Laser Machined Plastic Laminates: Towards Portable Diagnostic Devices for Use in Low Resource Environments


Abstract: Despite significant progress in development of bioanalytical devices cost, complexity, access to reagents and lack of infrastructure have prevented use of these technologies in resource-limited regions. To provide a sustainable tool in the global effort to combat infectious diseases the diagnostic device must be low cost, simple to operate and read, robust, and have sensitivity and specificity comparable to laboratory analysis. In this mini-review we describe recent work using laser machined plastic laminates to produce diagnostic devices that are capable of a wide variety of bioanalytical measurements and show great promise towards future use in low-resource environments.

Keywords: Low-resource environment bioanalytics · laser ablated plastic laminates · electrode arrays · DNA detection · orthogonal biosensor

1 Introduction

Despite substantial progress made in recent decades in advancing public health, significant challenges remain in extending these benefits to the developing world [1]. More than half of the deaths in undeveloped countries are attributed to infectious diseases, whereas deaths due to infectious diseases in high-income countries are less than 5% [2,3]. The first step in treatment and prevention of disease is accurate and timely diagnosis. In resource-limited environments on-site diagnostics can lead to significantly improved treatment and prevention of disease [4]. Additionally, improved diagnosis of infectious disease in low-resource settings would prove valuable in preventing and combating global pandemics.

Sophisticated molecular biotechnologies are commonly used for diagnosis in developed regions; however, numerous limitations (cost, lack of skilled laboratory staff, inability to transport samples long distances, limited access to reagents and materials, and lack of infrastructure) prevent use of these technologies in resource-limited regions [5–8]. Therefore, efforts must be made to develop diagnostic platforms that can be used in low-resource settings. Specifically, devices must be designed to ensure sustainability through low cost (less than $10 per use), functionality in harsh environments, little or no reliance on power or refrigeration, and require little or no expertise to obtain accurate results. Such devices should also have a hand-held form factor and require no bulky laboratory equipment or similar critical infrastructure for operation, facilitating portability and field use. All these requirements must be accomplished while maintaining laboratory quality, specificity, and sensitivity [9–11].

Herein we review our recent work toward the development of portable diagnostic biodetection devices with the goal of future use in low-resource environments. Engineering advancements reviewed address a number of critical criteria for field-deployable diagnostics that may enable use in undeveloped settings. Devices are prepared from plastic laminates machined via laser ablation which allows for relatively low cost and rapid prototype development. Integration of conducting materials (Au, Pt, ITO) within the plastic laminates enables a wide variety of electrochemical measurements. Added functionality can be incorporated into the platform permitting power-free valve actuation, on-device storage of reagent or sample solution, and control over temperature for on-
device denaturation of double stranded DNA (dsDNA) and renewing of the DNA probe surface for subsequent reuse. Multianalyte detection from dsDNA and electrochemiluminescent measurements is also shown.

We further demonstrate use of living cells within a plastic laminate device as the biosensing element for small molecule detection. Cells are stabilized via silica gel encapsulation, permitting storage under ambient conditions for months with high retention in cellular viability and sensing activity. Finally, we demonstrate orthogonal mode biodetection via electrochemical, fluorescent and colorimetric signals. The colorimetric approach requires no external equipment or expertise to obtain an accurate result, and multiple detection modes can reduce false positive and negative readings, increasing confidence in the sensor output. These accomplishments were achieved using low cost and robust plastic laminate technology. We conclude with prospects of this technology, as well as key requirements that remain in order to advance these capabilities for biodetection in low-resource environments.

2 Fabrication Technique: Laser Ablation of Plastic Laminates

Sustainable bioanalytical platforms for use in low-resource environments must be low-cost (less than $10 per use) and robust. Such platforms can be rapidly fabricated using laser ablation of plastic laminates [3]. Primary materials used in device construction are poly-(methyl methacrylate) (PMMA), and polyethylene terephthalate (PET, e.g. Mylar™) with thicknesses typically ranging from 0.06 mm–2.0 mm, although thinner and thicker laminates may be used. To form a three dimensional microfluidic platform laminate layers are joined together using adhesive on one or both sides. The adhesive is typically pressure-sensitive acrylic-based (0.013 mm thick, double sided) and can be cut directly, or bonded to the plastic laminate prior to cutting. Other materials used for construction are Kapton, glass, acrylic, PDMS, printed and coated paper, paraffin, high-strength neodymium magnets, carbon, pyrolyzed photoresist, printed circuit boards, and thermally evaporated conductors including Ti, Au, Pt, and ITO [12–14].

The basic microfluidic channel is fabricated by cutting the channel pattern into a plastic sheet coated with acrylic adhesive on both sides using a CO₂ laser ablation system, which forms the side walls, defining the channel’s three dimensions. Top and bottom walls are formed from plain plastic sheets. Into one or both of these plain sheets fluidic vias can be cut to allow for access to external tubing or additional channels in the device. Alignment of the plastic layers is accomplished with registration holes and pins. Layers are bonded together by pressing between two flat steel plates in a hydraulic press.

This fabrication method is useful for rapid prototyping as changes can be quickly and inexpensively made. Multilayer sensor packages can be assembled in 30–60 min from pre-cut and pre-processed layers. Fabrication including all processing steps requires 2–8 hours. Design changes can be made and implemented in minutes to hours. This rapid device prototyping and manufacturing capability is not available using standard Si/glass processing, plastic casting, injection molding or embossing techniques. Further, use of the PMMA and PET laminates for device construction affords a low profile and robust package. Material cost for the plastic laminate device is similar to that reported for other plastic microfluidic devices, typically in the single US dollar range for simple devices, to approximately 10 US dollars for high complexity platforms.

2.1 Integrated Components and Added Functionality

To enable selective and sensitive biodetection within plastic laminate based 3D microfluidic platforms functional material components can be incorporated on given plastic layers within the device. As shown in Figure 1, individually addressable electrode arrays, power-free valves, removable storage chambers, and resistive heating elements, among others, can be integrated within the plastic laminate device.

2.1.1 Individually Addressable Electrode Arrays of Au and ITO

Individually addressable electrodes on PET layers were prepared by patterning of conducting materials using electron beam evaporation through a stencil mask [15]. The stencil was made from a 0.5 mm thick quartz disk into which the pattern was cut using the same CO₂ laser ablation system used to ablate the plastic laminate. Alignment holes and pins ensured adequate registration between stencil and substrate during the deposition. This method of electrode patterning permits simple modification to the electrode configuration without the cost and time associated with photolithography mask design and fabrication.

Shown in Figure 2A is an array of nine individually addressable gold working electrodes and a common pseudo-
reference gold electrode on a 0.08 mm thick PET layer. As this method did not yield precisely defined features it was necessary to add a PET mask over the electrodes, defining the area of the working and reference electrodes exposed to the channel (also shown in Figure 2A). This layer also served to insulate working electrode traces that crossed the path of other channels in the device.

The pseudo-reference electrode was designed in a four leg comb layout to ensure symmetric potential drop across all nine working electrodes, which permitted consistent and quantitative electrochemical measurements between all working electrodes. The Pt counter electrode in this 3-electrode electrochemical cell resides on the top of the channel, above the working and reference electrodes as shown in a later figure (Figure 6, cross-section).

An indium tin oxide (ITO) working electrode array on PET laminate is shown in Figure 2B. ITO is desirable as an electrode material as it allows for a wider positive potential window in aqueous electrolyte solution over metallic electrodes and is transparent, allowing for optical detection assays (e.g. electrochemiluminescence). In order to minimize resistive losses from ITO traces and avoid cracking of the oxide deposited onto the flexible plastic laminate, the array was prepared by depositing small area ITO working electrode pads in contact with Au electrode leads.

Electrochemical responses of the Au and ITO working electrode arrays were investigated using the ferricyanide (Fe(CN)₆³⁻⁻⁻⁻) redox probe. Cyclic voltammograms at each of the working electrode materials are shown in Figure 2 (right panel). Similar peak potentials and peak currents were obtained across all electrodes with the average formal potential \( E^0 \) for \( \text{Fe(CN)}₆³⁻⁻⁻⁻ \) measured at \(-138 \pm 2 \text{ mV} \) vs. the common pseudo-reference [15]. The average variation in response between electrodes within a channel was less than 4%; variation between all electrodes in the array was less than 7%. This ability to integrate individually addressable electrode arrays of Au, Pt, and ITO enables a wide variety of electrochemical measurements and assays within the plastic laminate platform.

### 2.1.2 Simple and Power-Free Valving

Many biological assays rely on initial separation of materials/reagents from the introduced analyte solution, and time dependent transport of assay constituents through the microfluidic device. Simple to operate magnetic-adhesive based valves were developed to control fluid flow through the microfluidic platform and allow for on-demand access/release of stored reagent while maintaining the overall device seal. As shown schematically in Figure 3, the initially closed valve consists of a via connecting two chambers on different planes in the plastic laminate device. A thin ring of adhesive serves as a seat for a neodymium disk magnet that seals the via, preventing transport between the two chambers. The valve is actuated by bringing an external magnet in contact with the outer surface of the device (Figure 3, opened). This applied magnetic force unseats the internal valve magnet from the adhesive ring, exposing the underlying connection port between the two chambers, which allows transport of the contents between chambers in different planes without power or pumps. This valve design is low cost ($0.32 materials cost per valve) and requires no technical training to operate.
Photographs of a proof-of-concept device with stored reagent before and after valve actuation are presented in Figure 3 (bottom panel). Initially, chambers containing dyed water, 5% bleach and air were all sealed by two magnetic-adhesive valves. Opening the valves allowed the fluids to transport between all three chambers, resulting in loss of the blue pigment upon reaction with bleach.

### 2.1.3 Storage Chambers with Pressure Actuated Check-Valves

While on-chip detection has distinct benefits in terms of small sample size, portability and short assay time, capture and storage of the analyzed sample for further laboratory or forensic analysis may be required. An on-chip sample storage scheme is presented in Figure 4 [16]. The storage system was fabricated from three layers of a 0.127 mm thick polyetherurethane (PEU) formulation that was elastic, flexible and slightly tacky to itself. The self-tackiness facilitated sealing of the check valve in the final, filled state, but also caused difficulty in filling the device. To assist in chamber filling, a CO₂ laser under low power was used to cut grooves into the non-sealing surfaces to limit contact surface area. This was particularly necessary around the valve opening and the large surface of the storage area. Both high-power cutting and low-power laser welding were used to fabricate the valve and storage chambers (see Figure 4, schematic). The storage device was then integrated with microfluidic channels using pressure sensitive adhesive, sealing around the inlets.

The upper-left cross section schematic in Figure 4 shows three states of the storage system. The first is the unused state where the internal (P_i) and inlet/ambient (P_o) pressures are equal. When filling with sample solution, the inlet pressure is higher than the internal pressure. This state is typical during pressurized microchannel flow and can be concurrent with on-chip processing or analysis. Internal pressure continues to increase during filling as the elastic support expands. The integrated check valve is closed in the third state where the inlet pressure is less than the internal storage pressure. This setup allows for capture, storage and subsequent sample processing in parallel adjacent channels. Photos of an arrayed storage system are shown in the three states adjacent to the respective schematic cross-section. The bottom photo shows the arrayed storage system removed from the microfluidic platform. Two of the storage chambers leaked through the check valve upon removal due to high strain exerted on the flexible chambers and valves. Further design modifications are expected to optimize performance and provide leak free seals upon removal.

### 2.1.4 Temperature Control

An important requirement for many biological sensors is control of temperature within the device. For example, double stranded DNA (dsDNA) from real-world samples must be denatured to produce single stranded DNA (ssDNA) that can hybridize with surface immobilized ssDNA capture probes. Importantly, the process used to modulate temperature must be carefully controlled such that the solution does not vaporize and introduce bubbles that can disrupt fluidic flow and lead to non-uniform coverage of the probe detection surfaces, or damage the sample.

A Pt resistive heater was fabricated using the method described in section 2.1.1. This heater was integrated into a plastic laminate device above the top wall of the microfluidic channels (see Figure 6, layer 4 and cross-section view). This resistive heater was used to simultaneously control the temperature in all three microfluidic channels in the device using a resistance thermal detector (RTD) element, and a proportional/integral/derivative (PID) closed-loop algorithm [15]. Figure 5A shows a typical thermal profile and the corresponding heater output for on-chip denaturation of dsDNA (85°C, 5 min treatment). The temperature in the microfluidic channel reached 95% of the set point within 25 seconds and closely maintained the set temperature (standard deviation of ± 0.43°C). This temperature was sufficiently high to dena-
Review

ture the dsDNA without vaporizing the analyte solution or damaging DNA immobilized onto Au electrodes at the bottom of the channel.

Following dsDNA denaturation, power output to the resistive heater was set to zero and the channels returned to room temperature under passive cooling, reaching 30°C in 3.5 min. This slow cooling allowed the analyte ssDNA to hybridize with surface immobilized capture probe DNA, as shown in Figure 5B (red). The current response measured from ssDNA (Figure 5B, blue) was greater than that measured for the same concentration of dsDNA denatured on-device, which was likely due to competition for hybridization between surface immobilized probe DNA and the in-solution complimentary DNA stand that was just denatured. Without employing the on-device heater, current responses for the dsDNA target (Figure 5B, white) were similar to that for the negative ssDNA control (random sequence, Figure 5B, green). These data confirm that denaturation of dsDNA was required for detection and that the on-device heater was effective at denaturing dsDNA, while permitting subsequent hybridization with the probe surface.

The integrated heater can also be used to renew the capture DNA probe surface. This was done via a high temperature water flush (stripping) to remove captured target and detection probe DNA. As shown in Figure 5C, currents measured following stripping and surface regeneration were similar to those obtained from arrays without any prior exposure to the analyte sample and detection assay components, demonstrating nearly complete regeneration of the probe DNA electrode surface. Signals obtained from the dsDNA target (blue bars) decrease slightly with subsequent runs, falling by 3% from run 1 and 2, and another 12% from run 2 to 3. (Run 2 was performed 1 day after run 1. Run 3 was performed 4 days after run 1.) The signal from the random ssDNA sequence control (red bars) also rose slightly between runs 1 and 3. Although these data indicate the number of effective reuses is limited as the sensor degrades/ages, stripping via the integrated heater can allow for reuse of the plastic laminate device.

2.2 Bioanalytical Platforms

Integration of functional components described in section 2.1 permitted the development of devices capable of performing diverse bioassays. The fabrication and performance of a bioanalytical platform for multitarget DNA detection, and a platform capable of orthogonal detection mode cell-based biosensing, are described.

2.2.1 Parallel Microfluidic Channel Fixture for DNA Detection

A hand-held, low-cost, and robust multitarget electrochemical DNA sensor was developed that allowed for detection from dsDNA samples. The sensor was prepared by integrating a gold electrode array, microscale fluid delivery network, Pt resistive heater, and RTD into a plastic laminate device [15]. A photograph, schematic cross-section and exploded layer-by-layer representation of the integrated system are shown in Figure 6. The device was comprised of 9 layers (three PMMA, six PET). For simplicity, only layers 0–4 are shown in Figure 6. Layers 5–9 were structural layers that formed the fluidic interface with external tubing for loading analyte solution. Layer 0 was used for interfacing the plastic laminate device electrically with a 34 conductor card-edge connector linked to the hand-held heater and potentiostat circuitry.

The microfluidic platform contained three parallel channels, each with three individually addressable working electrodes and a shared pseudo-reference electrode. To allow for multitarget detection from dsDNA within the plastic laminate system, each of the three electrodes in a channel was selectively functionalized via bias-assisted grafting of aryl diazonium salts [17–19]. This approach resulted in individual electrodes within each channel modified with capture DNA probe against a sequence relevant for 1) breast cancer (BC), 2) colon cancer (CC), and 3) a non-specific binding control surface.

The average response taken from 3 independent arrays following on-device dsDNA denaturation and electro-
chemical detection assay is shown in Figure 7. When presented with the target BC and/or CC DNA sequence(s) significant chronoamperometric responses were obtained for each electrode that was functionalized with the respective complementary DNA probe. In contrast, on-chip negative control surfaces (Figure 7, orange bars) showed substantially lower currents that were similar from channel to channel across the array. Signals from non-complementary ssDNA probe modified electrodes (channel A, green bar; channel B, blue bar) were slightly higher than the response from the non-specific binding control, but were clearly distinguishable from the complementary probe surfaces and consistent with non-specific binding currents measured previously (see Figure 5B).

The three channel configuration could be connected in series, permitting on-chip replicates performed in each channel, or for detection of up to eight unique targets with a single negative control surface. Alternatively, the three channels could be used over different time periods allowing the device to be used three times before surface renewal or device replacement is required. Although simple to operate and read using a hand-held potentiostat and heater electronics, challenges remain for use in low resource environments. These include the requirement of reagent, washing and labelling steps, and potential sample preparation (e.g. DNA extraction) prior to introducing the sample to the array.

This plastic laminate platform was further used to initiate electrochemiluminescent (ECL) reactions. ECL occurs upon the electro generation of chemical species that undergo subsequent electron transfer reactions that form excited states which emit light [20]. Incorporation of ECL labels in biomolecule detection systems for highly sensitive immunoassay and DNA detection are widely reported, and impact areas as diverse as microbiology, immunology, virology, and drug development [21].

ECL reactions were initiated at, and imaged through, ITO working electrodes as shown in Figure 8. A photograph of the individually addressable ITO-Au electrode array (layer 1), with the electrode mask (layer 2) and microfluidic channels (layer 3), is presented in Figure 8A. To generate the ECL reaction, ruthenium bipyridine \((\text{Ru(bpy)}^2+)\) was used with tri-n-propylamine (TPA) serving as co-reactant. A cyclic voltammogram of the \(\text{Ru(bpy)}^2+\) and TPA system measured within the plastic laminate device (Figure 8B) shows a strong oxidation peak at 1.4 V vs. the on-device pseudo-reference electrode, corresponding to the dual oxidation of both \(\text{Ru(bpy)}^2+\) and TPA. At potentials of 1.4 V and higher a visible bright orange-red emission was observed through the ITO working electrodes, as shown in the ECL reaction photo and white-light photo overlay in Figure 8C. It was possible to initiate ECL reactions in an addressable manner with one reaction in channel A, one reaction in channel B, and 2 reactions in channel C, as an example (Figure 8C). The ECL reaction light intensity imaged through the underside of an ITO electrode vs. applied potential is also presented in Figure 8D. Initiating and imaging this ECL reaction was not possible on the Au working electrode array due to the limited potential window of Au, and the opaque working and counter electrodes. These data demonstrate the utility of the plastic laminate platform to utilize ECL, in addition to potentiometry, as a potential detection tool for bioanalysis.
2.2.2 Orthogonal Living Cell-Based Biodetector

An orthogonal detection mode bioanalytical platform was prepared by integrating nanomaterial-modified transparent electrodes and living cells within a plastic laminate cartridge [22]. This device was capable of generating fluorescent, electrochemical, and colorimetric signals from a single target analyte in a complex solution. Coupling these three signal transduction methods can substantially reduce the impact of potential interferants found in real-world samples including molecules that are redox active at the potential range used for electrochemical assay [23], or molecules that are autofluorescent/cause high solution turbidity that can confound optical measurements [24]. This orthogonal detection mode sensor can thus provide complementary information regarding the analyte sample properties, potentially reducing false negative/false positive readings and increase confidence in the sensor output. The orthogonal detection cartridge was comprised of nine plastic layers and an ITO coated microscope slide patterned to provide a three-electrode electrochemical cell at the bottom of each well with the device. Shown in Figure 9 is a photograph of the assembled cartridge and schematic electrode layout, cross-section, and layer-by-layer views.

Orthogonal detection mode sensing occurred via co-entrapping multiple cell lines engineered to provide distinct responses to the target analyte. *Saccharomyces cerevisiae* yeast cells that express and secret glucose oxidase (GOx), and yeast cells that express yellow fluorescent protein (YFP), were entrapped in an equal molar ratio between the ITO surface and well bottom (see Figure 9 cross-section). The entrapping matrix was a glycerol-silica thin film that permitted storage of the living cells within the device under dry, ambient conditions for months with sufficient retention in cell viability and activity for effective bioassay (40% viability after 60 days).

Upon addition of analyte solution to a given well, cells were simultaneously monitored via chronoamperometry (H$_2$O$_2$ produced from excreted GOx) and fluorescence measurements (YFP emission). As H$_2$O$_2$ reduction on ITO surfaces is unfavored, the ITO surface was functionalized with gold nanoparticles (AuNPs) and Prussian Blue (PB). As shown in Figure 10, functionalization occurred within the fully assembled plastic laminate device (prior to introduction of living cells) and was selective, modifying the working electrode (WE) only. This modification significantly enhanced H$_2$O$_2$ reduction currents while maintaining an optically transparent electrode surface required for fluorescence measurements.
Fig. 11. Orthogonal biosensing of a single model analyte via fluorescent, electrochemical, and colorimetric detection modes. a) Simultaneous electrochemical (blue) and fluorescence (red) measurements. Insets: Photographs of colorimetric assay results following overnight incubation. b–c) Fluorescence microscopy images of YFP expression following an overnight treatment with (b) YP + 2% gal or (c) YPD. (Scale bar = 30 μm). d) Chronoamperometric measurements of H$_2$O$_2$ produced by GOx turnover following treatment with YPD (red) or YP + 2% gal (blue). (Reprinted with permission from [22]. Copyright 2012, Wiley.)

Orthogonal assay response with time is shown in Figure 11a. The amperometric signal (blue trace) initially increased more rapidly than the fluorescence response (red trace). This was not unexpected as the amperometric signal is based on the catalytic production of H$_2$O$_2$ by GOx, while the fluorescent signal is proportional to the total concentration of YFP expressed by the cells. The amperometric signal reached a maximum, and then declined between 6–9 hours from introduction of the analyte sample, which was attributed to depletion of enzyme substrate and electrochemical consumption/breakdown of H$_2$O$_2$. The fluorescent signal, however, continued to increase, reaching a maximum between 12–18 hours from introduction of analyte sample. This signal was stable, remaining even after cell death. Typical fluorescent and amperometric responses to treatment with the target analyte are shown in Figure 11b and 11d. The high selectivity of the engineered cells for galactose over glucose (monosaccharide epimer, negative control) in complex media (YPD) is also shown (Figure 11c, 11d). These results highlight the selectivity possible from this cell-based sensing device within a potentially confounding matrix. Use of media to introduce the target, however, is not required; statistically identical responses to the target analyte in complex media, phosphate buffer, or minimal salt solutions were also measured.

Colorimetric detection was also demonstrated. ABTS (2,2'-azino-bis[3-ethyl-benzothiazoline-6-sulfonic acid) assay was performed after a 9 hour treatment of the encapsulated cells with the analyte solution. Assay solution and silica matrix remained clear when cells were treated with YPD (negative control), while the assay solution and silica matrix developed a strong, dark green-blue color that was easily discernible by eye (Figure 11A, insets). This provides an additional orthogonal detection mode in which the assay can be performed without instrumentation, power, or significant training, potentially simplifying detection in low-resource environments.

Finally, we demonstrated co-entrainment of disparate classes of living cells within an ITO-glass/plastic laminate cartridge permitting multianalyte biodetection. Eukaryotic S. cerevisiae cells engineered to express YFP in the presence of galactose, and gram-negative prokaryotic Escherichia coli ‘riboswitch’ cells engineered to express green fluorescent protein (GFP) in the presence of theophylline, were co-entrapped in a glycerol-silica matrix thin film. Multianalyte detection from the S. cerevisiae and E. coli cells co-entrapped in the cartridge is shown in Figure 12. Treatment of the cells with galactose and/or theophylline in complex medium (YP) resulted in strong...
Review

YFP and/or GFP fluorescence, clearly showing the cell-based sensor distinguished between the two targets. Further, these results showed for the first time that both eukaryotic and prokaryotic cells remain viable and active, even when immobilized together at intimate proximities within the glycerol-silica matrix.

3 Conclusions and Future Prospects

Laser machined plastic laminate technology is a low-cost and rapid means of developing robust and versatile bioanalytical devices. Various devices have been developed using this technology that independently show key engineering advances, which include: (i) incorporation of conducting materials (Au, Pt, ITO) within the plastic laminate, producing high quality individually addressable electrode arrays; (ii) a low cost and simple to operate valve that requires no power to actuate; (iii) plastic laminate storage chambers with pressure actuated check-valves for capture of analyte solution permitting subsequent forensic analysis; (iv) incorporation of a resistive heating element for dsDNA denaturation and (v) surface renewal allowing reuse of the DNA sensor; (vi) selective functionalization of electrode arrays with ssDNA probes for multianalyte electrochemical DNA detection; (vii) initiation and imaging of electrochemiluminescent reactions within the plastic laminate device; (viii) selective functionalization of electrodes within plastic laminates with AuNPs and Prussian Blue for enhanced catalytic electrochemical detection; (ix) incorporation of diverse living cell types within plastic laminate devices stabilized in a sol-gel matrix for months under dry, ambient conditions; (x) simultaneous electrochemical and fluorescence measurements for orthogonal biodetection; (xi) colorimetric biodetection requiring no power, instrumentation or skill to perform and read the assay; and (xii) multianalyte cell-based biosensing of small molecules. These accomplishments were achieved via integration of additional functional material components within the laser ablated plastic laminate system.

To provide a sustainable tool in the global effort to combat infectious diseases the diagnostic device must be low cost, simple to operate and read, improve safety for the operator, have little to no reliance on power or refrigeration, and have sensitivity and specificity comparable to laboratory analysis. Laser machined plastic laminates, especially with colorimetric assay and integrated magnetic-adhesive based valving, show great potential for future application in low resource environments where power, skill, reagents, and other resources are scarce. Such systems will require continued design, development, and field testing in order to reliably operate under these stringent requirements and become practical for on-site diagnosis and combating global pandemics.

Acknowledgements

We thank Jacklyn Murton (Sandia) and Prof. Julie Lovchik (University of New Mexico) for magnetic-adhesive valve testing. This work was funded by the Sandia Lab Directed Research and Development (LDRD) Program and the Defense Treat Reduction Agency (DTRA) Chem. Bio. Basic Research Program (grants B0844671, B0947321, B01144531). This work was performed, in part, at the Center for Integrated Nanotechnologies, an Office of Science User Facility operated for the U.S. Department of Energy (DOE) Office of Science (user project U2012A0083, PI: JCH). CJB acknowledges funding from the Air Force Office of Scientific Research grant FA9550-10-1-0054, and the U.S. Department of Energy Office of Science, Office of Basic Energy Sciences, Division of Materials Sciences and Engineering. Sandia National Laboratories is a multi-program laboratory operated by Sandia Corporation, a wholly owned subsidiary of Lockheed Martin Company, for the U.S. Department of Energy’s National Nuclear Security Administration under contract DE-AC04-94AL85000.

References


Received: May 21, 2015
Accepted: May 22, 2015
Published online: July 14, 2015