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1 MXene Nanoflakes Decorating ZnO Tetrapods for Enhanced 2 Performance of Skin-attachable Stretchable Enzymatic 3 Electrochemical Glucose Sensor

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14

15 **ABSTRACT:**

16 Continuous painless glucose monitoring is the greatest desire of more than 422 million diabetics
17 worldwide. Therefore, new non-invasive and convenient approaches to glucose monitoring are more
18 in demand than other tests microanalytical diagnostic tools. Besides, blood glucose detection can be
19 replaced by continuous glucose monitoring of other human biological fluids (e.g. sweat) collected
20 non-invasively. In this study, a skin-attachable and stretchable electrochemical enzymatic sensor
21 based on ZnO tetrapods (TPs) and a new class of 2D materials - transition metal carbides, known as
22 MXene, was developed and their electroanalytical behavior was tailored for continuous detection
23 glucose in sweat. The high specific area of ZnO TPs and superior electrical conductivity of MXene
24 ($\text{Ti}_3\text{C}_2\text{T}_x$) nanoflakes enabled to produce enzymatic electrochemical glucose biosensor with enhanced
25 sensitivity in sweat sample ($29 \mu\text{A mM}^{-1} \text{cm}^{-2}$), low limit of detection ($\text{LOD} \approx 17 \mu\text{M}$), broad linear
26 detection range ($\text{LDR} = 0.05 - 0.7 \text{ mM}$) that satisfies glucose detection application in human sweat,
27 and advanced mechanical stability (up to 30% stretching) of the template. The developed skin-
28 attachable stretchable electrochemical electrodes allowed to monitor the level of glucose in sweat
29 while sugar uptake and during physical activity. Continuous in vivo monitoring of glucose in sweat
30 obtained during 60 min correlated well with data collected by a conventional amperometric blood
31 glucometer in vitro mode. Our findings demonstrate the high potential of developed ZnO/MXene
32 skin-attachable stretchable sensors for biomedical applications on a daily basis.

33

34 **KEYWORDS:** MXene; ZnO tetrapods; glucose sensor; glucose oxidase; sweat glucose monitoring.

35

36 **1. INTRODUCTION**

37 With the increased demand for real-time monitoring of health-related small molecular weight
38 biomarkers, the portable medical device industry has grown rapidly over the past decade (Cucciniello
39 et al., 2021). These devices have an advantage over conventional medical detection tools since they
40 are user-friendly and require non-invasive biological fluids such as sweat, saliva, or tears (Chen et

1 al., 2017). With remote patient monitoring capabilities and non-invasiveness, portable medical
2 devices are good candidates for painless harvesting of physiologically relevant data in real-time
3 mode. For example, skin-attached, stretchable portable devices can meet the requirements of painless
4 and continuous glucose monitoring in diabetic patients, eliminating the daily inconvenience of finger-
5 pricking. While real-time glucose monitoring can minimize the risk of serious complications caused
6 by diabetes, and also can provide insights into body glucose dynamics, especially after mealtimes,
7 during physical activity and at rest.

8 Although many approaches to non-invasive glucose monitoring are currently proposed, portable and
9 wearable devices are still poorly implemented on a large industrial scale, despite their grown potential
10 and advantages (Lee et al., 2017). Such a slow implementation is mainly due to the unclear correlation
11 between sweat and blood glucose levels (Lee et al., 2017; Sempionatto et al., 2021). Nevertheless,
12 this drawback does not negate the fact that sweat glucose still can qualitatively indicate hypo- or
13 hyperglycemia and the necessity for self-care intervention to control and stabilize blood glucose
14 levels. Besides, the skin-attachable sweat glucose sensor performance can be improved by developing
15 novel and effective transducers. In this case, significant attention should be given to the materials and
16 composites with enhanced catalytic activity towards glucose oxidation and superior electrical
17 conductivity, particularly in complex matrices, i.e. blood or sweat (Zhou et al., 2019).

18 Among all known semiconductors, the nanoscale zinc oxide (ZnO) is considered to be one of the
19 most suitable for biomedical applications. A variety of morphologies, the enhanced surface to volume
20 ratio, high isoelectric point (IEP, ~ 9.5), high electron transport capacity, and biocompatibility are just
21 a few of its many superior physical-chemical properties (Myndrul et al., 2020; Pavlenko et al., 2020;
22 Tereshchenko et al., 2016). Today, ZnO still meets scientists' expectations and is increasingly used in
23 a variety of detection techniques, such as optical (Myndrul et al., 2021), electronic (Wang et al., 2019),
24 SPR (Kuranaga et al., 2020), SERS (Y. Zhai et al., 2019), and piezoelectric (bio)sensors (Zhao et al.,
25 2015). Among ZnO nanostructures, ZnO tetrapods (ZnO TPs) occupy an important position on
26 biosensing development because of their higher active surface area provided by the large pods and
27 highly porous 3D interconnected networks, which provide more sites for analyte adsorption and, as a
28 consequence, better sensitivity to small molecular weight compounds detection (Lei et al., 2012a;
29 Mishra and Adelung, 2018a). It was recently reported that the high catalytic activity of ZnO TPs
30 towards the sulfonamide antibacterial degradation, which was 2-3 times higher when compared with
31 ZnO nanoparticles or nanoflowers (Yi et al., 2018). The enhanced surface to volume ratio and high
32 catalytic activity were the reasons for the higher sensitivity of TP ZnO to L-lactic acid than that of
33 ZnO nanorods with similar analytical performance (Lei et al., 2012a; Zhao et al., 2014). However,
34 the high electrical resistivity of ZnO TPs is a significant drawback for electrochemical sensing
35 applications. Yet, there is still a possibility to improve the electrochemical sensor performance by
36 integrating ZnO TPs with high conductive 2D nanomaterials (Theerthagiri et al., 2019).

37 Although the history of 2D materials is short and has not even crossed the 20-year mark (since the
38 synthesis of graphene in 2004) (Novoselov et al., 2004), have emerged in a wide variety of scientific
39 areas. 2D materials have become popular due to the enhanced optoelectronic properties that differ
40 from those of conventional bulk materials (Kang et al., 2019). Until now, particular attention has been
41 paid to the high catalytic activity of those atomically thin structures driven by enhanced surface area
42 and carrier transport. In this context, 2D materials are used to enhance the water splitting and CO₂
43 reduction performances (Fan et al., 2021). while high stability and biocompatibility make them
44 attractive for medical and biological applications (Tao et al., 2019). For example, biosensors based
45 on 2D sheets of graphene, MoS₂, and MXene (transition metal carbides, e.g. Ti₃C₂T_x)
46 (VahidMohammadi et al., 2021) demonstrate enhanced sensitivity towards proteins, bacteria, H₂O₂,
47 and glucose detection.(Zeng et al., 2021) However, the low electrical conductivity of MoS₂, high
48 hydrophobicity of both graphene and MoS₂, and laborious functionalization can complicate the
49 biosensing performance.(Kalambate et al., 2019) In contrast, MXene has advantages over graphene
50 and MoS₂ since it possesses outstanding metal-like conductivity and hydrophilic surface with
51 termination groups ($-OH$, $=O$ and $-F$), facilitating the functionalization process. Composites based

1 on MXene and ZnO exhibit unique properties that enable application in various applications,
2 including solar cells (Hou and Yu, 2021), water splitting (Sreedhar and Noh, 2021), electromagnetic
3 wave absorption (Qian et al., 2017), photocatalytic performance (Khadidja et al., 2021), light-emitting
4 diodes (Lu et al., 2020), and sensing (Yang et al., 2021). Moreover, due to the enhanced catalytic
5 activity of ZnO/MXene composites (Lv et al., 2021), it may be an ideal candidate for biosensor
6 applications, especially for electrochemical glucose detection.

7 In this report, catalytically active composites of ZnO TPs/MXene were prepared and deposited on a
8 stretchable electrode as an electroactive transducer layer for the qualitative analysis of glucose in
9 Phosphate-buffered saline (PBS), artificial and human sweat. To the best of our knowledge, this is the
10 first proposed protocol for decorating ZnO nanostructures with MXene nanoflakes and not vice versa.
11 The structural, chemical, electrochemical and optical properties of produced composites were
12 analyzed. ZnO TPs/MXene/GOx nanocomposites exhibited enhanced catalytic activity towards
13 glucose oxidation in PBS and artificial sweat compared to the pristine ZnO TPs and MXene. The
14 higher catalytic activity of ZnO TPs/MXene resulted in a better sensitivity and a lower limit of
15 detection (LOD) for glucose in PBS and artificial sweat. Moreover, the low applied negative potential
16 of -0.24 V allowed to avoid the impact of the interference species that could contribute to the response
17 of ZnO TPs/MXene/GOx-based electrode. The ZnO TPs/MXene stretching experiments have shown
18 the high mechanical stability (up to 30% of strain) of developed electrodes. We also performed in
19 vivo measurements of glucose in sweat, and we demonstrated the good correlation of obtained data
20 with one collected by a conventional amperometric blood glucometer.

21

22 **2. EXPERIMENTAL SECTION**

23 **2.1. Materials.**

24 (3-Aminopropyl)triethoxysilane (APTES, purity $\geq 98\%$) glucose oxidase (GOx) (№ G7141), D-(+)-
25 Glucose (№ 8270), Nafion 117 (~5%) containing solution (№ 70160), Glutaraldehyde (GA, 50 wt. %
26 in H₂O), Sodium hydroxide (NaOH, grade $\geq 98\%$), Zn powder (№ 324930, purity of 99,9%), ascorbic
27 acid (AA) (№ PHR1008), urea (purity $\geq 99\%$), uric acid (UA, purity $\geq 99\%$, crystalline) dopamine
28 (№ H8502), sodium chloride (NaCl, grade $\geq 99\%$), potassium chloride (KCl) (№ P3911), glycine
29 (purity $\geq 99\%$), Ammonium hydroxide solution (NH₄OH) (№ 221228), were purchased from Sigma
30 Aldrich. PBS, pH 7.4 was used as the diluent for glucose oxidase and glucose dissolution.
31 Polyethylene terephthalate (PET) was provided by Policrom Screens S.P.A. (Carvico, BG, Italy).
32 Silver reference electrodes and contacts on all produced electrodes were printed using LOCTITE®
33 ECI 1010 E&C (Henkel); working and reference electrodes were printed using LOCTITE® EDAG
34 407C E&C (Henkel). Lithium fluoride (LiF, 98.5% grade) powder and hydrochloric acid (HCl, 37
35 wt. %) were purchased from Alfa Aesar.

36

37 **2.2. Instruments.**

38 Morphologies of the ZnO TPs, MXene and ZnO TPs/MXene were investigated by scanning electron
39 microscopy (SEM) (JEOL, JSM7001F) with a dispersive energy X-ray (EDX) analyzer and
40 transmission electron microscopy (TEM) (JEOL ARM 200F) high-resolution transmission electron
41 microscope (200 kV) with an EDX analyzer. The structural state of samples was analyzed by means
42 of X-ray diffraction (XRD) (PANalytical, X'pert3pro MRD diffractometer) working with a Cu lamp
43 ($\lambda = 1.5418 \text{ \AA}$). Raman spectrum was measured by means of a Renishaw micro-Raman spectrometer
44 with a confocal microscope. The samples' optical properties (absorbance, photoluminescence) have
45 been studied with Ocean Optics spectrophotometer QE65pro. Electrochemical measurement and
46 glucose detection, including cycling voltammetry and chronoamperometry regimes, were performed
47 using potentiostats ER466 and GAMRY 620. The sonication of the MXene sheets to divide them into
48 MXene nanoflakes was performed by Sonifier ultrasonic processor (SFX 20:0.55).

2.3. Preparation of ZnO TPs.

ZnO tetrapods (ZnO TPs) were obtained by the simple catalyst-free oxidative-metal-vapour-transport method. The method was based on thermal evaporation of Zn powder at 1000°C for 1h in the air in a ceramic crucible (Tawale et al., 2010). After the reaction, the white powder of ZnO was deposited, which confirmed the oxidation of zinc metal powder.

2.4. Preparation of MXene.

Ti₃C₂T_x MXene was prepared by etching Ti₃AlC₂ (MAX-phase) ternary carbide using MILD method in a solution of lithium fluoride (LiF) in hydrochloric (HCl) acid (Alhabebe et al., 2017). Etching mixture was prepared from 40 mL of 12 M HCl (37%), 10 mL of DI-water and then 3.2 g of LiF are dissolved in this solution. The mixture was placed in a plastic container with a volume of 50 mL. Then 2 g of Ti₃AlC₂ powder with particle sizes of less than 40 μm is gradually added into etching solution under continuous stirring for 24 hours at 25°C. Then MXene slurry was rinsed with DI-water via repetitive centrifugation. As-prepared MXene slurry is further processed to obtain a colloidal solution of separated MXene flakes using mild delamination procedure assisted by intercalation of Li⁺ ion between Ti₃C₂ sheets with following the separation of MXene into colloidal solution in water (Alhabebe et al., 2017). We used the following protocol for the delamination: 2 g of lithium chloride (LiCl) were added to 40 mL of DI-water in 50 mL plastic container. Etched MXene slurry is added to the prepared solution. The process is performed at 35 °C for 24 hours under stirring. After that, MXene slurry is rinsed via repetitive cycles of centrifuging. Purity of MXene is confirmed by XRD analysis. After synthesis MXene sediment is stored at -18°C.

2.5. Fabrication of skin-attachable, stretchable electrodes.

The skin-attachable electrodes made of Silver/Silver-NaCl/Carbon in PTU (125 μm thickness) were produced by a conventional screen printing method, property of Eastprint-USA (<http://www.eastprint.com/electrodes-biosensors.html>) under design property of Biosensor srl (<https://www.biosensor-srl.eu/>).

2.6. Preparation of ZnO/MXene based electrode.

The ready-to-use MXene sheets were then placed in a PBS solution and sonicated (for 20 minutes at 70% ultrasonic processor power) to crush into MXene nanoflakes for use in the ZnO TPs decorating process. Once ready, MXene nanoflakes were then mixed with ZnO TPs in proportion 1:2 (1 mg/ml MXene and 2 mg/ml ZnO TPs in DI water) to prepare combined ZnO TPs/MXene structure. The sample was then investigated on SEM in order to confirm the coverage of ZnO TPs by MXene nanoflakes, and the result showed the inhomogeneity of the coverage (SI, Fig. 1S). In order to improve the uniformity, ZnO TPs were treated with 2% ATPES solution (in anhydrous EtOH 99.6%) to achieve functional groups on their surfaces. This process was followed by 3-times rinsing of ZnO TPs (2 times in EtOH and 1 time in PBS) to eliminate APTES remains. The functionalized ZnO TPs were then mixed again with MXene in 1:2 (1 mg/ml MXene and 2 mg/ml ZnO TPs in DI water) to prepare combined ZnO TPs/MXene structure. The process of mixing was performed under sonicating conditions (for 2 hours at RT). After the sonication, the sample was rinsed in DI water and centrifuged 3 times to remove unbound MXene nanoflakes and remove excess water.

The prepared ZnO TPs/MXene sample was mixed with Nafion (1 mL of 2% EtOH solution, neutralized to pH = 7, by use of 0.1 M NaOH). The droplet (8 μL) of this mixture was placed on the working electrode of the stretchable sensor and dried for 1 h at 65 °C. Once cooled, the sensor's working electrode was then cross-linked by GA (30 min, RT) and GOx was added (8μL of 20 mg/mL in PBS, for 12 hours at 4°C) to achieve a selective layer toward glucose. After that, the skin-attachable and stretchable sensor were coated by the additional layer of Nafion polymer (5 μL of 2 % solution in EtOH neutralized to pH = 7) and stored for 1 day at 4°C before use. The sketch diagram (SI, Fig. 2S) depicts MXene nanoflakes and ZnO TPs fabrication stages, the deposition of ZnO TPs/MXene

1 composite on the skin-attachable, stretchable electrode and selective layer formation using GOx and
2 Nafion.

3 For the comparative analysis of the electrochemical and structural properties and the glucose
4 detection parameters, three types of electrodes were prepared and titled as ZnO TPs/GOx-based
5 electrode, MXene/GOx-based electrode, and ZnO TPs/MXene/GOx-based electrode. The stability of
6 all prepared electrodes was tested in different pH.

7

8 **2.7. Electrochemical glucose detection in PBS and artificial sweat.**

9 The fabricated electrodes were initially tested in PBS (pH 7.4) with a change of glucose concentration
10 in the experimental cell. The electrochemical properties of ZnO TPs/GOx-, MXene/GOx-, and ZnO
11 TPs/MXene/GOx-based electrodes were studied using CV measurements at scan rate of 40 mV/s.

12 Chronoamperometry was performed to determine glucose using ZnO TPs-based electrode with an
13 applied potential of -0.43 V (*vs* Ag), MXene/GOx-based electrode with applied a potential of -0.21
14 V, and for the ZnO TPs/MXene-based electrode the applied potential was equal to -0.24 V.
15 Chronoamperometry measurements for the ZnO TPs/MXene/GOx-based electrode were carried out
16 in artificial sweat (pH 6.5) at the same applied potential (-0.24 V) as for PBS. Artificial sweat was
17 prepared according to the protocol EN1811:2012 and consisted of NaCl (0.5%), KCl (0.1%), and urea
18 (0.1%), while the pH was adjusted to 6.5 by use of NH₄OH (Meng et al., 2021). The selectivity of
19 ZnO TPs/MXene-based electrode to glucose in artificial sweat was evaluated by means of
20 chronoamperometry with the sequential addition of 0.1 mM AA, 0.5 mM urea, 50 μM UA, 0.05 mM
21 dopamine, 10 mM NaCl, 10 mM KCl, and 0.1 mM glycine.

22 All experiments were repeated at least three times and all parameters (LODs, sensitivities, electron
23 transfer rates, etc.) were calculated as means±standard error.

24 **2.8. On-body sweat analysis**

25 On-body sweat analysis was performed in a healthy subject aged 30 years. During the real-time
26 glucose monitoring, the electrode was tightly attached to the wet skin, the contacts were connected to
27 the potentiostat, and the data were displayed on a laptop screen. Chronoamperometry measurements
28 were carried out at the same applied potential (-0.24 V) as it for artificial sweat. The volunteer was
29 asked to take sweets and do exercises on a command. Additional invasive glucose measurements (8
30 discrete points) were done using a commercially available glucometer (VGM01, VivaCheck Ino).

31 **2.9. Electrode stretching tests.**

32 Stretchability was tested using a lab-made device capable of stretching the firmly attached electrode.
33 The discrete values of the applied strain varied from 0 to 35%. The response was measured by
34 potentiostat operating in amperometric mode. The 3-electrode system was sandwiched between slides
35 to ensure a constant analyte value during experiments. Various glucose concentrations (0.1 mM, 0.2
36 mM, 0.5 mM, 0.7 mM, and 1 mM) were tested on the electrodes with 0%, 15%, and 30% elongation
37 to evaluate the stability of the electrodes under various strains.

38 **3. RESULTS AND DISCUSSION**

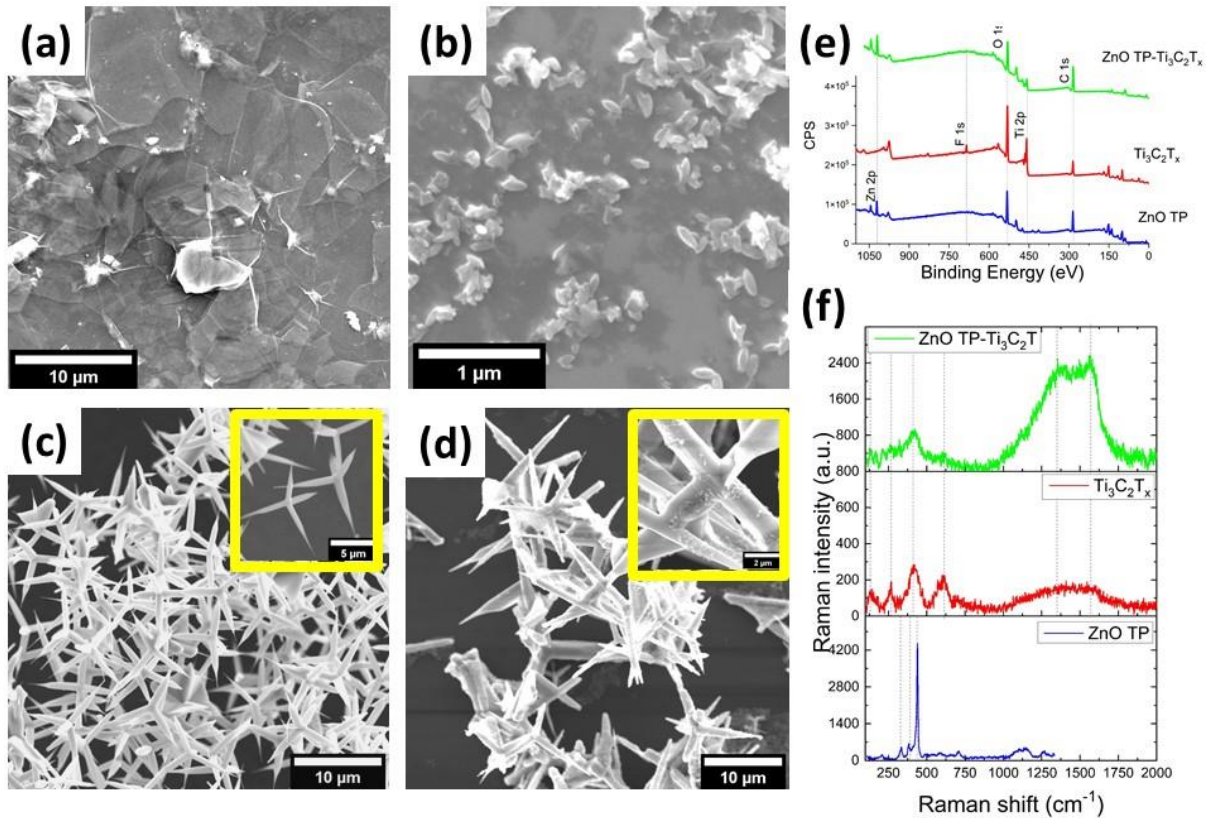
39 **3.1. Choice of Materials**

40 As mentioned in the “Introduction” section, ZnO is one of the widely used materials in sensor and
41 biosensor technologies, and such popularity is due to its high stability, biocompatibility, and enhanced
42 surface to volume ratio (when considering hierarchical structures). Since the stability of the
43 transducer, as well as its biocompatibility, are the most important parameters during the
44 electrochemical biosensor performance, ZnO was considered the most suitable material to be used in
45 our experiments. ZnO in the shape of tetrapods (ZnO TPs) has the highest active surface when
46 compared with the other ZnO morphologies, and therefore, the use of ZnO TPs can enhance the

1 efficiency of detection performance (Lei et al., 2012b; Mishra and Adelung, 2018b). The choice of
 2 MXene nanoflakes for ZnO decoration was based on the assumption that MXene would facilitate the
 3 charge transfer rate between the electrolyte and ZnO, as it was reported for the ZnO nanorods and
 4 MXene sheets (Liu and Chen, 2020). The enhanced charge transfer rate usually results in a better
 5 catalytic activity of the transducer towards glucose oxidation (Zhou et al., 2019).

6

7 3.2. Characterization of ZnO TPs/MXene nanocomposite.



8

9 **Figure 1.** SEM images of (a) MXene, (b) MXene nanoflakes, (c) ZnO TPs and (d) ZnO/MXene deposited on the Si
 10 substrate; (e) survey XPS spectra and (f) Raman spectra of produced samples (ZnO TPs, MXene nanoflakes and
 11 ZnO/MXene).

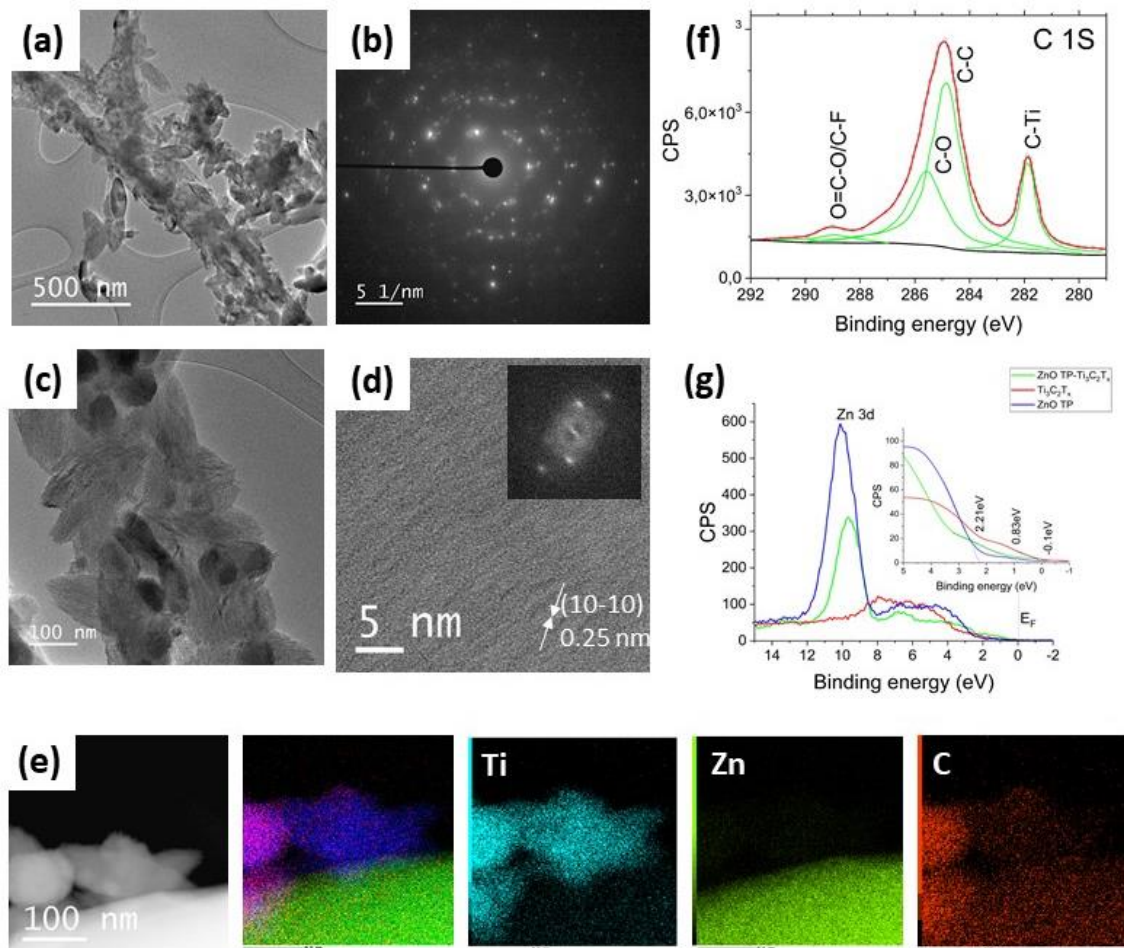
12

13 MXene $Ti_3C_2T_x$ flakes and then nanoflakes were successfully synthesized from a titanium aluminum
 14 carbide (Ti_3AlC_2) MAX phase using a method described in the Experimental section. Fig. 1a, b show
 15 SEM images of as-synthesized and nanoflakes of MXene ($Ti_3C_2T_x$), respectively. The lateral size of
 16 the delaminated MXene flakes is approximately 7-8 μm , whereas the nanoflakes have a leaf-shaped
 17 morphology with an average diameter around 100 nm, indicating the use of a high-power ultrasonic
 18 treatment may induce structural changes. Fig. 1c indicates that as-synthesized ZnO TPs are well
 19 faceted and consist of four "legs" connected to a central nucleus with an average length of $8 \pm 2 \mu m$
 20 and a diameter of $400 \pm 150 nm$ (SI, Fig. S3). ZnO TPs have hexagonal morphology, which suggests
 21 a single crystalline structure with preferred growth direction along the c-axis of the hexagonal unit
 22 cell. The zeta-potential values for ZnO TPs and MXene nanoflakes were $35 \pm 8 mV$ and -27 ± 1 ,
 23 respectively, which provides a strong electrostatic interaction between those two components.
 24 Despite this, it was necessary to carry out an additional chemical treatment (APTES treatment, see
 25 Experimental section) to obtain a conformal layer of MXene nanoflakes over ZnO TPs (Fig. 1d).
 26 Otherwise, an uneven distribution of MXene nanoflakes on the ZnO surface was observed (SI, Fig.
 27 S1).

1 To analyze the chemical composition and electronic states of produced nanocomposites, XPS analysis
2 and Raman spectroscopy were applied. The survey spectra for ZnO TPs, MXene nanoflakes and
3 ZnO/MXene nanocomposites are displayed in Fig. 1e. The spectrum of the ZnO TPs sample displays
4 the signals of Zn, O and C. The XPS analysis of core level Zn 2p spectra (the difference Zn 2p_{3/2} - Zn
5 2p_{1/2} = 23.1 eV) indicated the wurzite phase of ZnO (SI, Fig. S4) (Iatsunskyi et al., 2017). Four main
6 elements (oxygen (O), titanium (Ti), carbon (C) and fluorine (F)) were presented in the MXene
7 samples confirming the formation of Ti₃C₂T_x with functional surface groups of -OH and -F. After
8 the formation of ZnO/MXene nanocomposite the peak relating to the F element vanished, indicating
9 the formation of the interface between ZnO and MXene-Ti₃C₂. The detailed XPS analysis of the
10 fabricated ZnO/MXene nanocomposite is provided below and in the supporting information.

11 Fig. 1f shows the Raman spectra of produced ZnO/MXene nanocomposites and pristine components.
12 As-synthesized ZnO TPs exhibits Raman peaks at 440, 390, and 338 cm⁻¹ corresponding to E₂(high),
13 E₁(TO), A₁(TO) modes of ZnO wurzite phase crystal, respectively (Zhang et al., 2009). The Raman
14 spectrum of pristine Ti₃C₂T_x nanoflakes consists of a minimum of five vibrational modes: 125, 255,
15 425, 608, and 1520 cm⁻¹. Recently, it was shown that a Raman spectrum of Ti₃C₂T_x consists of many
16 peaks in the 100-800 cm⁻¹ range, and it can be distinguished into several spectrum regions (Sarycheva
17 and Gogotsi, 2020). According to the Sarycheva *et al.* (Sarycheva and Gogotsi, 2020), one may
18 conclude that the Raman mode at 125 cm⁻¹ is attributed to the resonant plasmonic peak. Two Raman
19 modes at 255 and 425 cm⁻¹ are associated with out-plane (A_{1g}) and in-plane (E_g) vibrations of C, Ti
20 atoms and surface groups (-OH, -F). The peak at 608 cm⁻¹ indicates the carbon vibration (E_g, A_{1g}).
21 The broad Raman peak at 1520 cm⁻¹ is attributed to the D and G bands of amorphous and graphitic
22 carbon. Peaks relating to anatase and rutile phases (140, 512 cm⁻¹ and 393, 587 cm⁻¹) were not
23 observed, indicating the presence only -OH group on the surface of MXene nanoflakes. As for
24 ZnO/MXene nanocomposites, one can observe the changes of Raman spectrum. Apart from graphitic
25 carbon bands, there are four weak peaks at 133, 170, 213, and 414 cm⁻¹. It is clearly seen that the
26 resonance peak shifted to higher wavenumbers. On the other hand, Raman modes associated with in-
27 plane vibrations of functional groups shifted to the red region. These shifts of both modes and the
28 high-intense D band can be explained by an increased concentration of defects that are formed during
29 the production of ZnO/MXene nanocomposites (Sarycheva and Gogotsi, 2020).

30



1

2

3 **Figure 2.** (a-d) TEM and SAED images of ZnO/MXene nanocomposites; (e) EDX element map scanning images of the
 4 interface between ZnO and MXene nanoflakes; (f) the carbon core-level XPS spectrum of ZnO/MXene nanocomposites;
 5 (g) Valence-band XPS spectra of ZnO TPs, MXene nanoflakes and ZnO/MXene nanocomposites.

6

7 The detailed structural and chemical properties of ZnO/MXene nanocomposites were analyzed using
 8 TEM and XPS (including Valence Band XPS). Fig. 2a demonstrates the separate "leg" of ZnO TPs
 9 covered by MXene $\text{Ti}_3\text{C}_2\text{T}_x$ nanoflakes. The uniform distribution of nanoflakes is clearly seen over
 10 the whole surface of ZnO (Fig. 2c, SI, Fig. S5). The corresponding selected area electron diffraction
 11 (SAED) pattern demonstrates the polycrystalline nature of ZnO/MXene nanocomposites (Fig. 2b).
 12 Fig. 2d shows the high-resolution TEM image of the separate MXene nanoflakes. The lattice
 13 parameter of the $\text{Ti}_3\text{C}_2\text{T}_x$ corresponding to $(10\bar{1}0)$ planes is measured to be about 0.255 nm. EDX
 14 elemental mapping images for the interface of ZnO- $\text{Ti}_3\text{C}_2\text{T}_x$ are shown in Fig. 2e. The uniform
 15 distribution of zinc, titanium, and carbon confirms the formation of the interface between the ZnO
 16 surface and $\text{Ti}_3\text{C}_2\text{T}_x$ MXene nanoflakes.

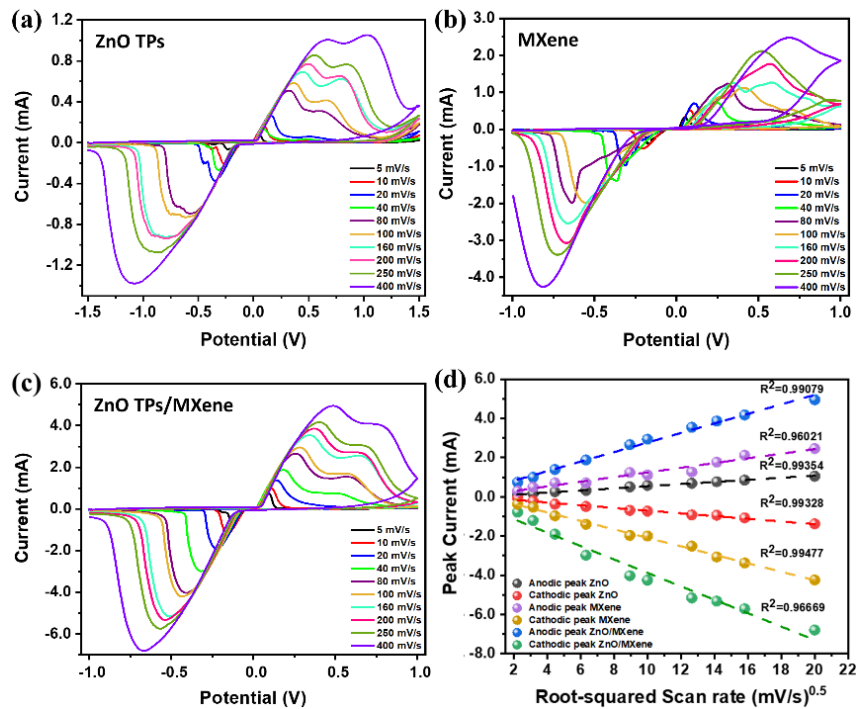
17 The high-resolution XPS spectra of C 1s and other elements for ZnO/MXene nanocomposites were
 18 analyzed (Fig. 2f; SI, Fig. S6). The core level C 1s peak was fitted with four main components at
 19 281.9, 284.8, 285.6, and 289.0 eV, which were attributed to C-Ti, C-C, C-O and C=O-O/C-F,
 20 respectively (Näslund et al., 2020). O 1s deconvoluted spectra (SI, Fig. S6) displays three main
 21 components at 529.7, 531.5, and 532.9 eV corresponding to O^{2-} , -OH surface groups and absorbed
 22 water, respectively (Näslund et al., 2020). Recently, it was shown that Ti-O bonds are the most stable
 23 among the terminated groups of $\text{Ti}_3\text{C}_2\text{T}_x$ MXene (Liu and Li, 2018). One may conclude that most
 24 terminated groups after the formation of ZnO/MXene nanocomposites are Ti-OH bonds, which was
 25 also confirmed by the high-resolution XPS Ti 2p peak (SI, Fig. S6).

1 One may suggest, both chemical and physical interactions between ZnO surface and MXene may
 2 lead to changes in the energy bands of obtained nanocomposites. To study this, the valence band (VB)
 3 XPS analysis was used, which can provide valuable information about the modification of the
 4 ZnO/MXene electronic structure. Fig. 2g demonstrates the VB spectra for ZnO TP, $\text{Ti}_3\text{C}_2\text{T}_x$
 5 nanoflakes and ZnO/MXene nanocomposites. Typically, the VB spectrum determines the density of
 6 states (DOS) below the Fermi level. In the case of ZnO TPs samples, it consists of two
 7 components/bands at about 10 eV and 5 eV corresponding to Zn 3d and O 2p orbitals, respectively
 8 (Iatsunskiy et al., 2017). The broad VB XPS peak at 8-3 eV for pristine $\text{Ti}_3\text{C}_2\text{T}_x$ MXene may be
 9 associated with the C 2s – Ti 3d hybridization region and the terminated groups on the surface
 10 (Magnuson et al., 2018). The VB maximum (VBM) for those samples was evaluated using the
 11 standard method as previously described (Iatsunskiy et al., 2021). The VBM values were estimated
 12 to be 2.2 eV below and 0.1 eV above the Fermi level (E_F) for ZnO TPs and MXene, respectively. A
 13 high DOS above E_F is correlated to the low resistivity of $\text{Ti}_3\text{C}_2\text{T}_x$ MXene. After the formation of
 14 ZnO/MXene nanocomposite, one may observe the shift to the lower energies of Zn 3d peak,
 15 confirming the chemical interaction of MXene nanoflakes with a ZnO surface. In addition, the
 16 estimated value of VBM for the ZnO/MXene nanocomposite becomes significantly lower (0.83 eV)
 17 in comparison to as-prepared ZnO TPs. It may be suggested the decreasing of the overall resistivity
 18 what must provide the enhanced electrocatalytic efficiency. Besides, according to the "d-band center
 19 theory", the modification of VBM can optimize the catalytic ability for the produced nanocomposites
 20 (Hammer and Nørskov, 2000).

21

22 3.3. Electrochemical properties of the fabricated electrodes.

23



24

25

26 **Figure 3.** (a) CV curves of ZnO TPs/GOx-based electrode, (b) MXene/GOx-based electrode, (c) ZnO TPs/MXene/GOx
 27 electrodes-based electrode scanned in 0.3 mM of glucose in PBS (pH = 7.4) at different scan rates from 5 to 400 mV/s.
 28 (d) The calibration plots of the anodic and cathodic peak currents vs the root-squared scan rate (for ZnO/GOx-,
 29 MXene/GOx-, and ZnO/MXene/GOx-based electrodes).

30

1 In order to study the electrochemical behavior of prepared skin-attachable electrodes based on ZnO
 2 TPs/GOx, MXene/GOx, and ZnO TPs/MXene/GOx, the CV curves were recorded at different scan
 3 rates, and obtained plots are shown in Fig. 3. The CV of ZnO TPs/GOx-based electrode (Fig. 3a)
 4 indicates that the peak current increases as the scan rate rises, pointing to the quasi-reversible redox
 5 reaction (Hu et al., 2011). The quasi-reversibility was also observed for electrodes based on
 6 MXene/GOx (Fig. 3b) and ZnO TPs/MXene/GOx (Fig. 3c). Fig. 3d shows that the peak current vs.
 7 the square root of scan rate is linear (R^2 vary from 0.96021 to 0.99477) for all samples, which
 8 indicates a typical diffusion-controlled electrochemical behavior (Tian et al., 2015). In addition, the
 9 electron transfer rates between ZnO TPs-, MXene-, and ZnO TPs/MXene-based electrodes and
 10 immobilized GOx were calculated according to the Laviron formulation using the following
 11 equations (Laviron, 1979):

$$12 \quad E_{p,c} = E^{o'} - \frac{2.3RT}{\alpha nF} \log \left[\frac{\alpha nFv}{RTk_s} \right] (1)$$

$$13 \quad E_{p,a} = E^{o'} - \frac{2.3RT}{(1-\alpha)nF} \log \left[\frac{(1-\alpha)nFv}{RTk_s} \right] (2)$$

14 where $E_{p,a}$ is the potential of the anodic peak, $E_{p,c}$ is the potential of the cathodic peak, $E^{o'}$ is the
 15 standard surface potential calculated by averaging the anodic and cathodic potentials at slow scan
 16 rates, v is the scan rate, α is the electron-transfer coefficient, k_s is the electron-exchange rate constant,
 17 R is the ideal gas constant, T is the absolute temperature (K), F is the Faraday constant, and n is the
 18 number of electrons transferred. Electron transfer parameters (α , and k_s) were determined by
 19 plotting E_p vs. the log of the scan rate (Laviron plots) (Oztekin et al., 2011). Using the slopes of linear
 20 portions of the Laviron curves (not shown here) for the ZnO TPs/GOx-, MXene/GOx-, and ZnO
 21 TP/MXene/GOx-based electrodes, the electron transfer constants were calculated as 0.46 ± 0.09 , 0.67
 22 ± 0.11 , and 0.4 ± 0.04 , respectively, while the electron-exchange rate constants were determined equal
 23 to $0.48 \pm 0.03 \text{ s}^{-1}$, $0.36 \pm 0.08 \text{ s}^{-1}$, $1.01 \pm 0.11 \text{ s}^{-1}$, respectively. The analysis of the obtained k_s values
 24 shows that ZnO TP decorating by MXene facilitate the transfer of electrons between the GOx and the
 25 electrode. Generally, the higher value of electron-exchange rate indicates the lower resistivity of
 26 electrode modifier. The electrochemical impedance spectroscopy (EIS) confirmed the lower
 27 resistivity of ZnO TP/MXene in comparison to the pure ZnO TP electrode (SI, Fig. S7).

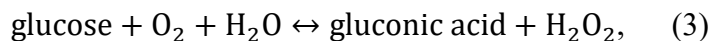
28 The above results comply well with the recently reported results for $\text{Ti}_3\text{C}_2/\text{ZnO}$ composites, where
 29 the interaction between ZnO and Ti_3C_2 (MXene) facilitates the electron transport (Wang et al., 2016).
 30 Authors also have shown that the rate capacity, in turn, was directly related to the enhanced surface
 31 to volume ratio of $\text{Ti}_3\text{C}_2/\text{ZnO}$ and an increase in the number of three-dimensional channels for the
 32 fast electron transport and the fast ion diffusion (Wang et al., 2016). Moreover, another 2D conductive
 33 material, like graphene, was already used to decorate ZnO nanomaterials, which led to a 3-fold
 34 increase in catalytic activity due to an increase in the electron transfer rate by 3-4 times as compared
 35 to a bare ZnO-based electrode (Xu et al., 2011). Thus, on the basis of the foregoing, it was suggested
 36 that an increase (2-3 times) in the electron exchange rate for the ZnO TPs/GOx -based electrode may
 37 lead to its better catalytic activity towards glucose oxidation. Additional experiments were also
 38 carried out to confirm

39

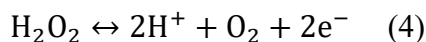
40 **3.4. CV-based sensor performance.**

41 CV-based measurements were performed to test the catalytic oxidation of glucose on GOx produced
 42 electrodes. For this purpose, various glucose concentrations were sequentially added to the
 43 electrochemical cell containing PBS with ZnO TPs/GOx-, MXene/GOx-, and ZnO TPs/MXene/GOx-
 44 based electrodes (one type of electrode per experiment). CV curves were measured in PBS solution
 45 under stirring conditions to achieve a homogeneous glucose distribution in the electrochemical cell.
 46 Depending on the electrode, the range of the recorded potential varied from -1 V to 1 V for ZnO
 47 TPs/GOx- and ZnO TPs/MXene/GOx-based electrodes (SI, Fig. S8a, b), and -0.6 to 0.6 V for
 48 MXene/GOx-based electrodes (SI, Fig. S8c), and the scanning rate was chosen equal to 40 mV/s. Fig.
 49 SI, Fig. S8d demonstrates the peak current vs the natural logarithm of the glucose concentrations

1 (sensitivity: $0.54 \pm 0.09 \text{ mA mM}^{-1} \text{ cm}^{-2}$, measured in the CV mode for ZnO TPs/MXene/GOx-based
 2 electrode within the physiological range of blood glucose (0 to 20 mM). The sensitivities for ZnO
 3 TPs/GOx-, and MXene/GOx-based electrodes were calculated as $0.45 \pm 0.06 \text{ mA mM}^{-1} \text{ cm}^{-2}$, and
 4 $0.43 \pm 0.07 \text{ mA mM}^{-1} \text{ cm}^{-2}$ indicating the lower glucose sensitivity in comparison to ZnO
 5 TPs/MXene/GOx-based electrode. It is seen that the cathodic and anodic currents gradually increase
 6 with the glucose concentration, indicating the glucose electrocatalytic oxidation behavior for ZnO
 7 TPs/GOx-, MXene/GOx-, and ZnO TPs/MXene/GOx-based electrodes. The typical reactions
 8 occurring at the working electrode can be described as following (Kim et al., 2014):

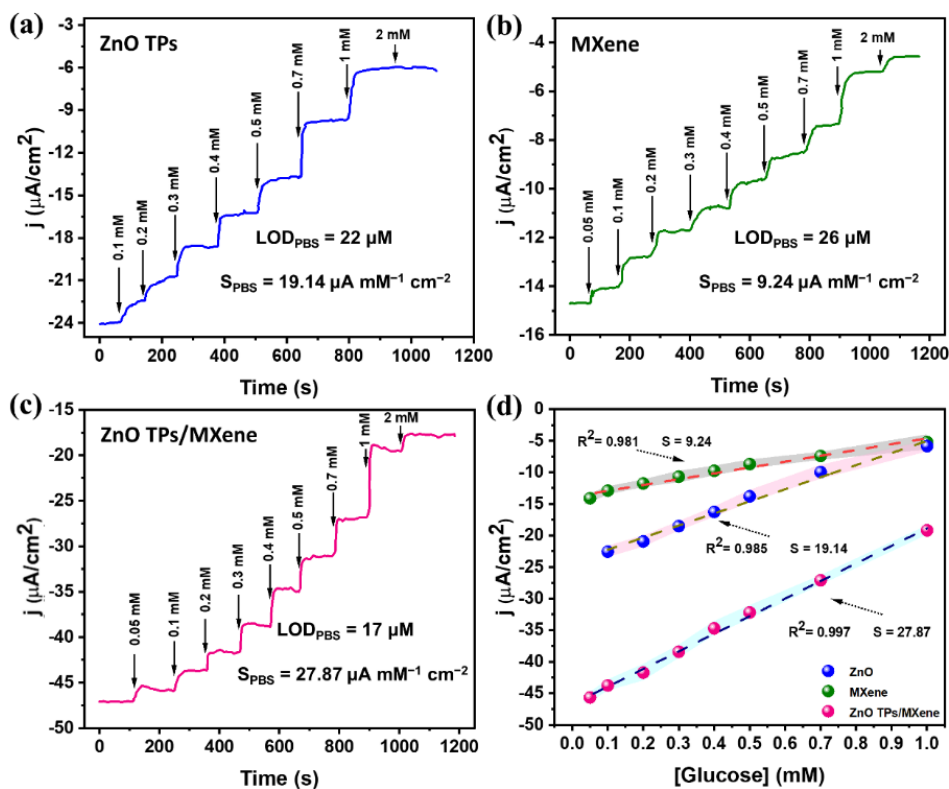


10 while, the reaction occurring at the counter electrode can be expressed by the following equation
 11 (Kim et al., 2014):



13 Thus, the detection of glucose can be carried out in both the CV mode and the linear sweep
 14 voltammetry (LSV) mode (Tian et al., 2015), measuring signals corresponding to hydrogen peroxide
 15 oxidation or dissolved oxygen reduction (free oxygen is consumed during the oxidation of glucose to
 16 gluconic acid by GOx in the surroundings of the working electrode). The later approach is based on
 17 the decrease of the cathodic current (oxygen reduction reaction – ORR) upon the addition of glucose.
 18 The operation in the cathodic regime (negative potentials) allows one to eliminate the effect of
 19 interference species. Besides, it is clearly seen the shift of the reduction potential of ZnO
 20 TPs/MXene/GOx-based electrode to the lower values (-0.21 V) in comparison to the ZnO TPs/GOx-
 21 based electrode. According to this, one may conclude that MXene enhance the catalytic activity of
 22 produced nanocomposites.

23 3.5. Chronoamperometry-based sensor performance in PBS (pH = 7.4).



26
 27 **Figure 4.** (a) Chronoamperometry measurements in PBS with different concentrations of glucose for ZnO TPs/GOx-
 28 based electrode at applied potential -0.32 V, (b) MXene/GOx-based electrode at applied potential -0.24 V, (c) ZnO

1 TPs/MXene/GOx-based electrode at applied potential -0.21 V. (d) Calibration curves for the ZnO TPs-, MXene-, and
2 ZnO TPs/MXene/GOx-based electrodes.

3

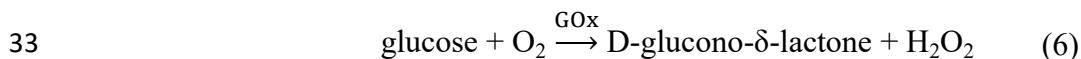
4 After a comprehensive analysis of the structural, optical properties, and the electrochemical behavior
5 of the fabricated ZnO TPs/GOx-, MXene/GOx-, and ZnO TP/MXene/GOx-based electrodes, all the
6 necessary parameters were determined for the subsequent chronoamperometric detection of glucose.
7 Fig. 4a-c represents chronoamperometry curves during the sequential addition of glucose to PBS
8 solution for the ZnO TPs/GOx-, MXene/GOx-, and ZnO TP/MXene/GOx-based electrodes,
9 respectively. One may observe the well-known stairs-like behavior of the current response vs. the
10 concentration of glucose biosensors (Chen et al., 2017; He et al., 2019). Working potentials for
11 glucose detection varied depending on the electrode. Fig. 4d shows the linear calibration curves
12 obtained from the stair-like chronoamperometry curves that show current densities vs the glucose
13 concentrations. The sensitivities of the ZnO TPs/GOx-, MXene/GOx-, and ZnO TP/MXene/GOx-
14 based electrodes towards the glucose were determined from the slopes of the calibration curves and
15 were estimated to be $19.14 \pm 1.4 \mu\text{A mM}^{-1} \text{cm}^{-2}$, $9.24 \pm 1.2 \mu\text{A mM}^{-1} \text{cm}^{-2}$, and $27.87 \pm 2.8 \mu\text{A mM}^{-1} \text{cm}^{-2}$,
16 respectively. The LODs were calculated using eq. (5) (Myndrul et al., 2021).

$$17 \quad LOD = 3.3 \cdot \sqrt{n} \cdot \sigma / b, \quad (5)$$

18 where σ is the standard deviations of the negative control at small concentrations, n – number of tests,
19 b is the slope of the curve, depicted in Fig. 4d.

20 The values of LODs were estimated to be $22 \pm 1.3 \mu\text{M}$, $26 \pm 1.8 \mu\text{M}$ and $17 \pm 1.7 \mu\text{M}$ for ZnO
21 TPs/GOx-, MXene/GOx-, and ZnO TP/MXene/GOx-based electrodes, respectively. One can assume
22 that the higher sensitivity and the lower LOD of ZnO TP/MXene/GOx-based electrode can be
23 attributed to its enhanced electron transfer rate if compared to ZnO TPs/GOx- and MXene/GOx-based
24 electrodes. Recent references show that the sensitivity of the prepared ZnO TPs/MXene/GOx-based
25 electrode is competitive with the reported sensitivities of skin-attachable or screen-printed glucose
26 electrodes (see Table S1).

27 The mechanism behind these types of amperometric glucose sensors is based on the change in the
28 concentration of free oxygen surrounding the working electrode. The reduction current depends on
29 the activity of the electrode towards the oxygen reduction reaction since a higher oxygen
30 concentration provides a high reduction current, and contrary. In this particular case, the oxygen
31 concentration is governed by the oxidation of the glucose at the hydroxyl (flavin adenine dinucleotide
32 (FAD)) group of GOx, which occurs with free oxygen consumption (González-Gaitán et al., 2017):

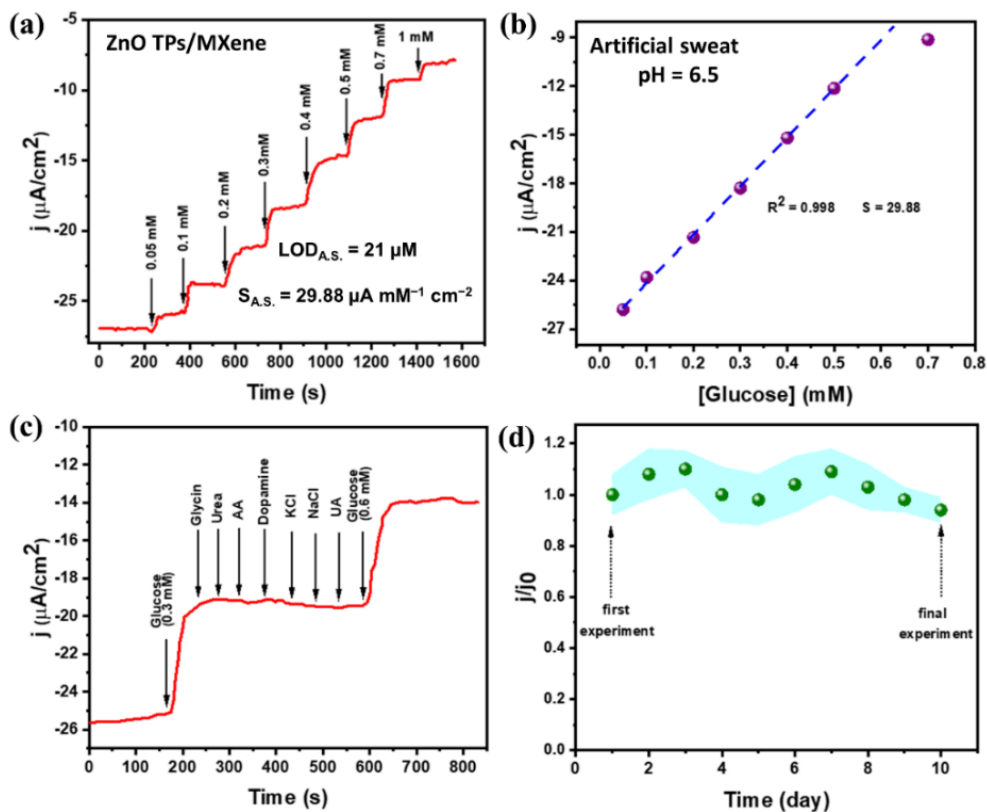


34 The GOx catalyzes the oxidation of glucose to D-glucono-1.5-lactone with the consumption of O_2
35 and reduction of FAD to FADH_2 . Since FADH_2 in an aqueous solution tends to be oxidized, it gives
36 the protons to reduce water molecules to H_2O_2 . Depending on the applied potential, either the anodic
37 current for measuring the rate of H_2O_2 oxidation or the current promoted by the O_2 reduction in the
38 cathodic area can be used (Soto et al., 2019).

39 **3.6. Chronoamperometry-based sensor performance in artificial sweat (pH = 6.5).**

40 The sensor was initially intended to be used for qualitative analysis of sweat glucose. Therefore, it
41 was important to test the produced electrode under sweat-like conditions. For this purpose, ZnO
42 TP/MXene/GOx- based electrode was chosen as the most efficient one. All experiments were
43 performed in artificial sweat with pH ~6.5 (detailed information in the experimental section). The
44 amperometric detection showed a similar stairs-like response of current density to glucose
45 concentrations (Fig. 5a). The LDR (in artificial sweat) of the glucose sensor lies between 0.05 and
46 0.7 mM. The sensitivity to glucose was extracted from the slope of the calibration curve and was
47 about $29.88 \pm 2.4 \mu\text{A mM}^{-1} \text{cm}^{-2}$ (Fig. 5b). The value of LOD was calculated to be $21 \pm 1.1 \mu\text{M}$.

1



2

3

4 **Figure 5.** (a) Chronoamperometry measurements for ZnO TP/MXene/GOx-based electrode in artificial sweat containing
 5 different concentrations of glucose (applied potential -0.24 V). (b) Calibration curve indicating a linear relationship
 6 between the current density of ZnO TP/MXene/GOx-based electrode and glucose concentration. (c) Selective glucose
 7 detection by use of ZnO TP/MXene/GOx-based electrode. (d) Long-term stability and multiple use of the ZnO
 8 TP/MXene/GOx-based electrode.

9

10 One of the main characteristics of the sensor is its selectivity with respect to various interfering agents
 11 contained in the analyte of interest. Fig. 5c shows the selectivity tests of ZnO TP/MXene/GOx- based
 12 electrode to detect 0.3 mM and 0.6 mM of glucose under the interference of different agents such as
 13 0.1 mM AA, 0.5 mM urea, 50 μM UA, 0.05 mM dopamine, 10 mM NaCl, 10 mM KCl, and 0.1 mM
 14 glycine. As can be observed, ZnO TP/MXene/GOx- based electrode provides highly selective
 15 glucose detection because of the specific interaction of the glucose and GOx treated ZnO TP/MXene
 16 composites. Moreover, the high selectivity may also be explained by the low applied potential (-0.24
 17 V) insufficient to oxidize interfering agents (such as AA, UA, etc.) (González-Gaitán et al., 2017).

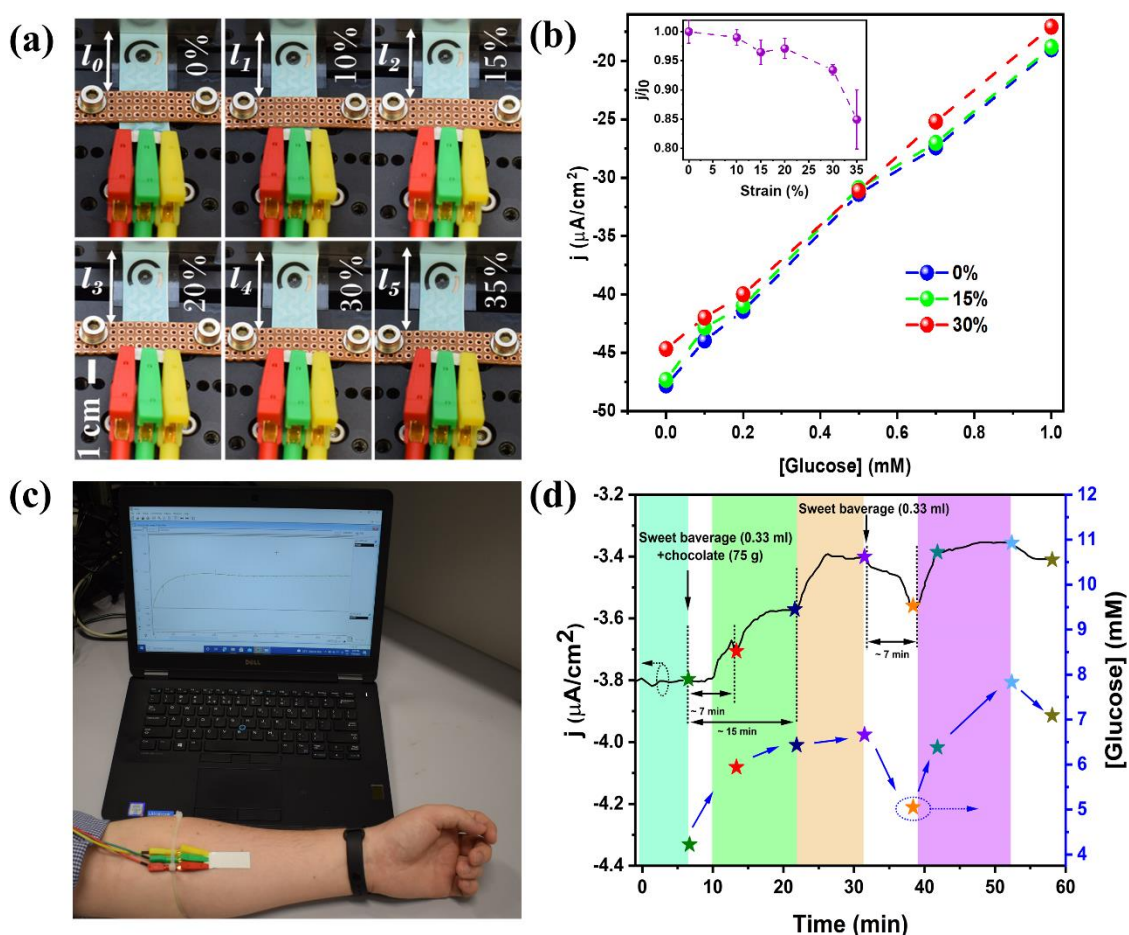
18 Fig. 5d represents the long-term stability of the ZnO TP/MXene/GOx- based electrode for glucose
 19 detection during 10 days of the experiments. After each experiment (one per day) the electrode was
 20 rinsed in PBS and deionized water, and then stored in a refrigerator at 4°C ready to be reused for
 21 further tests. The maximum deviation of the current density was about 10%, which barely affects the
 22 sensitivity of the ZnO TP/MXene/GOx- based electrode up to 10 days. Besides, the reproducibility
 23 of produced ZnO TP/MXene/GOx- based electrodes was studied. Twelve sets of samples/electrodes
 24 were produced using the same protocol. The deviation of sensor performance parameters was about
 25 of 10% (SI, Fig. S9).

26 3.7. Sensors performance under different applied strains.

27 Considering skin-attachable sensor performance, studying its stability under various mechanical
 28 stresses is important. This research analyzed the sensor's performance of ZnO TP/MXene/GOx-

1 based electrodes under different applied strains ranging from 0% to 35%. This value of strains range
 2 is much enough for flexible sensors to enable its application as a skin-attachable device. Fig. 6a shows
 3 the experimental setup for analysis of the stretch abilities of produced sensors. The working, auxiliary,
 4 and counter electrodes were sandwiched between slides to prevent the spread out of the analyte and
 5 to maintain a constant analyte volume during the experiment. The inset of Fig. 6b shows the relative
 6 change of the sensor response, where j and j_0 current densities before and after applied strains,
 7 respectively. It can be seen that the ZnO TPs/MXene/GOx electrode demonstrates a stable response
 8 (95% – 100%) up to 30% elongation, while significant instability occurs at 35% elongation. It might
 9 be explained by the fact that the deformation of ZnO TPs/MXene matrix leads to a decrease in the
 10 number of electron path-ways and, consequently, to the current reduction. The glucose detection
 11 under application of strains up to 30% (Fig. 6b) showed nearly identical calibration curves (current
 12 densities vs glucose concentrations) with similar slopes, indicating similar sensitivities and LODs. It
 13 may be noted, that the developed ZnO TPs/MXene/GOx-based sensors demonstrate the better or
 14 comparable with other stretchable sensors based on CoWO₄/CNT (Oh et al., 2018), and v-Au NW-
 15 electrodes, where the 30% elongation caused a significant current reduction (~5 times) (Q. Zhai et
 16 al., 2019).

17



18

19

20 **Figure 6.** (a) Digital photographs of the stretchable ZnO TPs/MXene/GOx-based electrode at various strains (0% - 35%).
 21 (b) Performance of the glucose sensor (in artificial sweat) at various applied strains (the insert graph represents the change
 22 of ZnO TPs/MXene/GOx electrode current density vs applied strains. (c) Digital photograph of skin-attachable sensor
 23 performance for sweat glucose qualitative analysis. (d) Graph representing the current density changes of the ZnO
 24 TPs/MXene/GOx-based sensor under the sweets consumption.

25

26

3.8. Qualitative glucose monitoring using stretchable ZnO TPs/MXene/GOx electrode.

1 In the final stage of the research, the developed glucose sensor based on ZnO TPs/MXene/GOx
2 nanocomposites were applied for human sweat analysis. The sensing skin-attachable pathway is
3 depicted in Fig. 6c. The change in current density was recorded in real-time, during sweets
4 consumption and exercising after the signal stabilization (blue area, Fig. 6d). The sweets were
5 followed by exercises (squats) to stimulate sweating, which is associated with higher post-meal blood
6 glucose levels (Lee et al., 2018). It is considered that sugar from beverages is absorbed in the intestine
7 within 10-15 minutes (Wolever, 2003), which corresponds to the decrease in current density and its
8 saturation within 15 minutes (the light green area in Fig. 6d).
9 Such a behavior of the current density correlates well with the experiments in PBS and artificial
10 sweat, thereby pointing to the increment of the glucose content (post-meal glucose) in a naturally
11 produced sweat. The second decrease in current density (light orange area) may be due to the sugar
12 from the chocolate as it takes longer time (15-30 min) to reach the bloodstream (Shaw et al., 2020).
13 After the two-stage decline (up to 31th min), the current density increased (31-38 min), indicating a
14 drop in the sweat glucose. This decrease in post-meal sweat sugar may be related to muscle glucose
15 uptake during exercise (SyLOW et al., 2017). Then, the second dose of sweet beverage was consumed
16 (at 32 min) to increase the glucose level. It is seen that the current density decreased again, thereby
17 confirming the similar behavior of the skin-attachable sensor after the glucose level increased. In
18 parallel with the qualitative glucose monitoring in sweat, the conventional approach for glucose levels
19 detection has been performed by using a commercially available blood glucometer. It is obvious that
20 the sweat glucose real-time monitoring and discrete measurements by means of glucometer had
21 similar trends as shown in Fig. 6d. Moreover, the real-time measurement of the current density of the
22 amperometric glucose sensor correlates well with the other sensors for post-meal glucose monitoring
23 (Bae et al., 2019; Chen et al., 2017; Zaharieva et al., 2019). In addition, qualitative glucose monitoring
24 was performed in a second volunteer, and the results demonstrated the similar trend (SI, Fig. S10).

25

26 CONCLUSIONS

27 In this study, ZnO TPs/MXene-based electrodes were fabricated and used for on-body qualitative
28 glucose monitoring in sweat. The synthesis of ZnO TPs/MXene was performed in accordance with
29 the three-step manufacturing process, including the synthesis of MXene and ZnO, with the following
30 decoration of ZnO TPs with MXene nanoflakes. The fabricated sensors exhibited high sensitivity of
31 $27.87 \mu\text{A mM}^{-1} \text{cm}^{-2}$ and $29.88 \mu\text{A mM}^{-1} \text{cm}^{-2}$ for tests in PBS and artificial sweat, respectively.
32 Moreover, none of the interfering agents affected the sensor response, which indicates the high
33 selectivity of the produced ZnO TPs/MXene/GOx-based electrodes. The long-term stability of the
34 sensor was studied during a 10-days experiment, and the result showed that the sensor response
35 remains stable within this period. The ZnO TPs/MXene/GOx-based electrode demonstrated
36 mechanical stability of up to 30% elongation, with an insignificant change of the detection
37 parameters. The real-time monitoring of post-meal glucose levels in sweat had the same trend as when
38 measured with a conventional blood glucometer under similar conditions. The developed skin-
39 attachable and stretchable electrodes based on ZnO TPs/MXene nanocomposites demonstrate a
40 prospective approach towards non-invasive, continuous glucose monitoring in sweat for healthcare
41 applications.

42

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5

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