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MCM-based microlaboratory for simultaneous measurement of several biochemical parameters by spectrophotometry

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Abstract This paper reports on the concept, fabrication and characterization of a multi-chip module microlaboratory. The application is in the spectrophotometric analysis of human physiological fluids in a clinical setting. The system is composed of three stacked parts: (1) a central microfluidic system die containing the microchannels, which is fabricated by applying MEMS techniques to an SU-8 layer, (2) an optical filtering system on the top side, fabricated using a dielectric thin-films multilayer and (3) a detection and readout system on the bottom side, which is fabricated in a CMOS microelectronic process. The system modularity and emphasis on microfabrication provides potential for low unit cost. The application is the simultaneous and quantitative measurement of the concentration of four biochemical parameters in human physiological fluids by spectrally selective optical absorption. The intensity of the light transmitted through the physiological fluid results in an electrical output signal in the form of bit streams, which allows simple computer interfacing. A simple white light source is used for illumination, due to the optical filtering system, which makes the microlaboratory portable. The quantitative measurement of chloride, creatinine, total protein and uric acid in urine is successfully demonstrated.

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1 Introduction

Biochemical analysis of physiological fluids of a patient is a standard routine procedure in a clinical setting for diagnosing the patient. Depending on the stage of treatment, the purpose is in detecting or confirming the illness, measuring the progress towards recovery or confirming full recovery (Todd et al. 1984). The symptoms are the starting point of the diagnosis. Most diseases leave a molecular fingerprint in the physiological fluids, which can be measured to significantly improve the precision of the diagnosis (Tüdos et al. 2001; Connolly 1995). At the present, a patient has to go to a clinical laboratory for this purpose, even in case of small routine tests or periodic checkups. Tests results may take hours or even days to become available to the medical practitioner (Strasinger and Di Lorenzo 2001). As a consequence, a reliable diagnosis cannot be performed within the consultation time. Moreover, the sample should be subjected to a huge logistics system to avoid mix-up within the many samples send to the laboratory. This situation is changed drastically once the physician can perform the analysis at his office and be provided with immediate results with precise information about the state of the patient. This is the feature pursued with the microlaboratory reported in this paper.

Disciplines such as microelectronic device fabrication, electric circuit design and analytical chemistry have to be combined to make the development and optimization of microsystems for biochemical analyses possible, which have been revolutionizing the field of biochemical sensing

(Minas 2008). The advantages associated with shrinking clinical analyses systems include: reduced sample size, higher degree of integration and hence enhanced potential for automation, reduced response time, potential for improved analytical performance, reduced chemicals storage and hence laboratory safety and a considerable reduction in sample handling, logistics and costs (Irawan et al. 2007).

In clinical diagnostics, the spectrophotometric analysis (the study of the interaction of electromagnetic radiation with chemical compounds) is one of the most commonly used analytical techniques for physiological fluids analysis. The spectrophotometry by optical absorption is often used to determine the concentration and/or amount of a particular biochemical parameter in physiological fluids samples (Strasinger and Di Lorenzo 2001). The measurement is based on the measurement of the optical absorption spectrum over a well-defined part of the visible spectrum. Since the absorption peak wavelength is parameter-specific, the sample can in principle be analyzed by scanning the different wavelengths.

However, many of the analytes (an analyte is the parameter being analyzed) of interest for clinical analysis do not have chromophores that absorb light in a useful part of the visible range. Suitable chemical compounds are available (reagents) to transform these analytes into colored products that do have adequate absorbance (Sigma-Aldrich 2008). The combined effect of a particular biochemical parameter present in the sample with a suitable reagent does provide the desired effect: (1) an absorption maximum at a parameter-specific wavelength in the visible spectral range and (2) an absorbance at this wavelength that is directly proportional to the concentration of that particular biochemical parameter in the sample.

The application of the particular microlaboratory presented here is the spectrophotometric analysis of physiological fluids, especially the simultaneous measurement, by optical absorption, of the concentration of four biochemical parameters in those fluids. With this device, the physician will have instantaneous access to critical information needed.

2 Microlaboratory design and fabrication

The reported microlaboratory comprises three main microsystems in a stacked multichip module (MCM)-based microsystem: the microfluidic system (Fig. 1(a)), the optical filtering system (Fig. 1(b)) and the detection and readout system (Fig. 1(c)). Operation is based on optical absorption at a well-defined wavelength of the visible spectrum. The device is illuminated from the top side using a white light beam, which is guided through the detection chambers containing the samples to be analyzed (Fig. 1(d)). The impinging light is filtered by the optical passband filters, to yield a narrow spectral band centered at the wavelength at which the colored mixture that is being analyzed has its absorption maximum. The intensity of the selected spectral component transmitted through the fluid is measured using underlying photodetectors, which are vertically aligned with the optical filters (Fig. 1(d)). This optical intensity is proportional to the analyte concentration. A Sigma-Delta ($\Sigma\Delta$) converter is integrated with the photodetectors to convert the analog signal into a digital signal for further computer interfacing (Candy and Temes 1992). The optical detection microsystem avoids the need for expensive readout optics and may enable low-cost disposable devices, which would improve the use of spectrophotometric analysis in clinical diagnostics.

2.1 Microfluidic system

The practical limitation to the shrinking of clinical analysis systems (i.e. the sample volume or the size of the microfluidic device) is set by the specified detection limit in concentration measurement of the desired analyte (Petersen et al. 1998). The concentration determines how many target molecules are present for a particular sample volume (i.e. the product of the cross-sectional area of the channel and the illuminated channel length). Sample volumes that are too small may not contain any target molecules and, thus, will be useless for detection purposes. Common human physiological fluids analysis, require analyte concentrations between 10^8 and 10^{15} molecules per nanoliter (Manz et al. 1990). Consequently, a nanoliter sample volume is realistic only if the detection system can measure the detection limit of 10^8 molecules. Whether actual shrinking to the nanoliter range is feasible also depends on practical aspects, such as the channel height (or optical pathlength) required to achieve measurable absorption (which depends on the analytes absorption coefficient), as well the minimum channel width and the photodetector area, which are both set by lithographic constraints. Typical minimum dimensions for the optical pathlength is 500 μm. The reduction of this optical pathlength makes solutions with a relatively low absorption coefficients difficult to detect, even with a high-accurate detection system. For the four biochemical parameters reported as the target application of the microlaboratory, the total protein in urine is the one that has the lowest target molecules (on average 3 μmol/L) and also the lowest concentration (less than 15 mg/dl). For a 3 mg/dl concentration of total protein in urine there are 3.61×10^8 molecules per nanoliter (enough for reproducible measurements of analyte concentrations (Strasinger and Di Lorenzo 2001)).

Another essential requirement for any practical fully integrated microlaboratory in a single-chip for spectropho-

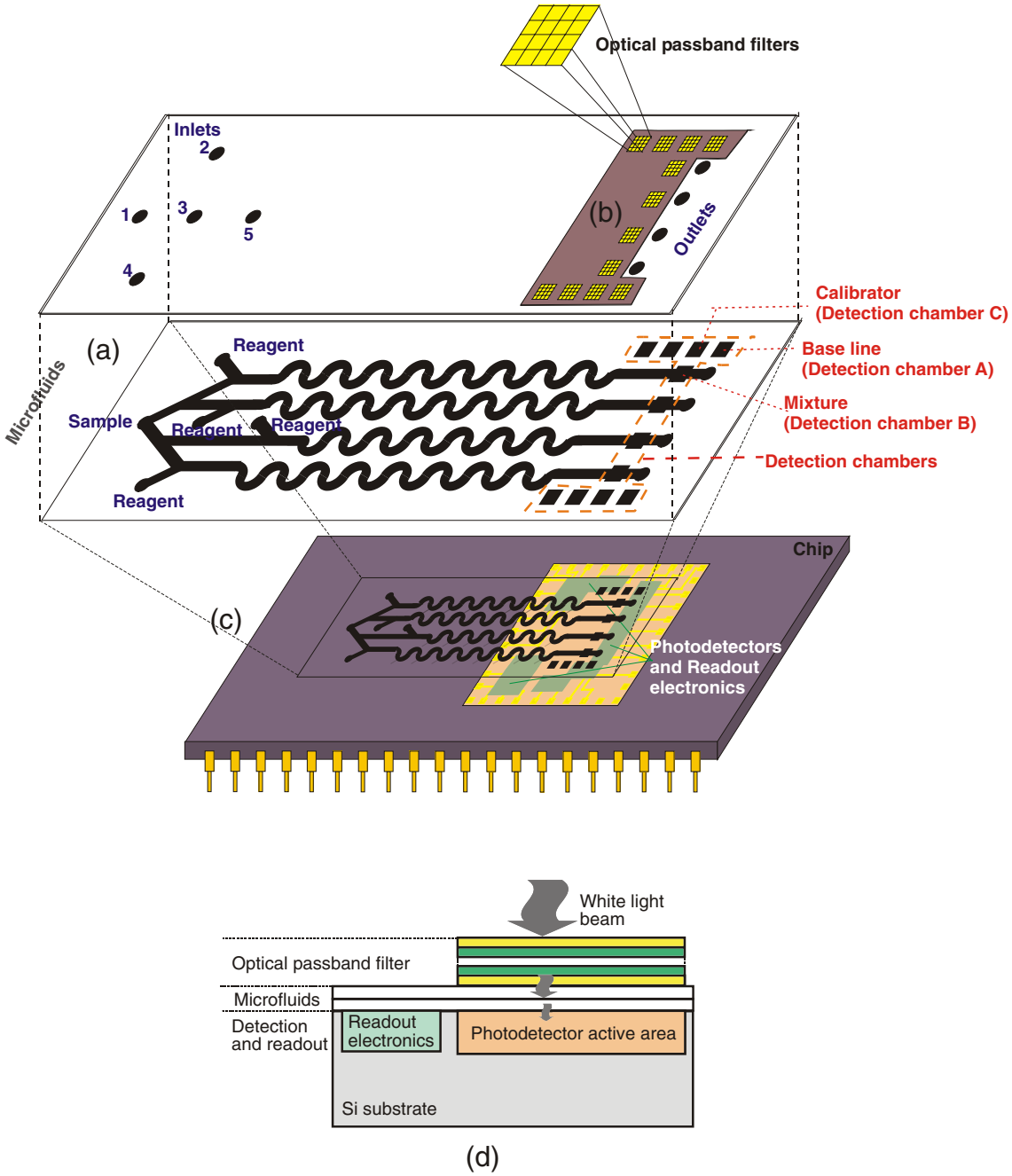

Fig. 1 Microlaboratory structure: (**a**) microfluidic system, the black squares are the detection chambers; (**b**) optical filtering system; (**c**) detection and readout system; (**d**) cross-section of the structure for a single optical-channel

tometric analysis based on colorimetric detection is the ability to mix two or more fluids thoroughly and efficiently (i.e., in a reasonable amount of time). The microscale conditions have distinctive properties due to its small dimensions and typically low volume flow rate. First, liquid flow is laminar, not turbulent (it is generally operated at Reynolds number of less than 1). Second, shrinking the dimensions of a microlaboratory means that both molecular and thermal diffusion times are significantly reduced (Simpson et al. 1998). As a result, diffusion can be used for mixing fluids, e. g., to bring reactants together across small channels, as, for example, in the molecular binding assays presented as the target application of this micro-laboratory. Diffusion time varies with the square of the distance (a practical example is 17 min for diffusion over 1 mm, while only 100 ms is required for a 10 μm distance).

Finally, the small volume of a microsystem does facilitate rapid dilution or injection of reactants into the system (Huang et al. 2006). The fluidic properties of the reagents and the samples used in the application discussed

in this report are highly suitable for mixing in a reasonable amount of time by diffusion only (i.e. have a high-diffusion coefficient) (Sigma-Aldrich 2008).

The microfluidic system was designed for allowing the simultaneous measurements of four biochemical parameters in urine samples. The microfluidic system is composed of two wafers (Fig. 1(a)). The first has the holes for injecting and draining of the liquids (inlets and outlets) and the second includes the microchannels (for fluids transportation and mixing) and the detection chambers. The structure is similar to a serpentine, which provides complete and homogeneous mixing of the reagent with the sample driven by diffusion in a fast way. There is just one inlet for the sample (inlet 1 in Fig. 1(a)) and four inlets for the specific reagents (inlets 2, 3, 4 and 5 in Fig. 1(a)) of the biochemical parameters analysis. Each main channel is 400 μm wide and 500 μm deep. The longest main channel is 16 mm long. All detection chambers are 600 μm×600 μm wide and 500 μm deep (this depth is crucial for the optical absorption measurements), which gives a liquid volume of 0.18 μl. This quantity allows having more than 10^8 target molecules of each of the four biochemical parameters reported in this paper (for the 3 mg/dl of total protein concentration, there are yet 649.8×10^8 molecules).

The analysis of each biochemical parameter comprises the simultaneous measurement of the absorbance of three fluids that are present in three detection chambers (for example, detection chambers **A**, **B** and **C** in Fig. 1(a), which are for measuring chloride). One contains the reagent, detection chamber **A**, and it is needed to obtain the baseline reference and to calibrate the light source fluctuations. The other, detection chamber **B**, allows the mixed solution analysis; it contains the reagent plus the analyte samples. The third, detection chamber **C**, is needed to calibrate the analyte concentration (with a well-known concentration standard).

The use of those three absorbance measuring channels eliminates the need for initial system calibration. Such a feature is usually referred to as 'self calibration', which highly adds to the system performance. This feature allows the use of a non-calibrated external polychromatic light source, such as a regular lamp connected to the power supply. Any fluctuation in intensity or spectral distribution of the light source is compensated by the simultaneous detection of the optical signals from: the fluid samples, the baseline reference and the calibrating channel.

The microchannels are fabricated using a layer of photoresist SU-8 deposited on a glass substrate. The SU-8 photoresist is an epoxy-based material that offers good properties, such as high-mechanical strength, good adhesion on many different substrate materials and biocompatibility (Ribeiro et al. 2005). This epoxy-based material provides low sidewall roughness and deep rectangular vertical profile of the microchannels, which is suitable for optical absorption measurements (the Lambert-Beer's law is directly applicable). Moreover, it also provides a low-cost fabrication process, UV lithography semiconductor compatibility and does not require expensive masks (Gelorme et al. 1989). The negative mask to be used for patterning the microchannels structure over the glass substrate is fabricated from a regular transparent foil (like the one used in printed circuit boards). Therefore, the microfluidic system can be a disposable system, which avoids the cost associated with cleaning of the microchannels after use. The SU-8 photoresist chosen is the SU-8-2150, which has a high viscosity and it is the most appropriate for the required channels depth. Figure 2 shows the fabricated microfluidic system.

2.2 Optical filtering system

Several analytes present in human physiological fluids are measured using spectrophotometry by optical absorption (mainly the ones for routine tests analyses). Seventeen-ketosteroids, chloride, uric acid, cholesterol, glucose, magnesium, creatinine, urea, hemoglobin, β-glucuronidase, bilirubin, leucine aminopeptidase, calcium, oxalate, total protein and albumin are 16 examples of those analytes. The absorption maximums of the colored mixtures produced by those analytes range from 480 nm to 600 nm and narrowly spaced at 8 nm on the average (Sigma-Aldrich 2008). Since the regular white light source features all the visible components, the optical passband filters on top of the microlaboratory must be highly selective. Their selectivity, defined as the FWHM (Full-Width-Half-Maximum), should be less than 6 nm to avoid misidentification in an analysis by possible overlapping absorbance spectra of mixtures. It should be noted that this problem is aggravated by the fact that the bandwidth of the absorption spectrum of each analyte extends over about 100 nm. In addition, the optical filters should be designed to yield a narrow passband around the value of the wavelength for which the colored mixture being analyzed has its absorption maximum. Instead of fabrication of four optical filters, a 4×4 optical passband filter array has been developed (see Fig. 1(b)). The

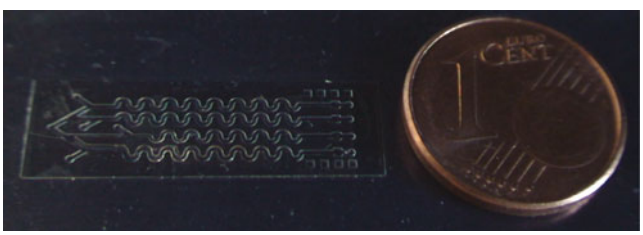

Fig. 2 A photograph of the SU-8-based microchannels structure

advantage of this approach is that it allows the concentration measurement of the 16 analytes listed above with the same opto-electronic detection device (optical filter plus detection and readout systems), while channel shifting can be used in case of a systematic error in the thickness of filter layers.

A Fabry-Perot structure (Macleod 2001) with dielectric mirrors was used. In a Fabry-Perot optical filter the thickness of the resonance cavity determines the tuned wavelength. That structure is the most suitable for providing high-transmittance and low FWHM, once dielectric mirrors, when properly designed and fabricated, offers high-performance characteristics (high reflectivity with low absorption losses) (Macleod 2001). When using dielectric layers, at least two different materials with high and low refractive index and alternately deposited are needed to obtain narrow passband optical filters. TiO_2 and SiO_2 (with refractive index of about 3.0 and 1.5 in the visible spectrum, respectively) were chosen for the fabrication of the dielectric mirrors. SiO_2 has been selected because the wavelength dependence of its refractive index for the spectral band between 480 nm and 700 nm is almost constant (1.465 to 1.457, respectively). TiO_2 has been selected due to fabrication constraints (the deposition process is well-characterized). Table 1 shows the thickness of the dielectric layers that better features the optical characteristics, the feasibility and the fabrication process for the array of the 16 highly selective Fabry-Perot optical filters. This optimized result was obtained with the help of thin-film optics software (TFCalc3.5). The 16 spectral bands are achieved using 11 dielectric layers and changing only the SiO_2 cavity (layer 6), between 140 nm and 230 nm, with 6 nm steps. The simulated optical filters have a FWHM less than 5 nm and a transmittance higher than 85% (more details about the simulations can be found in (Minas et al. 2006)). The complete structure of the 16 optical filters is arranged in a 4×4 array for matching the detection chambers layout. The dielectric films are deposited by Ion Beam Deposition that features high-quality thin-film fabrication (Macleod 2001). The layout design for the deposition process of the several resonance cavities is crucial to minimize the masks number and the deposition steps. The optical filter array fabrication steps and its performance are described in detail in (Minas et al. 2006). In summary: fabrication requires only four masks, used with different deposition times and 15 deposition steps.

2.3 Detection and readout system

The photodetectors convert the light intensity that is transmitted through the colored mixture into a photocurrent. These are arranged in an array (with the same layout as the optical filters) and this array is placed underneath the detection chambers and vertically aligned with the optical filters array (Fig. 1(c) and (d)). The device will be more cost-efficient if the photodetectors could be integrated with its readout electronics in a single-chip and fabricated in a standard CMOS process (without extra masks or steps). Figure 3(a) shows the three vertical junction photodiodes that are possible in a standard *n-well* CMOS process. At a particular wavelength, their quantum efficiency varies according to their junction depth. The quantum efficiency of a photodiode reveals how well it collects the incident light (Sze 1981). In the visible spectrum, the blue light (of about 450 nm) is more efficiently collected by a shallower junction (*p+/n-well* and *n+/p-epilayer*) and the red light (of about 650 nm) by a deeper junction (*n-well/p-epilayer*), see Fig. 3(b). Therefore, it was chosen the *n+/p-epilayer*

Table 1 Layer materials, thicknesses and tuned wavelength of the 16 highly selective Fabry-Perot optical filters array

Filter number		1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16
Tuned wavelength (nm)		**480**	**488**	**495**	**503**	**512**	**520**	**528**	**536**	**544**	**552**	**560**	**568**	**575**	**583**	**591**	**600**
Layer	Material	Layers thickness (nm)															
1	TiO_2	45															
2	SiO_2	95															
3	TiO_2	45															
4	SiO_2	95															
5	TiO_2	45															
6	SiO_2	140	146	152	158	164	170	176	182	188	194	200	206	212	218	224	230
7	TiO_2	45															
8	SiO_2	95															
9	TiO_2	45															
10	SiO_2	95															
11	TiO_2	45															

Mirror

Cavity

Mirror

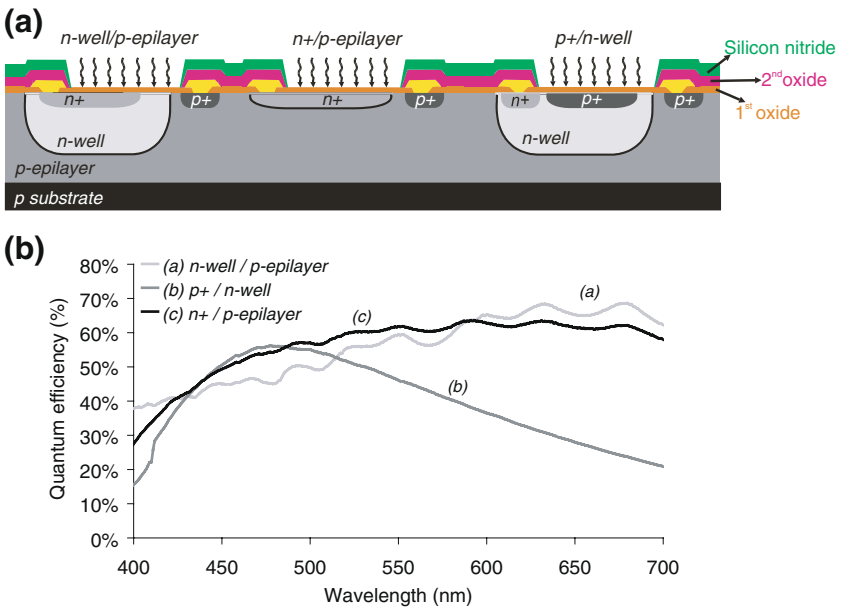

Fig. 3 (**a**) The three vertical junction photodiodes structures available in a standard *n-well* CMOS process: *n-well/p-epilayer*, *p+/n-well* and *n+/p-epilayer*. (**b**) Measured quantum efficiency of these three photodiodes structures

junction, because it provides the best possible quantum efficiency in the desired spectral range (wavelengths between 480 nm and 600 nm, Fig. 3(b)), due to the different doping concentration between the *n* and the *p* side, which extends the *p* side depletion area more deeply.

Due to technological constraints imposed by the CMOS process, the multi-layer coating is restricted to combinations of the stack of the dielectric layers used and selectively removed by mask design on top of the photodiode pn-junction. In the standard *n-well* CMOS process used, there are three dielectric layers available above the pn-junction. They act as a thin-film stack and influence the optical transmittance for each wavelength independently (Minas et al. 2005). Therefore, the second oxide layer and the silicon nitride layer were removed (without affecting the CMOS process) to eliminate the wavelength dependence of the transmission through these layers. The first oxide layer should remain to protect the photodiode active area. Indeed, removal of the first oxide layer would introduce random surface roughness (see Fig. 4), which would adversely affect the optical performance.

An additional photodiode for measuring the photodiode dark current is introduced in the circuit. The dark current is the current that flows in a photodiode when there is no optical radiation incident on it. It is usually measured and subtracted from the flux. Since the dark current is temperature dependent, one measurement at the beginning of the experiment is usually not sufficient. Thus, in the reported circuit, a dark current compensation channel is implemented. This photodiode is completely covered with metal.

The circuit used for readout of the photodiode is an implementation of a Sigma-Delta ($\Sigma\Delta$) converter (Fig. 5). The output format is a bit stream signal, which makes it suitable for use in small data-acquisition and control systems. This signal is proportional to the photodiode current and hence proportional to the intensity of the light transmitted through the fluid. Each photodetector array has its own readout circuit. This scheme of having parallel measurement of the optical absorption of the three fluid detection chambers avoids the errors that can be introduced due to the light source fluctuations when it is measured one optical channel at a time. Further computer processing perform additional calculations of these three output signals and, based on Lambert-Beer's law, a concentration value of the analyte is achieved. Additional information about the CMOS $\Sigma\Delta$ converter can be found in (Fernandes et al. 2008).

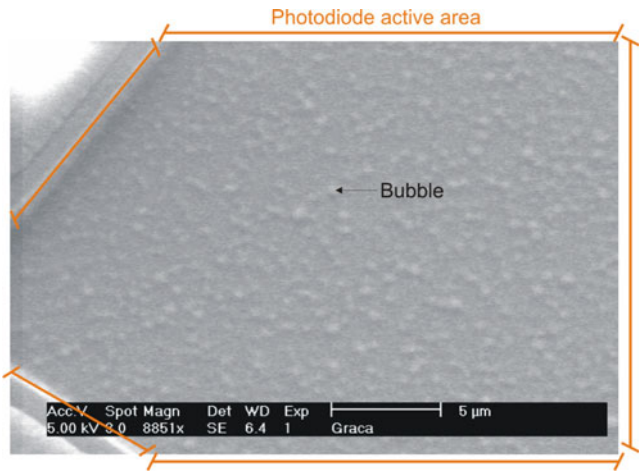

Fig. 4 A SEM photograph of the CMOS photodiode active area without the first oxide. The surface roughness is shown, which results in scattering and affects the responsivity of the photodiode

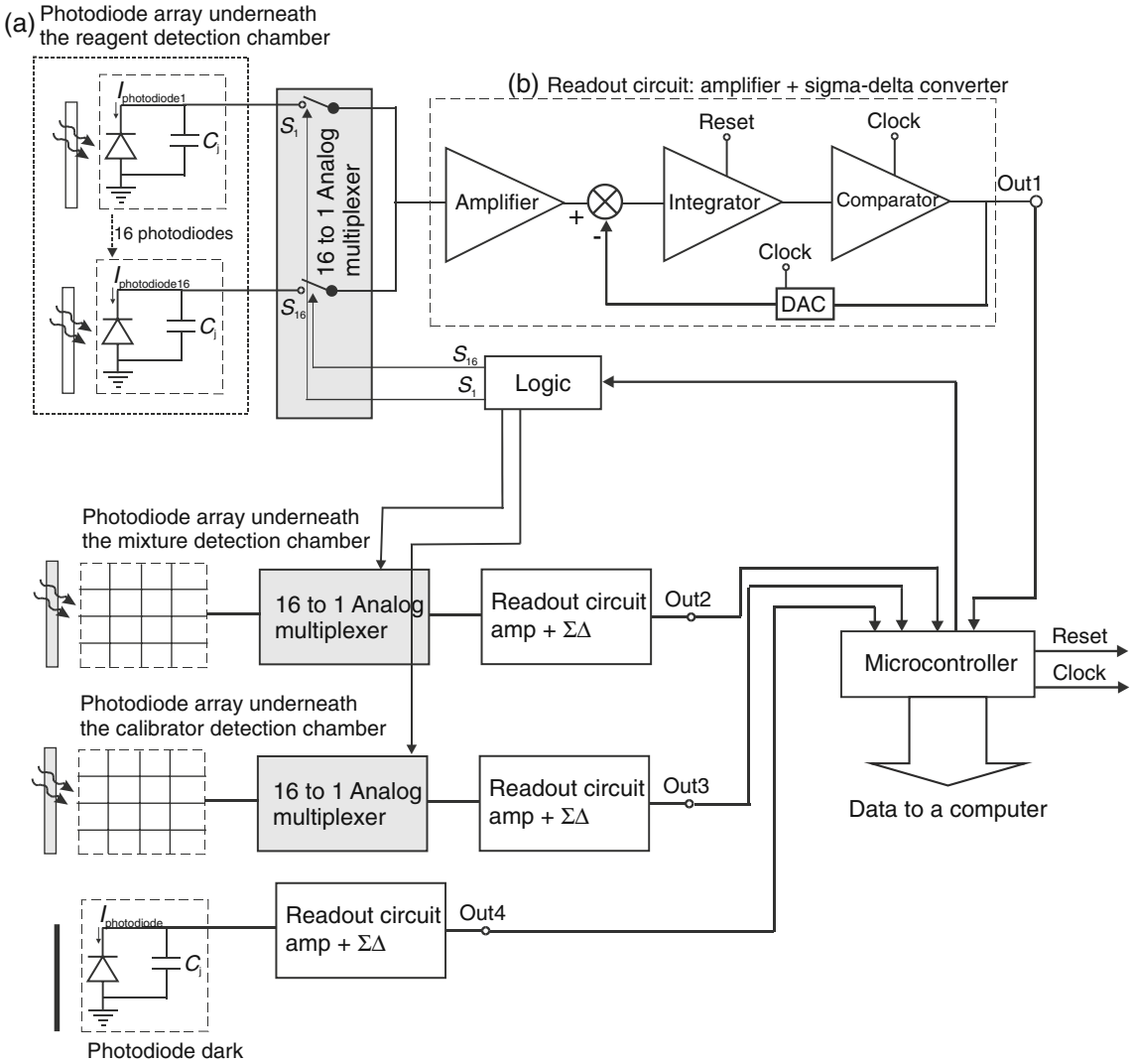

Fig. 5 Block diagram of the readout circuit of the photodiode array with a detail of (**a**) the connection of one photodiodes array and its equivalent circuit and (**b**) one Sigma-Delta converter

The complete microlaboratory is presented in Fig. 6. The CMOS chip has been fabricated through a 0.7 μm, double-metal, *n-well* CMOS process. The area of each photodetector is 130 μm×130 μm and each ΣΔ converter is 118 μm×48 μm. The microfluidic die is glued above the CMOS chip. The optical filters array and the CMOS chip are the same for several analyses of the programmed 16 biochemical parameters.

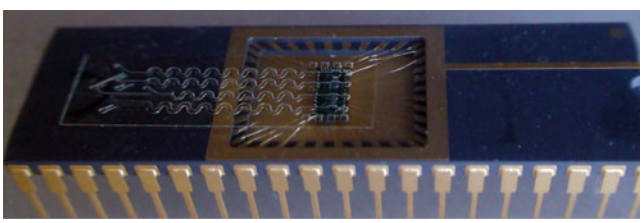

Fig. 6 Photograph of the microlaboratory

3 Experimental

The electrical characterization and the performance of the on-chip detection and readout circuits were presented in (Fernandes et al. 2008).

Prior to actual measuring, the compliance with the prescribed procedures that are described in the reagents protocol (Sigma-Aldrich 2008) need to be verified. Use with volumes down to 3 ml of sample volume was confirmed acceptable for proper operation of the microlaboratory. The experiments involve the quantitative measurement of four analytes in urine: chloride, creatinine, total protein and uric acid. Total protein has the lowest value for the concentration of analytes per nanoliter in urine, whereas uric acid has the highest value (Strasinger and Di Lorenzo 2001). Combining the measurement results on these two analytes indicates

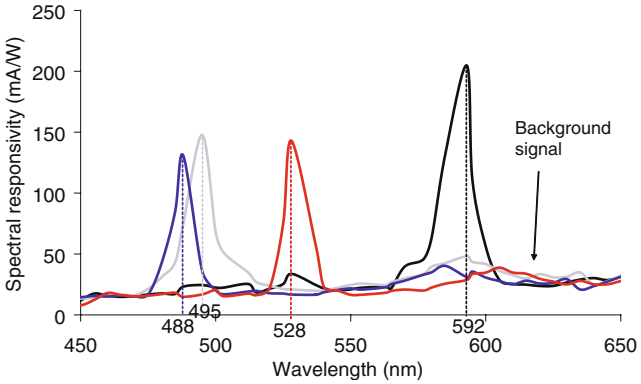

Fig. 7 Measured spectral responsivity of four optical filters using an empty detection chamber (optical effect of photodiode included)

Table 2 Microlaboratory results obtained for randomly selected urine samples (between run and within run $n=5$)

	Chloride (mmol/L)	Creatinine (mg/ml)	Total protein (mg/dl)	Uric acid (mg/dl)
Adult 1	90	1	0	25
Adult 2	20	2	5	45
Adult 3	50	1.5	10	20
Adult 4	40	1.6	4	40

whether the dynamic range of operation of the system is sufficient. The sensitivity, the linearity and the reproducibility coefficients variation should be the same as the ones obtained on well-known standards with the commercial equipments available at clinical laboratories. Moreover, the spectral measurements of Fig. 7 confirm that the performance of the optical filters on top of the microlaboratory is sufficient to allow the use of a polychromatic light source for illumination. Therefore, the microlaboratory can be illuminated using a 200 W halogen lamp. The background signal as well as some small variations in the tuned wavelength (deviation of 0.9 nm from the designed due to imperfections of the incident light wave and roughness of the surface of the Fabry-Perot optical filters) do not significantly affect the measurements. These small errors are made negligible by compensation using simultaneous light intensity measurement in the three detection chambers.

The complete and homogeneous mixing of a urine sample with the specific reagent for the quantitative measurement of the analyte concentration was performed in about 20 s. The reagents used are from Sigma-Aldrich (Sigma-Aldrich 2008). The urine samples were references each with a well-defined concentration of several analytes

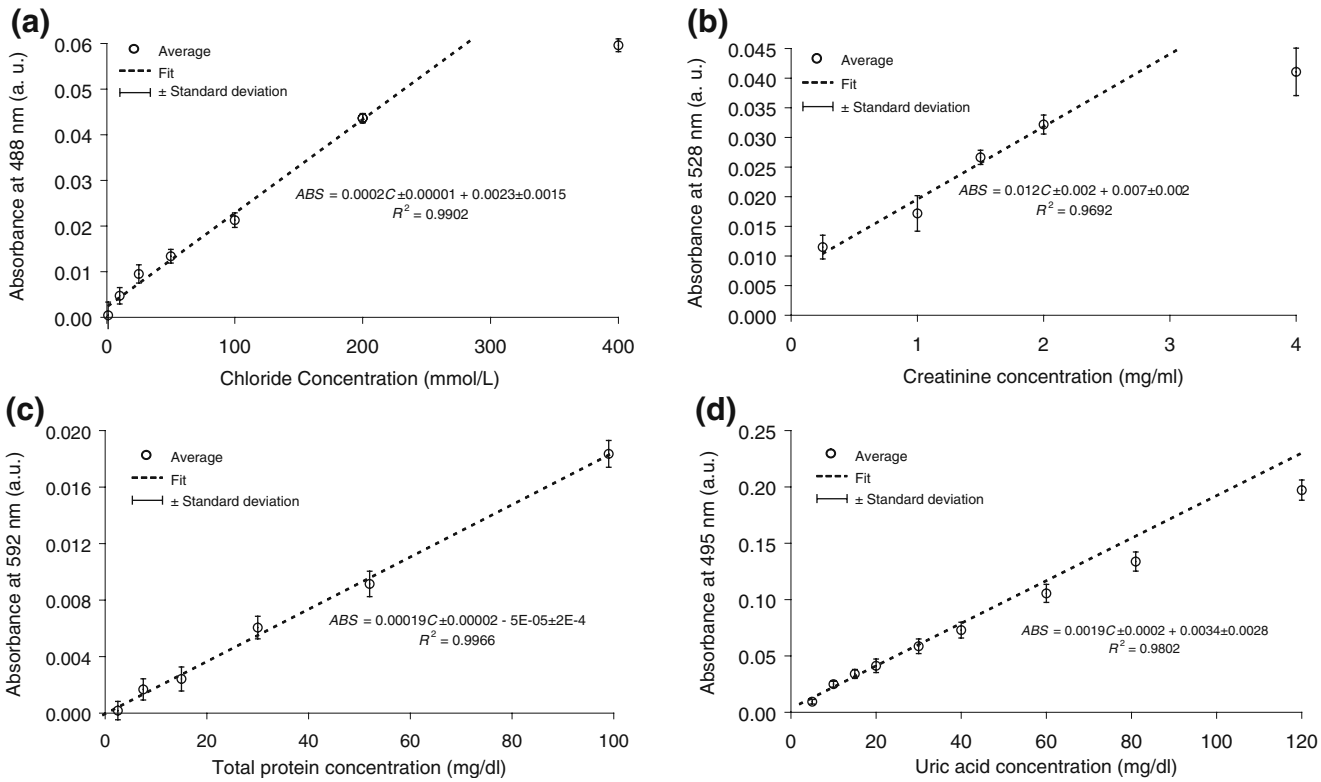

Fig. 8 Absorbance as a function of the biochemical parameters concentration using only a white light source as the microlaboratory illumination, a 200 W halogen lamp, (average±standard deviation, samples number=10)

in a range around the level that can be typically found in human urine (usually called "normal and abnormal controls"). Measurements were performed over a range that includes normal and typical abnormal analytes concentration values in a human being. Figure 8 shows the calibration curve of the four biochemical parameters measured on-chip. The data was obtained using the digitalized measured values of the transmitted light measured in the photodiodes.

The following conclusions can be drawn from the results obtained: (1) a linear response is demonstrated over a concentration range of 1–200 mmol/L for chloride, 0.25–2 mg/ml for creatinine, 0–100 mg/dl for total protein and 0–30 mg/dl for uric acid (with correlation coefficients above 0.9692). These linear concentration ranges are in agreement with the ones reported in the reagents protocol (Sigma-Aldrich 2008). For a concentration exceeding the upper range of the range with a linear response the sample should be diluted and re-assayed multiplying the result by the dilute factor. This procedure is described in the reagent protocol.;[1] (2) the intensity of the color produced by the mixture is directly proportional to the analyte concentration, which confirms that the measurement principle satisfies the Lambert-Beer's law; (3) the analyte absorption spectra show a maximum peak at a specific wavelength; (4) the reproducibility mean coefficient variation of ten replicate analyses for each concentration are compatible with the prescribed procedures. Except for creatinine, whose value is 20%, the reproducibility mean coefficient variation for the three biochemical parameters is less than 10%; (5) the microlaboratory achieved sensitivity corresponds to a relative resolution of 3.3% for those four analytes, which is enough for human being urine values and for evaluating the state of health.

Therefore, the results confirm the direct proportionality between intensity of the color produced by the mixture and the analyte concentration. Moreover, the results are in agreement with measurements performed with traditional macroscopic instruments on the same well-known standards and using state-of-the-art laboratory equipment. Therefore, the performance in terms of precision, reliability and sensitivity of the analysis performed is comparable to that in state-of-the-art systems used in clinical analysis laboratories. It should be noted that the main objective of this work is to demonstrate that the same performance as in state-of-the-art macroscopic equipment available at clinical laboratories can be achieved in a microsystem. A higher sensitivity is not required by the application.

[1] The reagent protocol reports that the measurements should be done in the linear range and when outside dilutions should be done (Sigma-Aldrich 2008).

An increased sensitivity could be obtained when using different optical techniques, such as fluorescence, refractometry or surface plasmon resonance. However, such an improvement in sensitivity would be at the expense of complexity of the optical microsystem.

After providing the proof-of-concept with results in Fig. 8, some initial tests on actual urine samples were performed and the results are presented in Table 2.

4 Conclusions

The microlaboratory offers unique opportunities for spectrophotometric analysis in clinical diagnostics. The general advantages of microlaboratories include portability and small sample volume. Specific advantages of the optical technique presented are the use of white light illumination made possible by the integration of highly selective optical filters, based on Fabry-Perot thin-films optical resonators. The possibility to use a simple halogen lamp is perhaps the most significant benefit. Usually, these systems require a special light source, which limits portability and adds to costs. The highly selective optical filtering system and the self-calibration capability implemented in this microsystem enables a biochemical analysis at any location with instantaneous results, without the use of complex and expensive analysis systems, which makes it portable, broadly applicable and a powerful tool in hospitals, operating rooms, physician's offices, clinical laboratories as well as in patient homes.

The application of the described microlaboratory was the concentration measurement of biochemical parameters in human's physiological fluids. However, besides the healthcare sector, this device can also be used for other sectors, where spectrophotometric analyses are required: textile in dyers; environmental, monitoring the air and the water quality for potential toxins and pesticides; screening foods for promptly identifying drugs abuse and forensic investigation.

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