URINE OUTPUT MONITORING
A Simple and Reliable Device for Monitoring Critical Patients' Urine Output

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Abstract: Currently, critical care units are equipped with sophisticated commercial monitoring devices capable of sensing most of the patient' physiological parameters, and of automatically supervising whether the values of the parameters lie within a preestablished range set by the clinician. The automation of these tasks has discharged the healthcare staff of a considerable workload. It also avoids human errors, which are common in repetitive and monotonous tasks.

In all likelihood, urine output is the most relevant physiological parameter that has yet to be sensed or supervised automatically. This paper presents a patent-pending device capable of sensing and supervising urine output. The device uses reed switches that are activated by a magnet that is attached to a float in order to measure the amount of urine collected in two containers. When a container fills, it is emptied automatically using a siphon mechanism and urine begins to collect again. An electronic unit sends the state of the reed switches via Bluethooth to a PC. From this information, the PC calculates the urine output and supervises the achievement of therapeutic goals. The end result is a fully automated, simple, inexpensive and accurate urine meter.

1 INTRODUCTION

Current critical care units are equipped with sophisticated commercial monitoring devices that allow clinicians to sense nearly any physiological parameter that may provide information relevant for interpreting the patient's state. In most cases, these devices can also supervise that the values of the physiological parameters they sense remain within a preestablished range set by the clinician. This range represents the values considered as normal for each parameter. If a parameter does not fall within its acceptable range, the corresponding sensing device alerts the healthcare staff by means of an audible warning (Otero et al., 2009b) (Hande et al., 2006).

These devices discharge the healthcare staff of a considerable workload, since they need not continuously supervise if the physiological parameters of every patient lie within the acceptable range. They also avoid human errors, which are common in any repetitive and monotonous task such as the one at hand (Jungk et al., 2002) (Mora et al., 1993).

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Arguably, the most relevant physiological parameter which still is measured and supervised manually by healthcare staff is the patient's urine output. Urine output is the best indicator of the state of the patient's kidneys. If the kidneys are producing an adequate amount of urine it means they are well perfused and oxygenated. Otherwise, it is a sign that the patient is suffering from some complication. Urine output is required for calculating the patient's water balance, which is essential in the treatment of burn patients (Mitra et al., 2006). Finally, it is also used in multiple therapy protocols to check whether the patient reacts properly to treatment (Rivers et al., 2001). When urine output is too low the patient is said to have oliguria. If the patient does not produce urine at all, then he/she is said to have anuria. Sometimes, urine output can be too high; in these cases the patient is said to have polyuria.

Currently, in critical care units, urine is collected in a graduated container that is connected to the patient's bladder through a Foley catheter. Periodically the nursing staff manually records the reading of the container of every patient, and operates a valve which releases the urine into a larger container. In critical care units of first world countries, this procedure is usually performed every hour, 24 times a day, 365 days a year. In the case of emerging countries, often only the burn patients –for whom urine output monitoring is of paramount importance– have this parameter registered every hour, while the remaining critical patients have it recorded every 2 or 3 hours. This disparity in criteria is due to the more reduced availability of healthcare staff in the critical care units of emerging countries.

It can even be argued that the monitoring interval currently employed in first world countries –once every hour– also is a compromise between avoiding risk states for the patient and relieving the nursing staff of an excessive burden. A system capable of automatically monitoring urine output would decrease the workload associated with this task and, at the same time, it would permit supervision to take place on a more continuous basis.

This paper presents a patent-pending device capable of sensing and supervising urine output (Akinfiev et al., 2010). Section 2 reviews related work. Our device is described in Section 3. It uses reed switches that are activated by a magnet that is attached to a float in order to measure the amount of urine collected by two containers. An electronic unit checks the state of the reed switches and sends it to a PC, which supervises the achievement of the therapeutic goals that have been established for urine output. Section 4 discusses the results of this work. Finally, a series of conclusions and lines of future extension are given.

2 RELATED WORK

Automating the supervision of urine output is a problem that has only begun to be addressed by the biomedical engineering community very recently. There are several problems that have contributed to this delay. On the one hand, it requires devices capable of measuring very small amounts of flowing liquid (up to 3-5 milliliters per hour). This precludes the use of common industrial solutions such as ultrasound sensors (Johnson, 1978) or commercial flowmeters. On the other hand, any component of the device that is in contact, or may be in contact, with the urine cannot be reused on other patients and must be replaced approximately every 4-7 days for hygienic reasons. Therefore, any component of the device that is or may be in contact with the urine must be easy to dispose of, and should have a low price. This precludes the use of expensive high precision laser based solutions (Ishida, 1990). Furthermore, contact with the urine also means that the component is in indirect contact with the patient's bladder through a Foley catheter. Therefore, the components that come into contact with the urine have to be sterilized before use. Finally, urine contains Uric acid, Sodium, Potassium, Chlorine and other components which make it corrosive, especially for metals.

The first device proposed to automatically measure urine output was Urinfo 2000, developed by the Israeli company Medynamix (Hersch et al., 2009). Urinfo 2000 was designed to automate the hourly urine output measurement, but not to take more frequent measures. Its operation is based on counting the number of drops of urine produced by the patient, and from this count it estimates urine output. The average error of the device when used to take hourly measurements was 8% ($\pm$25 ml). Urinfo 2000 cannot supervise therapeutic goals and its readings are not transmitted to a central station/PC. Thus it requires the nursing staff to take the urine output measures from the device's display, placed next to the patient's bedside.

The authors of this paper proposed a device to automate the supervision of urine output in (Otero et al., 2009c). Several technical and legal problems precluded us from moving this device beyond the laboratory validation phase. Using the knowledge we gained building it, we developed a second device with the main objective of conducting a series of clinical studies based on more continuous and accurate monitoring of urine output throughout the stay of patients

in a critical care unit (Otero et al., 2010) (Otero et al., 2009a). This device uses a high precision scale to measure the weight of a commercial urine meter. On the scale's pan there is a support frame made up of Bosch profiles that isolates the scale from force transmission from the patient's bed, and guarantees that the urine flows properly through the urine meter's input tube. The maximum measurement error of this device is under 1.5%.

Currently this device is in use at a research unit associated with the University Hospital of Getafe, in Spain. In this research unit a series of experiments aimed at the study of sepsis in an animal model –pigs– are being conducted. The aim of these experiments is twofold. On one hand, we would like to gain a better understanding of the pathophysiological mechanisms underlying systemic and renal hemodynamics during sepsis –hence the interest in a continuous and accurate monitoring of urine output. On the other hand, we would like to define the optimum monitoring interval for urine output, and to determine the level of accuracy required for this task.

The high accuracy and acquisition rate of this device make it ideal for carrying out clinical studies. However, this device was not intended to be used in the clinical routine. Its size and operation make it somewhat tedious to be used in a critical care unit. Conversely, in using this device we have learned that its accuracy –1.5%– and measurement interval –up to 10 sec.– are superior to what is required for properly supervising urine output. Our experience shows that an accuracy of 5% and a monitoring interval of 5-10 minutes is enough. Therefore, these parameters can be relaxed to obtain a cheaper and simpler device. That is the goal of the device presented this paper.

3 THE DEVICE

In this section we first describe the general operating principles on which our device is based. Although we will use the device to measure urine output, the principles on which it is based are general and could be used to measure the volume of any liquid flowing through a tube. Then we describe the prototype we have built, and the calibration procedure we have carried out to determine its operating parameters. Finally, we describe how the supervision of therapeutic goals is carried out.

3.1 General Operating Principles

In our device the urine arising from the patient's bladder is collected through an input tube in a container.

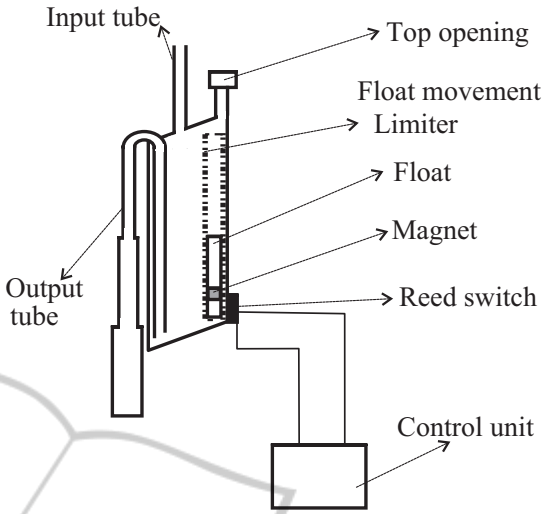

Figure 1: Diagram of the small container of our prototype.

This container is equipped with an output tube that empties its contents when it gets filled by taking advantage of the siphon principle (see Figure 1). One end of the output tube will be placed in the bottom wall of the container. Ideally the bottom wall of the container will be inclined or conical and the tube will be in the lowest part. This favors a more complete evacuation of the container. Both ends of the output tube are always open. When the container begins to receive liquid, the liquid goes up inside of the output tube until it reaches the elbow located in its top part. When it reaches the elbow, liquid begins to fall down the portion of the tube located on the exterior of the container. By the principle of the siphon, the liquid will flow down the tube and the container will be emptied.

For the system to function properly, the output tube should not be too wide. If the tube is too wide, when the water starts flowing, air could rise through the tube towards the container, breaking the siphon principle. On the other hand, to drain the container as quickly as possible, it is desirable that the tube be as wide as possible. A compromise between these two goals can be achieved using a variable diameter tube, so that its diameter grows (staggered or progressively) from the elbow of the tube to the end which is outside the container (see Figure 1). In this way, the principle of the siphon will work properly because the tube is narrow enough in its upper part, but when the water begins to fall it finds a wider tube which provides less resistance, thereby increasing flow rate.

The container must also be equipped with a filtered top opening that is used to equalize the internal and external pressures without risk of bacterial contamination. This opening should be located at a height

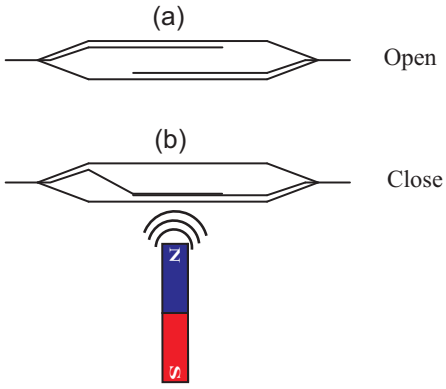

Figure 2: Operation of a reed switch. In the presence of a magnetic field, the switch is closed. In the absence of magnetic field, it is open.

H_1 above the elbow of the output tube. The liquid will not start flowing immediately when it reaches the elbow height; but it will continue to accumulate in the container until the height of the column of liquid which is above the elbow height overcomes the surface tension of the liquid against the walls of the output tube. The surface tension increases as the diameter of the output tube decreases. The height H_1 and the effective volume of the container V^S (the volume of water contained just before the draining starts) must be determined experimentally.

Ideally, the upper wall of the container will also be inclined or conical in shape. The top opening to equalize the internal and external pressures should be placed in the uppermost part of the wall (see Figure 1). This prevents bubbles from forming in the upper wall of the container, bubbles that would occupy volume and distort the measures.

One or more reed switches are placed on the outside of the container that will receive the urine. A reed switch contains two or more magnetizable, flexible, metal reeds hermetically sealed in a tubular glass envelope whose end portions are separated by a small gap. Under these conditions, the switch is open (see Figure 2a). A magnetic field properly applied will cause the reeds to bend, and the contacts to pull together, thus closing the switch (see Figure 2b).

A float within a structure designed to limit its movement so that the float can only move vertically is located inside of the container, near the container wall where the reed switches are placed (see Figure 1). The float has a magnet attached which interacts with the reed switches closing them when the magnet is approximately at the same height as each of the reed switches.

An electronic unit is connected to the reed switches, continuously checking their state. For the general case in which there are N reed switches on the outside wall of the container, the procedure for measuring the volume of liquid flowing into the container is as follows. At least one reed switch should be located in a position such that when the container is empty, the magnet located on the float closes the reed switch. As the liquid begins to flow into the container, the float, and therefore the magnet, begins to raise. At some point, the magnet will stop interacting with the first reed switch and, therefore, it opens. At that point, a volume V_1 of liquid has flowed into the container. When enough liquid has been accumulated in the container, the magnet will close the second reed switch. At this point an additional volume V_2 of liquid has flowed into the container, being the total amount of liquid accumulated $V_1 + V_2$. When the magnet moves higher, the second reed switch opens again and an additional volume V_3 of liquid has flowed –being the total volume of liquid $V_1 + V_2 + V_3$.

In general, when the reed switch N is closed, we will add the volume V_{2N-2} to the the volume of liquid that has flowed, and when the reed switch N is opened again, we will add the volume V_{2N-1}. When the container is emptied, the float with the magnet will go back to the bottom of the container, and therefore it will close the first reed switch. At this point, the effective volume of the container –V^C– has flowed, and the measurement procedure is resumed.

The volumes V_{2N-2} and V_{2N-1} for each reed switch, and the effective volume V^C must be determined experimentally.

3.2 Our Prototype

The general operating principle described in the previous section suffers from a problem: the volume of liquid that flows into the container from the time the container begins to empty through the siphon mechanism, until it is completely empty, will not be measured. Depending on the specific application and characteristics of the container, this may or may not be tolerable. The problem that concern us, the measurement of the amount of urine produced by a patient, requires an early warning about deviations from the therapeutic goals. Thus, a small volume container must be used –approximately 5 ml in the prototype we have built. A container of such a small size needs to be emptied a large number of times, which may make the inability to measure the liquid that flows during the discharge of the container intolerable if the patient has polyuria, i.e., the patient is producing a large amount of urine. If the patient is producing normal amounts of urine, or if he/she has oliguria, the amount of urine that will flow during discharge of the

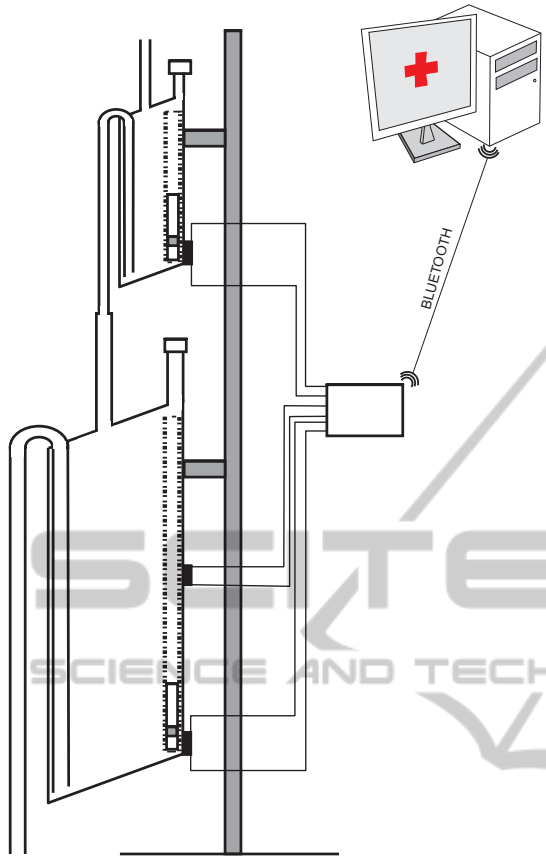

Figure 3: Diagram of our prototype device.

container will be nil or negligible.

It could be argued that in the case of patients that have polyuria it is not required to have a high degree of accuracy in the measurement of the urine output. The more urine is produced, the less important it is to have an accurate measure; only when the patient is producing small amounts of urine it is important to measure accurately (Otero et al., 2010). Therefore, the operation of the device as is described in the previous section could be acceptable. In any case, this problem can easily be solved using two different size containers, each of them working according to the principle presented in the previous section. The output of the smaller container is connected to the input of the larger container. Thus, although the volume of liquid that flows during the discharge of the first container is not measured in the small container, it will be measured in the larger one.

Chaining two containers as described here causes another problem. It can happen that when the small container releases its content, the larger one is nearly full. In the worst scenario, the first drop that falls from the small container would trigger the draining of the large container. Thus, the content of the small container would not be measured in the large one. The maximum error this may cause is $(V^S/V^L)\%$, being V^S and V^L the effective volumes of the small and large container, respectively. On average, the large container will start to drain when the small container is emptied halfway. Therefore, the average error caused by this effect will be $((V^S/2)/V^L)\%$. As long as $V^L >> V^S$, this error will be small.

The prototype we have built uses two containers. The first one was built to have a volume of approximately 5 ml, and the second a volume of approximately 375 ml (see Figure 3). The first container is equipped with a single reed switch that is activated when the float is on its lowest position. The staggered diameter of its output tube increases in three steps. The output tube is connected to the input tube of the larger container, which is equipped with two reed switches. The tube of the large container is connected to a 2.5 liters plastic bag that collects the liquid once it has been measured.

An electronic unit continuously checks the status of the reed switches, and reports any change via Bluetooth to a Java application that runs on a PC. From these changes the application calculates the volume of urine that has flowed and displays a chart with this information.

3.3 Device Calibration

The calculation of the effective volumes of the containers –V^S and V^L– must be determined experimentally. By effective volume we mean the volume of liquid that triggers the emptying of the container through the siphon mechanism. This volume will be slightly higher than the volume corresponding to the height of the top of the siphon mechanism because the liquid does not begin to flow until the pressure overcomes the surface tension force of the liquid against the walls of the output tube. The smaller the diameter of the output tube, the greater the height in the column of liquid required to overcome the surface tension will be.

The effective volume of each container was determined separately. A saline solution with properties similar to those of urine and an dropper were used to simulate the flow of urine (see Figure 4). Each of the containers were placed so that they would release their content on a bowl placed on the plate of a high-precision industrial scale –a PGW 4502e, built by Adam Equipment Inc. This scale has an accuracy guaranteed by the manufacturer of 0.01 g. Given that the density of the saline solution was known, we can determine the volume of liquid that the container releases into the bowl from the weight.

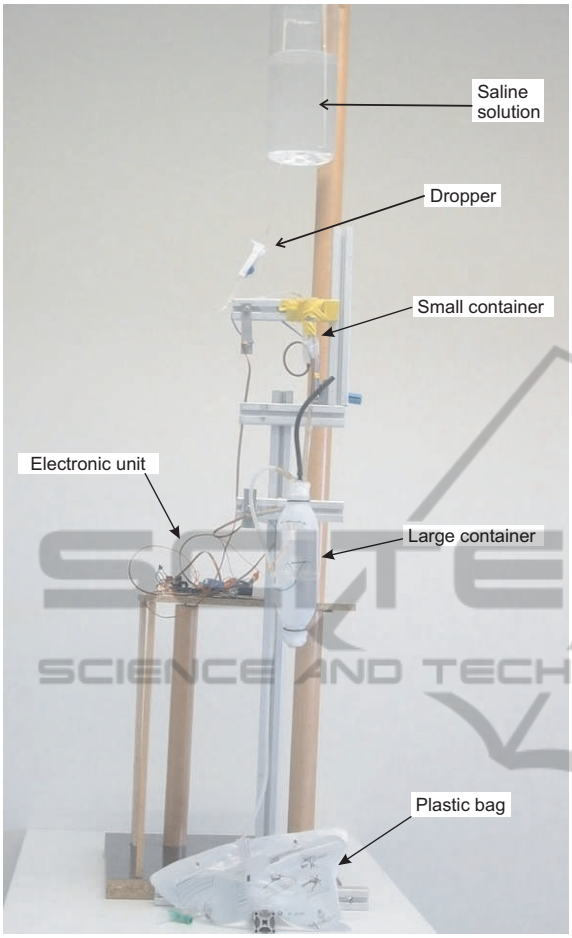

Figure 4: Picture of the prototype device with the saline solution and eye dropper used in its validation.

The PGW 4502e is equipped with a serial port that permits querying for readings. We built a program that periodically takes measurements from the scale. The program together with the dropper allow us to automate the process of carrying out multiple measures of the volume of liquid released by the containers.

Using this set up we took 200 measurements of the volume of liquid released by the small container, and 50 measures of the volume of liquid released by the large container. From these measures we calculated the effective volumes of the containers: 5.87 ± 0.32 ml and 376.72 ± 1.11 ml (mean $\pm$ standard deviation). At the time of writing, the authors are working to determine the volumes V_1^L, V_2^L and V_3^L, i.e., the volumes corresponding with the opening of the first reed switch of the large container, with the closing of the second reed switch, and whit the opening of second reed switch, respectively. This is challenging because it requires measuring with high accuracy the volume of liquid in the container before the liquid is released, without interfering with the normal operation of the device. These volumes are not necessary for the system to work correctly, but they would allow us to correct the measurements obtained from the small container –which are less precise–, not just one time but four times for each release cycle of the large container.

3.4 Therapeutic Goals Supervision

The state of the reed switches placed in the containers is sent from the electronic unit to a Java application installed on a PC. From these states and from the values of V^S and V^L, the amount of liquid flow can be calculated. The Java application allows the healthcare staff to inspect a graph showing urine output, and to set the therapeutic goals for the urine output. These therapeutic goals are represented with the aid of the Fuzzy Set Theory, a tool which has proved its value for handling and representing medical knowledge (Barro et al., 2001).

We shall introduce some basic concepts of the Fuzzy Set Theory. Given a discourse universe $\mathbb{R}$ we define a *fuzzy value* C by means of a possibility distribution π^C defined over $\mathbb{R}$ (Zadeh, 1975). Given a precise value $x \in \mathbb{R}$, $\pi^C(x) \in [0,1]$ represents the possibility of C being precisely x. A *fuzzy number* (Kaufmann and Gupta, 1984) is a normal ($\exists x \in \mathbb{R},\ \pi^C(x) = 1$) and convex ($\forall\ x,\ x',\ x'' \in \mathbb{R},\ x' \in [x,x''],\ \pi^C(x') \geq min\{\pi^C(x), \pi^C(x'')\}$) fuzzy value. Normality and convexity properties are satisfied by representing π^C, for example, by means of a trapezoidal representation. In this way, $C = (\alpha, \beta, \gamma, \delta)$, $\alpha \leq \beta \leq \gamma \leq \delta$, where $[\beta, \gamma]$ represents the core, $core(C) = \{x \in \mathbb{R} |\ \pi^C(x) = 1\}$, and $]\alpha, \delta[$ represents the support, $supp(C) = \{x \in \mathbb{R} | \pi^C(x) > 0\}$.

A fuzzy number C can be obtained from a flexible constraint given by a possibility distribution π^C, which defines a mapping from $\mathbb{R}$ to the real interval $[0,1]$. A fuzzy constraint can be induced by an item of information such as *"x has a high value"*, where *"high value"* will be represented by $\pi^{C=high}$. Given a precise number $x \in \mathbb{R}$, $\pi^{C=high}(x) \in [0,1]$ represents the possibility of C being precisely x; i.e., the degree with which x fulfills the constraint induced by *"high value"*.

Physicians are accustomed to expressing the therapeutic goals for urine output in milliliters of urine produced per kilogram of patient body mass per hour $-ml/kg \cdot h$. Our tool allows them to indicate the weight of the patient -*P*- and the therapeutic goals represented by the trapezoidal possibility distribution π^U. π^U can be interpreted as a computational projection of the piece of clinical knowledge "adequate UO". The minimum and maximum values acceptable for urine

output are the beginning and the end of the support of the distribution, respectively. If the patient produces less urine than the amount corresponding with the beginning of the support, the patient is clearly in oliguria. If he/she produces more urine than the amount corresponding with the end of the support, the patient is clearly in polyuria. The beginning and end of the core are the limits of the interval within which ideal values of urine output lie.

If u_i is the urine output in $ml/kg \cdot h$, the degree to which the therapeutic goals established by the clinician are being met is $\pi^U(u_i)$. If $\pi^U(u_i) = 1$, the urine output is within the range of ideal values. If $\pi^U(u_i) = 0$, either the urine output is less than the minimum acceptable value -the patient has oliguria or anuria-, or greater than the maximum acceptable –the patient has polyuria. In both cases, the program produces an audible warning. The closer $\pi^U(u_i)$ is to 1, the closer the amount of urine produced by the patient is to the ideal value, and the closer $\pi^U(u_i)$ is to 0, the closer the patient is to oliguria or polyuria.

In the tool $\pi^U(u_i)$ is represented by a color code used when drawing the graph of urine output. Red represents the null compatibility –the patient is clearly in oliguria or in polyuria–, followed by orange, yellow, green and black, which represents the total compatibility -the urine output lies within the range of ideal values. Therefore, the urine output graph provides instant visual feedback on the patient's state.

The tool also generates an audible warning when the small container is not filled within the maximum time allowed by π^U. This time is given by $\alpha \cdot P/V^S$, α being the beginning of the support of π^U, P the patient's weight, and V^S the volume of the small container.

4 DISCUSSION

Given that the effective volume of the small container is 5.87 $\pm$ 0.32 ml (mean $\pm$ standard deviation), in 95% of cases the amount of urine that is really released when the container is emptied will be in the range 5.87 $\pm$ 0.64 ml (mean $\pm$ 2·standard deviation). Therefore, 95% of the measures derived from the small container will have an error of 10.9% or less. The purpose of this small container is to provide an early warning if the patient's urine output is not within the therapeutic goals: depending on the patient's weight and state this container should be filled within 5 and 12 minutes, if the patient is within the limits of the therapeutic goals. If this does not occur, the healthcare staff is alerted by an audible warning.

To obtain more accurate measures of the total urine output, we rely on the second container. In 95% of cases, the amount of urine released by this container will be within the range 376.72 $\pm$ 2.22 ml. On the other hand, on average half of the effective volume of the small container will not be registered. Thus, the error in the measures in 95% of the cases will be 5.87/2+2.22=5.16; 1.37% of its volume. This error is low, especially when compared to errors committed by the nursing staff when they take measures visually, which has been reported to be as high as 26% (Hersch et al., 2009).

Our device is capable of providing feedback on the status of the patient's kidneys more frequently than is currently available in critical care units – hourly. For a patient of 80 kilos, which should produce at the very least least 40 milliliters of urine per hour, our device could warn of a deviation from the therapeutic goals in less than 10 minutes – the time in which the small container should be filled if the patient produces urine within the therapeutic goals. Therefore, it has the potential of improving patient outcome. Given that it provides feedback on the patient's state at shorter time intervals, it can allow the clinician to react more promptly to complications in the state of the patient.

The containers needed to build our device are not more complex than the containers used in commercial urine meters, which often require a small container embedded within a large container, valves that communicate both container with each other and with a plastic bag, mechanisms to prevent the containers from overflowing, etc. However, the price of our manufacturing our device is slightly higher than the price of the commercial urine meters because of the addition of four pieces: two floats, and two magnets. The rest of the pieces that are part of our solution do not have to be discarded because they are not in direct contact with the patient's urine, nor do they suffer significant degradation caused by its operation. Thus, their cost can be amortized over a long period of time and their impact on the overall cost of the solution is negligible.

Despite its higher cost, the device has the potential of producing significant economic savings for the institutions that provide healthcare services. Currently, a nurse must visit each of the patients's beds of the critical care unit to manually record urine output every hour. The nurse must put on gloves since he/she is going to manipulate body fluids, walk to the patient's bed, take the measure visually, write it down, open the valve that releases urine from the graduated container to the plastic bag, wait for the urine to drain, close the valve and check if the plastic bag needs to be emptied. This procedure takes at least 2 minutes.

In a small critical care unit with only 10 patients, this means 20 minutes per hour; 8 hours a day. Typically, nurses work in shifts of six hours. Therefore, each day one and a third nurses' shift are required only for tasks related to monitoring urine output. Even a partial automation of these tasks has the potential of yielding significant economic savings.

5 CONCLUSIONS AND FUTURE WORK

We have built a device capable of automatically sensing and supervising the urine output of critical care patients. The device comprises two containers of different volumes, a small one that receives the urine coming from the patient's bladder, and a greater volume container in which the first container releases its content when it gets full. Both containers release their content automatically when they are filled using a siphon mechanism.

The containers are equipped with reed switches that are activated by a magnet that is attached to a float located inside the containers. These reed switches allow us to identify the instants at which they get filled with urine. An electronic unit sends via Bluethooth the information provided by the reed switches to a PC which calculates the urine output from the switches' state, and supervises the achievement of the therapeutic goals established by the clinician. The error in measuring the patient's urine output is under 2%. The large container is the one which allows us to obtain this high accuracy, while the small one permits an early warning of deviations from the therapeutic goals.

The cost of our device is slightly higher than the cost of the commercial devices currently used in monitoring urine output. However, the device has the potential to save costs for the institutions that provide health services by freeing a considerable amount of time for the healthcare staff. Furthermore, it provides a more continuous supervision of the urine output than is currently carried out in critical care units, which may help improve patient outcomes.

As future work, we intend to take advantage of all the state changes of the reed switches of the large container to correct the urine output measures while the large container is been filled. Currently, this correction is only performed when the large container releases the urine. We also will start to use our device in animal tests conducted in a research unit associated with Getafe University Hospital. After this phase, we intend to use it in a pilot test in the Intensive Care Unit of this hospital.

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REFERENCES

Akinfiev, T.; Apalkov, A.; Otero, A.; Palacios, F. Device for measuring the amount of liquid that flows and procedure for its measurement. Patent Pending, 2010. ES P201031227.

Barro, S., Marín, R., Palacios, F., and Ruíz, R. (2001). Fuzzy logic in a patient supervision systems. *Artificial Intelligence in Medicine*, 21:193–199.

Hande, A., Polk, T., Walker, W., and D.Bhatia (2006). Self-powered wireless sensor networks for remote patient monitoring in hospitals. *Sensors*, 6(9):1102–1117.

Hersch, M., Einav, S., and Izbicki, G. (2009). Accuracy and ease of use of a novel electronic urine output monitoring device compared with standard manual urinometer in the intensive care unit. *Journal of Critical Care*, 24:629–633.

Ishida, S. (1990). Liquid level indicator using laser beam. us patent 4938590.

Johnson, S. J. (1978). Liquid level measurement device. United States Patent 3693445.

Jungk, A., Thull, B., and Rau, G. (2002). *Fuzzy Logic in Medicine*, chapter Intelligent Alarms for Anaesthesia Monitoring Based on Fuzzy Logic Approach. Physica-Verlag.

Kaufmann, A. and Gupta, M. (1984). *Introduction to Fuzzy Arithmetic*. Van Nostrand Reinhold Company Inc.

Mitra, B., Fitzgerald, M., Cameron, P., and Cleland, H. (2006). Fluid resuscitation in major burns. *ANZ Journal of Surgery*, 76:35–38.

Mora, F., Passariello, G., Carraylt, G., and Pichon, J. (1993). Intelligent patient monitoring and management systems: A review. *IEEE Engineering in Medicine and Biology*.

Otero, A., Akinfiev, T., Fernández, R., and Palacios, F. (2009a). A device for automatic measurement of critical care patient's urine output. In *6th IEEE International Symposium on Intelligent Signal Processing*, pages 169–174.

Otero, A., Akinfiev, T., Fernández, R., and Palacios, F. (2010). A device for automatically measuring and supervising the critical care patients urine output. *Sensors*, 10(1):934–951.

Otero, A., Félix, P., Barro, S., and Palacios, F. (2009b). Addressing the flaws of current critical alarms: a fuzzy constraint satisfaction approach. *Artificial Intelligence in Medicine*, 47(3):219–238.

Otero, A., Panigrahi, B., Palacios, F., Akinfiev, T., and Fernández, R. (2009c). *Recent Advances in Biomedical Engineering*, chapter A prototype device to measure and supervise urine output of critical patients, pages 321–324. Intech.

Rivers, E., Nguyen, B., e Havstad, S., Ressler, J., Muzzin, A., Knoblich, B., Peterson, E., Tomlanovich, M., and Group, E. G.-D. T. C. (2001). Early goal-directed therapy in the treatment of severe sepsis and septic shock. *New England Journal of Medicine*, 345:1368–1377.

Zadeh, L. (1975). The concept of a linguistic variable and its application to approximate reasoning. *Information Science*, 8:199–249. Part 1.