without aqueductal stenosis and no significant ischemic or white matter lesions. All participants were considered to be in a mental condition compatible with giving informed consent, and a written consent was collected from all participants. The Regional Ethical Review Board in Umeå approved the study. The study is registered in ClinicalTrials.gov no: NCT01374048.

Pressure measurement and data sampling

The study design has previously been described (12, 16, 17), and it involves recording ICP using an intraparenchymal catheter tip sensor (Codman MicroSensor^{TM} Johnson & Johnson Professional, Raynham, MA, USA) inserted into the roof of the right ventricle. Surgery for implantation of catheters was performed under general anesthesia and performed by either one of two neurosurgeons. Overnight ICP registrations were performed, and the next morning, a CSF infusion test was performed with simultaneous recordings of ICP and lumbar space pressure (LP). LP was measured in lumbar subarachnoid space using an 18-G needle connected via a catheter to a transducer (Becton Dickinson, Franklin Lakes, NJ, USA). The lumbar measurement system was calibrated to zero, at the midpoint between the highest and lowest points of the patient's head in the supine position. A second needle was inserted next to the pressure measurement needle for pressure control by CSF volume alteration, using an infusion apparatus (12, 16, 17). Pressure data were sampled at 100 Hz using an acquisition card (MIO16X50; National Instrument, Austin, TX, USA) and recorded on a computer. During measurements, patients were awake and positioned in the supine position. After recording the baseline (or resting) pressure, the pressure was elevated and kept constant on two (three patients) or three (seven patients) excess pressure levels (Fig. 1). After finishing the top level of 45 mmHg, the pressure was released to baseline and the session finished with drainage to zero ICP. Five to six pressure levels were obtained per patient (Fig. 1), resulting in a total of 57 pressure levels for the 10 patients. Because of the noisy signal at the drain level, these data were omitted, resulting in 47 pressure levels used in the analysis. After data acquisition, computed tomography was performed to rule out complications.

Bench test

The characteristics of the pressure measurement systems were compared in a bench test. Pressure waves were generated in a fluid filled pressure chamber by connecting an external voltage signal to a pressure wave generator (Model 601A Blood pressure system calibrator, Biotech Instruments, Inc., Burlington, VT, USA). Data were sampled at 100 Hz using the same setup as in the *in vivo* experiments.

The pressure signals were modeled after a physiological signal. A typical cardiac-related pulsation was chosen from the baseline registration of intracranial pressure of one patient in the *in vivo* experiment. The original pulsation had a mean value of 17.6 mmHg, amplitude of 6.3 mmHg, and a heart rate of 86 beats per minute (bpm) (1.4 Hz). These values are representative of the patients. To test amplitude dependence of the pressure measurement systems, this pulsation was rescaled to amplitudes ranging from 1 to 25 mmHg in steps of 1 mmHg, with an additional low value of 0.5 mmHg. Similarly, to test for frequency dependence, the experiment was repeated for resampled signals of 1 and 2 Hz (corresponding to heart rates of 60 and 120 bpm). To test for mean pressure dependence, measurements with a 1.4-Hz signal using constant amplitude of 3 mmHg and varying mean pressure from 0 to 21 mmHg (3 mmHg increments) were performed. The complete protocol was repeated six times with new sensors in each repetition.

Waveform analysis

A program for automatic waveform analysis was developed in Matlab® (Mathworks Inc., Natick, MA, USA). Two main methods of calculating pulse wave amplitudes (time and frequency domain) exist in the literature (18–21). Both were

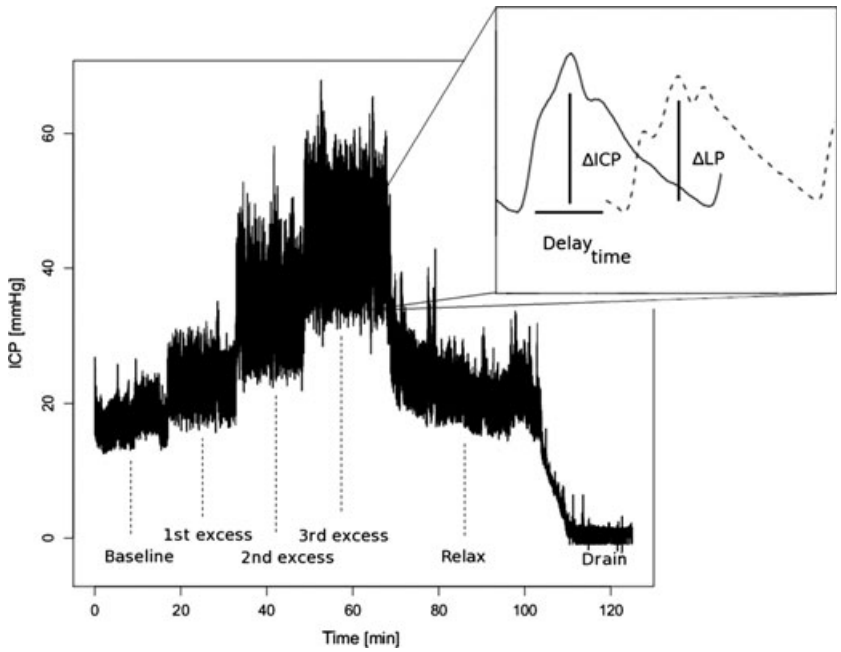

Figure 1. The intracranial pressure (ICP)-curve during Ringer infusion and drainage in one patient. First the baseline pressure was recorded. Then infusion was performed, keeping the mean pressure constant on two or three levels (1st, 2nd and/or 3rd excess phase). Then pressure was released to baseline (relax phase). The recording ended with CSF withdrawal to zero ICP (drain phase). A dashed lumbar CSF pressure (LP) curve is added for illustrative purposes. The computer algorithm finds the local minima and maxima for every pressure wave and calculates ICP amplitude (ΔICP), lumbar CSF pressure amplitude (ΔLP) and the delay between the diastolic peak on the ICP and LP curve ($delay_{time}$).

evaluated in this study, referred to as method 1 (time domain analysis) and method 2 (frequency domain analysis). In method 1, local maxima and minima, corresponding to the systolic and diastolic peaks of each pressure wave, were identified (Fig. 1). The program calculated the following parameters: ICP amplitude (ΔICP), lumbar CSF pressure amplitude (ΔLP), and time between intracranial and lumbar pulse ($delay_{time}$). The parameters are reported as means for each pressure level. In method 2 (frequency domain analysis), peak-to-peak amplitudes were calculated using the Fast Fourier Transform, as the amplitude of the fundamental frequency, corresponding to the heart rate (ΔICP_f, ΔLP_f) (18, 19). Also, amplitudes of the first and second over tones were identified (ΔICP_{OT1}, ΔICP_{OT2}, ΔLP_{OT1}, and ΔLP_{OT2}). The frequency analysis was performed on successive 6 s time windows and reported as means for each pressure level.

Statistics

The differences between lumbar and intracranial variables and bench-test variables were examined with Bland–Altman plots (22). As the normality constraint was not met, Spearman correlation was consistently used to calculate correlations between lumbar and intracranial variables. To explain the variation in the difference between the intracranial and lumbar variables, we applied a general linear model (GLM) that included mean ICP as a covariate ($k*ICP_{mean}$) and a patient-dependent factor (m_{Pat}). The general equation appeared as:

$$dA_i = k_i * ICP_{mean} + m_{Pat\_i}$$

Five equations were applied: Equation index $i = 1$, refers to method 1, and dA_1 equals $\Delta ICP - \Delta LP$, k_1 gives the slope of the regression line, and m_{Pat_1} relates to a patient-specific offset determining dA_1. The same reasoning can be transferred to the other equations, targeting amplitude difference using method 2 ($i = 2$), amplitude difference of first and second over tones (i = OT1 and OT2) and $Delay_{time}$ (i = DEL). Significance threshold was set to 0.05. Statistical analysis was performed using SPSS version 18.0 (SPSS Inc, Chicago, IL, USA).

Results

Amplitudes

As shown in Fig. 2, the correlation between ICP and LP amplitudes using both method 1 and 2

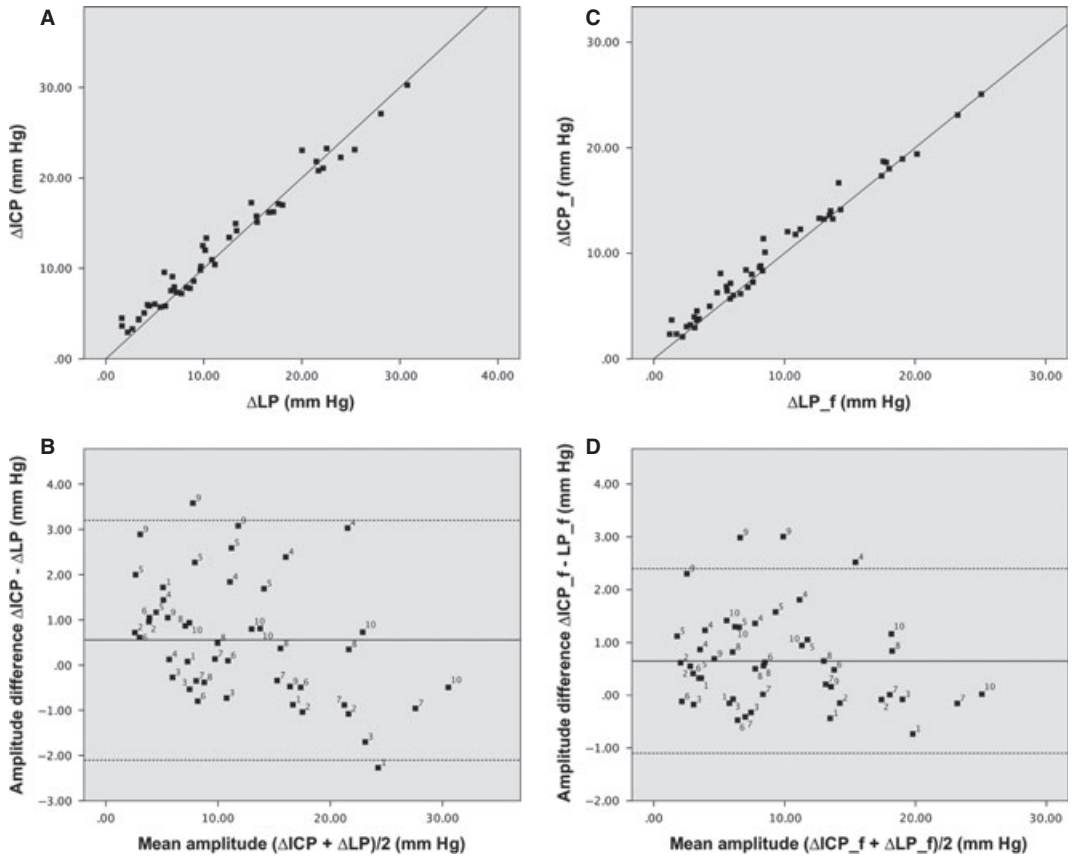

Figure 2. Agreements in the time and frequency domain analysis. ICP, intracranial pressure; LP, lumbar CSF pressure; ΔLP, lumbar CSF pressure amplitude; ΔICP, ICP amplitude; ΔLP_f, frequency domain lumbar CSF pressure amplitude; ΔICP_f, frequency domain ICP amplitude. (A) shows the agreement between ΔICP and ΔLP (method 1) ($r = 0.98$, $P < 0.001$). The line represents the entity line. (B) shows the amplitude difference as a function of mean amplitude ($r = -0.47$, $P < 0.01$). (C) and (D) shows the corresponding agreement in the frequency domain (method 2) (ΔICP_f and ΔLP_f) ($r = 0.98$, $P < 0.001$ and $r = -0.16$, $P = 0.27$). The numbers indicate patients.

was very high ($r = 0.98$, $P < 0.001$ and $r = 0.98$, $P < 0.001$). The Table 1 summarizes averages of the pulse pressure data on each pressure level.

At baseline, the amplitude difference (ΔICP–ΔLP) was 0.9 (SD = 1.0, $P < 0.05$) mmHg using method 1, and 0.7 (SD = 0.8, $P < 0.05$) mmHg using method 2. The corresponding results for the highest pressure level were −0.2 (SD = 1.6, $P = 0.6$) mmHg (method 1) and 0.4 (SD = 0.9, $P = 0.2$) mmHg (method 2). According to the Bland–Altman plots, the amplitude difference decreased with increasing amplitude using method 1 ($r = -0.47$, $P < 0.01$) (Fig. 2B), but not using method 2 ($r = -0.16$, $P = 0.27$) (Fig. 2D).

In the GLMs, the patient-dependent factors m_{Pat} represent individual offsets for every patient. To be precise, in equations indexed 1 and 2, m_{Pat} represents a theoretical individual amplitude difference (ΔICP-ΔLP) extrapolated to zero ICP. In method 1, the patient-dependent factor m_{Pat_1} significantly contributed to explain the variation in pulse pressure difference, with all values being positive (0.1–3.0 mmHg) and five of them being significantly different from zero (pat 4 = 2.7, pat 5 = 3.0, pat 8 = 1.4, pat 9 = 2.9, pat 10 = 1.5 mmHg). Using method 2, the patient-dependent factor was significant for the same patients (pat 4 = 1.6, pat 5 = 1.1, pat 8 = 0.7, pat 9 = 1.9, pat 10 = 1.0 mmHg). The slope was negative for method 1 ($k_1 = -0.033$, $P < 0.01$), but non-significant for method 2 ($k_2 = -0.002$, $P = 0.85$). However, looking at the overtones, the slopes were again negative ($k_{OT1} = -0.017$, $P < 0.01$ and $k_{OT2} = -0.015$, $P < 0.01$).

A Bland–Altman plot for the bench test using a frequency corresponding to 1 Hz (60 bpm) (Fig. 3A) shows that the amplitude difference ($ICP_{sensor} - LP_{sensor}$) was slightly positive at low amplitudes using method 1, and significantly negative at high amplitudes ($r = -0.73$, $P < 0.001$). In contrast, there was a positive correlation using method 2 ($r = 0.73$, $P = < 0.001$). For higher frequencies (84 and 120 bpm), the correlation in

Table 1 Average pulse pressure data on each pressure level.

Measure	Baseline Mean (SD), n = 10	1st excess Mean (SD), n = 8	2nd excess Mean, (SD), n = 9	3rd excess Mean (SD), n = 10	Relax (SD) Mean (SD), n = 10
No. pulses	649 (365)	450 (181)	515 (188)	581 (298)	581 (307)
Mean ICP (mmHg)	18.7 (5.1)	26.7 (3.4)	34.8 (3.7)	44.3 (5.8)	20.6 (3.9)
ΔLP (mmHg)	5.2 (3.5)	10.1 (3.2)	15.6 (4.7)	21.9 (5.4)	6.1 (2.4)
SD_{LP}	1.5 (1.8)	2.5 (1.9)	3.7 (2.6)	4.2 (2.6)	1.4 (0.7)
ΔICP (mmHg)	6.1 (3.1)	11.0 (2.5)	16.3 (3.9)	21.7 (4.8)	6.7 (1.8)
SD_{ICP}	1.4 (1.8)	2.5 (1.9)	3.6 (2.5)	4.0 (2.0)	1.3 (0.6)
ΔICP−ΔLP (mmHg)	0.9 (1.0)*	0.9 (1.6)	0.7 (1.6)	−0.2 (1.6)	0.6 (0.7)*
ΔICP_f (mmHg)	4.1 (1.8)	9.1 (2.5)	13.3 (3.3)	17.9 (4.1)	5.4 (1.9)
ΔICP_{OT1} (mmHg)	0.9 (0.3)	1.8 (0.9)	2.8 (1.5)	3.9 (2.0)	1.0 (0.5)
ΔICP_{OT2} (mmHg)	0.3 (0.3)	0.5 (0.4)	0.6 (0.5)	0.7 (0.7)	0.3 (0.2)
ΔLP_f (mmHg)	3.4 (2.0)	8.0 (2.6)	12.4 (3.8)	17.5 (4.5)	4.9 (2.0)
ΔLP_{OT1} (mmHg)	1.0 (0.6)	1.9 (0.8)	3.2 (1.6)	4.4 (2.1)	1.2 (0.6)
ΔLP_{OT2} (mmHg)	0.3 (0.1)	0.5 (0.3)	0.9 (0.7)	1.1 (0.9)	0.5 (0.5)
$\Delta ICP_f - \Delta LP_f$ (mmHg)	0.7 (0.8)*	1.0 (1.1)	0.9 (1.1)*	0.4 (0.9)	0.5 (0.5)*
$\Delta ICP_{OT1} - \Delta LP_{OT1}$ (mmHg)	−0.1 (0.5)	−0.1 (0.5)	−0.4 (0.6)	−0.6 (0.5)*	−0.3 (0.3)
$\Delta ICP_{OT2} - \Delta LP_{OT2}$ (mmHg)	0.1 (0.3)	−0.0 (0.3)	−0.3 (0.3)*	−0.4 (0.3)*	−0.1 (0.4)
$\Delta ICP - \Delta ICP_f$ (mmHg)	2.0 (1.9)*	2.2 (0.8)*	3.0 (1.1)*	3.8 (1.2)*	1.2 (0.3)*
$\Delta LP - \Delta LP_f$ (mmHg)	1.8 (2.3)*	2.1 (1.0)*	3.2 (1.2)*	4.4 (1.2)*	1.1 (0.4)*
$Delay_{time}$ (ms)	39 (26)*	29 (21)*	24 (17)*	18 (14)*	35 (22)*

ICP, intracranial pressure; ΔLP, lumbar CSF pressure amplitude; ΔICP, ICP amplitude; SD_{LP}, standard deviation of lumbar CSF pressure amplitude single waves; SD_{ICP} standard deviation of ICP-amplitude single waves; ΔLP_f, frequency domain lumbar CSF pressure amplitude; ΔICP_f, frequency domain ICP amplitude; Rise $time_{LP}$, time between local minima and maxima of the lumbar pressure pulse; Rise $time_{ICP}$, time between local minima and maxima of the ICP pulse; $Delay_{time}$, time between intracranial and lumbar pulse.

*Indicate significance on 5% level for a two tailed t-test.

method 1 was even more negative ($r = -0.86$, $P < 0.001$ and $r = -0.94$, $P < 0.001$) (Fig. 3B and C), and the maximum pulse pressure difference between the sensors increased from 0.4 mmHg at 60 bpm to 1.3 mmHg at 120 bpm. Using method 2, the correlation decreased at higher frequencies ($r = 0.46$, $P < 0.001$ and $r = -0.13$, $P = 0.11$). There was no dependence of amplitude difference on mean pressure in either method 1 or 2 ($r = -0.20$, $P = 0.63$ and $r = 0.12$, $P = 0.77$) (Fig. 3D).

Delay

The $delay_{time}$ was 39 ms for baseline pressure and decreased for higher pressures (see Table 1). The observation of a smaller delay at higher pressures was supported by the GLM for $delay_{time}$, where k_{DEL} was significant at −0.63 ms/mmHg ($P < 0.001$). The m_{Pat_DEL} was significant in this GLM as well, with all individual values being positive (30–81 ms) and significant.

Outcome

In a follow-up at three to 6 months after shunt surgery, gait ability had improved in nine patients. In one patient, no improvement was seen. No patients presented any clinical or radiological complications following the experimental procedure or the shunt surgery.

Discussion

Differences between pulse pressure amplitudes measured by intracranial and lumbar approach can originate from true physiological differences between the measurement sites and/or from differences in sensor systems detecting the amplitudes. In addition, the calculated amplitudes will depend on the signal analysis that is applied to the measured data. Amplitudes are typically determined either straightforward from the time series analysis where the pressure difference between peaks and troughs is detected or by frequency domain analysis were the amplitude of the fundamental frequency is determined with Fast Fourier Transform. In this study, we have, over a wide pressure interval, compared the direct intracranial pulse pressure amplitude in brain tissue, with the indirect measurement accessed through lumbar puncture. We found that intracranial pulse pressure was accessible through lumbar puncture, with a slight underestimation at baseline and moderately elevated pressure, which disappeared at the highest pressure level. To determine the contribution to measurement errors because of using different sensor systems, we per-

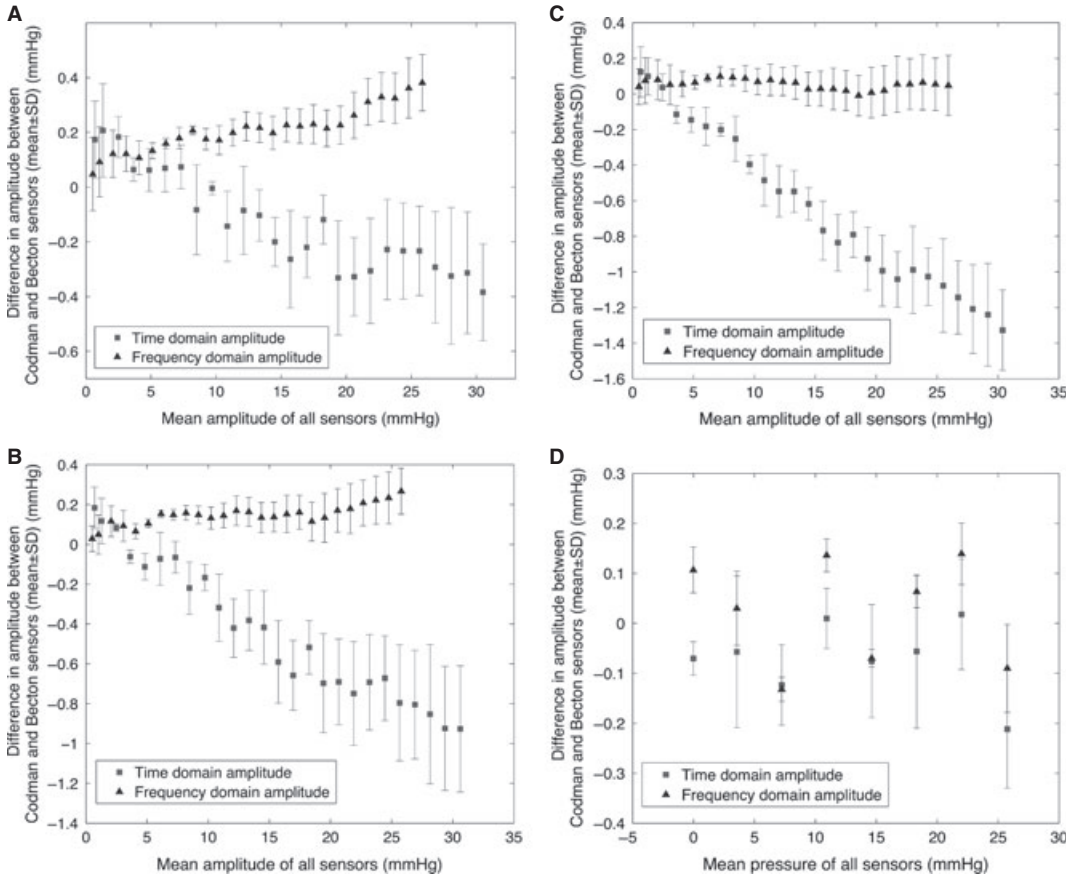

Figure 3. Bench test amplitude difference as a function of mean pressure amplitude and simulated heart rate. Both time domain (method 1) and frequency domain (method 2) data are shown. The data represent means for six repeated measurements with different sensors. Figure (A) shows the amplitude difference for the 60 beats per minute (bpm) simulated signal ($r = -0.73$, $P < 0.001$ and $r = 0.73$, $P < 0.001$). (B) shows the 86 bpm signal ($r = -0.86$, $P < 0.001$ and $r = 0.46$, $P < 0.001$) and (C) the 120 bpm signal ($r = -0.94$, $P < 0.001$ and $r = 0.13$, $P = 0.11$). (D) amplitude differences as a function of mean pressure ($r = -0.20$, $P = 0.63$ and $r = 0.12$, $P = 0.77$).

formed bench-test measurements. They revealed that differences were negligible at baseline pressures, but increased at higher amplitudes and frequencies, explaining why the *in vivo* amplitude differences decreased at these levels. At low ICP, the pressure waves were attenuated on their way down through the spinal canal, and for clinical purposes, this error should be accounted for. At elevated ICP, the amplitude differences were small enough to be of no clinical relevance. The frequency domain analysis (method 2) omitted the higher frequencies and therefore produced lower pulse amplitudes, which for a physiological analysis give an underestimation of the biomechanical load on the tissue.

Pulsatility in disease

Cardiac-related pulsatility exerts cyclic stress on tissue and contributes to diseases like atherosclerosis (23), macular degeneration (24), renal microvascular injury (25), and liver cirrhosis (26). In the brain, abnormal pressure pulsations have been linked to cerebral white matter changes (4) and may be a better marker of severity and outcome than mean ICP in conditions like normal pressure hydrocephalus (NPH) (27), traumatic brain injury (28), subarachnoid hemorrhage (29), and hydrocephalus in children (30). In addition, there is an emerging concept of cerebral pulse wave encephalopathy, linking diseases like vascular dementia, Alzheimer's disease, and NPH (5, 6, 31). The utility of amplitude measurements is particularly strong in the field of NPH, where invasive monitoring of ICP amplitude is used to predict shunt response (20). A more accessible method to study intracranial pulse pressure would benefit both research and clinical management.

Pulse pressure amplitudes

As anticipated, there was a high correlation between ICP and LP amplitudes using both method 1 and 2 (Fig. 2A and C). Still, at baseline

pressures, lumbar amplitudes were lower (Table 1). In the bench test, it was observed that measurement differences from both method 1 and 2 were negligible at baseline pressure, so clinical differences were physiological. The underestimation compared to intraparenchymal measurements would mean that any limit for 'elevated ICP amplitude' would have to be adjusted, if measured via lumbar space. This would have clinical relevance in, for example, selection of shunting candidates among NPH patients, where such limits are being used today (20).

At the highest excess level, the differences in amplitudes using method 1 decreased, also supported by a significant negative correlation between amplitude difference and mean amplitude in the Bland–Altman plot (Fig. 2B), and a negative slope in the corresponding GLM ($k_1 = -0.033$, $P < 0.01$). These changes were not observed for pulse wave amplitudes in method 2 [non-significant correlation in Fig. 2D, and corresponding GLM ($k_2 = -0.002$, $P = 0.85$)]. Instead, the changes were present in the consecutive two overtones of the signal ($k_{OT1} = -0.017$, $P < 0.01$; $k_{OT2} = -0.015$, $P < 0.01$). This can be explained by the results from the bench tests, where there was only a small amplitude difference using method 2, but using method 1, at high amplitudes, the lumbar amplitudes exceeded the intracranial ones (Fig. 3A). This effect was more pronounced at higher frequencies (Fig. 3B–C). Thus, it is likely that part of the amplitude difference in method 1 was attributed to differences in the characteristic of the measurement systems, distorting the higher frequency parts of the signal, and that these dissimilarities mask the physiological difference, especially at higher pressures. However, the effect was quite small when compared to the pressure amplitude. In the field of NPH, clinically relevant amplitude limits for selecting patients to operate are 4, 5, and 6 mmHg (27). Bench-test difference between sensor readings in this range is between 0.1 and −0.2 mmHg (Fig. 3) and can be regarded as negligible. The highest measured amplitudes in the patients were about 20 mmHg, and here the difference between sensors increased to about −1 mmHg. Contrary to our findings, Eide and Brean found that the difference between intracranial and lumbar pulse amplitudes increased after elevating the CSF pressure (13). The discrepancy in results may be related to different measurement systems used for the lumbar measurements.

In the general linear models, the factors m_{Pat} can be considered to represent patient-specific differences in intracranial and lumbar amplitudes. Importantly, these factors were all positive in method 1 (eq. indexed 1), and they remained positive, but smaller, in method 2 (eq. indexed 2). The positive m_{Pat} fits with the earlier stated general lumbar underestimation of the pulse pressure. The variability of m_{Pat} values is likely due to patient-specific differences in physiological characteristics. On top of this, there are also differences within individual patients, with changes in amplitude difference of up to 2 mmHg in spite of only small changes in mean amplitude (e.g., patient 9 Fig. 2B–D), which are more likely attributed to measurement-related variability.

Physiological background

The physiological background to the lumbar underestimation is likely dampening of the pulse wave traveling down the spinal canal. This was also supported by the delay of the lumbar pressure wave compared to the intracranial (Table 1), a delay decreasing with increasing pressure according to the general linear model ($k_{DEL} = -0.63$ ms/mmHg). This observation is likely related to the stiffer craniospinal compartment at higher pressures, increasing wave velocity. The dampening depends on factors like Pressure Volume Index (32), as well as length and cross-sectional geometry of the spinal canal. Further studies are needed to investigate the relation between the amplitude difference and the physiological features of the spinal canal, for instance the presence of stenosis.

Frequency vs time domain amplitudes

As method 2 omits higher frequency data, which normally contributes to the amplitude, its amplitudes appear smaller. The degree of underestimation compared to the time domain amplitudes depends on the amplitude of the higher frequencies in the pulse wave (giving it a 'peaky' waveform) and varies between patients (27). The differences are quite large, between 2 and 4 mmHg (Table 1), clearly demonstrating that comparison of pulse pressure amplitudes in method 1 and method 2 is not viable. This was also acknowledged in a previous study (27). Including higher harmonics in the analysis will relieve the discrepancy.

Conclusion

Intracranial pressure waves were measurable through lumbar puncture. There was a difference in amplitudes from ICP and lumbar measurements

at low pressures, likely due to dampening of pulse waves in the spinal canal. Measurement system discrepancy was present mainly at the higher pulse pressures and was statistically but not clinically significant in our setting. Amplitudes calculated from frequency transformations based only on the fundamental frequency underestimate the amplitudes and do not assess the real biomechanical strain on tissues. Hence, we advocate analysis of data in the time domain and emphasize the necessity of understanding the frequency characteristics of the sensor system used. Lumbar pressure amplitude measurement with fluid catheter is an alternative to direct ICP measurement with a catheter tip sensor, but the thresholds for what should be interpreted as elevated amplitudes need to be adjusted.

Acknowledgements

We would like to give special thanks to research nurse Doris Kjellgren for all her assistance during the study.

Conflicts of interests and sources of funding

Niklas Lenfeldt has worked as a consultant for Likvor AB; a company that has commercialized the apparatus investigating CSF dynamics in this study. Jan Malm is listed as an inventor on a patent regarding CSF dynamic investigation apparatus, for which he has received royalties from Likvor AB. Anders Eklund has received honorary for lecturing from DePuy, Inc. and has patent interest regarding CSF dynamic investigation apparatus, for which he has received royalties from Likvor AB. Anders Behrens, Sara Qvarlander, and Lars-Owe Koskinen report no disclosures.

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