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An Impedance Sensor to Monitor and Control Cerebral Ventricular Volume

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Abstract

This paper presents a sensor for monitoring and controlling the volume of the cerebrospinal fluid-filled ventricles of the brain. The measurement principle of the sensor exploits electrical conductivity differences between the cerebrospinal fluid and the brain tissue. The electrical contrast was validated using dog brain tissue. Experiments with prototype sensors accurately measured the volume content of elastically deformable membranes and gel phantoms with conductivity properties made to match human brain. The sensor was incorporated into a fully automatic feedback control system designed to maintain the ventricular volume at normal levels. The experimental conductivity properties were also used to assess the sensor performance in a simulated case of hydrocephalus. The computer analysis predicted voltage drops over the entire range of ventricular size changes with acceptable positional dependence of the sensor electrodes inside the ventricular space. These promising experimental and computational results of the novel impedance sensor with feedback may serve as the foundation for improved therapeutic options for hydrocephalic patients relying on volume sensing, monitoring or active feedback control.

Keywords

Cerebrospinal fluid; CSF; Pressure Shunts; Volume Sensor; Hydrocephalus

1. Introduction

Cerebrospinal fluid (CSF) is a colorless fluid with a consistency similar to that of blood plasma. CSF is secreted from the arterial blood in the choroid plexus and the brain parenchyma. It circulates through the cerebral ventricular and subarachnoid spaces and is reabsorbed into the venous blood of the sagittal sinus. When the normal CSF pathways are obstructed, CSF accumulation causes the enlargement of cerebral ventricles. The ventricular enlargement is often accompanied by increased intracranial pressures that lead to severe medical impairments such as the compression of brain, atrophy of the neural tissues or impaired blood flow. Acutely, it may even cause coma or death [1]. Hydrocephalus is a serious health problem for infants, adults, and the elderly with a treatment cost of around \$1 billion per year in the United States

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alone [2]. Over 150,000 people are diagnosed with hydrocephalus each year in the U.S. with more than 25,000 patients undergoing shunting operations. Shunting involves the removal of the excess CSF in the ventricles using a catheter; unfortunately, it often fails. The average functional period of a shunt is only five years, even though patients need them for their entire life. In children, the failure rate of shunts was 50% over five years. In adults, the reported complication rate is almost 35% with a 2.7% mortality rate [3]. These statistics show a clear need for improvements to hydrocephalus treatment.

Shortcomings of Existing Hydrocephalus Treatment

Existing shunt treatment for hydrocephalus involves the removal of excess CSF from the ventricles using a pressure valve [4]. When the intracranial pressure (ICP) rises above a preset level, a pressure-sensitive valve opens thus allowing the CSF to drain from the ventricles through a catheter into the peritoneal cavity. Pressure shunts only open at elevated ICPs. However, the ICP varies widely according to changes in the body position and activity. In addition, some patients suffer from enlarged ventricles despite ICP's within normal ranges. This syndrome is known as *normal pressure hydrocephalus*. Moreover, the lack of correlation between the ICP and enlarged ventricles may be responsible for the overdrainage or underdrainage of CSF that often occurs with existing shunts. Overdrainage can lead to the collapse of ventricular spaces caused by excessive CSF removal. Underdrainage occurs due to insufficient CSF efflux thus increasing the ventricles again. In light of these shortcomings, it is plausible to consider measuring the ventricular volume in addition to pressure measurements. The objective of this work is to demonstrate the feasibility of measuring ventricular volumes directly based on differences in electrical conductivity between the CSF and brain tissue. *Direct volume measurements* would also permit active volume control using feedback principles.

The paper is organized as follows. A protocol to demonstrate the principle of the fluid volume sensor including volume measurements on brain surrogates and preliminary feedback control are outlined in section two. The electric field properties of the multidimensional CSF-filled brain cavities are analyzed with rigorous finite element analysis. The results of the conductivity experiments are provided in section three. Section four discusses experimental sensor tests and critically assesses performance in light of the requirements for a novel therapy. The validation in section five examines size changes and sensor positional dependence in real human brain geometries, as a function of electric field and sensor configurations simulated with a rigorous finite element model.

2. Methods

2.1 Measurement Principle of the Sensor

Fig. 1 shows the vision of a new hydrocephalus treatment based on direct volume measurements. A conductance catheter to measure the size of a fluid-filled cavity consists of excitatory and measurement electrodes. An alternating current is introduced through the excitatory electrodes, thus creating an electric field extending in the fluid-filled cavity as well as the surrounding tissue. The measurement electrodes are positioned at a fixed distance of each other and detect voltage drops across the fluid-filled space. If the conductivity of the surrounding tissue is sufficiently different from the conductivity of the fluid in the cavity, then the detected voltage drop is a function of the fluid-filled space. Dynamic volume increase can be detected as decreasing voltage drops across the measurement electrodes [5]. This principle for measuring cerebral ventricular volume is similar to the principles of the cardiac conductance catheter [6] - [11]; a difference in that our sensor is intended for control of a cavity size, not only measurement.

The volume of CSF-filled ventricles, V_f , is directly related to the measured voltage difference, ΔU , as in Eq. (1) with α , a calibration factor; σ_{csf} the specific conductivity of CSF; L , the distance between the electrodes; I , the current emitted; G_{brain} , the conductance of brain tissue; and G_{CSF} , the conductance of CSF [6], [7].

$$V_f = \alpha \cdot \sigma_{csf} \cdot L^2 \cdot (G_{CSF} - G_{brain}) = \alpha \cdot \sigma_{csf} \cdot L^2 \cdot \left(\frac{I}{\Delta U} - G_{brain} \right) \quad (1)$$

A similar principle known as the *electrical impedance tomography* is a technique for detecting brain conductance changes to map brain function [12]-[14]. However, despite earlier efforts in conductance measurements, to date no-one has successfully implemented a sensor for direct volume measurement of the CSF-filled cerebral ventricles surrounded by brain tissue.

2.2 Three Dimensional Conductance-Volume Measurements

In order to test sensor performance in realistic ventricular geometries, we created surrogate brain phantoms using three-dimensional molds of agarose gel (Sigma Aldrich) designed to match the conductance properties of brain tissue and filled them with artificial CSF. By adding salt, the formulation of the gel was adjusted to closely match the electrical conductivity of brain tissue. Four molds of different size were created to cover the expected range of ventricular enlargement. The silicone-insulated catheter held a sensor consisting of four equidistant platinum/iridium electrodes separated by a distance of 1.5 mm. The prototype sensor specifications are summarized in Table 1. The outer pair were excitatory; the inner pair served as measurement electrodes. The alternating current source consisted of a function generator (BK Precision) in series with a 10 k ohm resistor, as shown in Fig. 2a. This signal was connected to the inverting input of an operational amplifier (Texas Instruments), which maintained the amplitude constant. A wider frequency range was tested but reported results used a frequency at 1 kHz [15]. The actual current across the excitatory electrodes was verified by measuring the voltage drop across the resistor. The voltage drop of the measurement electrodes was detected and sent to an instrumentation amplifier (Analog Devices) with a gain of 124 dB. This signal was also connected to an oscilloscope for voltage measurements. Volume sensing experiments were conducted within safe levels of current stimulation without causing neuron activation or hydrolysis [14].

2.3 Experimental Setting for the Feedback Control System

The feasibility of directly maintaining and restoring normal ventricular sizes was tested by means of a closed loop feedback system with the volume sensor and a mini-pump actuator (Mk III, Hobbico Inc., Champaign, IL). The equivalent circuit diagram is shown in Fig. 2b. Input voltages ranging from 1-5 V with a frequency of 100 kHz generated an electric field of ~0.0025 V/m in the fluid space between the two measurement electrodes. 100 cc of artificial CSF were initially filled into an elastic rubber balloon; a baseline reading for the calibration of the sensor was taken. Artificial CSF in the range of 5-350 cc volume was added by a second miniature pump to emulate CSF accumulation leading to ventricular expansion. The sensor processed voltage data with a sampling rate of four readings per second to the controller. The digital controller compared the actual volume with a desired set point to compute the desired control action. The controller output operated the pump actuator. Several control schemes including proportional, proportional integral, as well as on-off control were implemented with the *virtual instrument* panel of LabVIEW (National Instruments).

2.4 Computer-Aided Sensor Optimization

In order to estimate leakage currents and optimal distance of the sensor electrodes as well as to assess the dependence of volume measurements as a function of catheter position, a rigorous

finite element analysis was conducted. The solution of the scalar field equations, $V(\vec{x})$, in the cavity and surrounding brain tissue allowed calculation of the equipotential lines and the prediction of the voltage drop at the measurement electrodes for any geometry of the cavities or different catheter configuration. In the simulations, electrical properties of fluid and brain tissue were taken from the experiments described above. The measured electric potential resided in a computational domain with the electrical conductivity of the CSF, $\sigma_1 = 2.01 \text{ S/m}$, and $\sigma_2 = 0.172 \text{ S/m}$ for the brain. Neumann boundary conditions as in eq. (2) and in Table 3 were applied with a current density of 16.66 A/m^2 at the upper excitatory electrode, A, while the lower electrode, D, was grounded. The measurement electrodes were represented as surfaces B and C; the measurement electrodes were considered perfectly insulated. The finite element model in eq. (2) was solved with ADINA [21].

$$\sigma \vec{\nabla}^2 V(\vec{x}) = 0$$

with $\vec{n} \cdot \vec{J} = \sigma \vec{\nabla} V(\vec{x}) = 16.66 \text{ A/m}^2$ at the excitatory electrode A
 $V=0$ at electrode D

(2)

3. Results

3.1 Static Conductance-Volume Measurements

Before animal or human experimentation, we wanted to optimize the electronic properties and design parameters of the novel volume sensor using expedient bench-testing on brain surrogates with realistic geometry and electronic properties equal to real brain tissue. In a series of experiments, we exactly determined the amount of salt needed to obtain the same contrast ratio of conductivity between artificial CSF and gel as found between artificial CSF and real brain tissue. To emulate the complex shape of real ventricles in human brains, we cast three-dimensional ellipsoidal cavities with realistic shapes depicted in Fig. 3.

3.2 Dog Brain Conductivity Experiments

Static experiments were performed to validate the sensor principle in real brain tissues. These experiments provided a means to validate the electrical conductivity ratio between CSF and brain tissue. In a set of separate experiments, the average conductivity of CSF was found to be 2.01 S/m . This value is about *twelve times higher* than that of brain tissue. This contrast ratio between CSF and brain tissue was found to be in good agreement with published data [16]-[18].

3.3 Monitoring and Feedback Control of Ventricular Sizes

The sensor performance in dynamically enlarging elastic cavities was examined. We used elastic rubber balloons to monitor and control the exact dynamic enlargement in bench-scale experiments. Sensor calibration was performed through static conductance-volume experiments as described in section 2. Up to 100 cc of artificial CSF were added and the total volume of fluid was recorded and measured every five seconds. Through the LabVIEW control module, the sensor was able to measure and store the measurement during the fluid addition. Fig. 4a compares the volume inferred from the sensor reading with the actual fluid volume over a period of 100 seconds. Speed and accuracy of the sensor in detecting the volume change is suitable for active feedback control described next.

Closed-loop feedback control experiments were conducted to dynamically measure and actively control the size of distensible fluid-filled spaces with feedback control. The enlargement of the ventricle as it occurs in hydrocephalus was simulated as a disturbance caused by fluid addition. The controller was programmed to operate the pump to remove the

excess fluid and to restore the volume desired set point. The fluid volume measurements were automatically stored and sent to the controller four times per second. The actual volume of balloon cavity was also carefully measured to quantify the sensor performance and the controller response. The fluid volume was initially maintained at 150 cc for the first 10 seconds. This phase was followed by CSF accumulation leading to ventricular enlargement through the addition of 150 cc to 220 cc of fluid over the next 10 seconds. The controller was then switched on and its dynamic response activated the pump to restore the set point volume of 150 cc in about 20 seconds. A typical controller response is shown in Fig. 4b. The readings show a slight offset between the actual and measured values. The simple on-off controller also indicates a wavy response as expected. However, the desired set point volume is reached in spite of the measurement inaccuracy. For control of ventricular sizes in hydrocephalus, only the restoration of the desired ventricular size is important; larger measurement errors in high volume ranges do not affect the accuracy close to the set point.

3.4 Simulated Conductance-Volume Relationship

In order to better quantify the sensor sensitivity, we also simulated the electric field measured by our prototype sensor in our surrogate brain models. The simulated field potentials are reported in frames E-H of Fig. 3. Fig. 5 shows the comparison between experimental measurements and simulated voltage drops. These results show that the simulations agree with the experimental values. Both methods demonstrate that the conductance increases with volume. The variance is less than 2%. The trend also saturates at larger volumes as reported previously [11].

Sensor positional dependence was also investigated by repeating the calculations for potential differences at the measurement electrodes subject to horizontal displacements of the sensor as shown in Fig. 6. Even for extreme sensor dislocations, see position E, large ventricular volumes always fall into the high conductance range.

4. Discussion

Current hydrocephalus shunting valves open only with elevated intracranial pressures. Their high failure rate motivates research for alternative treatment options. We propose direct measurement of ventricular volume. The measurement principle for this novel technology requires an electrical conductivity difference between CSF and brain tissue. Our experiments demonstrated a conductivity ratio between CSF and brain tissue of twelve; this value is comparable to earlier literature results [16]-[18]. This contrast between CSF and brain tissue is deemed sufficient for effective volume sensing.

Our experimental setting tested the prototype sensor using ventricle-shaped geometries and specially formulated agarose gel filled with artificial CSF and similar contrast ratio as in the real brain. The conductance of the gel mimicking the soft brain tissue was obtained by varying the gel formulation by salt addition. The ease-of-use allowed us to cast realistic shapes for an 8-fold size change. In this range the sensor detected the volume with a variance less than 7.5 % (N=5). The dislocations in the experiments were deliberately caused by displacements in the order of 1 mm of the sensor position. This small positional dependence should be satisfactory for clinical use. We also confirmed the insensitivity to position of the measurement principle by a rigorous finite element analysis performed for the four cavities in Fig. 3, frames A-D. The calculation of the entire electric field induced by the excitatory electrodes produced voltage drops in the simulation that were very similar to the actual experimental measurements. The difference between the simulation and the real experiment was on the order of one standard deviation of the experimental error as shown in Fig. 5. These results support the use of computer simulation as a practical method to optimize the sensor parameters in a rational fashion.

Four intermediate ventricular sizes were chosen to capture changes from normal up to 8 times the normal ventricular volume. These volume ranges are clinically relevant for the human lateral ventricles [19]. For clinical treatment of hydrocephalus with feedback, the volume sensor should be sensitive enough to detect the onset of ventricular enlargement. A sensor based on feedback control need not track precisely the entire size range from a normal of 15 mL to a hydrocephalic case of 300 mL, because control intervention should prevent excessive enlargements of the patients' ventricles. Therefore, sensor signal saturation at the high end of volume increase does not invalidate the feasibility of the proposed feedback treatment option.

A major concern for proper sensor function is its positional dependence. The volume measurement should be largely insensitive to normal catheter positional changes inside the ventricles; reasonable dislocations should have a modest effect on the volume reading. A rigorous estimate of expected positional dependence of the volume sensor technology inside an expanding ventricle in our surrogates reported in Fig. 6 showed that positional dependence was mild even with a single pair of measurement electrodes. The sensor design utilizes a ring electrode, thus making insensitive measurements of directional changes in volume. This supports the mild dependence on catheter position. Initially, the sensor was placed in the center of the cavity [8]-[11]. The experiments were repeated by positioning of the sensor gradually closer to the cavity wall to assess positional dependency. Future sensor designs could warrant an array of planar measurement electrodes if directional changes in volume are of interest.

The envisioned feedback treatment was implemented conceptually to demonstrate the feasibility of maintaining desired fluid volume with the help of feedback based on dynamic volume sensing. A simple on-off control scheme was successful in restoring desired setpoint volume in response to CSF accumulation disturbance. In the experiment, CSF was inserted by a second pump to emulate pathological CSF accumulation causing hydrocephalus. Although the actual volume exhibited some fluctuation due to the simplicity of the control scheme, the sensor performed satisfactorily. The oscillations in the tracking response can be reduced easily by implementing more advanced control schemes.

Saturation effects can be drastically reduced by adding more measurement electrodes with advanced calibration; an effort we will investigate in future work. The feedback control experiments were performed with a higher excitation frequency than the agarose gel experiments. Frequency dependence was measured in previous studies, and a frequency that provided the largest conductance measurement was implemented in our balloon model. In the agarose gel model, we used a lower frequency due to the large conductivity ratio.

Our work is to improve existing therapy for hydrocephalic patients. However, there are limitations of our proposed technology. Our sensor is intended for implantation into a single ventricle. We have focused on a design specifically for the lateral ventricle and frontal horn. Surgeons may want to implant the sensor into a different part of the ventricular system that does not have as much expansion. The limitations on any particular site may be resolved with patient-specific sensor design based on CT and MRI data. Our device is intended to reduce the number of shunt revisions. Existing shunts may malfunction due to excessive CSF removal, causing over drainage with ventricular collapse. The proposed sensor needs to be tested clinically, to see if it prevents this from occurring.

In this article, we have assumed that the conductivity of CSF and brain tissue remain constant throughout hydrocephalus. However, CSF conductivity might change due to hemorrhage or infection. The large contrast between CSF and brain tissue will diminish the change in current density within the cavity if this occurs. There may be a small percentage of leakage current as discussed earlier, but our calculations suggests its impact to be small compared to the conductance measurement due to volume enlargement. Our finite element model enables us

to assess the effect of varying system parameters such conductivity properties or leakage scenarios on sensor accuracy.

This work has focused on volume changes that occur in the ventricles. In instances such as normal pressure hydrocephalus, the sensor may not be useful if the ventricles do not change size. Nevertheless, improvements to sensor design and instrumentation may allow for greater sensitivity. Potentially, the sensor might be able to detect periodic changes in the ventricular volume due to pulse pressure changes in the ventricles. This would reflect brain compliance, which changes in normal pressure hydrocephalus. Detecting of brain compliance changes might be another useful future application for the sensor technology introduced in this article.

5. Validation of Sensor Performance in Simulated Normal and Hydrocephalic Subjects

Even though at the early stages of the project we cannot justify experimentation on human subjects, we wished to assess the expected sensor performance and positional dependence in real human brain geometries. To realize this assessment without harm to patients, we performed a rigorous finite element analysis of the induced fields of different sensor configurations on real patient ventricular models. The computations helped optimize our sensor design parameters for best performance in hydrocephalus humans. Frame A of Fig. 7 shows normal ventricular spaces in a 32 year old normal subject. An image of a fully developed hydrocephalic brain is shown in Frame D. It was obtained from a 62 year old patient suffering from communicating hydrocephalus. The patient data were collected according to IRB protocol at the University of Illinois at Chicago and the University of Chicago. The images were reconstructed using image reconstruction tools [22]. The ventricular sizes were determined using a snake algorithm where T1-weighted volume images were converted to milliliters [24]. These brain images were used as accurate representations of the computational domain for finite element analysis. We also prepared models for intermediate ventricular sizes to simulate the transition of pathological stages B, C, and D with ventricular sizes of 42, 75, and 160 mL respectively (not shown). For the brain geometry of each of the disease states, we implemented finite element models according to eq. (2), with equal geometrical dimensions and electrical properties of our sensor, CSF, and brain tissue as given in Table 2. Triangular meshes were used with an average of 59,568 elements for each stage. The proper mesh size was determined by trial and error to obtain mesh independent simulation results. A current source of $0.88 \text{ A} / \text{m}^2$ was applied to an outer electrode pair to produce a current of $25 \text{ } \mu\text{A}$. Eq. (2) was used to rigorously calculate the voltage potential field throughout the computational domain. To emulate different sensor configurations, the potential difference at the surface of the measurement electrodes positions B and C were calculated. Although we could have accounted for the anisotropic conductivity tensor, σ , using DTI methods [23], here the electrical conductivity of the brain tissue was assumed to be isotropic and homogenous with a mean value of $\sigma = 0.172 \text{ S} / \text{m}$. The CSF conductivity was set at $\sigma = 2.01 \text{ S} / \text{m}$. The details of Fig. 7 show the isopotential contour lines achieved with a typical sensor. The chart also summarizes the predicted conductance measurement of the system during the simulated transition from the normal state to hydrocephalus. The conductance predictions correctly reflect the trend in size changes from normal ventricular size to hydrocephalus. The ventricular size change from 25 cc to 250 cc in hydrocephalus incurred a conductance change from 1 S to 1.45 S . The rigorous computations also allowed us to estimate the leakage current into the brain tissue. In the normal subject, leakage was also determined to be less than $5 \text{ } \mu\text{A}$ or less than 20% of the input current.

We believe that the accuracy of our computer-aided analysis would be satisfactory for clinical sensors with feedback control. The sensors performance can be compared to the MRI volume accuracy. The resolution in the horizontal and vertical direction in MR images is limited by a voxel size of typically $1 \times 1 \times 1.5 \text{ mm}$ in a 3T GE Sigma scanner (GE Medical Systems,

Milwaukee, WI) [19]. For the lateral ventricles, this MRI resolution is estimated to cause a volumetric variance of at least ± 0.276 mL, or $\pm 3.6\%$ of its size. Our prototype volume sensor variance was 8%, which was obtained by comparing measured to calculated volumes. This preliminary error is not much larger than the best state-of-the-art MRI technique. However, MRI scans are not practical for an on-line treatment. The sensor accuracy could further be improved with (i) higher gain amplifiers, (ii) more measurement electrode pairs, and (iii) optimal placement of excitatory electrodes along the full length of the cavity. The simulation approach would provide a systematic tool to address these improvements.

Positional Dependence of Sensor in Simulated Hydrocephalus

We also assessed the positional dependence of the sensor within simulated human lateral ventricles. In repeated simulations, the electrodes were displaced from the center towards the ventricular walls. For five different positions of 1 mm displacements from the actual center to the superior and inferior ventricle wall, the simulations gave similar volume readings with a variance of less than 2%. As a general trend, it was observed that the conductance decreased when the sensor moved closer to the wall due to an increase in leakage current. However, these changes were modest due to the high electrical contrast between CSF and human brain tissue. Overall we conclude from the gel brain phantom experiments as well as the rigorous electric field validation that the sensor position detecting the ventricular CSF space is only mildly dependent on the position, therefore a sensor placed in a human subject who maintains normal activities is expected to be sufficiently insensitive to the sensor position inside the ventricle.

6. Conclusions and Future work

This paper introduces a ventricular volume sensor and demonstrates experimental and theoretical results supporting the direct measurement of ventricular volumes based on the conductivity differences between CSF and the brain parenchyma. The conductance of CSF in the dog brain was found to be about *twelve times* higher than that of the brain tissue. Bench-scale experiments demonstrate the feasibility of an electrical conductance-based sensor for accurately measuring and controlling the size of elastically expanding cavities. Comparison of agarose gel experimental and simulation data provided verification of measurement principle for three dimensional cavities in phantoms emulating real brain ventricles. Preliminary experiments using a feedback control system demonstrated that the ventricular sizes could be restored even with simple control algorithms. Since our technique produces a continuous conductance-volume correlation, our novel device is expected to detect and monitor the volumetric change in real time and in-vivo. We also note that for feedback control a highly accurate measurement is not necessary. As soon as the ventricular size leaves the desired set point as detectable by a voltage drop, proper control action can be implemented immediately to reduce CSF accumulation before the ventricles grow too large.

In the future, we will conduct long term in-vivo studies in animal brain tissues for studying the biocompatibility and robustness of the sensor. Future directions also include optimal sensor design using simulation experiments to obtain accurate calibration curves. More measurement electrodes along the sensor axis will improve the accuracy of the measurements, which is also a part of our future work. The experimental and computational work that we have thus shown with our proposed sensor with feedback may serve as the foundation for improved therapeutic options for hydrocephalic patients relying on volume sensing, monitoring or active feedback control.

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Fig. 1. Proposed CSF volume sensor with feedback control. The volume sensor is shown in (1). It feeds voltage drops sensed at the measurement electrodes to a microcontroller unit (2) which also provides power and RF signal transmission. The micro pump (3) receives a control signal from the microcontroller to maintain the desired ventricular volume.

Fig. 2. Experimental setup of three-dimensional volume measurement experiments. Frame A shows a CSF-filled three-dimensional cavity molded into agarose gel with conductivity properties similar to the brain. B shows the conceptual circuit diagram and the control scheme for controlling the volume of fluid contained within the cavity. The reading was sent to a Flow Controller, FC, which controlled a micro pump. Feedback control was implemented using an on-off algorithm and a Flow Transmitter, FT.

Fig. 3.

A series of agarose gel cavities (Frames A-D) with increasing cavity size from 2 – 16 mL. The volume sensor was placed vertically, collinear with the main axis of the cavity. Simulations of the electric field and conductance of the brain phantoms and CSF were conducted using finite element analysis in Frames E-H. The simulated current density, represented as arrows in E-H, demonstrates that the voltage potential difference strongly correlates with the cavity volume.

Fig. 4.

Dynamic fluid volume measurements. Comparison of actual volume with volume sensing in elastic balloons in a dynamic experiment is shown in frame A. These results demonstrate that fast volume expansion was detected reasonably well by the sensor. Frame B shows response of the feedback controller to simulated ventricular enlargement. The controller successfully maintained the volume at a desired volume by discharging excess CSF through a micro-pump using an on-off control algorithm. The simple on-off control scheme maintains the set point, but exhibits characteristic variations due to the simple control law.

Fig. 5.

Comparison of Experimental and Simulated Volume Measurements. The experimental measurements obtained with a 25 μ A excitation current are depicted by their measured error bars ($\diamond$). The calculated voltage drops for simulated measurement electrodes are shown as a line (—). The experimental data has a variance of less than 2%. Simulated voltage drops match closely the experimental results.

Fig. 6. Simulated Positional Dependence. This figure summarizes simulated results to quantify the positional dependence of volume sensing. We used finite element analysis to simulate excitation with 25 μ A introduced by sensors with positional displacements, in an 8 fold increase in volume. Positions a – d were spaced 0.47 mm apart from each other with position A being the center of the cavity. Position e simulates a large dislocation of the sensor with an angle of 10°. Despite variability in the measurements due to the horizontal positioning, the conductance volume trends are conserved.

Fig. 7.

Case Study of conductance changes during the transition of hydrocephalus using our prototype design in simulations using finite element analysis. The sizes of the intermediate stages B-D (not shown) were artificially created by interpolation to emulate transitions from normal to hydrocephalus. The projections of volume changes were taken as size changes of the lateral ventricle with known volumes of the normal and hydrocephalic lateral ventricle. The simulations were repeated five times with different sensor positions located between 1 – 5 mm away from the central axis. An 8 fold increase from normal size correlates with our experimental and simulated analysis.

Table 1

Summary of sensor specifications including material, spacing interval, and other dimensions.

Parameter	Value
Number of electrodes	4
Electrode shape	Cylindrical
Electrode length	1.5 mm
Electrode Spacing	1.5 mm
Stimulating Electrodes	Platinum/Iridium
Insulating catheter	Silicone
Catheter Radius	1 mm

Table 2

Conductivity values of Cerebrospinal fluid (CSF) and brain matter (gray and white matter) in animal and human brain tissues. Measured vs. published data.

Biological Tissue	Sensor Measured values of Conductivity, σ	Published values of Conductivity, σ
Conductivity of gray and white matter (brain tissue)	0.172 ± 0.078 S/m	0.06-0.3 S/m [15]-[17]
Conductivity of Cerebrospinal fluid	2.01 ± 0.10 S/m	1.81-2.0 S/m [18]

Table 3

Boundary conditions and material properties used for computer simulations using Finite Element Analysis.

Condition	Gel Phantom Cavity	Human Brain Hydrocephalus
Neumann $\vec{n} \cdot \vec{J} \in S_{A,D}$	$\vec{n} \cdot \vec{J} = 16.66 \text{ A} / \text{m}^2$	$\vec{n} \cdot \vec{J} = 0.88 \text{ A} / \text{m}^2$
Dirichlet $V = S_{B,C}$	$V = 0 \text{ V}$	$V = 0 \text{ V}$
Brain Tissue Conductivity	0.172 S/m	0.172 S/m
CSF Conductivity	2.01 S/m	2.01 S/m