Microwave TDR for Real-Time Control of Intravenous Drip Infusions

Andrea Cataldo, Giuseppe Cannazza, Nicola Giaquinto, Amerigo Trotta, and Gregorio Andria

***Abstract*—This paper explores the use of a microwave-reflectometry-based system for the automatic control and real-time monitoring of the flow and of the liquid level in intravenous (IV) medical infusions. In medical and hospital contexts, other kinds of devices, mainly based on the optical detection and counting of the infusion drops, are used. Nevertheless, the proposed system is aimed at circumventing some typical drawbacks deriving from the adoption of these traditional methods, thus allowing an efficient alternative for automatically monitoring the instantaneous flow of IV medical solutions. To this purpose, the proposed system combines microwave time-domain reflectometry (TDR) measurements with a noninvasive sensing element (i.e., strip electrodes directly attached to the external surface of the infusion bottle). Experimental results confirm that, by using low-cost portable TDR devices, the solution flow process can be controlled with acceptable accuracy. Therefore, the proposed method can be regarded as a promising control tool for in-hospital patient management as well as for telemedicine programs.**

***Index Terms*—Biomedical monitoring, medical services, microwave measurements, microwave reflectometry, time-domain reflectometry (TDR).**

I. Introduction

INTRAVENOUS (IV) and intra-arterial access constitutes an excellent pathway for continuous administration of drug or pharmaceutical therapy. As a matter of fact, the delivery system should be able to regulate a certain drug concentration in the patient's body, which is needed for a desired therapeutic result. For this reason, the accurate control and monitoring of the infusion process are of paramount importance. In fact, although underinfusion may cause an insufficient therapy, overinfusion, on the contrary, may provoke serious toxic effects [1]. On such basis, the IV infusion processes require a constant and accurate control of the flow rate and of the correct dripping. Gravity IV infusion is widely used in many medical facilities; typically, the drip rate is initially regulated manually, and the infusion dripping is controlled visually by counting the drops for 15–30 s. The fixed drip rate may vary due to different causes such as dilation or contraction of the vein, decrease of the fluid pressure, variation of the patient's body temperature, and patient's motion.

In order to reduce the frequent intervention of medical personnel, which typically would occur every 15–20 min, different kinds of drip rate meters have been developed. The sensing technology of such systems typically relies on optical sensors, which detect the drops falling in the drip chamber [2]–[4]. Commonly, an infrared transmitter and a receiver are attached to the drip chamber, and the flow rate is monitored by counting the number of drops falling in a certain time. Although these systems based on the optical detection of drops are the most widely used, in [5], a different drop-sensing method is presented. It is based on a capacitive system formed by some noncontacting multiple electrodes wrapped around the infusion supply components.

For all these existing systems, the design choice of counting the number of drops implies some critical performance limitations. A first limitation concerns the need of performing adjustments related to the volume of the drops which principally depends on the geometric dimensions of the tubing. As a consequence, the infusion control accuracy is strongly limited by the viscosity and volume variation of drops (which is in the range of 0.05–1 mL), as well as by the flow rate variation [1]. Furthermore, when dealing with optical detection, since the sensing element is mounted on the transparent drip chamber, another critical issue is the correct alignment of the optical sensor so as to optimize the overall sensitivity performance. Finally, the light scattering response depends on the specific optical properties of the drip chamber and on those of the infusion liquids, which, in turn, can be of various natures, such as saline solutions, drugs, blood, or more complex mixtures.

On the other hand, the adoption of active pumping infusion devices [6], which combine electronics with a mechanism to generate flow, may lead to some improvement of performance with respect to the traditional gravity-based systems, owing to the accuracy enhancement in the flow rate control. These systems, however, besides being quite complex, are still based on the counting of drops and are therefore prone to the problems related to the variable volume of the drops.

An alternative approach to circumvent these performance limitations is monitoring the flow without counting the drops. The method presented in [7] is based on the use of ultrasound for finding the tube diameter and for measuring the flow rate. Nevertheless, one weak point of this method is the complexity of the system. In fact, the final results and, most importantly, the associated uncertainty derive from the combination of the measurements of the tube diameter and the flow rate.

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The limitations of the state-of-the-art systems for IV infusion control justify the investigation on alternative methods, involving the direct sensing of the liquid volume variations inside the solution container.

It is well known that the monitoring of liquid levels (or equivalently of volumes) can be performed through several methods, including (besides optical systems) ultrasonic or capacitive sensors, as well as microwave sensing [8]. In particular, microwave time-domain reflectometry (TDR) is proven to be a powerful and effective method for quantitative characterization of several liquids. A drawback of TDR measurements, however, is that they commonly involve the insertion of a specific sensing probe into the samples under test, thus monitoring the dielectric characteristics and the quantity of the investigated material [9]–[11].

Nevertheless, it is worth pointing out that, different from the traditional TDR-based liquid-level detection, a major requirement for the proposed application is the noninvasiveness of the sensing element, due to the necessity of not contaminating the medical preparation. Additionally, some other important aspects to be taken into account are the sensitivity, resolution, and accuracy of the measurements, as well as the instrumentation portability and the possibility of multiplexing measurements. In particular, multiplexing possibility would strongly reduce the implementation costs in the typical case of several in-hospital patients, who could be monitored simultaneously by a single instrumental unit.

On the basis of these considerations, the present work aims at developing an alternative monitoring method, based on the TDR technique, which can meet all the aforementioned requirements, thus guaranteeing a real-time, low-cost, and automatic control of the IV infusion process. The noninvasiveness of the sensing element is guaranteed by the adoption of two-strip electrodes, directly attached to the external surface of the solution container and contacted to the TDR coaxial cable through two feed points. Experimental results, reported for two cases, confirm the applicability of the method, which is based on the following three main steps.

1) First, TDR measurements are performed, and the reflected waveform corresponding to the probe portion in contact with the infusion bottle is acquired as a function of the apparent distance ($L_{\rm app}$). In this regard, it is worth recalling that $L_{\rm app}$ is defined as the distance traveled by the signal that propagates along the transmission line and the probe [10].
2) The reflected waveform is processed through a specific algorithm so as to remove the noise in excess and to interpolate the signal.
3) Finally, through the derivative of the signal, the clear peaks occurring at the beginning and termination of probe are suitably associated to the corresponding liquid-level variation.

As a consequence, the proposed method can be suggested as an excellent monitoring tool for IV infusion process, both for in-hospital patient management and for telemedicine programs [12].

On a side note, it is important to underline that the controlling program can be suitably designed to give an alarm if the IV drip rate gets out of a fixed range or when the liquid volume decreases below a certain value.

II. Basic Theory and Experimental Setup

In TDR, the system to be monitored is stimulated through an appropriate electromagnetic signal, typically a steplike voltage pulse, which is propagated through a probe; any impedance variation will cause the partial reflection of the propagating signal. The analysis of the reflection coefficient in time domain ρ allows the retrieval of the dielectric characteristics of the material under test, as well as of its quantitative parameters, such as in the case of the level of liquid materials. When considering the case of a liquid sample having a certain level L, at the air-to-liquid interface, due to the difference in dielectric permittivities, a significant change in the characteristic impedance of the probe occurs. Therefore, the measured reflection coefficient shows a significant variation which may be used to individuate the air-to-liquid interface. For homogeneous materials, the reflection coefficient remains approximately constant until the reflection caused by the probe termination occurs. The analysis of TDR waveform directly leads to the evaluation of $L_{\rm app}$ (also called electric distance), which can be considered as the distance that would be traveled by the signal, in the same interval of time, if the signal were propagating at the speed of light. Typically, in TDR liquid-level measurements, the apparent distance is evaluated through the individuation of the beginning and end of the probe; thus, it can be associated to the physical length L through the following equation:

$$L_{\rm app} = \sqrt{\varepsilon_{\rm app}} L = \frac{ct_t}{2} \tag{1}$$

where t_t is the travel time (round-trip time taken by the signal to travel between the beginning and end points), $\varepsilon_{\rm app}$ is referred to as apparent dielectric permittivity of the material in which the probe is inserted, and c is the speed of light [9].

For the intended application, since the noninvasiveness of the probe is a stringent requirement, the adopted design strategy was to attach the sensing element to the external surface of the bottle in which the medical liquid is contained. For this purpose, the most adequate and easy-to-implement probe configuration is the so-called two-strip probe. In fact, this adopted configuration can simultaneously guarantee an easy and low-cost fabrication, a good stability, and dimension control when it is attached, along with very low impedance mismatch and suitable connection to the coaxial cable. It is worth noting that, although a thickness of glass is interposed between the probe and the internal part of the bottle, the level variation, caused by the liquid-to-air interface change, can be suitably sensed by the reflection of the electromagnetic signal.

This *ad hoc* probe configuration, shown in Fig. 1, was specifically realized and improved with respect to the preliminary one presented in [11]; two copper strips, with a width of 3 mm and a mutual distance of 3 mm, were placed on a biadhesive tape. The two-strip electrodes were directly contacted to the inner and outer conductors of a coaxial cable, thus permitting the connection with the TDR instrument. In this regard, it is important to highlight that, in order to minimize the parasitic impedance

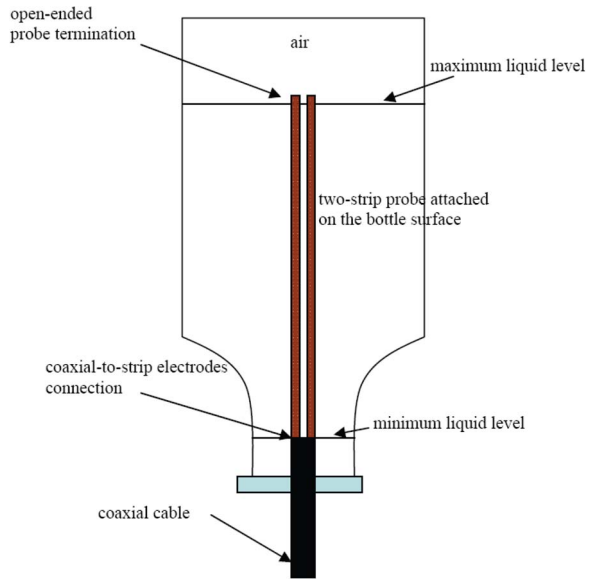

Fig. 1. Scheme of the experimental configuration: The two-strip probe is attached to the infusion bottle surface and is contacted with the coaxial cable.

mismatch at the coaxial-to-strip transition, the mutual distance between the two electrodes should be maintained as short as possible. Furthermore, another crucial issue associated with good measurement accuracy relates to the distance constancy between the strips, while no particular requirements are needed for the length. Finally, as for the termination of the strips, the open-ended configuration is the most adequate design choice. In fact, in addition to its realization simplicity, the high load impedance introduced by the open circuit can be promptly identified in terms of TDR reflected signal.

On the basis of these considerations, for the designed configuration, there are two clear discontinuities which will cause abrupt variations in the reflection coefficient measured by the TDR. The first one is located at the coaxial-to-strip connection, and the second one is at the liquid–air separation surface in the bottle. Therefore, the TDR method, in conjunction with the designed probe, is definitely adequate to determine the level of the liquid in the bottle. As a matter of fact, to accurately detect the level variation, the top surface of the liquid must be kept perpendicular to the bottle axis. For this reason, particular care is required in checking the alignment between the bottle and the holder.

The TDR measurements reported in this paper were performed through the HL1500 TDR unit, which generates a steplike signal with a rise time of approximately 200 ps. This fast rise time, which ensures a good spatial resolution, is associated to a corresponding frequency content of the signal (i.e., -3-dB bandwidth attenuation) of approximately 1.7 GHz. The steplike signal amplitude is ± 250 mV, and the maximum current value, during the measurement, is 270 mA; hence, the peak power consumption is 0.135 W or lower. The used instrument is a low-cost portable unit which also permits the measurement multiplexing. The instrument was interfaced to a PC, thus also providing the possibility of automatically repeating the acquisitions, and to process them properly in order to obtain the final level measurement.

III. Experimental Results

The experiments were conducted on two glass bottles containing different volumes of physiological solution of NaCl (0.9% in weight), which were 250 and 500 mL, respectively. Starting from the maximum volume condition, measurements were repeated for consecutive decreases of a reference volume of 5 mL. This reference volume was estimated with an uncertainty of 0.5 mL, owing to a graduated drip chamber, in which the infusion liquid dropped. It is worth mentioning that, through some specific preliminary experiments, this quantity of volume variation was determined as the minimum one for which the corresponding differences in TDR signal were suitably detectable. According to this assumption, 5 mL of IV solution, which corresponds approximately to 50–100 drops, can be considered as the resolution of the measure. However, in order to assess a rigorous metrological characterization of the method, a separate uncertainty analysis has been performed and reported in Section III-C. In this way, for each 5-mL-step volume flowing in the graduated drip chamber, the TDR measurement was acquired and processed. Consequently, the corresponding L_{app} is evaluated through the derivative of the signal, and it is directly associated to the reference volume of the liquid remaining in the bottle.

In the following sections, the proposed procedure is detailed, and the obtained results are reported for the experiments with the 250-mL bottle. Experiments with the 500-mL bottle are not substantially different, as regards waveform acquisition and processing method, and therefore, only the final results are reported for this case.

A. Acquisition of the TDR Waveforms

The HL1500 TDR unit acquires waveforms always in averaging mode, i.e., the provided waveform is the average of a minimum of 2 and a maximum of 128 single internal acquisitions. Since the final accuracy was a prominent issue, the maximum number of averaged waveforms (128) has been selected for all the measurements. Each waveform is made of 2048 samples.

In order to illustrate the generic response modality, two acquired TDR waveforms are shown in Fig. 2, which correspond to the maximum and the minimum liquid level in the 250-mL bottle, respectively.

In the figure, the two reflection points, corresponding to the coaxial-to-strip connection and the liquid-to-air separation surface, are clearly visible. The first one (corresponding to the first steep variation in the waveform, occurring approximately at 1.1 m) is reasonably constant for all the values of liquid level. The second step in the waveforms varies with the variation of the liquid-to-air surface. The signal amplitude corresponding to the open-ended termination of the probe exceeds the ideal value of $+1$. This effect is due to the additive contribution of the typical multiple reflections caused by the signal which propagates back and forth along the probe.

The apparent distance L_{app} between these two reflections is directly and uniquely related to the actual distance between the points and, ultimately, to the fluid volume V in the bottle. It goes without saying that the dependence between the evaluated

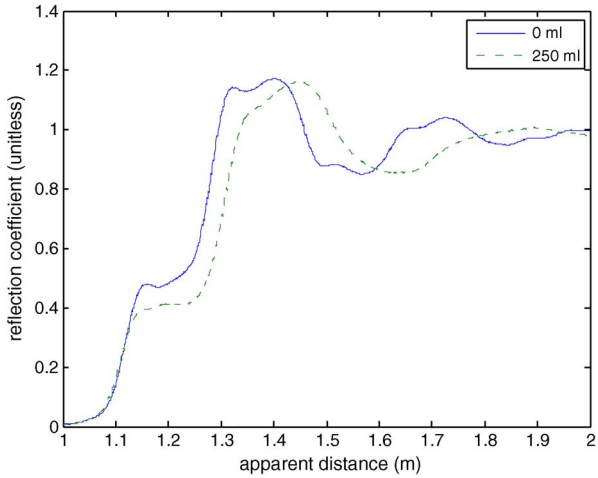

Fig. 2. TDR waveforms acquired for two different levels of the liquid in the 250-mL bottle.

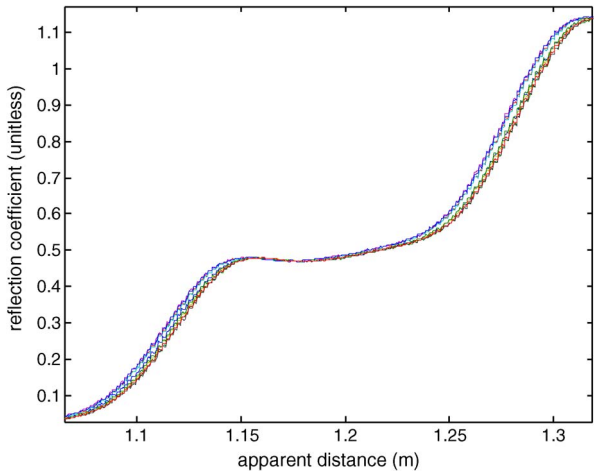

Fig. 3. Set of ten repeated TDR measurements for the 0-mL volume in the 250-mL bottle. The reported apparent distance range includes only the waveform portions corresponding to the two reflection points.

$L_{\rm app}$ and the corresponding volume is not linear, mainly because of the curved surface of the bottle.

In order to evaluate the repeatability of the measurements, for each of the considered liquid levels, the TDR measurement has been repeated ten times. An example of the typical results is shown in Fig. 3, which refers to the ten repeated measurements acquired for the 0-mL volume in the 250-mL bottle.

The accurate evaluation of the apparent distance $L_{\rm app}$ between the two significant reflection points can be performed through the so-called derivative method [9].

In practice, the clear peaks appearing in the first derivative curve of the TDR waveforms can accurately localize the main impedance discontinuities, corresponding to the different interfaces (i.e., the coaxial-to-strip connection and the liquid-to-air interface). However, due to the noise presence and the quantization effect, the derivative cannot be accurately evaluated by means of a simple difference between consecutive samples. In order to obtain a better first derivative, a specific algorithm, based on the well-known Nicolson method [13] for the spectral analysis of steplike signals, was developed. The algorithm yields satisfactory results for the intended application; it is particularly effective for the determination of the reflection points corresponding to the air-to-liquid interface variations.

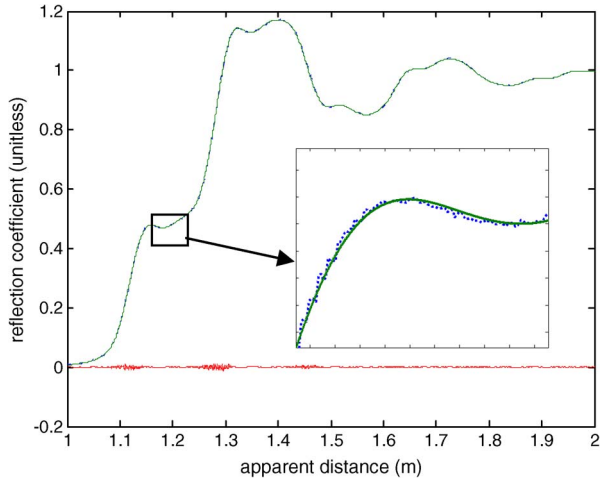

Fig. 4. Plot of an original and a filtered TDR waveform corresponding to the 0-mL volume in the 250-mL bottle. The residual difference (line in the neighborhood of the zero level) and a "zoom" of the two signals are also reported.

B. Processing Algorithm for the TDR Waveforms

The filtering and differentiation of the TDR waveforms have been performed through a proper implementation of the aforementioned Nicolson method. First of all, a linear function is subtracted from the TDR waveform so that, after the subtraction, the first and the last sample have the same value. After this "detrend" operation, the resulting waveform may be considered as a single period of a periodic signal, and therefore, any linear transformation of the signal (including filtering, interpolation, derivation, etc.) is a proper operation on its fast Fourier transform (FFT) coefficients.

As regards the filtering threshold, previous experiences with the same TDR unit confirmed that the useful frequency content of the acquired waveforms goes up to $B_{\max} = 2.7$ GHz. The frequency resolution of the FFT is

$$f_0 = \frac{c}{2 \cdot L_{\rm app}^{\rm tot}} = 0.15 \cdot 10^9 \text{ Hz} \tag{2}$$

where $c \cong 3 \cdot 10^8$ m/s is the velocity of light in vacuum and $L_{\rm app}^{\rm tot} = 1$ m is the total apparent distance corresponding to the acquisition interval of the waveform. Therefore, the maximum harmonic order (N_h) including the useful spectral information can be calculated as

$$N_h = \frac{B_{\max}}{f_0} = 18. \tag{3}$$

The TDR waveform corresponding to the 0-mL volume in the 250-mL bottle, resulting after the elimination of the harmonics above the said limit, is shown in Fig. 4, together with the residual difference (i.e., the difference between the filtered and the original waveform). It is visually clear that the selection of 18 FFT coefficients allows a very good reconstruction of the original waveform, at the same time eliminating almost all the noise.

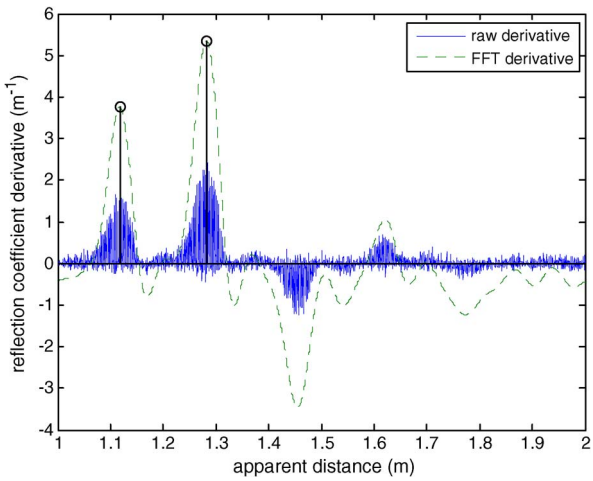

Fig. 5. Numerical derivative of the waveforms shown in Fig. 4. The raw derivative obtained by simple differences between consecutive samples is also reported for comparison. The maxima determined on the basis of the FFT derivative are highlighted.

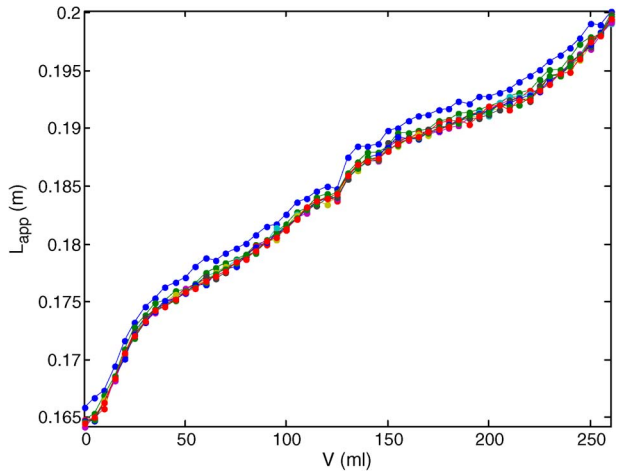

Fig. 6. Results of the ten separate measurements of $L_{\rm app}$, for a set of 53 liquid volumes in the 250-mL bottle. The range 0–260 mL has been covered, through steps of 5 mL.

The filtered waveform is interpolated and differentiated in the frequency domain. More specifically, the FFT coefficients are multiplied by $j2\pi f_0$ (first derivative), and a proper number of zero harmonic coefficients are added (frequency-domain zero padding), obtaining a total of 20 480 harmonic coefficients (ten times the original number of samples). Then, the inverse FFT yields the interpolated first derivative of the filtered waveform, in which the maxima are easily located. Fig. 5 shows the result of the whole processing applied on the waveform in Fig. 4. From this figure which, for comparison, also reports the raw derivative obtained by simple differences between consecutive samples, the performance enhancement in peak detection is clearly evident. In this way, the evaluation of $L_{\rm app}$ corresponding to the distance between the two maxima can be evaluated with high accuracy and repeatability. As a practice consequence, the liquid-level variation could be controlled by an automatic peak searching tool. This operative modality is independent from bottle geometries or infusion type, provided that a preliminary calibration procedure is performed.

C. Calibration Curves and Measurement Uncertainty

Fig. 6 shows the $L_{\rm app}$ evaluated for 53 different liquid levels, assumed as reference or "true" levels, in the 250-mL bottle. The reference levels are equally spaced with a step of 5 mL, ranging from 0 to 260 mL. As aforementioned, the TDR measurement has been repeated ten times for each level, and the figure reports the corresponding ten separate curves versus the reference liquid volume.

All the curves are very close among them, except a single one, which is above the others and corresponds to the *first* measurement made at each considered volume. This phenomenon, which has been systematically observed in all the experiments, is attributable to an intrinsic "settling effect" of the TDR instrument. Consequently, in view of a practical application involving the use of that instrument, the control software should be set so as to ignore the first measurement and to subsequently consider the second one. Under this hypothesis, it is legitimate to consider the nine curves following the first one (which is discarded), in order to derive the calibration curve and the measurement uncertainty.

The calibration curve is obtained as follows.

1) Let y_n, $n = 1 : N$, be the values of the reference liquid volume V. For the 250-mL bottle, the values are simply $y_n = (n-1) \cdot 5$ mL, and $N = 53$.
2) Let x_{ni}, $n = 1 : N$, $i = 1 : 9$, be the evaluated values of $L_{\rm app}$ for the nth reference liquid volume and the ith curve.
3) Let $x_n = \bar{x}_{ni}$ be the average over i of the values x_{ni}.
4) The calibration curve, which associates a proper volume $y = V$ to any given $x = L_{\rm app}$, is the nonlinear function $g(x)$ made by single linear functions connecting the points (x_n, y_n). As a matter of fact, being $g(x)$ a piecewise linear function, its evaluation is quite simple.

Aside from knowing the $y = V$ associated to a given $x = L_{\rm app}$, it is important to know the uncertainty $U(V)$ associated to the measured volume. The uncertainty may be estimated from the nine repeated measurements, using the following algorithm.

1) Evaluate $s_{xn} = \mathrm{std}(x_{ni})$, the sample standard deviation over the index i of the measurements x_{ni} corresponding to the reference volume y_n.
2) For a confidence level $l = 90\%$, evaluate $U_{xn} = s_{xn} \cdot t_{(1+l)/2}$, where $t_{(1+l)/2} = 1.86$ is the student's t quantile with $9 - 1 = 8$ degrees of freedom. U_{xn} is the expanded uncertainty in the determination of x_n corresponding to a given volume y_n.
3) To evaluate the uncertainty U_{yn} in the determination of the volume y_n corresponding to a given x_n, take the maximum between the two deviations

$$U'_{yn} = g(x_n + U_{xn}) - g(x_n) \tag{4}$$

$$U''_{yn} = g(x_n) - g(x_n - U_{xn}). \tag{5}$$

Deviations (4) and (5) are evaluated through two distinct linear functions if $x_n + U_{xn} \le x_{n+1}$ and if $x_n - U_{xn} \ge x_{n-1}$,

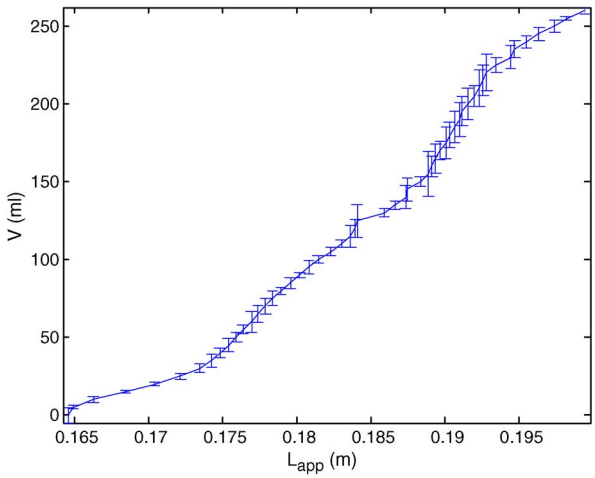

Fig. 7. Calibration curve for the 250-mL bottle. The curve represents the reference liquid volume V corresponding to the evaluated apparent distance $L_{\rm app}$. The error bars represent the student's t expanded uncertainty in the estimation of V, associated to a confidence level of 90%.

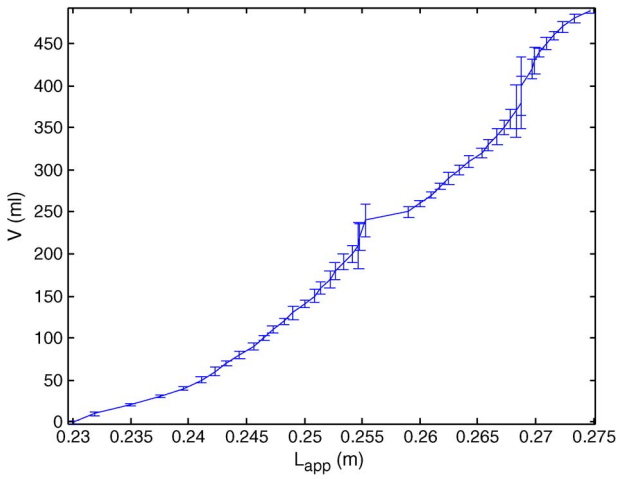

Fig. 8. Calibration curve for the 500-mL bottle, with the error bars representing the student's t expanded uncertainty in the estimation of V, with a confidence level of 90%.

respectively; otherwise, they are calculated using the proper piecewise linear function. The final result (i.e., the calibration curve and uncertainties in the volume measurements) is shown in Fig. 7. The uncertainty is 5.46 mL on average, ranging from only $U = 0.56$ mL (for $L_{\rm app} = 0.1684$ m, corresponding to $V = 15$ mL) to $U = 14$ mL (for $L_{\rm app} = 0.1889$ m, corresponding to $V = 155$ mL). Even if an improvement in accuracy is desirable and probably possible with improved hardware, the average uncertainty of 5.46 mL is acceptable, and the methodology appears quite promising on the basis of these preliminary results.

Finally, Fig. 8 shows the results obtained for the 500-mL bottle. In this case, the average uncertainty is 8.65 mL, and the minimum uncertainty is 1.41 mL. Unfortunately, for certain volume ranges, the measurement repeatability is slightly lower, resulting in a maximum uncertainty value of 34 mL. This effect is attributable to the fact that, for the 500-mL-bottle case, the 5-mL-step volume decreases are associated to variations in values of $L_{\rm app}$ smaller than those observed in the previous case. As a consequence, the calibration function $g(x)$ has a higher slope in the corresponding points (x_n, y_n); hence, according to (4) and (5), a higher uncertainty is observed.

D. Comparative Analysis on the Uncertainties of the Proposed and Existing Methods

The uncertainty values obtained in the previous section are summarized in Table I (percent values are referred to the full scale).

As a first observation, it should be noted that the obtained uncertainty relates to the characterization of the repeatability and not of the reproducibility of the measurements. As a matter of fact, a rigorous reproducibility analysis would require the consideration of independent prototypes of the measurement system and a standardized realization of the sensing element and of the attachment mode. This is outside the scope of this paper. However, on the basis of the performed experiments, a substantial increase in the overall uncertainty is not expected, provided that suitable reproducibility conditions are considered.

TABLE I
UNCERTAINTY VALUES OF THE METHOD

Bottle volume	minimum U		average U		maximum U	
250 ml	0.56 ml	0.22%	5.46 ml	2.2%	14 ml	5.6%
500 ml	1.41 ml	0.28%	8.65 ml	1.7%	34 ml	6.8%

Furthermore, a direct comparative analysis on the uncertainties of the proposed and existing methods is rather difficult. In fact, as stated in the Introduction, the existing systems and, consequently, the related performance specifications are usually based on the counting of drops [2]–[7] rather than the volume evaluation. Typical errors affecting the drop-counting systems are on the order of 2 drops/min, as reported in [4] and [6]. For a typical drop rate of 40 drops/min, this specification is equivalent to an error of 5%. This value is reasonably low at short times of the infusion process but becomes significantly higher as the infusion time increases. As a consequence, the accumulated error is 12.5 mL for a 250-mL bottle, and it is 25 mL for a 500-mL bottle. For a slower rate equal to 20 drops/min, the error is doubled.

On the basis of such analysis, it is clear that the development of detection method based on the evaluation of the volume variation can successfully lead to a substantial performance improvement. In this regard, the average uncertainty associated to the proposed system is on the order of 2%, with some additional reductions for low liquid levels (implying a detection improvement in the critical phase when the infusion process is approaching the end). This uncertainty limitation is substantially improved with respect to the performance of the existing systems.

As a final remark, for the proposed method, an additional reduction of the uncertainty (at least in some critical points of the calibration curve) is desirable, along with the assessment of the reproducibility performance. These are goals to be reached in a further future research.

IV. Conclusion

A microwave-reflectometry-based measurement method for real-time continuous monitoring of the IV drip infusion has been investigated. The monitoring system presents the advantages of high accuracy, modularity, low cost, and portability that make it particularly attractive for practical *in situ* medical applications.

The noninvasiveness of the sensing element is adequately accomplished by an *ad hoc* adhesive two-strip probe, whose adoption does not compromise the overall measurement accuracy. Furthermore, the procedure for evaluating the apparent liquid level, and for relating it to the corresponding liquid volume, is simple and easy to implement. These aspects make the proposed system more attractive with respect to the existing monitoring solutions, which are mainly based on sensing of the drops falling in the drip chamber. As a key issue, the evaluated uncertainty values are significantly lower than the typical errors associated to the traditional systems.

The direct relationship between the variation of the liquid volume and the measured data allows one to individuate specific calibration curves, which can be used to monitor in real time and *in situ* the evolution of the entire IV drip infusion process. The implementation of such a system (versatile, low cost, remotely controllable, and metrologically accurate) is quite promising, as it would allow a substantial improvement of the in-hospital patient management as well as of the telemedicine programs.

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