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Therapeutic Drug Monitoring of Amikacin Using a Surface Plasmon Resonance Biosensor

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Abstract

The therapeutic drug monitoring (TDM) of pharmaceutical drugs with narrow therapeutic ranges is of great importance in the clinical setting. It provides useful information towards the enhancement of drug therapies, aiding in dosage control and toxicity risk management. Amikacin is an aminoglycoside antibiotic commonly used in neonatal therapies that is indicated for TDM due to the toxicity risks inherent in its use. Current techniques for TDM such as high performance liquid chromatography (HPLC) and gas chromatography-mass spectrometry (GC-MS) are costly, time consuming, and cannot be performed at the site of action. Over the last decades, surface plasmon resonance (SPR) biosensors have become increasingly popular in clinical diagnostics due to their ability to detect biomolecular interactions in real-time.

We present an SPR-based competitive immunoassay for the detection of the antibiotic amikacin, suitable for TDM in both adults and neonates. We have obtained high specificity and sensitivity levels with an IC_{50} value of 1.3 ng/mL and a limit of detection of 0.1 ng/mL, which comfortably comply with the drug's therapeutic range. Simple dilution of serum can therefore be sufficient to analyze low-volume real samples from neonates, increasing the potential of the methodology for TDM. Compared to current TDM conventional methods, this SPR-based immunoassay can provide advantages such as simplicity, potential portability, and label-free measurements with the possibility of high throughput. This work is the foundation towards the development of an integrated, simple use, highly sensitive, fast, and point-of-care sensing platform for the opportune TDM of antibiotics and other drugs in a clinical setting.

Keywords: Therapeutic drug monitoring; Surface plasmon resonance; Amikacin; Immunoassay

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4 **1. Introduction**
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8 In the last decade research related to immunosensors has been focused towards the development
9 of point-of-care devices that can allow management of a particular issue whether it is food safety,
10 security, environmental, or medical related [1, 2]. Therapeutic drug monitoring (TDM) is a branch of
11 clinical chemistry and pharmacology that specializes in the measurement of therapeutic drug
12 concentrations in blood or other body fluids. TDM is indicated for drugs that have narrow therapeutic
13 ranges, i.e. drugs that present a high risk of toxicity when overdosed or ineffectiveness when underdosed
14 [3]. Moreover, neonate (newborn) patients have unique drug dispositions that indicate the need of TDM
15 in order to ensure a safe drug therapy [4]. This monitoring aims to enhance drug efficacy through toxicity
16 risk management, dosage control, and assistance in clinical diagnosis [5, 6].
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19 Antibiotics are drugs that are suitable for TDM and they are the most widely used drugs in
20 neonatal therapy due to newborns' high susceptibility to infections [7]. Amikacin (AK) is an
21 aminoglycoside antibiotic commonly used in neonatal therapies [8, 9], and is indicated for TDM due to
22 the toxicity risks inherent to its use as well as its narrow therapeutic range (1-30 $\mu\text{g/mL}$) [10, 11]. TDM of
23 Amikacin and other pharmaceutical drugs is mostly performed in blood serum or plasma [6, 12] and
24 given the low blood volume in neonates of approximately 80 mL/kg, small sample sizes are most
25 desirable in order to avoid detrimental impact to the patient due to blood loss [4]. Current conventional
26 methods for TDM are high performance liquid chromatography (HPLC) [13, 14], gas chromatography-
27 mass spectrometry (GC-MS) [15, 16], and other commercial immunoassays [17, 18] (if available for the
28 desired analyte), which are costly and time consuming. Additionally, these methods are complex and
29 cannot be performed at the site of action [6].
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32 Surface plasmon resonance (SPR) is a widely known optical technique thoroughly studied in the
33 last few decades, which uses evanescent waves to investigate surface phenomena and generate a signal
34 related to a change in refractive index at an interface. It has become popular in a variety of fields
35 including environmental protection, fundamental biological studies, food safety and clinical diagnosis
36 applications [19]. Advantages of SPR biosensing include real-time and label-free monitoring, simplicity,
37 high levels of sensitivity and fast-response measurements, making it a potential alternative to current
38 clinical and laboratory measurement techniques such as enzyme immunoassays (ELISA, EMIT) and
39 chromatographic techniques (LC-MS, HPLC) [19, 20]. While SPR biosensing has been used for the
40 detection of antibiotics in milk samples for food safety applications [21, 22], their application for TDM
41 purposes, especially in neonates, has not been approached. The antibiotic kanamycin has been monitored
42 using a fluorometric technique based on plasmon resonance light-scattering (PRLS) of gold nanoparticles
43 (AuNPs) [23], albeit in urine samples for monitoring of nephrotoxicity and ototoxicity.
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4 In this work we present a SPR-based competitive immunoassay for the detection of AK, suitable
5 for TDM in both adults and neonates. We have obtained high specificity levels and a high sensitivity
6 which complies with the drug's therapeutic range, demonstrating the effectiveness of this approach to
7 further point-of-care on-site TDM monitoring.
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10 11 12 **2. Materials and Methods**

13 ***2.1 Chemicals and Reagents***

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18 Reagents for the self-assembled monolayer (SAM) formation (16-mercaptohexadecanoic acid
19 (MHDA) and 11-mercaptoundecanol (MUOH)), common reagents (N-hydroxysuccinimide (NHS), 1-
20 ethyl-3(3-dimethylamino-propyl) carbodiimide hydrochloride (EDC), ethanolamine), inorganic salts for
21 buffer preparation (phosphate buffer saline (PBS), 2-(N-morpholino)ethanesulfonic acid (MES), Tween
22 20), bovine serum albumin (BSA) and AK were purchased from Sigma Aldrich.
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26 Polyclonal IgG anti-AK antibody was purchased from Abcam (England) and AK-BSA conjugate
27 was synthesized as described in Section 2.4.
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30 31 ***2.2 Instrumentation***

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35 SPR measurements were performed using a homemade SPR biosensor based on the Kretschmann
36 configuration under a prism-coupling scheme. A polarized 3 mW laser beam with an operational
37 wavelength of 670 nm is directed at a fixed angle towards the gold-coated sensing surface. The sensing
38 chips are 10 x 10 x 0.3 mm glass coated with 2 nm of chromium and 45 nm of gold, from Ssens
39 (Netherlands). SPR measurements are performed at a fixed angle of incidence. The optimum
40 measurement angle is selected during calibration to ensure maximum sensitivity to changes in the
41 refractive index, which in turn are caused by mass variations at the sensing surface. These variations are
42 detected as changes in the reflected light intensity at a photodiode, and are subsequently amplified and
43 converted to a digital signal, which we refer to as SPR signal (intensity variation of the reflected light at a
44 fixed angle).
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48 Sample sizes of 250 μ L are injected by a diaphragm pump to the flow delivery system, which
49 incorporates a flow cell with two independent channels. The system incorporates all the optics, electronics
50 and fluidic components necessary to function autonomously.
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53 54 55 56 57 ***2.3 Synthesis of the AK-BSA conjugate***

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4 AK-BSA conjugate was prepared based on the conjugation of the amino groups of AK to the
5 carboxylic groups present in BSA. Briefly, a solution of AK (11.11 mg/ml in MES buffer pH 5.0) and the
6 carrier protein BSA (10 mg/ml in MilliQ water) was incubated for 3 hours with 10 mg of EDC as a linker.
7 Next, purification of the conjugate was achieved by using PBS 10 mM (pH 7.4) as a purifying buffer and
8 a 30 kDa centrifugal filter. The solution was mixed with 250 μ L of purifying buffer in the filter and was
9 centrifuged at 12000 RPM during 120 s, after which the residues from the collection tube were discarded.
10 This process was repeated three times, yielding the final conjugate solution. Conjugate concentration was
11 determined by UV resulting in 14.3 mg/ml.
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19 ***2.4 Immobilization Procedure***

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22 Gold chips were prepared for immobilization by a wash with ethanol, followed by 20 minutes of
23 surface cleansing with UV-ozone (Bioforce Nanosciences, Inc, model UV-TC.220) to eliminate organic
24 contamination. A second rinse with ethanol was performed and finally the chips were dried with N₂.
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27 A covalent strategy was used for biofunctionalization of the surface. After chip cleaning, the
28 formation of an alkanethiol self-assembled monolayer (SAM) was carried out ex-situ by coating the gold
29 chip overnight with a mixed solution of MHDA:MUOH in ethanol, with a molar ratio of 1:1 and at a total
30 concentration of 250 μ M. The chip was rinsed with ethanol and dried with N₂, and mounted in the SPR
31 platform. Next, an AK-BSA conjugate was covalently coupled to the carboxylic groups of the SAM
32 through activation of the acid groups with a solution of 0.2 M EDC and 0.05M NHS in MES buffer (pH
33 5.0). Subsequently, after the concentration of AK-BSA conjugate in PBS buffer was injected, a 1 M
34 ethanolamine solution (pH 8.5) was used for the deactivation of unreacted carboxylic groups. This
35 procedure was done at a flow-rate of 14 μ L/min.
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44 ***2.5 SPR immunoassay format***

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47 An indirect competitive immunoassay format was used for the detection of AK levels given the
48 low molecular weight of the molecule (585.60 Da). Therefore, the AK-BSA conjugate was immobilized
49 and used as the sensing bioactive surface. After assay optimization, samples consisting of mixtures (1:1
50 v/v) of different analyte concentrations (ranging from 1 pg/mL to 5 μ g/mL in PBS or PBST (0.05%
51 Tween 20)) and a fixed antibody concentration were flowed over the sensing layer at a speed of 22
52 μ L/min. Samples were incubated for 10 min at room temperature before injection and monitoring was
53 performed in real time. Since antibody binding to the immobilized conjugate is inhibited by the presence
54 of analyte, lower analyte concentrations resulted in high SPR signals and vice versa. Standard calibration
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4 curves were constructed with at least 7 points (7 different AK concentrations) with three replicate
5 measurements each. Reutilization of the sensor surface was achieved using a solution of NaOH 45 mM as
6 a regeneration solution, during 120 s at a speed of 22 $\mu\text{L}/\text{min}$ after each sample injection.
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9 10 **2.6 Data analysis**

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14 Standard calibration curves were obtained by plotting the average SPR signal and standard
15 deviation of three samples for each AK concentration, versus the logarithm of the AK concentration of
16 the sample. These experimental points were fitted to the four-parameter logistic equation:
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$$19 \quad y = \{D + (A - D) / [1 + (x/C)^B]\}$$

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24 where y is the SPR signal (variation of reflectance at a fixed angle, % Reflectance), A is the maximum
25 signal (zero concentration of analyte), D is the asymptotic minimum (background signal), B is the slope of
26 the sigmoidal curve at the inflection point, and C is the concentration which gives 50% of the maximum
27 signal, IC_{50} .
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31 32 **3. Results and Discussion**

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36 Molecular weight of the analyte is a key factor when selecting the immunoassay configuration in
37 SPR biosensing. A direct assay configuration (which uses an antibody coated surface) may be the best
38 option in order to simplify conditions, even more with views of developing point-of-care devices, but it
39 becomes an unfeasible approach due to the low molecular weight of AK (585.60 Da), which inevitably
40 would lead to low detectability. Therefore, we proceeded to the optimization of an indirect competitive
41 inhibition assay (Fig. 1), in which samples consisting of a fixed concentration of antibodies incubated
42 with different concentrations of the analyte are then flowed over an antigen-coated surface. Compared
43 with direct assays, indirect approach avoids some drawbacks related to the immobilization of antibodies
44 to the solid surface and their potential loss of affinity, either due to their attachment which can affect the
45 native structure, or simply during assay development and optimization. This process usually requires
46 working under conditions that allow regeneration of the bioactive surface (i.e. basic or acid conditions
47 which disrupt the interactions between the antibody and the target molecule, therefore releasing it and
48 leaving the receptor ready for the next sample). Subjecting the antibody coated surface to these conditions
49 usually lead to an irreversible alteration of antibody structure, and therefore to its binding ability.
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51 Moreover, the direct approach has to deal with the proper orientation of the antibodies on the surface,
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4 which can also affect the overall assay performance. On the other hand, in an indirect assay the
5 immobilization of the target, either directly or through a carrier molecule (typically a protein) which
6 previously has been decorated with the analyte, provides a higher degree of robustness and reusability
7 even under mild or harsh regeneration conditions.
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12 In the indirect immunoassay, parameters such as antigen conjugate concentration, antibody
13 concentration and interaction buffer become crucial to obtain the best features in terms of sensitivity,
14 reproducibility and stability. These factors together with the regeneration solution were optimized to
15 achieve the best sensor performance.
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19 AK-BSA conjugate concentrations ranging from 5 to 30 $\mu\text{g/mL}$ of AK-BSA were tested over
20 gold substrates coated with a mixed SAM (MHDA:MUOH 1:1). As can be seen in Table 1 a gradually
21 increasing amount of conjugate was bound to the surface as the added antigen concentration is also
22 higher, until approaching saturation of the surface at 30 $\mu\text{g/mL}$ of AK-BSA.
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27 The concentration of fixed antibody to be used in the inhibition immunoassay influences
28 immunoassay characteristics such as the limit of detection (LOD), sensitivity, and dynamic range.
29 Additionally, in order to favor competition, a low concentration of free antibody will usually give better
30 results. Therefore, there is a trade-off between using low concentrations for a better assay sensitivity, and
31 using a concentration high enough (but well below saturation level of the antigens) to obtain discernible
32 signals for a wider range of analyte concentrations.
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37 Several antibody concentrations (up to 6 $\mu\text{g/mL}$) were tested and the signal obtained did not reach
38 saturation in any case. The average signal response when injecting 6 $\mu\text{g/mL}$ of anti-AK antibody over
39 chips immobilized with 5, 15 and 30 $\mu\text{g/mL}$ of AK-BSA was of 0.30, 0.53, and 0.78 respectively. As
40 could be expected, results showed an overall higher response to the same antibody concentration for
41 surfaces in which higher immobilization signals were obtained. Finally an AK-BSA concentration of 30
42 $\mu\text{g/mL}$ and an antibody concentration of 6 $\mu\text{g/mL}$ were deemed optimal, guaranteeing a sufficient
43 maximum SPR signal in absence of analyte while assuring a wide dynamic range for clear differentiation
44 between different analyte concentrations.
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50 In order to assess the stability of the assay and also to test the reusability and robustness of the
51 bioactive surface, we studied several regeneration conditions in order to remove the bound antibody. The
52 performance of several acid and basic solutions (i.e. HCl, glycine and NaOH at different concentrations)
53 was compared. An effective regeneration is crucial to ensure the reusability of the sensing surface and the
54 reproducibility of the measurements. Incomplete removal of antibody will produce disperse
55 measurements between replicates and an excessively harsh conditions might compromise the stability of
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4 the sensing surface reducing its life span. Regeneration with NaOH 45 mM solution injected for 120 s
5 was selected as optimum conditions to achieve complete removal of the bound antibody. These conditions
6 ensured the reusability of the surface over a span of 3 days and at least 40 measurement cycles, observing
7 then a decrease in the SPR signal of 11% (taking as reference the zero analyte concentration). Stability in
8 the day-to-day measurements was evidenced by consistence in the SPR signals, confirming the reliability
9 and robustness of the immobilized chips.
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14 With these initial conditions, calibration curves were performed, both in PBS and PBST (PBS
15 containing 0.05% Tween 20) Figure 2 shows sensograms indicating the SPR response to the binding of
16 different concentrations of AK, followed by their respective regeneration cycles. As expected, SPR
17 signals decrease in response to increasing AK concentrations.
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21 Significant better performance was obtained using PBST instead of PBS. When using PBS as
22 running buffer, lower reproducibility was observed (relatively high standard deviation between
23 measurements). On the other hand, the use of PBST resulted in a lower dispersion between replicates and,
24 most importantly, in a much higher sensitivity (see Figure 3).
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29 Table 2 summarizes immunoassay features for both buffers and as can be seen a LOD one order
30 of magnitude better was reached using PBST instead of PBS, with a slope of the linear area close to 1,
31 which may result in a more precise and accurate calibration curve.
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35 Specificity of the sensing surface was evaluated with anti-vancomycin antibody (6 $\mu\text{g}/\text{mL}$)
36 obtaining negligible SPR signals (data not shown). Vancomycin (Van) is an antibiotic also used in
37 newborn therapies for the treatment of infections caused by Gram-positive bacteria, and can usually be
38 coadministered with AK. Thus, the specificity of the current assay was also tested to assess any potential
39 cross-reactivity with Van, by performing the inhibition assay with a fixed concentration of anti-AK
40 antibody (6 $\mu\text{g}/\text{ml}$) and varying concentrations of vancomycin. As can be seen in Figure 3 no inhibition is
41 observed even at high concentrations of vancomycin (9-18% of the maximum signal), which confirms the
42 low specificity of the assay towards Van and its suitability for the detection of AK.
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49 According to the obtained assay features, this SPR-based approach can be very useful for the
50 determination of AK since the sensitivity levels obtained are far below the drug's therapeutic range (1-30
51 $\mu\text{g}/\text{mL}$) and comparable or higher than current TDM methods [24–26]. This high sensitivity entails
52 advantages for the detection of AK in real human samples by allowing sample dilution of approximately
53 5000 times, resulting in sample sizes of few nanoliters. Non-specific binding of matrix components
54 present in serum can sometimes produce analytical interference (positive signal) which masks the specific
55 signal leading to wrong results. This extremely high dilution fold, (which represents less than 0.02% of
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4 sample) becomes convenient in order to minimize the potential negative effects of the complex matrix on
5 the sensing surface. Moreover, in intended uses where the clinical samples come from blood drawn from
6 infants, the low detectability of the assay overcomes the problems associated with dilution of the sample,
7 which sometimes, in assays that show moderate to fair sensitivities, implies to fall below the
8 quantification or even the detection limit.
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14 **4. Conclusions**

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17 A highly sensitive, label-free SPR immunosensor for the therapeutic monitoring of AK has been
18 developed. The sensor uses a sample size of 250 μ L, but according to the high sensitivity obtained, which
19 allows high sample dilution to enter in the working range, very low patient's sample volume of few
20 nanoliters will be presumably required, making the sensor favorable for TDM in neonates while
21 maintaining quantification capability. The sensor is based on an indirect competitive immunoassay with
22 AK immobilized on the gold surface. The immobilization process and other immunoassay parameters
23 were individually optimized for the best performance. We have obtained a very good sensitivity with an
24 IC_{50} value of 1.3 ng/mL. A low LOD of 0.1 ng/mL was achieved, which comfortably complies with the
25 drug's therapeutic range and is adequate for the assay's application in TDM. Reproducibility in the
26 measurements has been also achieved, as well as reusability of the surface for at least 40 measurements
27 over a span of 3 days, confirming the robustness of the method. Finally, non-specificity of the assay
28 towards Van (a drug often coadministered with AK) was also demonstrated.
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38 This SPR immunoassay method provides advantages over current TDM conventional methods
39 (such as enzyme immunoassays and chromatographic techniques) allowing the fast, label-free and real-
40 time determination of AK concentrations, while achieving comparable or even lower detection limits. To
41 our knowledge this could be the first application of SPR in the therapeutic monitoring of antibiotics. This
42 assay demonstrates the potential of SPR as a method for point-of-care TDM of antibiotics (and other
43 drugs) in a clinical setting. Future work will involve the evaluation of the technique using real serum
44 samples, as well as the detection of other drugs indicated for TDM.
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59 **References**

1. Holford TRJ, Davis F, Higson SPJ (2012) Recent trends in antibody based sensors. *Biosens Bioelectron* 34:12–24.
2. D’Orazio P (2011) Biosensors in clinical chemistry - 2011 update. *Clin Chim Acta* 412:1749–61.
3. Pretorius E, Klinker H, Rosenkranz B (2011) The role of therapeutic drug monitoring in the management of patients with human immunodeficiency virus infection. *Ther Drug Monit* 33:265–74.
4. Koren G (1997) Therapeutic drug monitoring principles in the neonate. *Clin Chem* 227:222–227.
5. Gross AS (2001) Best practice in therapeutic drug monitoring. *Br J Clin Pharmacol* 52:5–9. doi: 10.1046/j.1365-2125.2001.0520s1005.x
6. Hiemke C, Baumann P, Bergemann N, et al. (2011) AGNP consensus guidelines for therapeutic drug monitoring in psychiatry: update 2011. *Pharmacopsychiatry* 44:195–235.
7. Chirico G, Barbieri F, Chirico C (2009) Antibiotics for the newborn. *J Matern Fetal Neonatal Med* 22 Suppl 3:46–9.
8. Pacifici GM (2009) Clinical pharmacokinetics of aminoglycosides in the neonate: a review. *Eur J Clin Pharmacol* 65:419–27.
9. Vučićević K, Rakonjac Z, Miljković B, et al. (2014) Pharmacokinetic variability of amikacin after once-daily and twice-daily dosing regimen in full-term neonates. *J Pharmacol Sci* 124:138–43.
10. Allegaert K, Anderson BJ, Cossey V, Holford NHG (2006) Limited predictability of amikacin clearance in extreme premature neonates at birth. *Br J Clin Pharmacol* 61:39–48.
11. Siddiqi A, Khan DA, Khan FA, Razzaq A (2009) Therapeutic drug monitoring of amikacin in preterm and term infants. *Singapore Med J* 50:486–9.
12. Roberts JA, Norris R, Paterson DL, Martin JH (2012) Therapeutic drug monitoring of antimicrobials. *Br J Clin Pharmacol* 73:27–36.
13. Helmy S a (2013) Simultaneous quantification of linezolid, tinidazole, norfloxacin, moxifloxacin, levofloxacin, and gatifloxacin in human plasma for therapeutic drug monitoring and pharmacokinetic studies in human volunteers. *Ther Drug Monit* 35:770–7. doi: 10.1097/FTD.0b013e318297b6b0
14. Simiele M, Baietto L, Audino A, et al. (2014) A validated HPLC-MS method for quantification of the CCR5 inhibitor maraviroc in HIV+ human plasma. *J Pharm Biomed Anal* 94:65–70. doi: 10.1016/j.jpba.2014.01.031
15. Athanasiadou I, Angelis YS, Lyris E, et al. (2014) Gas chromatographic-mass spectrometric quantitation of busulfan in human plasma for therapeutic drug monitoring: a new on-line derivatization procedure for the conversion of busulfan to 1,4-diiodobutane. *J Pharm Biomed Anal* 90:207–14. doi: 10.1016/j.jpba.2013.12.004
16. Nakajima M, Yamato S, Shimada K, et al. (2000) Assessment of drug concentrations in tears in therapeutic drug monitoring: I. Determination of valproic acid in tears by gas chromatography/mass spectrometry with EC/NCI mode. *Ther Drug Monit* 22:716–22.
17. Cézé N, Ternant D, Piller F, et al. (2009) An enzyme-linked immunosorbent assay for therapeutic drug monitoring of cetuximab. *Ther Drug Monit* 31:597–601. doi: 10.1097/FTD.0b013e3181b33da3

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2
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4 18. Tachi T, Hase T, Okamoto Y, et al. (2011) A clinical trial for therapeutic drug monitoring using microchip-based
5 fluorescence polarization immunoassay. *Anal Bioanal Chem* 401:2301–5. doi: 10.1007/s00216-011-5304-9
6
7
8 19. Homola J (2008) Surface plasmon resonance sensors for detection of chemical and biological species. *Chem Rev*
9 108:462–93.
10
11 20. Campbell K, Huet A-C, Charlier C, et al. (2009) Comparison of ELISA and SPR biosensor technology for the
12 detection of paralytic shellfish poisoning toxins. *J Chromatogr B Analyt Technol Biomed Life Sci* 877:4079–
13 89.
14
15 21. Fernández F, Hegnerová K, Piliarik M, et al. (2010) A label-free and portable multichannel surface plasmon
16 resonance immunosensor for on site analysis of antibiotics in milk samples. *Biosens Bioelectron* 26:1231–8.
17
18 22. Haasnoot W, Cazemier G, Koets M, van Amerongen A (2003) Single biosensor immunoassay for the detection
19 of five aminoglycosides in reconstituted skimmed milk. *Anal Chim Acta* 488:53–60.
20
21 23. Wang X, Zou M, Xu X, et al. (2009) Determination of human urinary kanamycin in one step using urea-
22 enhanced surface plasmon resonance light-scattering of gold nanoparticles. *Anal Bioanal Chem* 395:2397–
23 403.
24
25 24. Adaway JE, Keevil BG (2012) Therapeutic drug monitoring and LC-MS/MS. *J Chromatogr B Analyt Technol*
26 *Biomed Life Sci* 883-884:33–49. doi: 10.1016/j.jchromb.2011.09.041
27
28 25. Jin Y, Jang J-W, Lee M-H, Han C-H (2006) Development of ELISA and immunochromatographic assay for the
29 detection of neomycin. *Clin Chim Acta* 364:260–6.
30
31 26. Baietto L, D'Avolio A, De Rosa FG, et al. (2010) Development and validation of a simultaneous extraction
32 procedure for HPLC-MS quantification of daptomycin, amikacin, gentamicin, and rifampicin in human
33 plasma. *Anal Bioanal Chem* 396:791–8.
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4 **Figure Legends**
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7 **Fig. 1** Indirect competitive assay. (a) formation of the alkanethiol mixed SAM (b) immobilization of the
8 AK-BSA conjugate (c) Samples consisting of a mixture of a fixed antibody concentration and varying
9 analyte concentrations are incubated for 10 minutes and flowed over the sensing surface. Reutilization of
10 the surface is achieved by means of a regeneration solution
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13 **Fig. 2** Sensograms showing the interaction of free anti-AK fraction at different AK concentrations (1) and
14 the subsequent regeneration step with NaOH 45 mM (2) after signal stabilization
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17 **Fig. 3** Standard calibration curve for AK in PBS (black) and PBST 0.05% Tween (blue). Each point
18 corresponds to the mean value and standard deviation of triplicate measurements. In red, specificity study,
19 performed with vancomycin (Van) as analyte instead of AK, using anti-AK antibody under optimized
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Table 1: Immobilization of AK-BSA

[AK-BSA], ($\mu\text{g/mL}$)	Immobilization signal, (a.u)
5	0.33 ± 0.03^a
10	0.85 ± 0.03
15	0.97 ± 0.07
30	1.11 ± 0.05

^a: mean signal \pm standard deviation of at least 3 replicates

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Table 2. Amikacin immunoassay features

	LOD (IC₉₀)	IC₅₀	Dynamic range (IC₈₀-IC₂₀)	Slope
PBS	0.5 ng/mL (0.85 nM)	43.2 ng/mL (73.8 nM)	2.3– 809 ng/mL (3.9-1383 nM)	-0.41
PBST	0.1 ng/mL (0.17 nM)	1.3 ng/mL (2.3 nM)	0.3 – 5.4 ng/mL (0.51-9.2 nM)	-1.02

Figure 1

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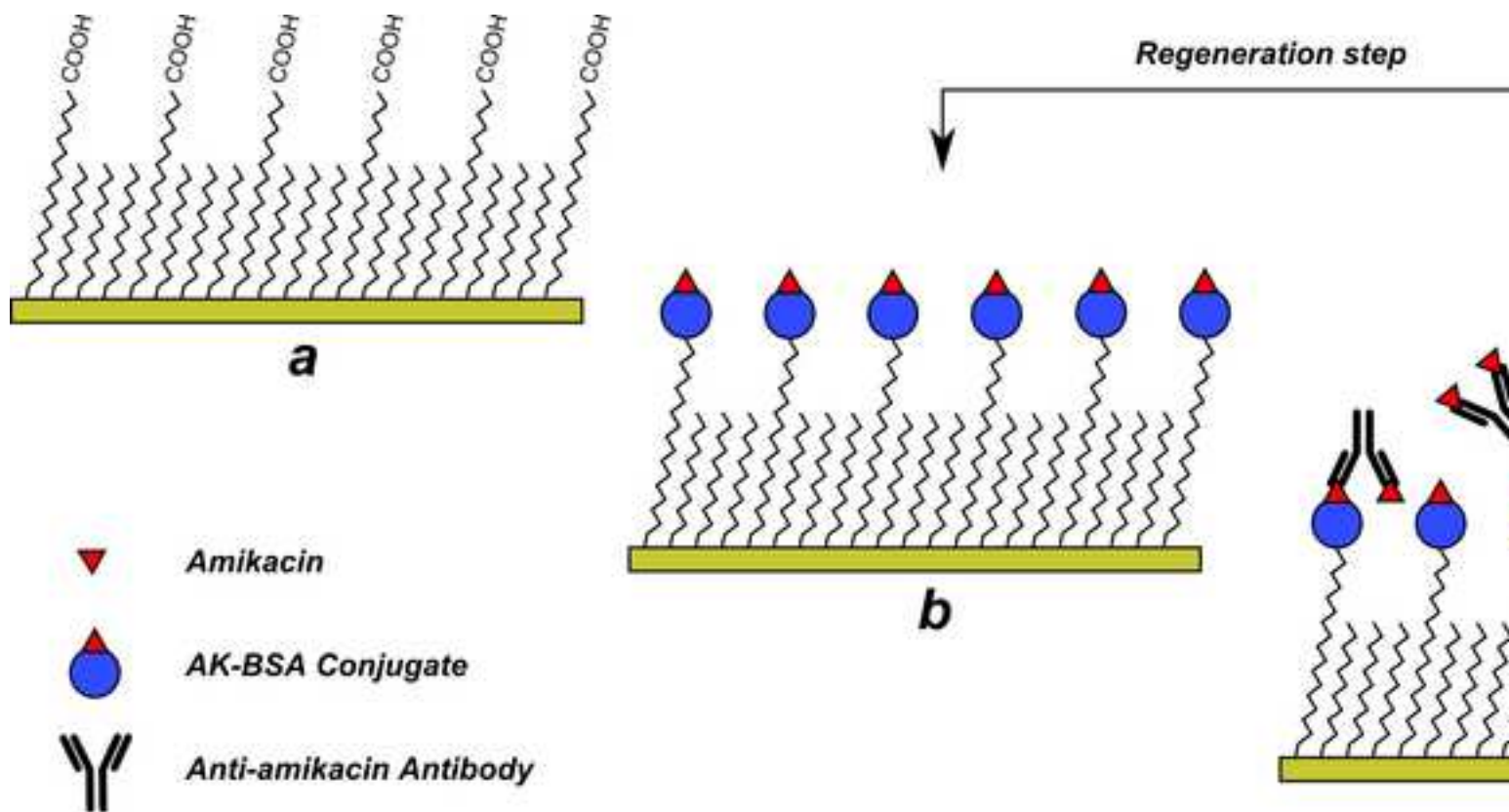


Figure 2

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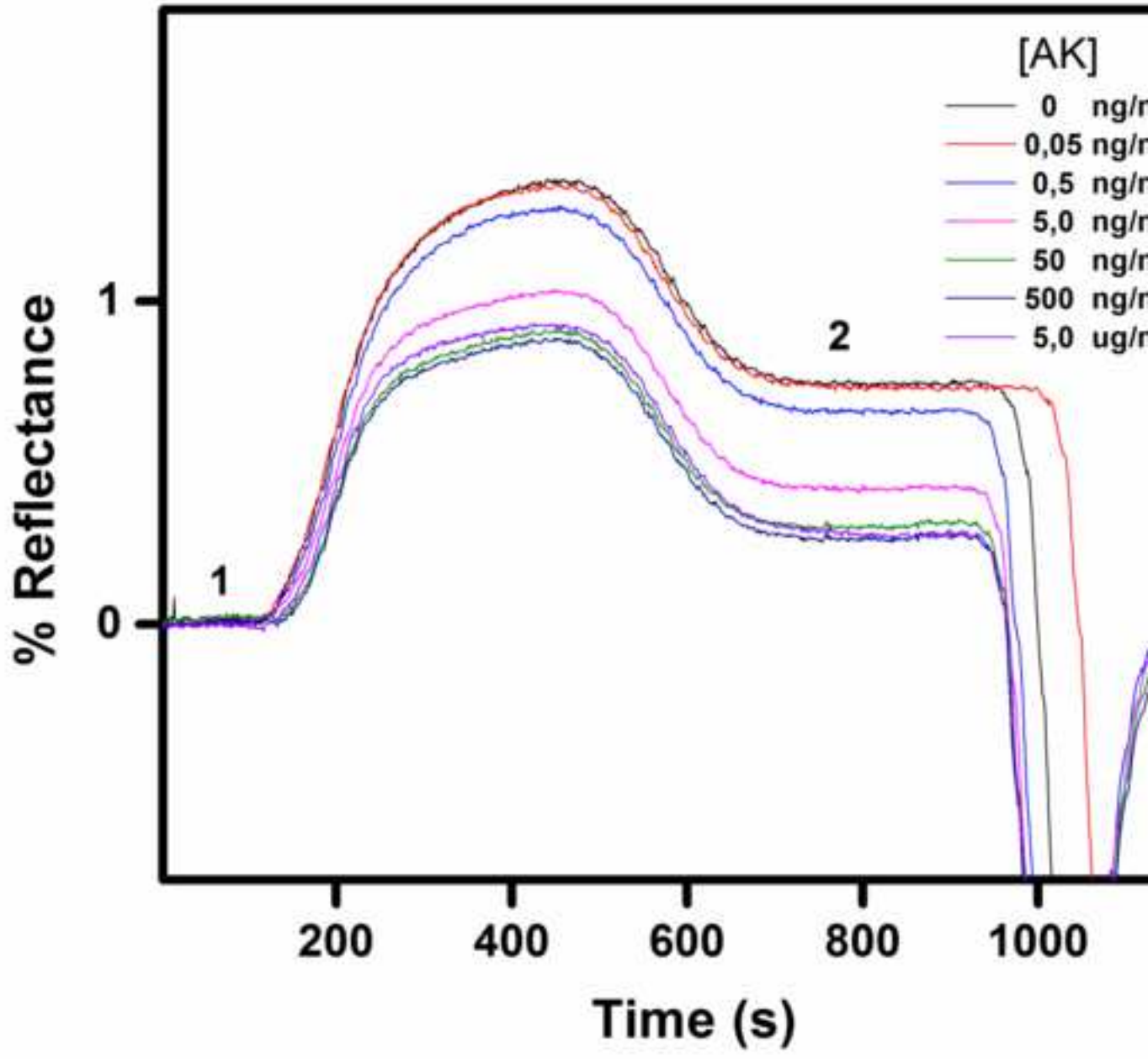


Figure 3

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