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Article

Rapid and Sensitive Point of Care Detection of MRSA gDNA by Nanoelectrokinetic Sensors

Rania Oueslati, [1], Yu Jiang [1], Jiangang Chen[2] and Jie Jayne Wu [1,*]

1 Department of Electrical Engineering and Computer Science, the University of Tennessee, Knoxville, TN 37996; oueslati.rania91@gmail.com; yjiang33@vols.utk.edu;
2 Department of Public Health, the University of Tennessee, Knoxville, TN 37996; jchen38@ utk.edu;
* Correspondence: jwu10@utk.edu; Tel.: 865-974-5494

Abstract: Biosensors have shown great potential in realizing rapid, low cost and portable on-site detection for diseases. This work reports the development of a new bioelectronic sensor called AC electrokinetics-based capacitive (ABC) biosensor, for the detection of genomic DNA (gDNA) of methicillin-resistant Staphylococcus aureus (MRSA). The ABC sensor is based on interdigitated microelectrodes biofunctionalized with oligonucleotide probes. It uses a special AC signal for direct capacitive monitoring of topological change on nanostructured sensor surface, which simultaneously induce dieletrophoretic enrichment of target gDNAs. As a result, rapid and specific detection of gDNA/probe hybridization can be realized with high sensitivity. It requires no signal amplification such as labelling, hybridization chain reaction, or nucleic acid sequence-based amplification. This method involves only simple sample preparation. After optimization of nano-structured sensor surface and signal processing, the ABC sensor demonstrated fast turnaround of results (~10 s detection), excellent sensitivity (a detection limit of 4.7 DNA copies /μL MRSA gDNA) and high specificity, suitable for point of care diagnosis. As a bioelectronic sensor, the developed ABC sensors can be easily adapted for detection of other infectious agents.

Keywords: Capacitive DNA sensor, methicillin-resistant Staphylococcus aureus, nanoelectrokicetics, point of care diagnostics

1. Introduction

Currently, on-site detection of infectious pathogens has unmet needs of finding a rapid, sensitive, specific and easy-to-do sensing method. Among infectious pathogens, antimicrobial resistance (AMR) is quickly becoming a global threat, which could make infections difficult to treat. Identification of AMR pathogens is essential to finding proper clinical intervention and reducing the spread of AMR. Staphylococcus aureus (S. aureus) is a major pathogen responsible for a variety of infectious diseases, and methicillin-resistant S. aureus (MRSA) is the most common cause of both hospital and community acquired blood stream infections. Currently, there are no economic and rapid strategies available to screen for MRSA infections. To further compound the problem, the number of samples for MRSA screening has increased dramatically over the past several years.

Existing approaches to detect specific pathogens include culturing methods and molecular diagnostic tests. Culture methods involve the growth of microorganisms on plates with predetermined culture medium under controlled conditions that allows the specific bacteria to multiply and form colonies for enumeration. Although it offers high sensitivity, detection of MRSA by microbial culture takes days to get results. Molecular diagnostics, such as DNA microarrays and polymerase-chain-reaction (PCR) based methods, detect target specific nucleic acid sequences from microorganisms of interest. These methods require trained personnel and sophisticated laboratory equipment. For point of care (POC) diagnosis, it is difficult to complete the complicated steps on site, such as labeling of the DNA samples, multiple washing and specific hybridization steps. Without a rapid

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and field deployable diagnostic method, it will possibly lead to delayed treatment and/or potentially inappropriate antibiotic treatment.

Recent years have witnessed great technological advances in biosensors and microfluidics, which together have contributed to the rapid growth of POC technologies. DNA biosensors are usually based on nucleic acid hybridization processes occurring on the sensor surface. It consists of a molecular probe immobilized on the sensor surface, which typically is a short DNA sequence that detect the target genetic section, enabling specific biomolecular interaction. In a DNA biosensor, the probe most frequently consists of approximate 15 to 50 oligonucleotides. [1] The sensor is a transducer which converts the interaction of the probe with the specific genetic sequence in the genome of pathogen into a measurable signal. [2] Successful recognition of a specific sequence of DNA requires a highly selective and specific probe layer. Subsequent application of a sample to the DNA sensor will result in hybridization (probe-target binding) if the target gene exists, whereupon a signal transduction can occur.

Based on the different types of transducers, main types of DNA biosensors include optical biosensors and electro-chemical biosensors. Electrical DNA sensors offer many advantages such as multiplexed detection, cost reduction, and compact instrumentation. In recent years, many electrical DNA sensors have been reported in the literature. [3] The electrical biosensors, based on the parameter measured, can be further classified as potentiometric, amperometric, and impedimetric sensors. [2] To sum up state-of-the-art in DNA sensors, Table 1 presents the figures-of-merit for the previously reported sensors. Here we report a DNA sensor based on the measurement of sensor capacitance. With the inclusion of phase information in capacitive sensing, it is expected to increase the sensitivity and specificity of a sensor. So far, not many capacitive biosensors have been developed for DNA/RNA classification and pathogen detection. In 1999, a label-free capacitive detection method for DNA detection has been reported by Berggren et al. with a limit of detection of 25 copies/μL containing single stranded fragments of cytomegalo virus. [4] Moreno-Hagelsieb et al. demonstrated the use of an Al/Al2O3electrode to detect as low as 50 ng/ml of 150 base pairs (bps) long biotinylated HIV DNA sequence. [5] In 2016, Rashid et al. developed a dengue virus serotype 3 biosensor based on direct electrochemical detection. It was able to detect 3.09 nM of target nucleic acid sequence using electrodes functionalized with a DNA probe. [6] Despite much research effort spent on nuclei acid biosensor development, significant challenges remain. A number of studies [4-7] reported poor sensor-to-sensor reproducibility. In addition, these approaches require complicated sensor operation such as labeling of the genomic material samples, as well as involves multiple washing and hybridization steps [5, 9-10], not very practical to serve for POC use.

Table 1. Comparison of electrochemicals sensors for S. aureus DNA detection.

Reference	Technique	Testing solution	LOD and DNA size	Dynamic range	Assay time	Sensor portability
[5]	Capacitance measurement	Labeled: Au-nanoparticles	0.2 nM dsDNA (150 bps)	0.2- 20 nM	7 min	Not portable (Lab testing)
[6]	Cyclic Voltammetry	10 mM PBS (pH 7.4) containing 5 mM $Fe(CN)_{63}/Fe(CN)_{64}$- and 0.1M KCl.	57 fM (after PCR)	-------	4 h	Not portable (PCR lab testing)
[7]	DPV	2 × SSC	23 pM (after PCR)	50 to 250 pM	2 h	Not portable
[8]	EIS	0.1 M PBS containing 5 mM [Fe(CN)6]3−/4−	100 fM DNA (short)	----	30 min	Not portable (Autolab measurement)
This work	Capacitance measurement	0.5x SSC	16 aM (2.8 Mbps)	0.4 fM-0.4 pM	10 sec	Portable: ABC readout system

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To further compound the problems, most reported sensors were developed to work with short DNA sequences of no more than several hundreds of bps such as those above, rather than working with whole genomic DNAs of millions bps from clinical samples. The detection outcomes of those sensors are usually unoptimistic when dealing with larger DNA or unpurified gDNA samples before enzyme digestion or PCR amplification. There exist several challenges for direct analysis of gDNAs with complex secondary structures and great molecular weights and lengths. Firstly, hybridization with the target sequence in gDNA is difficult owing to its complicated, possibly tangled, conformation. Secondly, it is difficult for gDNA to approach the probes, which is partially due to its electrical charges, and further worsened by transport limited reaction inherent to gDNA's large molecular size. Lastly, the extracted genome samples usually contain non-specific DNA fragments, which diffuse more readily than larger gDNAs and will lead to higher background noise due to non-specific surface adsorption. Therefore, while testing whole gDNAs is a more feasible strategy for POC, it is also more challenging to perform. Direct detection of bacterial gDNA, such as the one reported here with 2.8 Mbps, has not yet been reported.

To overcome these difficulties, we have developed an AC electrokinetics (ACEK) based capacitive (ABC) sensing method. ACEK is a type of nanofluidic phenomena taking place under AC electrical fields that can lead to directed movement of nano-particles such as biomolecules including DNAs towards electrodes, hence increasing detection speed and signal. [8-10] ACEK effects are induced when applying an AC electric field over electrode sensors in biofluids. In 2013, Cui et al. [12] reported an ACEK-enhanced capacitive immunosensor employing aluminum IDEs modified with a suitable mycobacteria antigen for POC serodiagnosis of infectious diseases. Here, this work reports an approach for sensitive and selective detection of gDNAs that will be highly valuable in controlling and preventing diseases outbreaks. Previously, our work has shown that dielectrophoresis (DEP) and other ACEK effects help to accelerate the binding process by increasing the diffusion of the target analytes onto the sensing surface [13-15]. Moreover, the ABC sensing method has been reported to demonstrate very promising performance for real time and sensitive detection of other nuclei acids [16-17]. This technology also boasts of a low-cost implementation. Its microelectrode sensors are built on printed circuit board (PCB) (~$1), making sensors low cost and disposable. A low-cost capacitance readout system has been developed, using microcontroller and integrated circuits, with a total cost less than $200.

2. Materials and Methods

2.1. Capacitive sensing of sensor surface nano-topology

In this work, disposable interdigitated electrodes (IDEs) have been adopted as the capacitive sensor. When immersed into an electrolyte, the IDE cell can be electrically modelled by an equivalent circuit as shown in Fig. 1. [19-20] The circuit is made up of three parts, the electrode's self-resistance (Rwire), interfacial impedances including the interfacial capacitance (Cint) in parallel to the charge transfer resistance (Rct) and Warburg impedance (Zw), and solution impedance including bulk resistance (Rs) and solution capacitance (Cs)[12]. At the electrolyte/electrode interface, the charge carriers change from ions in electrolytes to electrons in metals. This charge transfer process is often accompanied by electrochemical reactions. If electrochemical reactions are to be avoided, electrical charge transfer through the interface needs to be capacitive in nature, which is represented by Cint. [23] Cint is composed of the electric double layer capacitance and the capacitance caused by biomolecules deposition, which carries the information of specific binding occurring at the sensor surfaces. The charge transfer through the fluid bulk can be done in both resistive and capacitive manners. Cs is based on the dielectric properties of the sample fluid. Un-bounded biomolecules such as non-targets within the solution will cause a change in Cs. Based on the above analysis, Cin is useful for extracting the information of specific binding at the electrode surface, while Cs is strongly affected by sample properties

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and interferences. Cs, known as the parasitic capacitance, does not contribute positively to the detection, and should be minimized from the sensor response if possible. To do so, we capitalize on the fact that non-resistive components such as Cin and Cs in the circuit pose frequency-dependent impedance. As electrical current takes the path of the least resistance, there exists a frequency range over which the electrical current will predominately go through Rwire, Cint and Rs (as shown by red lines in Fig. 1, since other electrical components pose much larger impedance and their roles in electrical conduction can be neglected. As a result, in this frequency range, the sensor impedance can be simplified as a serial connection of Rs and Cint. By reading the capacitive signal only, the interfacial changes can be extracted directly.

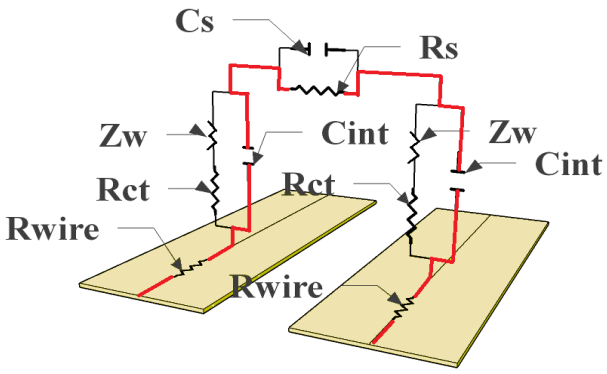

Figure 1. Equivalent circuit for an interdigitated electrode (IDE) when immersed into fluid.

Capacitive biosensors principle can be explained by electrically modeling the interface as a parallel plate capacitor, which consists of two conductive plates separated by a dielectric layer. In this situation, the dielectric layer is made up by the electrical double layer (EDL) and immobilized molecules on the electrode surface. The metal electrode is one plate, while the outer boundary of the EDL forms the other plate of the capacitor. According to (1):

$$C = \frac{\varepsilon_m * A}{d} \qquad (1)$$

where εm is the permittivity of the medium between the plates, A is the plate's surface area, and d is the thickness of the dielectric layer. Thus, the capacitance is proportional to the surface area.

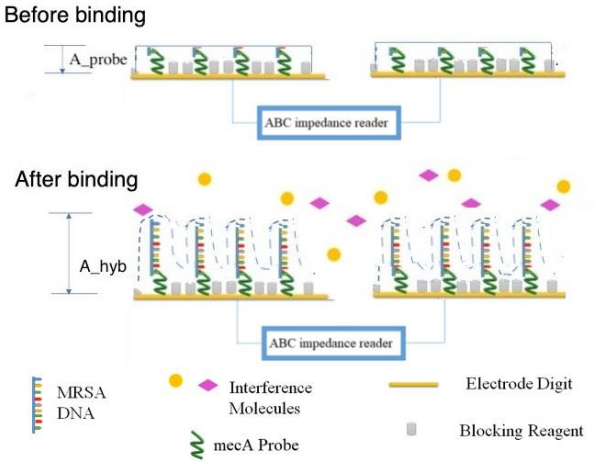

Figure 2. Change at electrode/electrolyte interface due to probe/target analyte binding. Cint increases as the surface area of the interfacial layer decreases.

As biosensors, the IDE surface need to be functionalized, i.e. immobilized with bio-recognition molecule, known as probe. Detection of the target DNA is achieved via

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specific binding between the target and the probe. When the binding occurs, the topology of the sensing electrodes changes. As linear gDNA hybridizes with the probe, the conductive gDNA becomes an extension of electrodes, and the Cint's area increases, hence an increase in Cint as conceptually shown in Fig. 2. [16]

Before the binding process, a layer of the probe molecules is immobilized onto the surface of the electrode. (see Figure. 2). The interfacial capacitance can be expressed as

$$C_{bef} = (\varepsilon_{\mathrm{m}} A_{probe}) / (\mathrm{d}), \quad (2)$$

where A_{probe} is the surface area of the EDL formed on the functionalized IDE surface, ε_{m} is the medium permittivity and d is the EDL thickness. When binding reaction occurs, the interfacial capacitance will change. As shown in Figure.2, as the binding between the probe and the analytes occurs, the surface of the dielectric layer increases. This change causes an increase in the interfacial capacitance of the electrode. The interfacial capacitance is then expected to change to:

$$C_{aft} = (\varepsilon_{\mathrm{m}} A_{hyb}) / (\mathrm{d}) \quad (3)$$

where A_{hyb} is the area of hybridized DNAs on the electrode surface. The change in the interfacial capacitance is consequently expressed as in (4):

$$\Delta C_{int} = \varepsilon_{\mathrm{m}} (A_{hyb} - A_{probe}) / (\mathrm{d}) \quad (4)$$

2.2. DNA enrichment by Dielectrophoresis (DEP)

Detection of biosensors typically relies on the diffusion of target macromolecules to sensors, which could lead to a long detection time (hours or even days) and low sensitivity. [21] To improve biosensor's sensitivity and response time, great research effort in micro-/nano- fluidics has been devoted to the enrichment of target biomolecules. [22] The enrichment mechanism adopted for our bacterial gDNA detection is based on ACEK enhanced capacitive sensing. A special AC measuring signal is applied between a pair of functionalized IDEs, which induces micro to nano-scale particle and fluid movement [23], so that biomolecules are convected towards the IDEs, and enriched in situ there.

ACEK nanofluidics refers to the induced movement of nano-bio-molecules and fluid by a small inhomogeneous AC electric field, which as a result is localized around electrodes [24]. For the test solution and target gDNA in this work, the dominant ACEK effect is dielectrophoretic (DEP) force. A great amount of research has been done on DEP for controlled manipulation of particles, binary separation, and characterization of particles, [25-26], however, using high voltages of several volts. Here, milli-volt level AC signal was used in order to achieve molecular selectivity in gDNA detection.

The target gDNAs are linearized by the testing solution, with 2.2-2.6 nm in diameter and 0.9-1.0 mm in length (~3 Mbps). The DEP velocity for a linear particle can be described as [27]

$$v_{cylDEP} = \frac{r^2 \ln\left(\frac{2l}{r}\right)}{18\eta} \varepsilon_m Re\left[\frac{\varepsilon_p^* - \varepsilon_m^*}{\varepsilon_m^*}\right] \nabla |E|^2 \quad (5)$$

where $\varepsilon_m$is the medium permittivity, η is medium viscosity, r the radius of DNA cross section, l is the length of DNA, ε_p^* and ε_m^* are particle and medium complex permittivity, respectively. Complex permittivity is defined as $\varepsilon^* = \varepsilon - j\frac{\sigma}{\omega\varepsilon}$ (where σ is conductivity and ω is angular frequency). $\mathrm{Re}\left[\frac{\varepsilon_p^* - \varepsilon_m^*}{\varepsilon_m^*}\right]$ gives the real part of the function. If $\mathrm{Re}\left[\frac{\varepsilon_p^* - \varepsilon_m^*}{\varepsilon_m^*}\right] > 0$ (usually when the fluid conductivity is higher than the particle conductivity at the operation frequency), particles will be attracted to electrodes, which is known as positive DEP (pDEP). [27] To ensure pDEP for gDNA, buffer with a conductivity of ~9 mS/cm was used in this work. Also based on (3), DEP velocity is strongly dependent on the DNA radius as well as the electric field gradient. Given the nm radius of nuclei acids and mV AC signal, only gDNAs will be enriched to the IDE surface for accelerate detection.

2.3. Reagents, sensor and sample preparation

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The sensors are prepared by immobilizing oligonucleotide probes on PCB based gold IDEs. The electrode digit is 400μm wide and 1cm long with a 200μm spacing between each. A detailed description of the PCB based gold IDE fabrication steps is provided in a previous paper. [28]

Cleaned electrodes are functionalized with thiol modified probe for 22 hours in a humidor under room temperature. The probe used in this work was a 5′ thiol modified oligonucleotide probe (5′-GTAGAAATGACT-GAACGTCCGATAA-3′). It was designed to specifically target mecA gene segment found only in specific MRSA. [31] Phosphate buffered saline (PBS) (0.05×) was used for probe incubation, which was prepared by 1:20 volume dilution of commercial physiological strength 1× PBS (containing 75 mM of sodium chloride, pH 7.4, Thermo Fisher Scientific, Waltham, MA, USA) in ultrapure water (Mili-Q). To prevent nonspecific binding due to uncovered sites on functionalized electrodes, the electrodes were then blocked with 1.0 mM 6-mercaptohexanol (Sigma-Aldrich, St.Louis, MO, USA) prepared in ultrapure water.

In this work, MRSA gDNA was used as a model, which is one of the most common causes of hospital and community-acquired blood stream infections. Quantitative S. aureus strain Mu50 DNA was purchased from ATCC (ATCC 700699DQ), which were prepared in ultrapure water at concentration of 2.7 ng/mL as the stock sample solution. The testing buffer was based on saline-sodium citrate (20xSSC) buffer, purchased from AccuGENETM (Lonza, Rockland, ME, USA). It was diluted in ultrapure water to make 0.5 × SSC. The gDNA from E coli was extracted and prepared in house and served as the control sample.

Three types of control tests were carried out to verify that the C_{int} change was caused by the target and not the interference molecules. One was testing the blank background solution (0.5×SSC) without target molecules using a functionalized sensor, and one was testing samples with target molecules on dummy electrodes (electrodes that were blocked without probe immobilization), so as to gauge the magnitude of response artifacts caused by the applied electrical field, and the third is to test samples with non-target molecules.

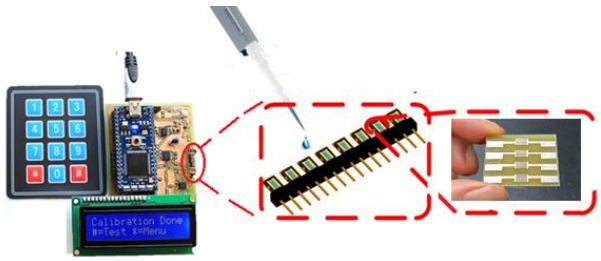

Figure 3. Photo of the handheld DNA ABC biosensing system. On the Au-PCB electrode, analytes are induced toward the binding sites on electrodes surface by ACEK effects. Hybridization between probe and DNA target increases the interfacial area of the electrode so a change at the interface (Cint) occurs. This change is detected and measured by the portable read-out system.

The IDEs were electrically connected to two contact pads, which were then used to connect to an impedance analyzer. Detection procedure was as follows. First, the double strands of MRSA DNA should be separated (denatured) and linearized to allow hybridization with the probes [11] on the IDEs. Thermal denaturation was used to obtain single strand DNA (ssDNA). [30] DNA stock samples were diluted and heated on a dry bath at 95°C. After 10 min, the tubes of the diluted samples were inserted in ice for 1 min. At After the samples returned to room temperature, they were added onto the sensors connected to the ABC system for tests, which imposed an inhomogeneous AC electric field on the sensor. As shown in Figure 3, the same ACEK signal was used to track the change in Cint at the surface of the IDE, quantifying the binding between the target and the probe DNA.

3. Results and discussion

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3.1. Nanostructured probe layer on electrode surface

Forming a good quality layer of probes on sensor surface is critical to obtaining sensitive and reliable biosensors. To achieve this objective, it is paramount important to ensure the cleanliness of electrode surface and effective immobilization of biomolecules on sensor surfaces.

Electrodes were washed sequentially with Acetone, Isopropyl Alcohol, de-ionized water and Ozone. As it affects different surface process (e.g., levels of cleanness and surface deposition), the cleaning step correlates strongly with the interfacial area or the binding sites at microelectrode surface. Hence, the sensor surface was closely monitored through the interfacial capacitances (Cint). This surface analytical method is simple and rapid and has been validated experimentally [19]. In our protocol, Cint was measured after each step of cleaning and incubation, using 0.1x (15mM Na+ PBS) solution at 100 kHz 5mVrms. We have developed a reference table of Cint at each step to ensure electrode quality. [31]

For surface-based biosensors, surface functionalization is a crucial step to realize biospecficity. This step consisted of immobilizing a layer of probe molecules onto the electrode surface. When functionalized properly, the detection of the binding signal became more effective, as the surface functionalization determines the number of available binding sites and so the change of the interfacial capacitance. It was typically desired to have the optimal surface coverage to obtain large sensor response. In this work, to improve MRSA sample detection and optimize the incubation step, three different probe concentrations were tested and their results are summarized in Figure. 4.

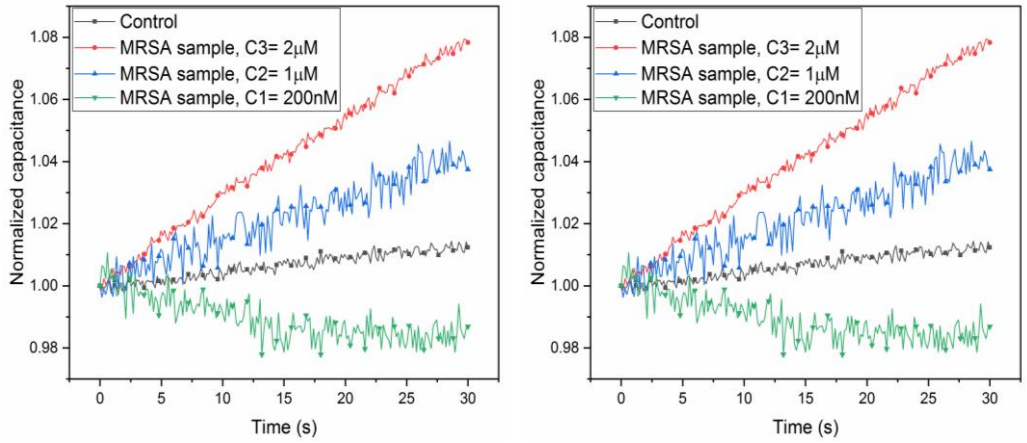

Figure 4. **(a)**Three different probe concentrations C1= 200 nM, C2=1 μM and C3= 2μM were tested; (**b)** Normalized Cint changes with time for control solution and MRSA sample under three different incubation conditions.

The sensor surface was incubated with 10μL of 200 nM, 1 μM and 2 μM mecA probe, respectively. The surface coverage by the probe was monitored by checking the normalized capacitance change before and after probe immobilization. As shown in Fig. 4a, by increasing the probe concentration from 200 nM to 2 μM, the capacitance change increased from 44% to 95 % during the probe incubation, which was expected as corresponding to a higher density of probe immobilization. Different coverage levels of oligo probes would lead to different sensor responses. A higher surface coverage of probes would typically yield a larger response. Figure 4(b) shows some typical responses for sensors functionalized at 3 different probe concentrations when tested using MRSA gDNA samples at 2501.4 copies/μL. As shown in Fig. 4b, the normalized capacitance changed quasi-linearly with time, indicating continuous hybridization between the probes and the MRSA gDNAs. The capacitance change rate was higher for sensors with higher probe concentrations. This could be explained by the higher coverage of the surface that leads to higher chance of

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probe-target hybridization at the electrode surface and then higher sensor response. It should be noted that for sensors functionalized at 200nM probe concentration, the capacitance change became negative when testing MRSA gDNA, which was supposed to be the result of target gDNAs depositing onto the electrode surface after hybridization. As a matter of fact, a lower surface coverage of nuclei acid probes would lead to lower charge density at the electrode surface, thus less repulsive force between the probe-covered electrode surface and the gDNAs. Because the AC electrical fields exerted attraction DEP force on gDNAs, given less charge repulsion from the electrodes, the captured gDNA would approach the electrodes closely and lead to an increase in the thickness of biopolymer layer at the electrode surface, causing a decrease in Cint for specific detection. The control experiments yielded almost no change in Cint as expected (<+/-1%/min).

The probe density also affected the linearity of the transient curves in Figure 4(b). The sensor with the highest probe density exhibited the best linearity, while the sensor functionalized at 200 nM showed slowing down of hybridization after 15 s, which could be attributed to a shortage of probes.

3.2. Optimization of assay time

Next, the assay time was optimized for this MRSA gDNA sensor. Results are shown in figure 5. Our previous work on protein and small molecule detection found that capacitance change rate taken over different time period correlates with analyte reaction / transport-controlled processes [32, 33]. Because DEP attraction decays with distance, gDNAs' travel towards the sensor is slower for those farther away from the electrodes. So, at dilute concentration of gDNA, the hybridization events become less frequent with time. However, a longer assay time may allow stabilization of signals. In this test, the same AC signal was applied to the sensor with three different sweep times (10, 20 and 30 s). Two MRSA concentrations (2501.4 and 25014.4 copies/μL) in 0.5 × SSC were measured for this optimization process. Using 30s, 2501.4 and 25014.4 copies /μL of MRSA gDNA have the responses of 4.44±0.35% and 8.23±0.53 % per minute. When the sweep time decreased from 20 s to 10 s, the response of the sensor to 2501.4 copies/μL MRSA gDNA increased from 5.40±0.47 %/min for 20 s to 5.82±0.18 %/min for 10 s. Similarly, the sensor response of 25,014.1 copies/μL MRSA gDNA increased from 8.30±0.26 %/min for 20 s to 9.04±0.13 %/min for 10 s. A shorter assay time ensures that the reaction is transport limited, and the response has a better correlation with the target gDNAs in the solution, by using this real-time detection mode. The assay time was set to be 10 s from this point on. Also, there was a very little responses change for the control sample, attesting to the specificity of this gDNA sensor.

3.3. Sensor sensitivity and specificity

Three concentrations of MRSA gDNA, 250.1 copies/μL, 2,501.4 copies/μL and 25,014.1 copies/μL, spiked in 0.5×SSC were tested and the results are shown in Figure 6. Each sample was tested three times. A separate chip was used for each test. The results are 3.60±0.53 %/min, 5.82±0.17 %/min and 9.04±0.13 %/min respectively, with the responses of buffer solution 0.5xSSC (background) to be 0.1±0.15 %/min. The sensor showed a logarithmic dependence on MRSA concentration ranging from 250.1 to 25,014.1 DNA copies/μL, and the calibration curve of the sensor was calculated to be $(d|C|/dt) = 1.0067\ln(x) - 2.249$ where x is the concentration of MRSA in 0.5xSSC. The limit of detection (LOD) was determined as 3 standard deviations from the average response of the background control. Hence, the cut-off value for sensor response ($d|C|/dt$) was calculated to be 1.6%/min. Using the calibration equation, the LOD was calculated to be 4.72 DNA copies /μL in 0.5xSSC.

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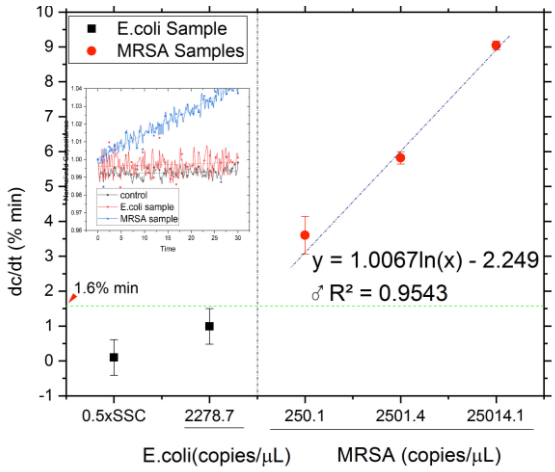

Figure 6. Response of MRSA and E.Coli samples in 0.5xSSC tested on Au-PCB electrodes incubated with 2 µM of mecA probe. LOD was determined by adding 3 standard deviations from the average response to the buffer control response (0.5xSSC).

As sensor specificity is crucial to biosensor operation, E.coli DNA, at a concentration of 2278.7 copies/µL, were tested to determine the sensor specificity. As demonstrated in Fig 6, the capacitance change rates for E.coli (2278.7 copies/µL) was 0.51±0.58 %/min, which is below the cut-off response, hence considered as being negative. In contrast, for MRSA gDNA at a similar concentration of 2,501.4 DNA copies /µL, the sensor response was close to 6%/min. Therefore, the sensor is specific to MRSA gDNA.

4. Conclusions

By optimizing the probe concentration and assay time for the detection of hybridization, rapid, sensitive and specific detection of MRSA gDNA has been achieved using a newly developed ABC biosensor. The sensor operation involves only heat treatment of double strains DNA (dsDNA), adding the sample to the electrode sensor, and then applying the specified AC signal for 10-30 sec. The LOD was calculated to be 4.72 DNA copies /µL. The work will eventually lead to a rapid and PCR-free biosensor platform for on-site sensitive and specific detection of genetic sequences for a wide range of pathogens and biomaterials. For future development, the capacitance measurement can be processed by a microcontroller to achieve automatic calculation of the capacitance change rates. Because electrical detection method is used here, the resultant sensors can be easily combined to form a multiplexed panel test, which is necessary for accurate detection and prognosis of many pathogens.

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