

BIOELECTRONICS

An in-body networking system for communication between wearable and implantable therapeutics

Ramy Ghanim^{1†}, Yoon Jae Lee^{2,3††}, Garan Byun^{2,4,5}, Joy Jackson⁶, Julia Z. Ding⁶, Jihoon Park¹, Markella Bibidakis³, Elaine Feller¹, Eugene Kim^{2,5}, Dilay Aygun¹, Anika Kaushik¹, Alaz Cig^{5§}, Sean Healy⁶, Camille E. Cunin⁷, Aristide Gumyusenge⁷, Woon-Hong Yeo^{2,4,5,6*}, Alex Abramson^{1,2,6,8*}

Networks of bioelectronic sensors and actuators enable closed-loop therapies that connect distant, related physiological dynamics. Currently, device interactions are limited by communication methods that inefficiently penetrate tissue and require bulky components. Inspired by the ionic signaling of the nervous system, we engineered a communication platform that uses tissue as a conductive medium to transmit signals among implants and wearables through epidermal, subcutaneous, intraperitoneal, and gastrointestinal spaces. Devices generate transient electric fields that selectively activate other devices when receiving pulses that switch on their specific transistor circuits. Implants are syringe-injectable, require negligible power consumption in listening states, and provide >10× greater tissue communication coverage than Bluetooth. In vivo in rats, we demonstrate coordinated, full-body networks of sensors and neural interfaces that enable wireless dual-limb motor control.

Smart bioelectronic therapeutics such as insulin pumps, pacemakers, neurostimulators, electronic pills, drug delivery implants, and prosthetics improve patient outcomes by linking sensors and actuators through closed-loop feedback control (1–11). Systems can achieve greater functionality by distributing additional coordinated, miniaturized nodes throughout the body to synchronously monitor and treat multiple interconnected physiological sites. However, current wireless communication approaches require bulky components that limit device use cases and patient acceptance, suffer from poor intratissue performance, and limit the number of coordinated independent nodes (12).

Interdevice communication commonly relies on energy-intensive radio frequency (RF)-based wireless methods. Commercial Bluetooth Low Energy (BLE) components require a minimum device width of 5 mm and reduce battery lifetimes by up to 90% (table S1). Large batteries and electronic components necessitate either a wearable framework or invasive surgical implantation, whereas devices reduced to <3 mm in diameter can be inserted at an outpatient clinic (13). Near-field communication (NFC) and inductive coupling necessitate precisely positioned epidermal or external transmitters. These limit patient adherence, require centimeter-scale alignment with targeted devices, and are limited by the specific absorption rate. External transmitters are also required for devices that implement more experimental, space-efficient communication methods such as magnetoelectronics (14), optoelectronics (15), and ultrasound (16). As a result, these systems are poorly

suited for deep targets, misalignment-tolerant operation, or multi-site control.

Here, we introduce SWANS, the Smart Wireless Autonomous Networking System. SWANS harnesses the conductive nature of body tissue to transmit communication signals ionically through 30+ cm of tissue, improving tissue communication coverage by >10× compared to RF methods. SWANS circuits can be made entirely of micrometer-scale passive electronic components. This allows devices to achieve microwatt-level power consumptions when in a listening state, enabling year-long device lifetimes. It also allows for millimeter-scale device dimensions, enabling entire systems to be injected through a syringe.

To communicate, SWANS devices emit low-frequency characteristic voltage pulses into tissue that generate transient electric fields across the body. The gradients are received by other devices in contact with the tissue, causing passive transistor switches to briefly turn on when their gate threshold voltage is met. Each device possesses a specific transistor circuit that only responds to its corresponding voltage and pulse length, enabling multiplexing. Through ionic signaling, SWANS enables cross-body bidirectional communication between wearables and implants as well as inter-implant communication.

In-body ionic communication has already been clinically implemented in FDA (US Food and Drug Administration)-cleared devices. One example is Abilify MyCite, which uses ionic conduction to transmit adherence data from an oral tablet to a wearable patch (17, 18). Prior preclinical studies have used ionic conduction to transmit data across tissue, but these systems typically link two devices and rely on megahertz-to-gigahertz signaling with comparatively large and power-intensive electronics (table S2).

To demonstrate the advantages of SWANS multiplexed cross-body network, we characterize our system through several tissue types in ex vivo swine, chicken, and rodent models. Moreover, in vivo in rodents, we engineer a SWANS-enabled network of wearable front-paw sensors and implantable hindlimb neural interfaces to modulate dual hind leg motor control.

In-body ionic communication for triggering bioelectronic therapeutics

SWANS reduces the size of implanted communication components (Fig. 1, C to E) and improves the battery life of implantable devices by using low-power transistor switches to regulate the device's DC output (table S1). SWANS maintains near-zero standby current until the transistor enters its active region, where power draw approaches that of conventional wireless protocols. By minimizing standby consumption, SWANS extends device lifetime by >15× relative to BLE and NFC (Fig. 1, F and G; fig. S1; and supplementary text). Although BLE device lifetimes can be improved by decreasing the passive current draw used for advertisement procedures (19), this approach effectively limits the feasibility of integrating devices into rapid-acting closed-loop systems. By contrast, the SWANS communication structure allows for compact devices with round-the-clock connectivity and effectively no passive current draw.

Each SWANS implant possesses two tissue-interfacing receiving pads connected to the gate pins of a transistor that detect emitted voltage gradient signals in the tissue and facilitate switching. By using different transistor types and passive circuit components, each SWANS implant responds only to characteristic pulse magnitudes, widths, and frequencies emitted by the wearable hub, enabling individual implant control. Here, we demonstrate nine such circuits (fig. S2 and supplementary

¹School of Chemical and Biomolecular Engineering, Georgia Institute of Technology, Atlanta, GA, USA. ²Wearable Intelligent Systems and Healthcare Center (WISH Center), Institute for Matter and Systems, Georgia Institute of Technology, Atlanta, GA, USA. ³School of Electrical and Computer Engineering, Georgia Institute of Technology, Atlanta, GA, USA. ⁴Parker H. Petit Institute for Bioengineering and Biosciences, Georgia Institute of Technology, Atlanta, GA, USA. ⁵George W. Woodruff School of Mechanical Engineering, Georgia Institute of Technology, Atlanta, GA, USA. ⁶Wallace H. Coulter Department of Biomedical Engineering, Georgia Institute of Technology and Emory University School of Medicine, Atlanta, GA, USA. ⁷Department of Materials Science and Engineering, Massachusetts Institute of Technology, Cambridge, MA, USA. ⁸Division of Digestive Diseases, Emory University School of Medicine, Atlanta, GA, USA. *Corresponding author. Email: aabramson@gatech.edu (A.A.); whyeo@gatech.edu (W.-H.Y.) †These authors contributed equally to this work. ‡Present address: Department of Computer Science, Georgia State University, Atlanta, GA, USA. §Present address: Department of Mechanical Engineering, Stanford University, Stanford, CA, USA.

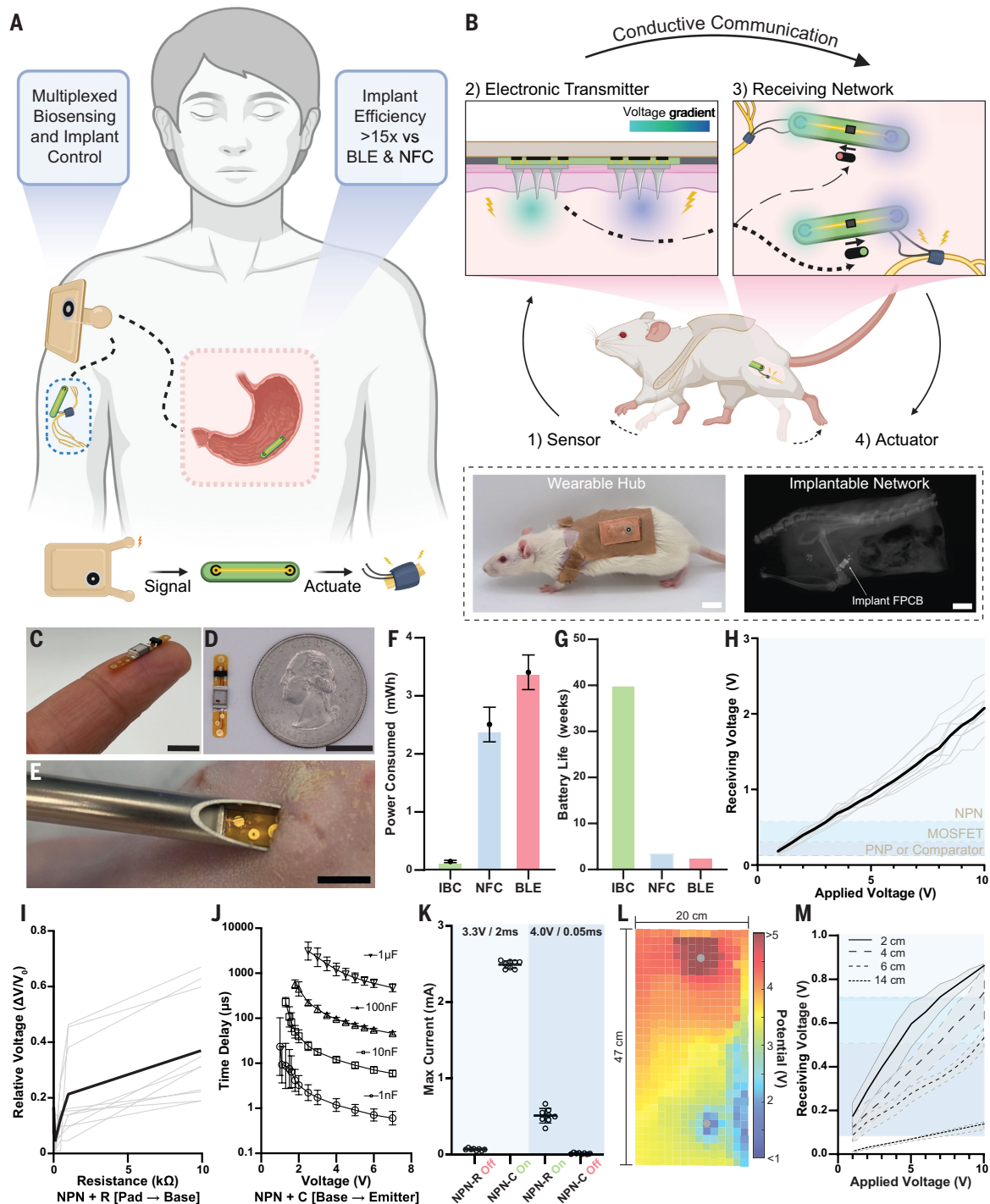


Fig. 1. In-body communication enables low-power networks of wearable and implantable therapeutics. (A) An overview of the Smart Wireless Autonomous Networking System (SWANS). (B) Example of SWANS use case, where wearable sensors detect front limb motion to independently control hind leg motor control. (C and D) SWANS implant (size: 3 mm by 1.1 mm by 17 mm). (E) Syringe implantation of SWANS receiver in rat subcutaneous space. (F and G) Theoretical (bars) and experimental (points, mean $\pm$ SD, $n = 3$ devices) power characteristics of implanted neuromodulation bioelectronics controlled via SWANS in-body ionic communication (IBC), near-field communication (NFC), and Bluetooth Low Energy (BLE). Battery lifetimes were calculated using platform-appropriate batteries (IBC: 26 mAh, 1.55 V; BLE: 20 mAh, 3.0 V; NFC: 20 mAh, 3.3 V). (H) Voltage measured across a transistor's gate when applying increasing amplitude pulses across ex vivo chicken breast tissue. Shaded regions denote threshold voltages (black line represents the average; gray lines represent individual tests; $n = 9$ trials). (I) Adding a resistor between the receiving pad and NPN transistor base (NPN-R) ex vivo in chicken breast increases the required applied voltage for switching (black line represents the average; gray lines represent individual tests; $n = 10$ trials). (J) Adding a capacitor between the base and emitter of an NPN transistor (NPN-C) ex vivo in chicken breast increases the required applied pulse length for switching (mean $\pm$ SD; $n = 6$ to 11 trials). (K) Low-voltage long pulses only trigger NPN-C switches ex vivo in chicken breast, whereas high-voltage short pulses only trigger NPN-R switches ($R = 2.2$ k Ω ; $C = 1$ μF ; individual points, mean $\pm$ SD, $n = 8$ trials). (L) Heat map of generated voltage potentials upon application of pulses (10 V, 1 Hz, 200 μs) using hypodermic needles (gray circles) in ex vivo skin-on, bone-in, pork belly ($n = 3$ trials). (M) Voltage measured across each transistor's gate at different depths in ex vivo pork belly (lines represent mean $\pm$ SD, $n = 9$ trials). Scale bars: (B, C, D), 1 cm; (E), 5 mm.

text). For example, an NPN bipolar junction transistor switches on when the potential difference between the base and emitter approaches 0.7 V, whereas circuits with lower switching thresholds require smaller applied voltage gradients. Distinct switching criteria therefore allow either selective activation of individual implants or, when a single pulse exceeds multiple switching thresholds, simultaneous activation of multiple devices.

We engineered the SWANS system to selectively communicate with multiple implanted devices placed anywhere within the body's conductive tissue by optimizing the voltage magnitude applied to the tissue as well as the separation distance and surface area of the receiving pads. In *ex vivo* studies on chicken breast tissue, the potential difference between receiving pads scaled linearly with the voltage applied by the signal injection probes. This allowed for regulation of multiple transistor types (Fig. 1H). Increasing the resistance between an NPN base and receiving pad increased the pulse voltage magnitudes necessary for switching by up to 30% ($n = 10$, Fig. 1I). Adding a capacitor between the NPN base and emitter increased the required pulse lengths necessary for switching by more than 100% (Fig. 1J). Combining these two findings, we demonstrated that long pulses with low-voltage magnitudes (3.3 V, 2 ms) triggered switches that possess a 1- μ F capacitor, whereas short pulses with high-voltage magnitudes (4 V, 0.05 ms) triggered switches with a 2.2-k Ω resistor. The pulses that turned on one switch did not turn on the other switch, demonstrating the capacity to selectively trigger implants (Fig. 1K).

To modulate switching times in SWANS devices, we designed SWANS implants containing organic electrochemical transistors. These implants actuated at extended timescales compared to NPN transistors after receiving inputs of sufficient pulse lengths and frequencies (fig. S3).

To further increase the number of independently actuatable SWANS devices, we implemented TS8831ST comparators that define discrete voltage windows for switching (0.10 to 0.20 V, 0.27 to 0.38 V, 0.50 to 0.61 V) (fig. S4). Comparator-based windows, compared to transistor-based thresholds, improved multiplexing but modestly increased device size and energy drain.

Body tissue possesses a high conductivity, spreading SWANS voltage gradients throughout the entire volume in all directions. In heterogeneous, skin-on, bone-in *ex vivo* pork belly, a 10-V, 1-Hz, 200- μ s pulse generates a voltage gradient across 30+ cm of tissue (Fig. 1L) and up to 14 cm deep (Fig. 1M). Similar experiments in *ex vivo* chicken breast confirm these findings (fig. S5). Implant communication components as small as 7 mm by 1 mm enabled transistor switching in these tissue models, demonstrating that SWANS can reach sizes much smaller than BLE and NFC systems. The orientation and location of the receiving pads relative to the wearable signal injection probes moderately affect the magnitude of the received voltage gradient. Still, a >0.1 -V gradient was detectable at nonaligned positions >30 cm from the transmission probes. Consequently, the working area of SWANS links is more than $10\times$ the size of that permitted by other wireless methods such as inductive coupling, BLE, and NFC, owing to RF attenuation (figs. S6 and S7).

COMSOL simulations (fig. S8) and equivalent circuit models (fig. S9) provide additional context regarding the effects of pad orientation and spacing on electric field generation in both rat- and human-sized tissue models. We also demonstrated that day-to-day static electricity events, such as discharge from a balloon, had no effect on SWANS actuations (fig. S10).

A platform for low-power, wireless communication between wearables and implantables

The SWANS architecture has three components: (i) wearable hubs that coordinate wearable sensor readouts and therapeutic actuation via in-body ionic communication; (ii) microneedle patches that interface between the wearables and the body; and (iii) syringe-implantable circuits that receive and transmit signals and perform therapeutic

actions (Fig. 2, A and B, and fig. S11). To enable sensor-guided signaling *in vivo*, we engineered a wearable flexible printed circuit board (fPCB) containing interchangeable physical or chemical sensors, a control unit, and a configurable voltage output capable of delivering up to 12 V (Fig. 2C). These signals are introduced into the tissue via stainless-steel micro-needles, bypassing the highly resistive stratum corneum and directly generating a voltage gradient beneath the skin (Fig. 2D). The resulting voltage gradient is then detected by a syringe-implantable fPCB containing two receiving pads, an interchangeable transistor switch, a power source, and an actuator such as a neurostimulator (Fig. 2E). Together, these components form a low-power, end-to-end communication platform, wirelessly linking wearable sensors and in-body therapeutics.

The SWANS wearable hub receives inputs from multiple sensors, processes these data, and transmits 0- to 12-V ionic communication pulses through tissue to control implantable bioelectronics. In a proof-of-concept demonstration, two laser-induced graphene (LIG)-based resistive strain sensors placed on the forelimbs of rats detect motion intent (fig. S12). When either signal exceeds a threshold, the hub delivers a limb-specific signal into the body. This signal selectively activates one of two distally implanted transistor-controlled devices positioned on the hindlimbs, triggering sciatic nerve stimulation in the corresponding leg. This architecture enables selective sensor-guided control over multiple implanted bioelectronic devices.

The length of the implantable fPCB was determined by the receiving pad separation distance required for transistor activation. The width was determined by the battery footprint needed to power the therapy. *Ex vivo* characterization at 4-cm tissue depths showed that 7-mm-long fPCBs enabled at least one independent node, whereas 17-mm-long fPCBs allowed for three independent nodes. In its minimum communication-only configuration, SWANS measures just 1 mm in diameter, the required width of the receiving pad. It is a factor of 5 smaller than the BLE-enabled nRF52832 (20). It is one to two orders of magnitude smaller than many NFC tags, whose operating distance scales with coil dimensions (21). After integration of an EIA1812 battery, the implant measures 3 mm by 1.1 mm by 17 mm and remains compatible with deployment through a 6-gauge needle. Larger battery variants support modular control of additional implantable actuators including pumps and micro-light sources (fig. S13).

In addition to wearable-to-implant links, SWANS enables inter-implant and implant-to-wearable communication. Because our implants respond to specific voltage pulses, we were able to demonstrate an implant actuation cascade. The wearable triggers the first implant at a low voltage. This implant then emits a higher-voltage pulse, which triggers a second implant to actuate (fig. S14A). We can also add sensors to our cascade, enabling wearable and implantable feedback from disparate body locations. For example, we engineered a thermistor-equipped sensing implant to receive signals from a wearable, detect a high-grade fever ($>40^{\circ}\text{C}$), and subsequently send an actuating signal to a second implant. Because the thermistor decreases its resistance at higher temperatures, the implant only switches on to send this relay signal at high temperature (fig. S14B). The same cross-tissue implant signaling approach enables implant-to-wearable communication, providing feedback that confirms implant actuation (fig. S15). Together, this architecture enables full-body networking between wearable and implantable sensors and actuators.

Owing to the highly resistive nature of the stratum corneum, we implemented stainless-steel microneedles to enable direct electric field generation in the subdermal space, thereby increasing the voltage gradient within tissue. Although noninvasive methods such as surface electrodes with conductive gels can reduce skin-electrode interfacial impedance, they require frequent replacement to maintain communication fidelity. Comparatively, microneedles at least 250 μm in height reduce tissue-electrode interfacial impedance by four times and increase generated subdermal voltage gradients for at least 1 week without replacement (Fig. 2, F and G, and fig. S16).

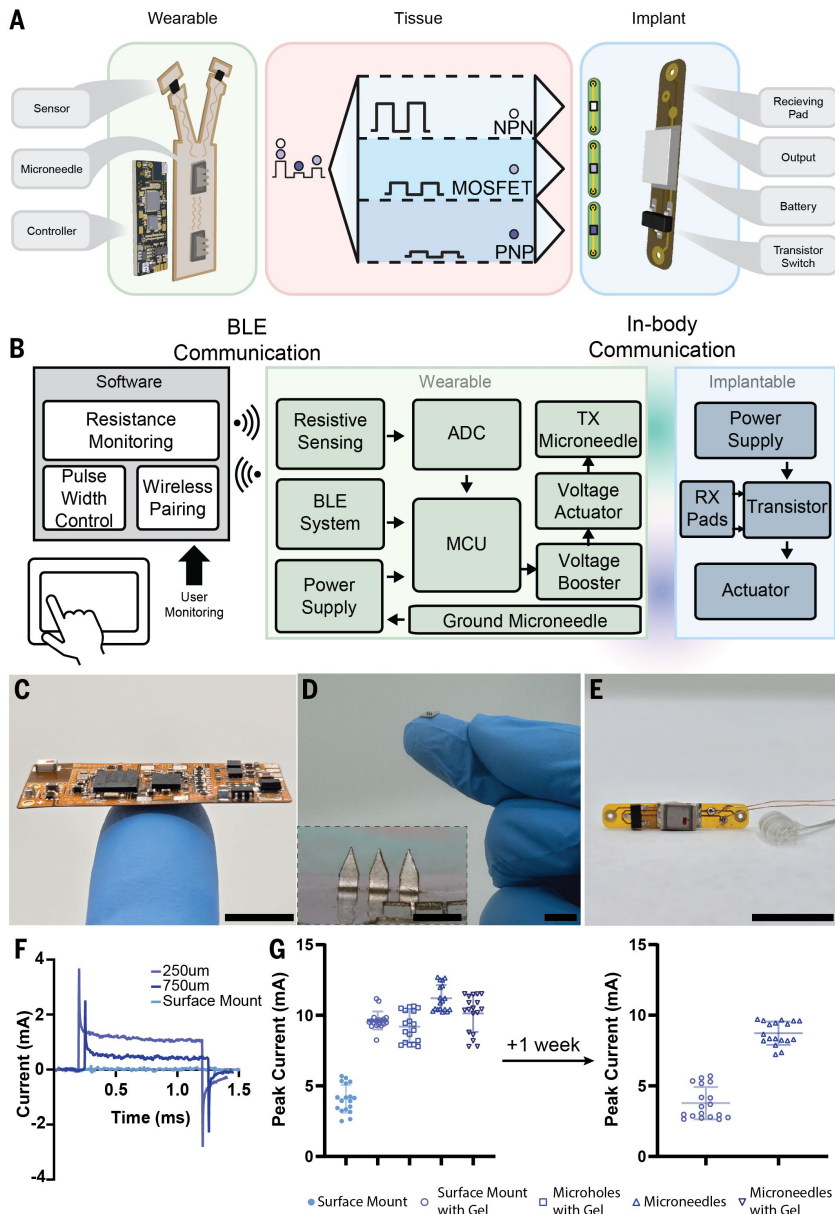


Fig. 2. In-body communication architecture. (A) A custom fPCB continuously monitors a strain sensor and automatically emits a variable (0 to 12V) square wave pulse to trigger the implantable network. Bioelectronics can be switched individually or synergistically. (B) Block diagram for closed-loop control and monitoring of an implantable actuator via in-body ionic communication. Components include analog-to-digital converter (ADC), microcontroller unit (MCU), transmitter (TX), and receiver (RX). (C) Photo of a wearable fPCB. (D) Photo of stainless-steel microneedles that enable the wearable hub's signals to bypass the stratum corneum (inset scale bar, 1 mm). (E) Photo of implantable fPCB with voltage-receiving pads and a nerve cuff. (F) Representative (4 V, 1 Hz, 1 ms) square wave pulses in vivo in rats using skin-interfaced microneedle electrodes with varying needle heights. (G) Peak injection current across various skin-interfaced electrodes in vivo in rats ($n = 18$ trials across three rats, SD plotted). Scale bars, 1 cm.

In vivo SWANS communication fidelity and biocompatibility

In vivo ionic communication in rats mirrored ex vivo trends (Fig. 3, A to C). Implant receiving voltages scaled linearly with applied signal and decreased as the wearable moved further away. Delivery of 7.5-V pulses from the wearable generated 0.87 ± 0.16 V at subcutaneous implants, sufficient to activate an NPN transistor in 26 out of 27 trials (Fig. 3A). PNP, comparator, and metal oxide semiconductor field-effect transistor (MOSFET) devices actuated at lower applied voltages. In the intraperitoneal space, MOSFET implants activated in nine out of

nine trials at 7.5 V, whereas PNP devices activated with less fidelity (Fig. 3B). Receiving pads placed in the stomach enabled PNP transistor and comparator switching at the same applied voltage, highlighting the potential to regulate gastrointestinal-resident electronics via SWANS (Fig. 3C).

SWANS implants successfully actuate in vivo in rats at varying orientations, locations, and depths within the tissue. We observed sufficient voltage gradients to trigger a transistor regardless of the spatial orientations of the transmitter and ground microneedles, including placements horizontally, diagonally, and vertically across the flanks of the rats (Fig. 3D). Current spreads in all directions from the transmitter, creating voltage gradients in distant areas, including the opposite flank of the rat. However, the voltage gradient is maximized when the wearable and implantable modules are superposed on each other, as ohmic losses are minimized.

Hematoxylin and eosin (H&E) and Masson's trichrome staining showed that SWANS stimulation (≤ 10 V, ≤ 10 -ms pulses at 1 Hz) did not produce cellular or tissue damage. Rat skin samples were collected after repeated microneedle insertion and electrostimulation at the same site for ≥ 100 pulses over 10 days, with microneedle replacement at each application (Fig. 3E). We also assessed 10-day chronic 250-μm microneedle implantation without electrical stimulation (fig. S16). Consistent with prior studies of electrical stimulation safety in humans and animal models, the SWANS current, pulse duration, voltage gradient, charge per phase, and charge density of $19 \mu\text{C}/\text{cm}^2$ fall within ranges that do not induce skin, neural, or muscle damage (22–26) (table S6). Low-current electrostimulation (≤ 5 mA, ≤ 10 V, ≤ 10 ms) produced no visible cellular damage or inflammation around the microneedle interface. As a positive control, prolonged stimulation (10 V, 100-ms pulses for 2 hours) produced clear tissue damage relative to experimental samples. Chronic microneedle implantation without stimulation produced localized scar formation consistent with prolonged tissue insertion.

We performed a 2-month chronic deployment of SWANS in vivo in rats to determine long-term safety (figs. S18 to S20) and signal fidelity during changing tissue conditions. Owing to the foreign body response, implant receiver pad impedance increased over time. Despite fibrotic encapsulation up to 1 mm thick around the implant, communication was maintained over the full period by increasing the applied voltage within safe limits until actuation (fig. S18). Rats maintained normal weight and showed no signs of distress over the period (fig. S18). After 2 months, skin, muscle, nerve, heart, and liver tissues showed no visual differences from untreated controls when stained for apoptosis (cleaved caspase-3), proliferation (ki67), or oxidative

stress (4-hydroxynonenal), aside from localized physical trauma associated with the microneedles (figs. S19 and S20). SWANS stimulation parameters (< 10 V, < 2 ms, 1 Hz) also did not alter natural electrical activity in the heart or peripheral nerves (figs. S21 and S22).

In vivo closed-loop neurostimulation

To demonstrate targeted therapeutic signaling across the body, we deployed multiple independent nodes that coordinated to enable bilateral triggerable neurostimulation in rats. Forelimb LIG sensors

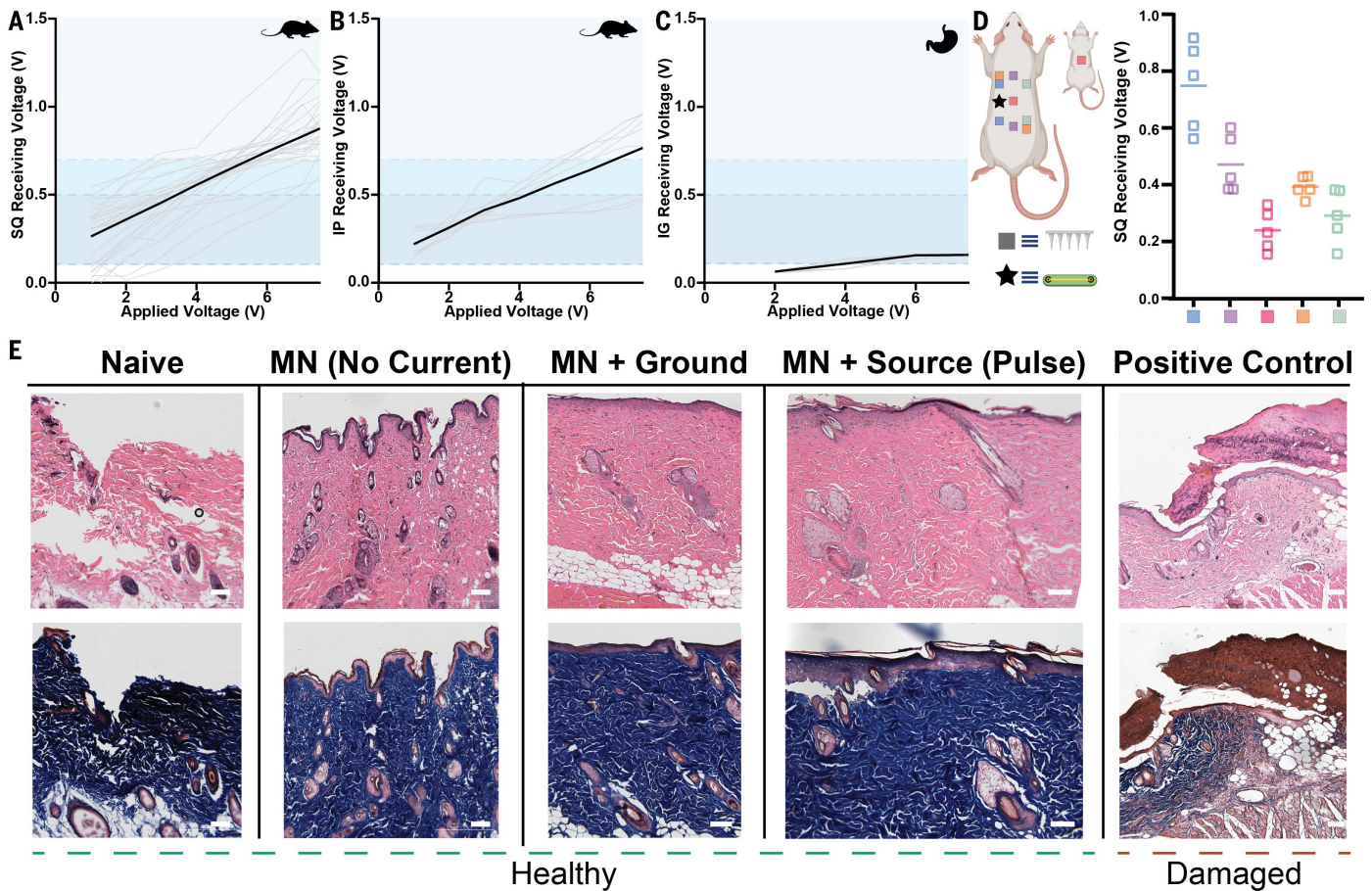


Fig. 3. In vivo IBC generates intrabody voltage gradients without inducing cell trauma. (A to C) Voltage measured across a transistor's gate when applying increasing amplitude pulses across in vivo rat tissue. Receiving pads were placed 2.5 cm apart subcutaneously (SQ) (A) or intraperitoneally (IP) (B), and transmitter and ground probes were placed 4 cm apart above the skin. In (C), receiving pads were placed 1 cm apart inside the stomach intragastrically (IG). Shaded regions denote transistor gate and threshold voltages [black line represents the average, gray lines represent individual tests; (A) $n = 27$ trials across three rats, (B) $n = 9$ trials across three rats, (C) $n = 10$ trials across one rat]. (D) Peak received voltage across receiving pads in vivo ($n = 5$ trials across one rat) when placed at varying orientations to the wearable signal injection probes. Receiving pads were placed 2.5 cm apart underneath the skin. Transmitter and ground probes were placed in spatially varied locations above the skin, including in line with the implant (blue), offset from the implant (purple), on the back and belly of the animal (pink), diagonally across the body (orange), and on the opposite side of the implant (green) ($n = 5$ trials; line represents the average). (E) Representative H&E (top) and Masson's Trichrome (bottom) stained skin samples upon variable electromechanical treatments performed every other day for 10 days (7.5 V, 2 ms, 1 Hz, 20 per day, 900- μ m microneedles (MN)). Scar tissue formation (brown) is noted when applying 10-V and 100-ms pulses for 2 hours as a positive control, while other histology appears healthy. Scale bars, 0.1 mm.

detected motion intent and prompted the wearable hub to transmit limb-specific voltage gradients that independently activated implanted stimulators on the corresponding hindlimbs (Fig. 4A). We confirmed that both NPN and PNP transistor switches on separate implants could be sequentially triggered to deliver sciatic nerve stimulation via our sensor-driven wearable hub ($n = 30$ trials over three rats; Fig. 4, B to D and F). In this setup, both NPN and PNP switches triggered leg motion at the higher-voltage pulses, whereas only the PNP turned on at the lower pulses. Implant activation triggered current flow through a sciatic nerve interfacing cuff, generating hind leg motion. As controls, we recorded that wearable pulses alone without implants or implants alone without wearable pulses had no impact on hind leg motion ($n = 30$ trials over three rats).

We also created two implants that triggered independently of each other. In the first, we added a resistor between the harvesting pad and the base of an NPN implant. In the second, we added a capacitor between the base and emitter of an NPN implant (Fig. 4, E and G; $n = 30$ trials over three rats). Owing to tissue impedance variability between rats, these implants and pulses required individualized settings to ensure

independent actuation. Only the capacitor (47 to 68 pF) implant stimulated leg motion during long pulses with medium voltages (1 to 2 ms, 3.2 to 8.1 V), and only the resistor implant (422 to 825 Ω) stimulated leg motion during a shorter pulse with higher voltages (0.15 to 0.25 ms, 4.1 to 10 V). By forming a bio-integrated network of dispersed wearable and implantable sensors and actuators, the SWANS communication protocol provides a platform technology to enable full-body coordinated control over bioelectronic therapeutic networks.

Discussion

SWANS establishes a framework in which dispersed wearable and implantable therapeutic devices communicate throughout the body via ionic tissue conduction. SWANS uses off-the-shelf components and can be implemented into any bioelectronic device. Compared with other wireless approaches, SWANS requires less power, operates over a larger working area, works without external antennae, reduces device alignment constraints, and is small enough to enable minimally invasive syringe-implantable bioelectronics. SWANS enables multiplexing

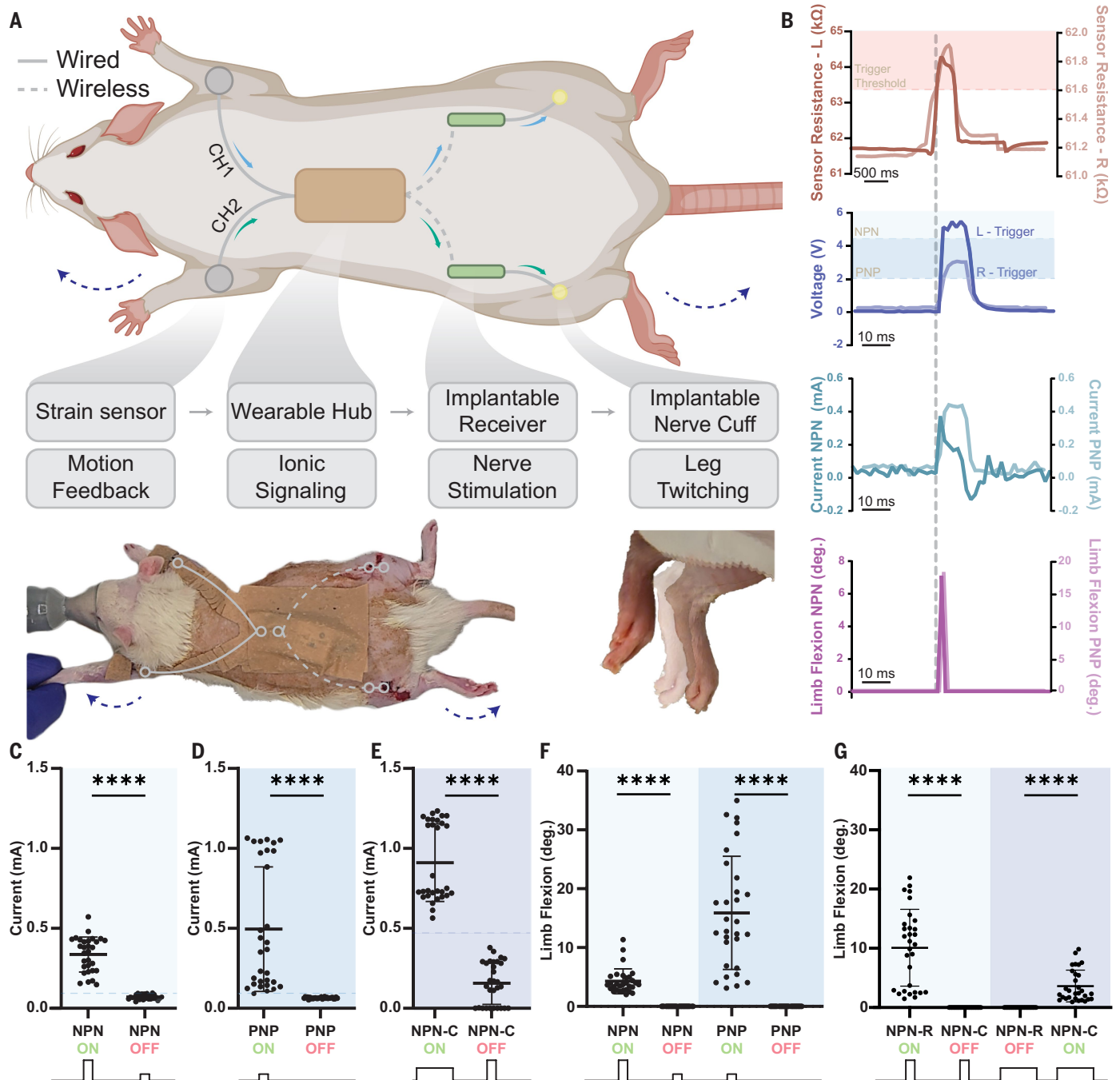


Fig. 4. Bio-integrated electronic networks for neurostimulation. (A) Schematic and image of rat outfitted with SWANS neurostimulator system with representative image of leg twitch from resting (translucent) to flexed (opaque). (B) Upon stimulus detection on either forelimb, the wearable board outputs different pulse magnitudes and widths, which trigger different implants to actuate. (C) Current outputs from NPN implants using high-voltage short pulses, (D) PNP implants using low-voltage short pulses, and (E) NPN + capacitor (NPN-C) implants using longer medium-voltage pulses. Points above dashed lines represent presence of leg flexion. (F) Triggering the NPN and PNP implant combination led to sequential triggering of PNP only followed by NPN + PNP with increasing voltage when placed in the same animal. (G) NPN + resistor (NPN-R) implants only triggered with high-voltage short pulses and NPN-C implants only triggered with medium-voltage long pulses when placed in the same animal. Pulse lengths and magnitudes varied across animals owing to variance in skin thickness and tissue impedance, as described in materials and methods ($n = 30$ trials over three rats with different rats tested in each panel, mean $\pm$ SD plotted; **** $P < 0.0001$ in unpaired t test with Welch's correction).

and logic-gated actuation by incorporating voltage-receiving circuits that respond to specific voltages and pulse lengths by adding resistors, comparators across a device's receiving pads. These components can also be replaced with sensors, allowing nodes to respond differently at various physiological conditions.

Our 2-month safety studies indicate that SWANS pulses do not induce detectable tissue injury under tested conditions. Further investigation is needed to define how repeated low-voltage ionic signaling and microneedle placement influence cellular behavior, tissue remodeling, and electrophysiological responses over longer timescales and

across disease states. Additionally, future experimentation is necessary to define communication performance in large animals and humans, as well as through heterogeneous tissue layers.

SWANS is particularly suited to applications requiring infrequent triggering and low data transfer. Implants maintain near-zero standby power, trading bandwidth for miniaturization and longevity, making SWANS a complementary choice to existing communication architectures.

REFERENCES AND NOTES

1. R. P. Wadwa *et al.*, *N. Engl. J. Med.* **388**, 991–1001 (2023).
2. S. M. Al-Khatib, *N. Engl. J. Med.* **390**, 442–454 (2024).
3. Y. S. Choi *et al.*, *Science* **376**, 1006–1012 (2022).
4. S. Kim *et al.*, *Small* **19**, e2206839 (2023).
5. H. Lorach *et al.*, *Nature* **618**, 126–133 (2023).
6. R. Farra *et al.*, *Sci. Transl. Med.* **4**, 122ra21 (2012).
7. N. Brasier *et al.*, *Nature* **636**, 57–68 (2024).
8. V. Nair *et al.*, *Science* **382**, eabn4732 (2023).
9. P. Hamilton *et al.*, *Seizure* **58**, 120–126 (2018).
10. T. Shu *et al.*, *Science* **389**, eadv3223 (2025).
11. J. M. Cochran *et al.*, *Patient Prefer. Adherence* **16**, 1805–1817 (2022).
12. R. Ghanim, A. Kaushik, J. Park, A. Abramson, *Device* **1**, 100092 (2023).
13. US Food and Drug Administration, “SUPPRELIN LA (histrelin acetate) subcutaneous implant prescribing information” (US Food and Drug Administration, 2011).
14. J. E. Woods *et al.*, *Nat. Biomed. Eng.* **10**, 532–544 (2026).
15. Y. Zhang *et al.*, *Nature* **640**, 77–86 (2025).
16. D. Seo *et al.*, *Neuron* **91**, 529–539 (2016).
17. H. Hafezi *et al.*, *IEEE Trans. Biomed. Eng.* **62**, 99–109 (2015).
18. H. Hafezi, T. Robertson, E. Snyder, B. Cozad (Google Patents, 2015).
19. S. R. Madhvapathy *et al.*, *Nat. Biomed. Eng.* **8**, 1040–1052 (2024).
20. K. Kwon *et al.*, *Nat. Biomed. Eng.* **7**, 1215–1228 (2023).
21. J. Kim *et al.*, *Small* **11**, 906–912 (2015).
22. K. H. Sim *et al.*, *Diabetes Metab. J.* **38**, 120–133 (2014).
23. H. S. Gill, D. D. Denson, B. A. Burris, M. R. Prausnitz, *Clin. J. Pain* **24**, 585–594 (2008).
24. S. F. Cogan, K. A. Ludwig, C. G. Welle, P. Takmakov, *J. Neural Eng.* **13**, 021001 (2016).
25. A. Scheiner, J. T. Mortimer, U. Roessmann, *Ann. Biomed. Eng.* **18**, 407–425 (1990).
26. R. M. Fish, L. A. Geddes, *Eplasty* **9**, e44 (2009).

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SUPPLEMENTARY MATERIALS

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Materials and Methods; Supplementary Text; Figs. S1 to S22; Tables S1 to S6;
References (27–60); MDAR Reproducibility Checklist; Data File S1
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An in-body networking system for communication between wearable and implantable therapeutics

Ramy Ghanim, Yoon Jae Lee, Garan Byun, Joy Jackson, Julia Z. Ding, Jihoon Park, Markella Bibidakis, Elaine Feller, Eugene Kim, Dilay Aygun, Anika Kaushik, Alaz Cig, Sean Healy, Camille E. Cunin, Aristide Gumyusenge, Woon-Hong Yeo, and Alex Abramson

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Editor's summary

Wearable and implantable bioelectronic devices are now used to routinely assess a person's health and, for some conditions, provide interventions as needed. An ongoing challenge is the power and device size required for wireless communication between devices. Ghanim *et al.* designed an electronic circuit that emulates the functionality of a nervous system. The researchers were able to remotely trigger electrical stimulation devices through pulsed signals transmitted through body tissue using less power than protocols such as Bluetooth. Implants could be injected using a syringe, and multiple devices could be controlled by using different signal patterns. —Marc S. Lavine

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