Abstract—Lack of bladder fullness sensation is an issue that arises in different neurogenic conditions and in addition to influencing patients’ quality of life, can result in serious kidney damage. We describe a wireless wearable sensor for detecting bladder fullness using near infrared spectroscopy (NIRS). The sensor has been tested in vitro and in vivo to verify its feasibility and is shown to be capable of detecting changes in bladder content noninvasively.

Index Terms—Biomedical monitoring, bladder dysfunction, near infrared spectroscopy (NIRS), wearable sensors.

I. INTRODUCTION

THE evolution of near infrared spectroscopy (NIRS) as a means of monitoring the hemodynamics and oxygenation of the bladder is recent [1]. Important information can be derived that contributes to the evaluation of patients with symptoms of bladder dysfunction, and understanding is growing of the distinct patterns of change in chromophore concentrations that occur in the context of disease. Wireless NIRS devices can now be utilized to evaluate bladder function in health and disease [1], [2], and NIRS is proving uniquely applicable to the study of bladder pathophysiology because of the anatomic and vascular characteristics of the organ, how the bladder’s microcirculation must function to maintain perfusion as it fills and contracts to empty, and because of the negative effect of disorders of detrusor muscle hemodynamics and oxygenation on normal voiding function [3].

One bladder-related symptom of concern not addressed by current NIRS monitoring studies is the inability to sense when the bladder is full. This symptom occurs in a number of conditions and for affected patients the problems that result range from accidental leakage of urine by day (incontinence) or bed wetting at night (enuresis), through to an inability to empty the bladder (urinary retention). While incontinence and enuresis are troublesome, can be embarrassing, and negatively affect a patient’s quality of life [4], urinary retention has potentially serious consequences, particularly in patients with abnormal bladder function secondary to spinal cord injury. In such cases, not being aware that the bladder is full can lead to back pressure developing in the urinary tract with a risk of serious damage to the kidneys. Unrecognized, this situation increases morbidity and contributes to a shortened life expectancy. However, even incontinence can have major social and fiscal consequences; in the elderly, loss of bladder control due to failure to sense the bladder is full and void voluntarily is the principal reason for elderly patients having to leave home and be taken into institutional care [5].

In all these cases, a wearable noninvasive device that monitors the fullness of the bladder and provides an alarm once the volume of urine in the bladder has reached a pre-set threshold would be beneficial. Individual patients would then, depending on their pathology, be able to empty their bladder voluntarily to avoid incontinence. In children with nocturnal enuresis, a problem that affects 20% of children over four years of age [6], a device that wakes the patient with an alarm once the bladder is full, but before incontinence occurs, has major advantages over current systems that only detect accidental voiding. Enabling the subject to wake, sense that his/her bladder is full, and void voluntarily before leakage occurs, would lead to conditioning over time to waking in response to the bladder being full, and resolve the enuresis.

Monitoring bladder size, volume, and content noninvasively can be done using ultrasonic scanning, and is common in clinical practice. This technique uses ultrasonic imaging to differentiate the urinary bladder from surrounding tissues and organs, produce volume information, and estimate urine level. The method gives accurate results, but does not lend itself to monitoring in ambulant subjects or home use as it requires powerful computational resources and a complex scanning control system. The requirements of gel application and control measurement also make most commercial ultrasonic scanners inappropriate for continuous wearable monitoring, although some portable and wearable ultrasonic sensors for bladder monitoring have been reported [6], [7].

Bioelectrical impedance analysis is another method that can be used to determine the volume of urine in the bladder. This technique is principally used for determining extracellular and total body water, and several skin surface electrodes are required on the abdomen at the level of the bladder for the changes in...
electrical impedance used for detection of urine volume to be measured [8]. This again may not be a practical method suited for ambulant monitoring.

We report the development and pilot testing in vitro and in vivo of a NIRS prototype for noninvasive optical monitoring of bladder filling to capacity using a compact wearable wireless system. We propose using this small, low-weight, inexpensive, wireless and easy to use device as a noninvasive method for monitoring the point in time when the bladder becomes full, with lower computational requirements and complexity compared to ultrasonic continuous measurement systems. Our method employs the absorption properties of human tissue and water in the near infrared (NIR) light wavelength range to measure changes in water content in the field beneath a NIRS device. Because the bladder rises out of the pelvis below the anterior abdominal wall as urine accumulates within the organ, this device can be used to detect when a bladder capacity previously defined by ultrasound is reached. When the bladder rises into the NIR light field as it fills, the water in the urine it contains results in high light absorption that generates an abrupt decrease in the light intensity sensed returning to the NIRS device. This event can be set to activate an alarm; potentially benefiting patients with any of the problems related to an inability to sense when their bladder is full.

II. DETECTION OF BLADDER FILLING TO CAPACITY USING NIRS

Light in the NIR wavelength range (650–1000 nm) has a high penetration depth in the living tissue. The major absorbing chromophores of physiologic interest in this wavelength window are oxygenated and deoxygenated hemoglobin (HbO2 and Hb) as indicated in Fig. 1. Water, which is the main compound in urine (95% [9]), also has an absorption peak at 975 nm and this peak can be used to detect urine content in the bladder and differentiate between an empty bladder, one with low volume, and a full bladder.

In NIRS, light in the NIR window is used to interrogate the tissue. A light source (emitter optode) is placed on the skin surface, with a detector (receiver optode) placed a few centimeters away. Changes in light attenuation due to absorption of the transmitted light by chromophores in tissue (HbO2, Hb and water) are detected by the receiver optode. The resulting changes in raw optical data are then converted to changes in chromophore concentration. A common model used for this purpose is the modified Beer-Lambert law [10]

\[
A(\lambda) = -\log \left( \frac{I}{I_0} \right) = \left( \sum_i \varepsilon_i(\lambda)c_i \right) BL + G
\]

where \(A(\lambda)\) is the light attenuation at wavelength \(\lambda\), \(I_0\) is the source intensity, \(I\) is detected light intensity, \(\varepsilon_i(\lambda)\) is the extinction coefficient of chromophore \(i\) at wavelength \(\lambda\) in \(\text{mol}^{-1}\text{cm}^{-1}\), \(c_i\) is the \(i\)th chromophore concentration in \(\text{molL}^{-1}\), \(L\) is the source-detector (SD) separation in cm, \(B\) is the differential pathlength factor and \(G\) is an additive term to take fixed scattering losses into consideration. This model is usually used in differential form to measure concentration changes in the tissue. The effective depth of penetration in this method is approximately half the SD separation distance.

Changes in tissue hemodynamics and oxygenation can be inferred from NIRS-derived changes in HbO2 and HHb. Monitoring such data has been widely used in the study of muscle [11], with applications in exercise science [12], rehabilitation [13], and medicine [14]. In urology, NIRS has been used to monitor bladder muscle function and to diagnose bladder dysfunction noninvasively by detecting patterns of changes reflecting variations in detrusor muscle hemodynamics and oxygen supply and demand [14].

Self-contained wireless NIRS devices have been utilized for a wide range of studies involving brain, muscle, and the bladder [1]. Such devices have the advantages of imposing no motion restriction, which means subjects can engage in active physical pursuits, and suitability for longer term monitoring in ambulant patients [1], [15]. Wireless NIRS devices often use Light Emitting Diodes (LEDs) as the light source. Although LED based NIRS systems have a broader spectrum compared to laser-based NIRS devices, they have the advantages of being small, low weight, inexpensive, compact and self-contained and can be applied directly on the skin surface without need for the fiber-optic cables required for laser systems.

The hypothesis for our NIRS-based method for monitoring the level of urine in the bladder and detecting bladder filling to capacity was that with an LED light source using a wavelength close to the absorption peak of water at 975 nm, a self-contained NIRS device placed on the abdominal skin would detect water (urine) when the bladder enlarged into the NIR light field. Ultrasound data indicates that as the bladder fills naturally, the dome of the organ rises within the abdominal cavity bringing the bladder and the urine it contains into the NIR light field [16]. The water contained in the bladder then absorbs light, causing a decrease in detected light intensity. Here, we describe a prototype of such a device as a proof of principle.

While this method is similar in concept to the method presented for continuous bladder monitoring using ultrasound [6], in our method it is the urine in the bladder (rather than the anterior wall of the bladder) which triggers the alarm. The level of bladder fullness that corresponds to the urine capacity that needs to be detected will depend upon the patient’s symptoms,
and his/her underlying medical condition. In later development phases, this capacity value can be defined for individual patients, and the fullness and position of the bladder beneath the abdominal skin that this volume corresponds to can be assessed by ultrasound. The NIRS device is then positioned on the abdominal skin so that it alarms when the bladder reaches the size that corresponds to the capacity required for that patient.

A. Electronics

The hardware consists of a $60 \times 70 \times 20$ mm wireless NIRS device, that is worn by the subject on the abdominal skin. The sensor can either operate offline by storing the data on board or in real time via a link to a base personal computer (PC) through a wireless USB dongle. The block diagram of the sensor is shown in Fig. 2. The sensor is made using commercially available components on a 2-layer printed circuit board and is enclosed in a custom made 3-dimensionally (3D) printed enclosure as shown in Fig. 3. The device weighs 55 grams. All components except the source LED and the detector are mounted using standard surface mounting technology. The source and detector are mounted on the front side of the enclosure using adhesive glue and are wired to the main PCB.

All the signal controls, sampling and processing are performed by a 16-bit low power microcontroller (MCU) (MSP430F2274 Texas Instruments, TX, USA) running at 16 MHz.

The source LED is a 950 nm LED (OSRAM Opto Semiconductors, 55 nm spectral half width, 16 mw nominal power) driven by a constant current driver, that in turn is controlled by a hardware timer. Even though the absorption peak of water is at 975 nm, the 950 nm source output is still highly absorbed by water as the spectral bandwidth of the source covers 975 nm wavelength. The light detector is a 5.22 mm$^2$ silicon photodiode integrated with a transimpedance amplifier (TIA) (OPT101 Texas Instruments). The responsivity of the detector is 0.45 A/W at 950 nm and the transimpedance amplifier is set to provide a gain of $6 \times 10^6$ V/A and a bandwidth of 2.5 KHz. The amplifier’s output is filtered and sampled by a 10-bit analog to digital converter (ADC) integrated on the MCU. Prior to sampling by ADC, the output of the amplifier is filtered by an active twin-T notch filter with center frequency at 60 Hz to remove interferences from AC power line coupling and ambient lighting followed by a first order lowpass filter with $f_c = 5$ kHz.

The sensor is powered by a 3.7 V, 850 mAh Lithium-ion polymer rechargeable battery that provides up to 20 hours of continuous monitoring. The battery voltage is regulated down to 3.3 V through a low dropout linear regulator. The battery is recharged through mini-USB connection.

In case of offline standalone operation, the sensor can log data on the 16 KB onboard flash memory storage. The data can be later downloaded into a PC for further analysis.

Two communication interfaces are supported: wired using USB 2.0 connection and wireless using wireless link and a wireless dongle connected to a PC.

The wireless link uses 868–915 MHz band for communication and transfers data at 250 kbps. A wireless module based on Texas Instruments CC110L radio transceiver is used (A110L, Anaren Microwave Inc, NY, USA). The MCU communicates with the module over the serial peripheral interface (SPI) bus at 250 kHz. The wireless link allows remote start and stop of data collection through PC, download of the logged data and real-time data streaming to the PC with a range of up to 20 m.

A triple axis accelerometer (ADXL345, Analog Devices Inc., MA, USA) is used to detect motion to remove motion corrupted
data segments. The accelerometer shares the SPI bus with the wireless module.

The sensor is encapsulated in a custom made 3D printed enclosure (Verowhite polyjet resin). An extruded feature that houses source and detector provides higher coupling with tissue and also reduces ambient light interference (see Fig. 3).

B. Firmware

The firmware controls the source LED timing and data sampling process, logs data and communicates with a PC for command reception or data transmission.

The scattering and attenuation of light in the tissue result in 6–7 orders of magnitude decrease in signal power. As a result, to have better signal-to-noise ratio (SNR) at the detector output, higher source optical power is desired to increase the number of photons that can reach the detector. However, to limit the total tissue exposure and minimize the possibility of tissue thermal overheating, the power has to be kept within a safe range. An average power limit of 2 mW can be considered safe and has been used as the limit for similar NIRS devices [17], [18]. To achieve this power level while having high instantaneous power, a source-switching scheme is employed as shown in Fig. 4. We chose a 30 ms delay between the LED activation times. We also empirically found 60 mW of instantaneous power to result in well detectable light levels as the light exits the tissue for our desired SD separation of 3 cm. As a result, the source LED needs to be activated for a maximum of 800 ms with instantaneous power of 60 mW which corresponds to a driving current of 370 mA (Fig. 5) in order to keep the average power below 2 mW. This scheme also reduces the total power consumption.

To ensure accurate timing of LED driving pulses and ADC conversion triggers, the LED is driven directly by hardware timer which is programmed to produce pulses every 30 ms. A separate timer triggers ADC conversion for sampling the LED light level. The transimpedance amplifier bandwidth of 2.5 kHz results in an approximate rise time of 140 μs for the LED pulses. Therefore, sufficient delay before sampling is necessary to allow the transients at the detector output to settle. We used a delay of 600 μs as shown in Fig. 6, which shows the signal at the detector output along with the sampling trigger signal.

The detector’s output signal is initially sampled at 83 kS/s and a total of 8 samples are recorded. These samples are then averaged and stored in a buffer. This sampling rate allows use of low order antialiasing filter and collection of sufficient samples during the LED activation time. The next sampling cycle occurs in 30 ms and follows the same pattern. This is equivalent to sampling the continuous optical signal at \( f_s \) = 83 kS/s, low pass filtering it with a moving average filter of length 8 and then down-sampling the result to approximately 33 Hz. The digital averaging helps reduce high frequency noise.

To prevent potential interference from ambient lighting, background light level is sampled as the baseline and subtracted from the detected light level. The firmware therefore takes a sample from background light level 800 μs after turning the source LED off in each sampling cycle. This delay ensures all transients have settled and the background light level is being sampled properly.

The baseline corrected value is placed in a data packet along with time stamp and transmitted wirelessly to the PC. In case of offline operation, it is logged onto the onboard flash memory.

The sensor operation can be controlled either by wireless commands through a PC or, for offline data collection, by user push button on the sensor.
C. PC Interface

A Graphical User Interface (GUI) based on Matlab (Mathworks, MA, USA) is used for remote control of the sensor, streaming and saving data to a file for long term monitoring, downloading data stored on the sensor’s memory and processing the signal in real time or offline (linear filtering, trend removal, etc.). A snapshot of the GUI with a sample data set is shown in Fig. 7. The top panel shows the real time trace of the signal or the loaded data. The bottom panels contain controls for wireless operation mode and USB wired mode.

III. Performance Evaluation

The sensor’s dark noise was measured by readings obtained by placing the sensor in a dark room with no light incident on the detector for one hour. The root mean square (rms) value of the noise in this setup was calculated and repeated for a couple of measurements to obtain an estimate of the prototype’s noise voltage. This value was calculated to be less than 470 μV. The noise equivalent power (NEP) was then calculated from dark noise measurements using

\[ P_t = \frac{V_n}{R(\lambda)G} \]  

(2)

where \( P_t \) is the incident light equivalent power in W, \( R(\lambda) \) is the responsivity of the detector at \( \lambda = 950 \) nm in A/W, \( G \) is the TIA gain in V/A and \( V_n \) is the noise voltage. The NEP is calculated to be approximately 180 pW. This defines the detection sensitivity for a signal to noise ratio of unity and is the minimum light level detectable by the sensor.

The long-term stability of the sensor was evaluated by continuous recording of data from a phantom using the sensor for 30 minutes after a warm up period of 1 minute. The aqueous phantom was prepared using method described in [19]. The phantom scattering and attenuation parameters are chosen to be close to those of abdominal tissue (in particular, abdominal fat with attenuation coefficient \( \mu_a = 3 \) cm\(^{-1}\) and reduced scattering coefficient \( \mu_s' = 3.3 \) cm\(^{-1}\) [20]). The phantom was made with 20% intra-lipid mixed with ink to obtain desired optical parameters. The difference between the initial and final reading normalized to the initial signal value was recorded as the drift. The device shows 1.5% drift over the period of 30 minutes.

IV. In-Vitro Setup

To verify the capability of the sensor in detecting bladder level changes in vitro, a simple setup, as shown in Fig. 8, was used. The setup was made to simulate the bladder, urine and the abdominal tissue during bladder filling and voiding. A latex balloon was submerged in a phantom prepared as described in the previous section in such a way that the balloon neck is attached to the top of the container. The balloon can be filled with water from the top using a syringe. The distance of balloon from the sidewalls was 1.5 cm when full and 6 cm when empty. The sensor was placed on the sidewall of the cylindrical container and secured with medical adhesive tape (3M, MN, USA). The data was recorded wirelessly.

Fig. 9 shows a sample recorded data when the balloon is filled and emptied. The intensity readings from the sensor were converted to attenuation as

\[ A = -\log \frac{I}{I_0}. \]  

(3)

An increase in amount of water in the optical path results in decrease in light intensity and therefore an increase in attenuation (A). As the balloon filling begins around \( t = 9 \) s, the absorbance increases up to the point where the balloon is filled around \( t = 10 \) s. Similarly, when voiding starts, the absorption decreases until the balloon is emptied.

The drop in the signal level between the end of filling and beginning of voiding is caused by motion of the balloon at the
Fig. 9. *In vitro* recorded data when the balloon is filled and emptied. Red (solid), green (dashed), black (dotted) and cyan (dashed-dotted) lines indicate beginning of filling, end of filling, beginning of voiding and end of voiding, respectively.

end of filling cycle as the result of ending water flow. The same occurs at the beginning of the voiding.

V. **In-Vivo Setup**

A. Materials and Method

Pilot data on 1 subject has been collected in 6 independent trials with the device during voiding to verify if the sensor is capable of differentiating between full and empty bladder. The sensor is placed 2-cm above the symphysis pubis across the midline during voiding as shown in Fig. 10 and is secured using medical adhesive tape. The absolute intensity reading from the detector is then converted to attenuation according to (3) and used for comparison between full and empty bladder. Data was transmitted wirelessly to a PC for recording. For this proof-of-principle test, the motion rejection feature of the sensor using the accelerometer was not used.

B. Results

Fig. 11 shows a typical attenuation signal recorded at SD separation of 3 cm. The red (solid), green (dashed) and black (dotted) vertical lines indicate permission to void, beginning and end of voiding, respectively. The signal shows a fall at the start of voiding and then plateaus around 15 s after beginning of voiding. This is possibly due to the fact that as the voiding begins, the bladder dome is in the light path between source and detector. As the bladder shrinks, the urine level in the light path reduces and light intensity at the detector increases (decrease in attenuation). At a certain point (in this case around 15 s after voiding begins), even though the voiding continues, the bladder is no longer within the range of the sensor’s light and therefore no further change in detected light intensity is observed.

Fig. 12 shows the light attenuation changes between full and empty bladder for 6 independent trials with urine volume ranging from 300 ml to 700 ml. A significant difference in light absorbance is observed between pre- and post voiding states as shown in Fig. 12 (paired t-test $p < 0.01$). The starting point or baseline is variable among trials as a result of differences in light coupling, geometry, etc. However, there is a consistent difference between pre and post voiding states in the trials as a result of change in bladder content (mean of the differences: 0.022 with standard error of mean of 0.0096).

For comparison Fig. 13 shows the concentration changes detected during a separate voiding session when using a general purpose desktop laser-powered reference spectrophotometer (OxyMon, Artinis BV, The Netherlands) with 971 nm laser for detection of water. The pattern of change in absorption is...
similar to those obtained with our prototype, even though the 971 nm signal is more sensitive to changes in water content.

Table I shows the overall system level parameters for the designed prototype. Active power consumption with and without radio refer to the cases when the device is linked to a PC and when the device is operating independently.

VI. DISCUSSION

We have developed a novel optical method for noninvasive monitoring of bladder capacity using a compact wireless NIRS prototype incorporating an LED with a wavelength of 950 nm and demonstrated the feasibility of using this device placed on the abdominal skin to detect a signal change that indicates when the bladder and the urine it contains have left the monitoring field of the device. Our data support our hypothesis that when the bladder fills and enlarges, urine within the bladder can be detected using NIRS with a light source close to the absorption peak of water at 975 nm. Further validation of our NIRS-based method to detect when an individual’s bladder capacity reaches a pre-defined limit is required, along with development of appropriate decision making process for activating filling alarm or slow drifts caused by the small variations in the position of the device which can occur during continuous monitoring. In such a scheme, the first channel would be placed over the bladder, with the second channel located further from bladder so as to differentiate motion/coupling-induced changes in the detected intensity from those caused by bladder water content changes. The changes caused by motion or changes in optical coupling will be highly correlated between the two channels, while changes caused by alterations in bladder capacity will only affect the channel located on the bladder.

The PC connectivity, in addition to providing an alternative method for device control as well as data processing and storing, can potentially be beneficial in cases were remote monitoring of a subject’s bladder activity is of interest. In urinary tract infection (UTI), for example, which is a common condition in spinal cord injury patients, the frequency of voiding increases and access to these information collected in normal daily life condi-
tions by the clinician is important in treatment of patients. In this case, the limited range of connection might limit the usage of wireless link to indoors use only. However, the same benefits can be offered by replacing the PC with a smart phone in the future.

For our device to reliably monitor ambulant subjects consistently and with the level of accuracy required for detection of bladder capacity in selected patient groups, additional trials and development are required. In particular the potential effect of different body postures and positions needs to be evaluated, even though MRI studies have shown that body position in young subjects does not affect the shape and position of the bladder significantly [23]. Ultrasound data also indicate that the anterior wall of bladder retains its position relative to the anterior abdominal wall as the bladder fills and empties naturally [16]. Data will also need to be collected in cohorts where the age range and diagnostic criteria match those of the patients for whom monitoring with a device such as ours is considered of potential benefit.

VII. Conclusion

We have designed and developed a compact wireless optical sensor prototype for continuous noninvasive monitoring of the bladder in patients who are unable to sense when their bladder is full. This is a significant clinical problem in individuals with abnormal (neurogenic) bladder function, such as patients affected by multiple sclerosis (MS), stroke and/or spinal cord injury, elderly patients with incontinence, and children with persistent enuresis. The device is capable of differentiating between when the bladder is empty or contains a small volume of urine and when it becomes full, by using the absorption properties of water at a wavelength of 950 nm. With such a device used as a sensor with an alarm, it is hence feasible to warn the subject when the volume of urine in his/her bladder reaches a predetermined threshold of bladder capacity. This would potentially enable patients at risk for urinary retention to protect themselves from renal damage, elderly subjects prone to incontinence to retain the ability to void voluntarily, and children with problematic enuresis to become conditioned to when they need to wake to void.

REFERENCES


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