nature nanotechnology

Article

https://doi.org/10.1038/s41565-023-01545-6

3D spatiotemporally scalable in vivo neural probes based on fluorinated elastomers

In the format provided by the authors and unedited

Table of Contents

Supplementary Discussions.

Supplementary Discussion 1. The cut-off frequency of a lossy transmission line.

We use the same model that we previously developed[1, 2], where a single interconnect behaves as a transmission line. The voltage (or current) attenuation of a harmonic signal of frequency f along a stripline of length L is:

$$\left|\frac{V(L)}{V(0)}\right| = \left|exp\left(-\sqrt{\frac{1}{f_c}(jf + f_d)}\right)\right| \quad (1)$$

Where $j^2 = -1$ and

$$f_c = \frac{1}{4\pi}\frac{h_c h_d}{L^2}\frac{\sigma_c}{\varepsilon_d} \quad (2)$$

$$f_d = \frac{1}{2\pi}\frac{\sigma_d}{\varepsilon_d} \quad (3)$$

f_c is the cut-off frequency of the lossless transmission line, with h_c and σ_c the thickness and conductivity of conductor, h_d, ε_d and σ_d the thickness on each side of the conductor, permittivity and conductivity of the dielectric encapsulation. In all simulations (Extended Data Fig. 1), we choose a relative permittivity of 3.2 for the dielectric and an ionic conductivity $\sigma_d = 5.18\text{x}10^{-8}$ S/m to model the built-in ionic conductivity in hydrocarbon-based dielectric elastomers[1].

In a neural probe, the ultra-high aspect ratio of individual electrodes limits f_c. Thus, dimensions of the probe and materials used as conductor and dielectric must be chosen carefully to ensure a sufficiently high cut-off frequency, *i.e.*, ensure transmission of signals of all frequencies below 10 kHz with minimal attenuation. We show that such a criterion totally excludes the use of low-conductivity conductors for scalable neural probes (Extended Data Fig. 1).

Supplementary Discussion 2. The trade-off between electrode density, cut-off frequency, and mechanical softness.

To minimize the bending stiffness of neural probes and for nanofabrication convenience, a planar transmission line geometry, stripline, is often chosen, with a metal line thickness below 100 nm. Lateral dimensions of the stripline being large compared to the thicknesses involved, most of the capacitive leakage current happens in the perpendicular direction. In practice, one solution to increase the density of electrodes is to decrease the thickness of dielectric encapsulation surrounding the nanometer-thin metal lines. However, for the reasons mentioned in the supplementary discussion 1, limiting capacitive leakage into the surrounding biofluids and crosstalk between vertically stacked metal lines to an acceptable level sets a theoretical lower bound to the encapsulation's thickness (Extended Data Fig. 1)[1]. Furthermore, sub-micron films of stiff dielectric polymer often have a non-ideal capacitive behavior due to pinholes and defects which further limits progress in this direction[3].

Instead of downscaling dimensions, *state-of-the-art* examples of neural probes for long-term neural interfaces, using stiff dielectric polymers (elastic modulus ~ GPa) as encapsulation[4, 5], realize high-density neural probes by stacking vertically two to three layers of metal electrodes, where the thickness of each dielectric layer is about 1.5-2 μm, to limit the capacitive leakage. We hypothesize that higher numbers of electrode layers have not been realized because the resulting stiffness of the neural probe would create large acute damage at implantation, generate a stronger long-term immune response, and make the probes very brittle during implantation[6-8]. Typical dielectric materials used as the passivation layer, such as parylene-C[9], polyimide[10], or SU-8[11-13], suffer from this limitation.

Supplementary Discussion 3. Bending stiffness of a multilayer neural probe.

The bending stiffness of a multilayer neural probe along its longitudinal direction can be estimated using a plate model, given that lateral dimensions are large compared to the thickness. For a homogeneous plate of elastic modulus E, Poisson's ratio ν, width w, and thickness h, the bending stiffness is:

$$K = \frac{Ewh^3}{12(1-\nu^2)} \tag{4}$$

We derive the expression of the bending stiffness for a multilayer plate with 2N-1 layers of metal and 2N layers of polymer described in Extended Data Figure 2a. Because the lateral spacing between interconnects is usually smaller and made of a softer material (the polymer) than the metal lines, we neglect its contribution to the bending stiffness and simplify the geometry to a 1D beam of alternating polymer and metal layers. We place a metal layer in the center to minimize its bending stiffness and build up layers symmetrically from the neutral axis of the beam.

The bending moment along the x direction (per unit width) is:

$$m_{xx} = \int_{-[Nh_d+(N-1/2)h_m]}^{Nh_d+(N-1/2)h_m} z\sigma_{xx}dz \tag{5}$$

The normal stress along the x direction, σ_{xx}, is given by Hooke's law under plane strain condition,

$$\sigma_{xx} = \frac{E}{1-\nu^2}(\varepsilon_{xx} + \nu\varepsilon_{yy}) \tag{6}$$

Where E is the elastic modulus.

In small deformations,

$$\varepsilon_{xx} = -z\frac{\partial^2 w}{\partial x^2} \tag{7.1}$$

$$\varepsilon_{yy} = -z\frac{\partial^2 w}{\partial y^2} \tag{7.2}$$

where w is the displacement in the z direction.

Thus,

$$z\sigma_{xx} = -z^2 \frac{E}{1-\nu^2}\left(\frac{\partial^2 w}{\partial x^2} + \nu \frac{\partial^2 w}{\partial y^2}\right) \qquad (8)$$

And the bending moment (per unit width) can be rewritten as:

$$m_{xx} = -\left(\int_{-[Nh_d+(N-1/2)h_m]}^{Nh_d+(N-1/2)h_m} z^2 \frac{E}{1-\nu^2}\, dz\right)\left(\frac{\partial^2 w}{\partial x^2} + \nu \frac{\partial^2 w}{\partial y^2}\right) \qquad (9)$$

The flexural rigidity (bending stiffness per unit width) is:

$$K = 2\int_0^{Nh_d+(N-1/2)h_m} z^2 \frac{E}{1-\nu^2}\, dz \qquad (10)$$

Which can be rewritten:

$$K = \frac{2E_m}{1-{\nu_m}^2}\left[\frac{h_m^3}{24} + \sum_{k=1}^{N-1}\int_{k(h_d+h_m)-h_m/2}^{k(h_d+h_m)+h_m/2} z^2\, dz\right] + \frac{2E_d}{1-{\nu_d}^2}\sum_{k=1}^{N}\int_{(k-1)(h_d+h_m)+h_m/2}^{k(h_d+h_m)-h_m/2} z^2\, dz \qquad (11)$$

And after integration:

$$K = \frac{2E_m}{1-{\nu_m}^2}\left[\frac{h_m^3}{24} + \frac{1}{3}\sum_{k=1}^{N-1}\left[(k(h_d+h_m)+h_m/2)^3 - (k(h_d+h_m)-h_m/2)^3\right]\right] + \frac{2E_d}{3(1-{\nu_d}^2)}\sum_{k=1}^{N}\left[(k(h_d+h_m)-h_m/2)^3 - \left((k-1)(h_d+h_m)+h_m/2\right)^3\right] \qquad (12)$$

The flexural rigidity, *K*, is plotted as a function of the number of metal interconnects layers in Extended Data Figure 2b. This calculation also highlights a dramatic jump of 4 orders of magnitude in bending stiffness when an elastomer-based probe contains more than one central layer of interconnects. Indeed, the further to the neutral axis the metal lines are, the higher their contribution to the bending moment is, which is reminiscent of the "H-shaped" cross-section beams used in civil engineering, where the bending stiffness-to-density ratio is optimized by using the same concept. The flexural rigidity of fluorinated elastomer-based neural probes is dominated by the metal layers that are located far from the neutral axis. In the geometry described in Extended Data Figure 2a, the PFPE-DMA-to-SU-8 ratio in flexural rigidity converges toward 0.28 for high numbers of interconnect layers (Extended Data Fig. 2c)

Given the extreme modulus mismatch between Au and PFPE-DMA (158,000 times difference), we recognize that there is a chance that the Euler-Bernoulli beam theory may break down and a split of neutral axis may occur. We therefore built a trilayer bending problem with two layers of gold sandwiching one layer of PFPE-DMA to double check. Following on previous work of split-of-neutral-axis theory[14, 15], we obtain the following x-dependent expression for the flexural rigidity when the beam is bent into an arc:

$$K = \frac{Eh_m^3}{6(1-\nu_m^2)} + \frac{Eh_m(h_m+h_d)^2}{2(1-\nu_m^2)}\left(1 - \frac{cosh\sqrt{(1-\nu_d^2)\frac{G_d\ x\ x}{E_m h_d h_m}}}{cosh\sqrt{(1-\nu_d^2)\frac{G_d\ L\ L}{E_m h_d h_m}}}\right) \quad (13)$$

where $G_d = \frac{E_d}{1+2\nu_d}$ is the shear modulus of the PFPE-DMA dielectric layer. This equation is plotted in Extended Data Figure 2e, which indicates that the flexural rigidity in the majority of the beam approaches the Euler-Bernoulli prediction.

Supplementary Discussion 4. Theory of the built-in ionic conductivity in dielectric elastomers.

We hypothesize that a dielectric elastomer containing ions behaves as an infinitely dilute solution, where molar conductivity is the sum of the contribution of individual ions, following Kohlrausch's law of independent ionic migration. Therefore, the ionic conductivity σ of the dielectric is the sum, over all conductive species in the media, of their electrical charge – mobility – concentration product:

$$\sigma = \sum_{all\ charged\ particles} q * Electrical\ mobility * Concentration \quad (14)$$

Where q is the elementary charge. Electrical mobility can be expressed as a function of the diffusivity, D, of the charged particle, by the Stokes-Einstein equation:

$$Electrical\ mobility = \frac{qD}{kT} \quad (15)$$

Where k is the Boltzmann constant and T the temperature.

At thermodynamic equilibrium, concentration of charged particles inside the material can be defined as the product of the concentration of the same in the surrounding media, C_{out}, multiplied by the solubility (or partition coefficient), S, of charged particles in the material:

$$Concentration = S * C_{out} \quad (16)$$

In the following, we suppose that the material is a dielectric polymer, and that the surrounding media is a biofluid solution. Given the composition of extracellular media in the brain, we assume for simplification that the biofluid only contains sodium and chloride ions, and that both ions have the same ionic diffusivity, D, and the same ionic solubility, S, inside the dielectric material. Therefore, combining equations 14 to 16, we obtain:

$$\sigma = 2\frac{q^2}{kT} D * S * C_{out} \quad (17)$$

The product $D*S$ is also named ionic permeability.

Supplementary Discussion 5. Determination of the ionic diffusivity, solubility, and conductivity by external electrolyte conductimetry.

1) **Crumpled thin films preparation.** Large areas (ranging from 150 to 300 cm^2) of dielectric thin films were prepared on glass slides following the preparation recipes described in the methods section, then soaked in deionized water to facilitate their peel off. After peeling off, the crumpled films were transferred in glass vials for the remainder of the experiment.
2) **Wash out.** Crumpled thin films were first immersed for 3 weeks in deionized water replaced regularly at ambient temperature to remove any impurities that could contribute to the ionic conductivity. The conductometer was used to confirm that the surrounding solution's conductance remained negligible (0-2 µS range) after 3 weeks, ensuring that the washout process was finished.
3) **Soaking in 10x PBS.** Samples were transferred to new glass vials in a large volume of 10x PBS solution, at a fixed temperature (4, 37 or 65 °C) for 3 weeks, to be fully soaked by ions. It was verified *a posteriori* that 3 weeks was a long time compared to the characteristic diffusion time of ions in the materials. Glass vials are tightly sealed to avoid evaporation of liquids and contact with the atmosphere.
4) **Surface wash and mass measurements.** Samples were thoroughly rinsed in two successive deionized water solutions (30 s in each) to remove 10x PBS solution remaining on the surface, then dried at 65 °C for 30 min, before mass measurements.
5) **Soaking in fixed volume of deionized water.** Samples were transferred to new glass vials containing 4 mL of deionized water and stored at a fixed temperature (4, 37 or 65°C). Conductance of the deionized water solution was measured regularly to determine the quantity of ions desorbed by each material over time. Glass vials are tightly sealed to avoid the evaporation of liquids and contact with the atmosphere.

The conductometer measures how many ions are soaked out of the thin films over time, at a given temperature. The value of the conductance can be converted back to the concentration of ions desorbed per volume of material, using the following formalism:

1) Convert conductance to conductivity in mS/m, given the aspect ratio of the conductance cell (all conductance values are given at 25 °C after automatic temperature compensation from the conductometer).
2) Convert conductivity to ion concentration in mol/m^3 (per volume of water), assuming an infinitely dilute electrolyte, where the only contributions are from sodium and chloride ions, with the respective molar conductivities of 5.011 and 7.634 $mS \cdot m^2/mol$ at 25 °C.
3) Multiply by the volume of water and divide by the volume of polymer to obtain the ion concentration desorbed per volume of polymer, C_{meas}.

Determination of the ionic solubility, *S*. The ion concentration's plateau (Fig. 2e, Extended Data Fig. 5c) corresponds to the concentration of ions inside the polymer thin film at equilibrium with 10x PBS solution. Dividing this concentration by the concentration of sodium chloride ions in 10x PBS solution (1.37 mol/L) obtains the ionic solubility *S*. *S* is a partition coefficient, without unit.

Determination of the ionic diffusivity, *D*. The volume of water surrounding the polymer during the de-soaking phase being kept large compared to the volume of polymer (*e.g.*, 30 to 200X depending on the sample), we assume that the concentration of ions outside of the thin film is always negligible compared to the concentration inside it. The corresponding one-dimensional boundary value problem (Extended Data Fig. 5b) is:

$$\frac{\partial C}{\partial t} = D\frac{\partial^2 C}{\partial x^2} \tag{18}$$

$$C(x = 0, t) = C(x = H, t) = 0 \tag{19}$$

$$C = C_0\,,\ 0 < x < H\,,\ t = 0 \tag{20}$$

where C is the concentration of ions inside the material, H the thickness of the thin film, and C_0 is the concentration of ions inside the material at time $t = 0$. C_0 is the product of the ionic solubility S, by the concentration of salt in the surrounding solution during the soaking phase ($[NaCl]_{10xPBS}$ = 1.37 mol/L). The theoretical solution of the boundary value problem (equations 8-10) is

$$C(x,t) = \frac{4C_0}{\pi}\sum_{n=0}^{\infty}\frac{1}{2n+1}\,sin\,sin\,(\frac{(2n+1)x}{H})e^{-D(2n+1)\pi^2 t/H^2}\ \ ,0 < x < H \tag{21}$$

where D is the ionic diffusivity[16].

The concentration of ions (per volume of polymer) measured by conductimetry in the surrounding solution, C_{meas}, is connected to C by the following relation:

$$C_{meas}(t) = C_0 - \int_0^H C(u,t)du \tag{22}$$

Therefore,

$$C_{meas}(t) = C_0\left(1 - \frac{4}{\pi}\int_0^1\sum_{n=0}^{\infty}\frac{1}{2n+1}\,sin\,sin\,((2n+1)u)e^{-D(2n+1)\pi^2 t/H^2}\ du\right) \tag{23}$$

We use the software Mathematica to fit the theoretical solution to the experimental data (Extended Data Fig. 5c) and extrapolate the ionic diffusivity, D.

Supplementary Discussion 6. Practical limitations to the 3D integration of interconnects in ultrasoft neural probes.

While we envision that the current fabrication process could be extended to more metal layers, the making of such probes may be challenging due to the following factors, ranked here by significance: (1) Achieving a high-yield lift-off process on soft structures thicker than 20 micrometers could be challenging. With the current design, this could limit the number of metal layers to a maximum of 8 or 9, considering a 2-micrometer thick dielectric encapsulation around each metal layer. (2) Progressive misalignment of the layers after multiple rounds of photopatterning could exist. However, this issue can be mitigated by introducing new sets of alignment markers throughout the fabrication process. (3) Mechanical stability of the comprehensive structure during its release from the wafer could be a concern. Additional layers of metal electrodes may cause stress accumulation, leading to debonding between layers after long-term exposure to biofluids. (4) Cyclic thermal processing of an assembly with more than 4 metal, and 5 polymer layers still requires validation.

Supplementary Tables

Polymer	Atoms/chain	Molecular weight /Da	Total atoms
PFPE-DMA	649	10755	9735
PDMS	1017	7504	15255
PHFIPA	1808	22347	27120
H-SEBS (83/17)	1242	6093	12420

Table S1. Number of atoms, polymer chains, and molecular weight used for molecular dynamics simulations.

Stage	Stage duration /ns	Total time /ns	Temperature /K	Pressure /bar
Heating	10	10.1	10 -> 300	1
Compression	10	20.1	300	1 -> 1000
Decompression	10	30.1	300	1000 -> 1
Heating	10	40.1	300 -> 500	1
Cooling	10	50.1	500 -> 300	1
Compression	10	60.1	300	1 -> 4000
Decompression	10	70.1	300	4000 -> 1
NPT	10	80.1	300	1

Table S2. Summary of the stages used to bring the polymer structures close to equilibrium.

Measurement technique	EIS	EEC
Description	• Measures the effect of ions penetrating inside a dielectric material on its impedance.	• Measures ions released from a dielectric material into an external solution of *initially* deionized water.
Setup	• Three-electrodes measurement with one side of the dielectric is in contact with 1x PBS solution, in which counter (Pt wire) and reference (Ag/AgCl) electrodes are immersed. The working electrode is on the other side of the dielectric. • AC • frequency range = [0.1; 10^6] Hz	• Conductometer immersed in the solution surrounding the dielectric material. • DC
Advantages	• Measure in an electrochemical environment similar to an implanted neural probe. • Allows for measurements as the dielectric polymer progressively soaked in a saline solution.	• Offers independent measurement of ionic diffusivity (*D*) and solubility *(S*). • Shows low sensitivity to defects (pinholes, cracks) in the thin dielectric film.
Limitations	• High sensitivity to defects (pinholes, cracks) in the thin dielectric film. • Requires an electrical modeling for the dielectric impedance and built-in ionic conductivity[1]. Here we use a resistor in parallel with the capacitance of the dielectric thin film to describe ionic conductivity. • Dielectric – electrolyte interfacial impedance can only be neglected at high frequencies in the modeling[1]. • Diffusivity (*D*) is estimated qualitatively by an impedance drop. Ionic conductivity (σ) is estimated through the modeling of the Nyquist plot[1]. Solubility is indirectly determined by using *D* and σ.	• Requires very large areas (> 150 cm^2) of thin films (< 5 μm) of dielectric material. • Requires a large volume of deionized water compared to the volume of dielectric material used to fall within the approximation of the boundary value problem. • Conductivity of the solution must be within the measurable range of the conductometer. • Unsuitable for non-crosslinked, thermoplastic elastomers such as PIB, which cannot keep their thin film geometry when crumpled. • Unsuitable for thin film elastomers such as PHFIPA and PPFHEA, which cannot be easily peeled off from the substrate.

Table S3. Advantages and limitations of electrochemical impedance spectroscopy and external electrolyte conductimetry.

Experiment	Conditions	Material	number of samples	Average thickness (µm)	mass (g)
EIS	10x PBS, 65°C	SU-8 2002 double layer	4	4.19	n/a
		PFPE-DMA	4	2.00	n/a
		PHFIPA	4	0.84	n/a
		PPFHEA	4	3.57	n/a
		H-SEBS	12	2.00	n/a
		PDMS	8	8.51	n/a
		PIB	4	8.45	n/a
		PI	3	2.00	n/a
	1x PBS, 37°C	SU-8 2000.5, double layer	4	0.84	n/a
		PFPE-DMA	4	1.25	n/a
EEC	1x PBS, 4°C	SU-8 2000.5 double layer	6	0.85	0.0422
		PFPE-DMA	8	2.37	0.0848
		H-SEBS	4	2.34	0.0689
		PDMS	6	6.06	0.1213
	1x PBS, 37°C	SU-8 2000.5 double layer	4	0.84	0.0497
		PFPE-DMA	8	2.05	0.0878
		H-SEBS	4	2.34	0.0698
		PDMS	6	6.06	0.1353
	1x PBS, 65°C	SU-8 2000.5 double layer	6	0.81	0.0222
		PFPE-DMA	8	2.05	0.0862
		H-SEBS	4	2.34	0.0698
		PDMS	6	6.06	0.1353

Table S4. Thin films thickness measured by contact profilometry.

Material	Average dielectric constant (mean ± S.D.)	Literature value	References
PFPE-DMA	1.99 ± 0.03	1.9-2.2	17-20
H-SEBS	2.59 ± 0.07	2.1-2.45	21-23
PDMS	2.46 ± 0.02	2.7-2.82	21, 24
PIB	2.39 ± 0.12	2.3-2.65	25
SU-8	4.55 ± 0.14	3.0-4.5	26-27
PI	3.40 ± 0.03	3.4-3.5	28-29

Table S5. Dielectric constant of different polymer thin films. Values are determined from averaged electrochemical impedance spectrums (frequency range 1-100 kHz) and compared to literature values.

Material	σ (S/m)	D (m^2/s)	S
PDMS	$5.30\ 10^{-8}$	$2.55\ 10^{-15}$	0.0404
H-SEBS	$7.62\ 10^{-9}$	$1.82\ 10^{-16}$	0.0814
PFPE-DMA	$6.34\ 10^{-10}$	$1.4\ 10^{-17}$	0.0881
SU-8	$1.62\ 10^{-9}$	$1.69\ 10^{-17}$	0.187

Table S6. Ionic conductivity, diffusivity and solubility in dielectric polymers measured by external electrolyte conductimetry.

Material	PDMS	SEBS	PFPE	SU-8
E_a (kJ/mol)	10.32	4.56	13.25	9.02
H_s (kJ/mol)	3.03	1.04	11.83	14.35

Table S7. Mean activation energy E_a, and enthalpy of solution H_s of ions inside dielectric polymers calculated from the Arrhenius model.

Material	Water Permeability ($m^2/s/Pa$)*10^{18}
Polydimethylsiloxane (Sylgard 184, 1:10 ratio)	182 ± 5 *
VHB Tape	40 ± 8 *
Polyisobutylene	2.56 ± 0.75 *
PFPE-DMA	4.03 ± 0.42 (this work, n = 4)

*: from reference [30].

Table S8. Water permeability of PFPE-DMA thin films compared to other elastomers. Value = mean ± S.D.

Technology	Number of interconnects (*N*)	Width (*W*) (µm)	Thickness (*H*) (µm)	Electrodes per cross-sectional area (*D*) (µm^{-2})	Dielectric insulation material	Elastic modulus (E_d) of dielectric (Pa)
This work	64 64	240 70	12 12	0.022 0.076	PFPE-DMA	5.0 10^5
Neuropixel 2.0[31]	1280	70	24	0.76	Silicon	6.2 10^{10}
Mesh nanoelectronics[11]	16	2000 (flat)	0.9	0.0088	SU-8	2 10^9
Mesh nanoelectronics[12]	16	2000 (flat)	0.9	0.0088	SU-8	2 10^9
Polyimide flexible probes[10]	61	1189	1.5 3	0.034 0.017	Polyimide	4 10^9
Polyimide flexible probes[4]	32	70	6	0.076	Polyimide	4 10^9
Polyimide flexible probes[5]	16	80	14	0.014	Polyimide	4 10^9
SU-8 flexible probes[32]	8 (NET-50) 4 (NET-10)	50 10	1 1.5	0.16 0.27	SU-8	2 10^9
PDMS + µ-structured PI/Pt/PI[33]	9	3500	200	0.000013	Polyimide / PDMS	3 10^6
Viscoelastic surface electrode arrays[34]	8	6000	250	0.0000053	Self-healing PDMS	9.0 10^5
3D-printed PDMS + PEDOT:PSS[35]	9	117	1000	0.000077	PDMS	1.1 10^6
PDMS + µ-cracked Au[36]	7	3000	120	0.000019	PDMS	1.2 10^6

[31]: The shank contains 1280 sites.

[11, 12]: The width of mesh nanoelectronics is the width of the device before release from the wafer.

[10]: The width is estimated in Figure S2.

[33]: The thickness is estimated from Figure 1 and the width from Figure S16.

[34]: Dimensions are given in Figure 5a and its caption.

[35]: The width is estimated in Figure S12 and thickness is estimated to be 50 (bottom PDMS) + 17 (conducting polymer) + 50 (top PDMS) = 117 µm in total.

Table S9. Number of electrodes per cross-sectional area, and elastic modulus of the dielectric encapsulation in flexible and soft brain implants in *state-of-the-art* examples. The density of interconnects is calculated as $D = N / (W*H)$.

Supplementary Video 1. SU-8 and PFPE-DMA neural probes with same geometry and thickness are poked and stretched with tweezers. The SU-8 probe fractures easily while the PFPE-DMA probe withstands large deformations.

References

1. Le Floch, P. *et al.* Fundamental limits to the electrochemical impedance stability of dielectric elastomers in bioelectronics. *Nano Lett.* **20**, 224-233 (2020).
2. Chipman, R. A. *Shaum's theory and problems of transmission lines*. (McGraw-Hill, 1968).
3. Ceyssens, F. & Puers, R. Insulation lifetime improvement of polyimide thin film neural implants. *J. Neural Eng.* **12**, 054001, 054001 (2015).
4. Musk, E. An integrated brain-machine interface platform with thousands of channels. *J Med. Internet Res.* **21**, e16194 (2019).
5. Chung, J. E. *et al.* High-density, long-lasting, and multi-region electrophysiological recordings using polymer electrode arrays. *Neuron* **101**, 21-31 (2019).
6. Lacour, S. P., Courtine, G. & Guck, J. Materials and technologies for soft implantable neuroprostheses. *Nat. Rev. Mater.* **1**, 16063 (2016).
7. Hong, G. & Lieber, C. M. Novel electrode technologies for neural recordings. *Nature Rev. Neurosci.* **20**, 330-345 (2019).
8. Nguyen, J. K. *et al.* Mechanically-compliant intracortical implants reduce the neuroinflammatory response. *J. Neural Eng.* **11**, 056014-056014 (2014).
9. Takeuchi, S., Ziegler, D., Yoshida, Y., Mabuchi, K. & Suzuki, T. Parylene flexible neural probes integrated with microfluidic channels. *Lab Chip* **5**, 519-523 (2005).
10. Guan, S. *et al.* Elastocapillary self-assembled neurotassels for stable neural activity recordings. *Sci. Adv.* **5**, eaav2842 (2019).
11. Liu, J. *et al.* Syringe-injectable electronics. *Nat. Nanotechnol.* **10**, 629-636 (2015).
12. Yang, X. *et al.* Bioinspired neuron-like electronics. *Nat. Mater.* **18**, 510-517 (2019).
13. Li, Q. *et al.* Cyborg organoids: Implantation of nanoelectronics via organogenesis for tissue-wide electrophysiology. *Nano Lett.* **19**, 5781-5789 (2019).
14. Li, L. *et al*. Integrated flexible chalcogenide glass photonic devices. *Nat. Photonics.* **8**, 643-649 (2014).
15. Li, S., Su, Y. & Li, R. Splitting of the neutral mechanical plane depends on the length of the multi-layer structure of flexible electronics. *Proc. R. Soc. A: Math. Phys. Eng. Sci.* **472**, 20160087 (2016).
16. Crank, J. *The mathematics of diffusion*. (Oxford university press, 1979).
17. Liu, J. *et al.* Fully stretchable active-matrix organic light-emitting electrochemical cell array. *Nat. Commun.* **11**, 3362 (2020).
18. Olson, K.R. *et al.* Liquid perfluoropolyether electrolytes with enhanced ionic conductivity for lithium battery applications. *Polymer* **100**, 126-133 (2016).
19. Timachova, K. *et al.* Mechanism of ion transport in perfluoropolyether electrolytes with a lithium salt. *Soft Matter* **13**, 5389-5396 (2017).
20. https://miller-stephenson.com/electrical-conducting-grease

21. Kong, D. *et al.* Capacitance characterization of elastomeric dielectrics for applications in intrinsically stretchable thin film transistors. *Adv. Funct. Mater.* **26**, 4680-4686 (2016).
22. Grigorescu, R.M. *et al.* Mechanical and dielectric properties of SEBS modified by graphite inclusion and composite interface. *J. Phys. Chem. Solids* **89**, 97-106 (2016)
23. Kim, B., Min, K., Kim, J., Hong, S. M. & Koo, C. M. The electromechanical properties of SEBS and SEBS-g-MA dielectric elastomer gels. *Mol. Cryst.* **519**, 77-81 (2010).
24. https://www.dow.com/content/dam/dcc/documents/en-us/productdatasheet/11/11-31/11-3184-sylgard-184-elastomer.pdf
25. Ruan, M. *et al.* Improved electromechanical properties of brominated butyl rubber filled with modified barium titanate. *RSC Adv.* **7**, 37148-37157 (2017)
26. https://kayakuam.com/products/display-dielectric-layers
27. https://memscyclopedia.org/su8.html
28. https://docplayer.net/54576661-Table-of-contents-product-bulletin-pi-2545-wet-etch-applications.html
29. https://www.dupont.com/content/dam/dupont/amer/us/en/ei-transformation/public/documents/en/EI-10142_Kapton-Summary-of-Properties.pdf
30. Le Floch, P., Meixuanzi, S., Tang, J., Liu, J. & Suo, Z. Stretchable seal. *ACS Appl. Mater. Interfaces* **10**, 27333-27343 (2018).
31. Steinmetz, N. A. *et al.* Neuropixels 2.0: A miniaturized high-density probe for stable, long-term brain recordings. *Science* **372**, 6539 (2021).
32. Luan, L. *et al.* Ultraflexible nanoelectronic probes form reliable, glial scar–free neural integration. *Sci. Adv.* **3** (2017).
33. Vachicouras, N. *et al.* Microstructured thin-film electrode technology enables proof of concept of scalable, soft auditory brainstem implants. *Sci. Transl. Med.* **11**, 514 (2019).
34. Tringides, C. M. *et al.* Viscoelastic surface electrode arrays to interface with viscoelastic tissues. *Nat. Nanotechnol.* **16**, 1019–1029 (2021).
35. Yuk, H., Lu, B. & Zhao, X. Hydrogel bioelectronics. *Chem. Soc. Rev.* **48**, 1642-1667 (2019).
36. Minev, I. R. *et al.* Electronic dura mater for long-term multimodal neural interfaces. *Science* **347**, 159-163 (2015).