



Electrospun piezoresistive pressure sensor integrated into a soft ball for portable, battery-free monitoring of hand grip strength

Abdulkadir Sanli 

Department of Bioengineering, Imperial College London, London SW7 2AZ, UK

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ABSTRACT

Hand grip strength is a critical biomarker for assessing muscular function and rehabilitation progress, but existing measurement systems are bulky, expensive, and confined to clinical settings, limiting accessibility. We report a portable, low-cost monitoring system using a soft spherical device embedded with a highly sensitive, scalable multiwalled carbon nanotubes (MWCNTs)-enhanced electrospun thermoplastic polyurethane nanofiber sensor. The sensor exhibits a 600 % relative current change at 10 N load and maintains stability over 500 cycles, enabling precise grip assessment. Integrated into an elastomeric ball, the system wirelessly transmits data via near-field communication, allowing real-time smartphone monitoring without batteries. Results confirmed reliable detection of inter-hand strength differences, demonstrating clinical applicability for rehabilitation and home-based care. This scalable, user-friendly platform bridges the gap between lab-grade sensing and practical rehabilitation, offering a versatile solution for personalized biomechanical monitoring.

1. Introduction

Hand grip strength has long been recognized as a reliable and non-invasive biomarker to assess muscular function, overall physical health, and disease progression [1,2]. It is frequently used as an indicator for conditions such as sarcopenia, frailty, neuromuscular disorders, malnutrition, and postoperative rehabilitation progress [3–5]. In geriatric populations and patients recovering from illness or injury, periodic assessment of grip strength is particularly critical for monitoring functional decline and guiding therapeutic interventions [2]. Despite its clinical relevance, current grip strength measurement systems are, however, largely confined to professional healthcare environments, often relying on bulky and expensive digital dynamometers that require trained personnel for operation and interpretation [6,7]. Such limitations hinder routine assessments and prevent seamless integration into home-based or decentralized rehabilitation programs.

Recent advances in flexible electronics [8,9] and smart materials [10] have sparked interest in developing compact, wearable, and user-friendly alternatives to conventional dynamometry. Portable soft devices, in particular, have emerged as promising solutions due to their conformability, lightweight design, and adaptability to diverse rehabilitation needs. For instance, iongel-based sensors have demonstrated excellent stretchability and biocompatibility, making them suitable for wearable applications [11,12]. Challenges, however, remain in

balancing performance, cost, and scalability for widespread adoption, as iongels often face issues such as lower mechanical strength, moisture sensitivity, and slower response times compared to CNT-based alternatives. Several research groups have explored soft pressure sensors and electronic skin platforms for physical rehabilitation monitoring [13–15]. For example, Zhang *et al.* [16] introduce a smart ball sensor using laser-kirigami graphene for personalized grip strength monitoring. The device adapts to different users, provides accurate real-time wireless data, and demonstrates strong performance in tests. The design, however, faces scalability challenges due to its non-replaceable integrated battery, which requires breaking the outer PDMS shell and the graphene-based spiral sensor once the battery is depleted. Li *et al.* [17] developed a flexible piezoresistive sensor based on multiwalled carbon nanotubes (MWCNTs)-decorated electrospun TPU fibrous networks for human motion detection. Compared to iongels or other soft composites, the MWCNTs-TPU fibrous network offers distinct advantages, including tunable mechanical properties, ease of fabrication, and robust piezoresistive response under cyclic loading—critical for repetitive grip strength assessments. While the fibrous structure offers excellent sensitivity (up to 2 kPa⁻¹) and flexibility (10 kPa), the ultra-thin, standalone form of the sensor limits its pressure range and mechanical durability under sustained or complex loading. Many other existing systems [18–21] demand sophisticated electronics, multi-step calibration procedures, and constant power sources, thereby making them expensive,

E-mail address: a.sanli@imperial.ac.uk.

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non-scalable, and impractical for long-term self-assessment. The integration of wireless modules, embedded batteries, and power-hungry microcontrollers complicates the device architecture and increases its weight, limiting its application in daily rehabilitation routines. From a user experience perspective, these systems often lack seamless data sharing capabilities and intuitive interfaces tailored to the needs of patients, caregivers, and clinicians.

In this study, we developed a portable, low-cost hand grip assessment system comprising a soft spherical device embedded with a force-sensitive MWCNTs-enhanced electrospun TPU fibrous network on interdigitated electrodes (IDEs). The system connects wirelessly via an NFC-enabled potentiostat readout circuit, eliminating the need for batteries—users can assess grip strength directly using a smartphone. A user-friendly Android application enables real-time monitoring, exercise logging, and secure data sharing with healthcare providers (e.g., hospitals or physicians) for rehabilitation tracking. A key innovation of our system is its inherent scalability – the electrospun sensing layer can be customized in size, shape, and stiffness, then integrated into virtually any soft device. This enables tailored solutions across the full spectrum of hand rehabilitation needs, from pediatric applications to spastic or arthritic hand conditions. Our proof-of-concept successfully demonstrates sensitive grip strength assessment, reliably detecting inter-hand strength variations. Beyond its immediate application in clinical and home monitoring, this technology establishes a versatile platform for biomechanical sensing. The easily adaptable sensing architecture opens new possibilities for personalized assistive devices and smart rehabilitation tools across diverse therapeutic applications.

2. Results and Discussion

2.1. Synthesis of MWCNTs-enhanced electrospun TPU nanofibers

In Fig. 1a, we illustrate the fabrication process of electrospun TPU nanofibers. First, we dried TPU granules in a vacuum oven at 80 °C for 3 h to eliminate absorbed moisture. We then dissolved 1.5 g of TPU

granules in 10 ml of Dimethylformamide (DMF) solvent and mixed the solution with a magnetic stirrer for 2 h at 60 °C to achieve complete dissolution without additives. For electrospinning, we loaded the homogeneous TPU-DMF solution into a 10 ml syringe mounted on a syringe pump. Using an electrospinning unit (Inovenso NS24XP), we processed the solution with the following optimized parameters: 15 kV applied voltage, 700 rpm collector speed, 1.2 ml/h feed rate, 20 cm needle-to-collector distance, and 2 h spinning duration. During deposition, we programmed the syringe tip to oscillate 40 mm laterally at 20 mm/h while maintaining environmental conditions at $25 \pm 2^\circ\text{C}$ and $60 \pm 5\%$ RH. After electrospinning, we carefully peeled the resultant TPU nanofiber mat from the aluminum foil-covered 10 cm diameter rotating collector and sectioned samples with dimensions 5 cm to 7 cm for subsequent functionalization.

We first prepared a stable dispersion by combining 1 wt% MWCNTs with SDS surfactant solution (density: 1 g/mL) (see Fig. 1b). We sonicated this mixture in a bath sonicator for 1 h and then centrifuged it at 1500 rpm for 30 min to remove large aggregates. We then immersed the electrospun TPU substrates in the supernatant MWCNTs dispersion and applied another 60 min of sonication to enhance nanoparticle adhesion throughout the fibrous network. After functionalization, we thoroughly rinsed the samples with distilled water to remove excess surfactant and dried them in an oven at 60 °C for 1 h to stabilize the conductive network ($\sim 100\ \mu\text{m}$ thick – see SI Fig. S1b). SEM characterization confirmed successful MWCNTs decoration, revealing well-dispersed carbon nanotubes uniformly coating the TPU nanofibers without notable aggregation (Fig. 1b). This homogeneous distribution was critical for achieving consistent piezoresistive response across the functionalized substrates [22,23].

Fig. 1c illustrates the multilayer sensor architecture, fabricated through three sequential steps: First, we screen-printed silver interdigitated electrodes (IDEs) with $\sim 500\ \mu\text{m}$ line spacing on a flexible polyamide substrate to create the conductive readout platform (see Fig. S1c). The measured resistance across each electrode structure was in the range of $0.34\ \Omega \pm 0.06\ \Omega$. Next, we carefully aligned and laminated the sensor

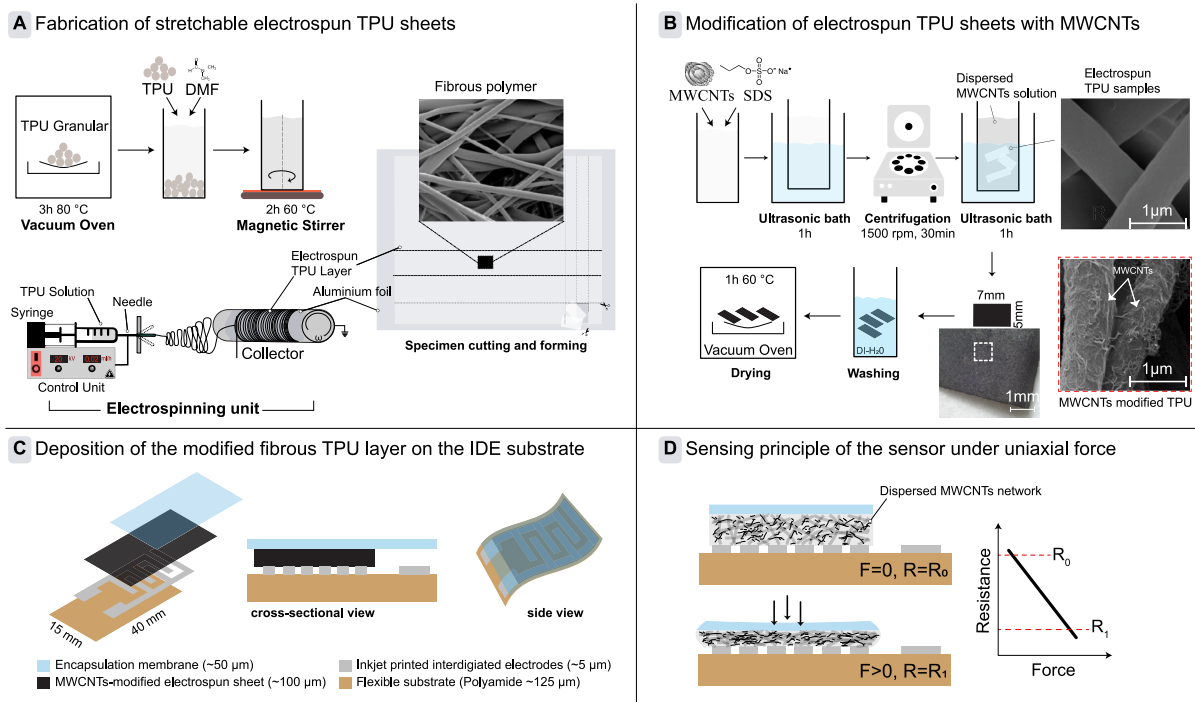


Fig. 1. Fabrication process of MWCNTs-enhanced electrospun TPU nanofiber-based piezoresistive sensor. (a) Electrospinning setup and parameters for TPU nanofiber mat production. (b) Functionalization of TPU nanofibers with MWCNTs via sonication-assisted dispersion and deposition. (c) Multilayer sensor assembly: screen-printed silver interdigitated electrodes (IDEs), aligned MWCNTs-TPU nanofiber mat, and protective encapsulation.

mat (MWCNTs-enhanced TPU nanofiber) ($\sim 1 \text{ cm}^2$ active area) over the IDEs, ensuring complete contact between the conductive fibers and electrode fingers. Finally, we encapsulated the assembly with a $50 \text{ }\mu\text{m}$ -thick encapsulation membrane (polyethylene/acrylic composition) using low-temperature lamination. This protective membrane securely maintains the sensor mat in contact with the IDEs while preserving its mechanical flexibility.

2.2. Characterization of standalone sensor

The standalone multilayer sensor structure was characterized using a customized mechanical test system comprising a Mecniesin Multitest 5-xt (100 N load cell) with a stainless-steel contact pin (7 mm diameter, 38.48 mm^2 contact area) ensuring full coverage of the sensor area (35 mm^2) for uniform force distribution (Fig. 2a). We initially set the test system to apply a uniaxial compressive force from 5 N to 100 N in 10 N increments ($\sim 143 \text{ kPa}$ to $\sim 2.86 \text{ MPa}$), with current responses recorded using an Autolab PGSTAT204 potentiostat in chronoamperometry mode (0.5 V applied potential). The average electrical conductivity of the sensor was calculated from the measured resistance values, revealing an initial conductivity of approximately 0.25 S/m at rest (0 N), which increased to about 0.32 S/m under a 5 N load (see Fig.S2). Through systematic evaluation of acquisition intervals (0.05–0.5 s, see Fig. S3), a 0.1 s sampling rate was selected as optimal for preserving signal fidelity while capturing dynamic response characteristics, enabling comprehensive evaluation of the standalone sensor performance before its integration into the soft ball.

The response of the sensor to uniaxial loading (Fig. 2b) exhibits a pressure-dependent behavior characteristic of its ultrathin ($\sim 200 \text{ }\mu\text{m}$) compliant design. Below 15 N (428.6 kPa), the device demonstrates remarkable sensitivity, achieving over 600 % relative current change at 10 N loading (286 kPa). In this low-pressure range (Region I), the sensor

is highly responsive to incremental loading steps, showing a clear and gradual signal increase with a sensitivity of 49.65 kPa^{-1} and strong linearity ($R^2 = 0.932$). Beyond 15 N (Region II), the response becomes less steep, with a reduced sensitivity of 2.07 kPa^{-1} while still maintaining a good linear fit ($R^2 = 0.9433$), suggesting suitability for a wider dynamic range of pressures (for details, see Fig. S6). This dual-region behavior enables both high-resolution detection at low forces and extended range performance for higher loads. This behavior originates from three coupled mechanisms: (i) strain-mediated reduction of inter-CNT distances enhances quantum tunneling conduction [24–26]; (ii) elastic deformation of the TPU nanofiber matrix promotes the formation of new conductive pathways by forcing MWCNTs into closer proximity [24,27,28]; and (iii) the interdigitated electrode geometry further amplifies these effects by maximizing the interfacial contact area, enabling effective percolation pathways [29,30]. Above the 15 N (428.6 kPa) threshold, the response of the sensor diminishes as the TPU network undergoes complete densification, reaching its mechanical limit. This saturation effect directly reflects the structural constraints inherent in the thin-film architecture, where full compression of the elastomer matrix eliminates capacity for additional deformation-mediated changes in CNT configuration. The transition at 428.6 kPa (15 N) thus defines the operational ceiling for high-sensitivity performance, establishing the device as particularly suited for applications demanding precise low-pressure detection below 300 kPa.

Complementary creep testing (Fig. 2c) demonstrated the ability of the sensor to reliably distinguish 1 N load increments across the 1–15 N range. Under sustained 5-minute constant loading intervals, the signal deviation remained below 6 % at low forces and gradually decreased to 3 % as the load increased to 15 N (see Fig. S5a), reflecting characteristic TPU viscoelastic relaxation while maintaining clear discrimination between adjacent force steps [31,32]. During cyclic durability testing (50 mm/min, Fig. 2d), the sensor exhibited progressive stabilization during

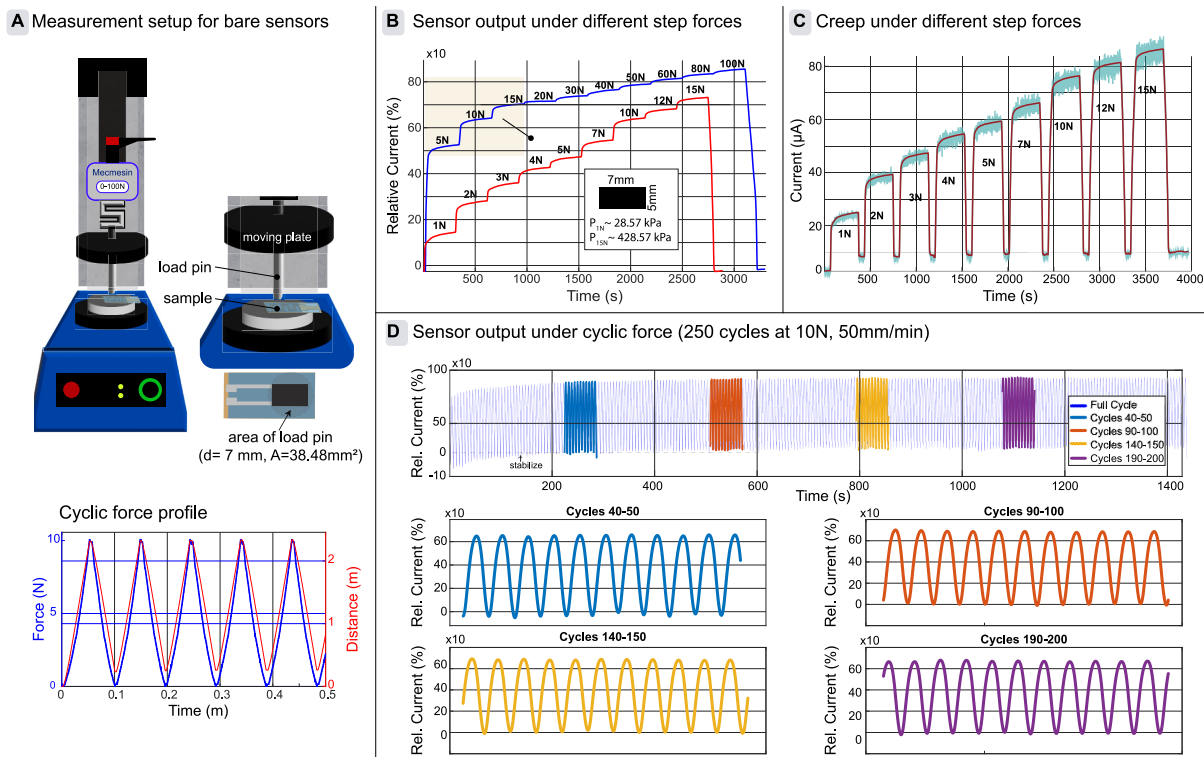


Fig. 2. Standalone sensor performance characterization under mechanical loading. (a) Customized test setup: Mecniesin Multitest 5-xt system with a 7 mm contact pin for uniaxial compression, coupled with Autolab PGSTAT204 for real-time current measurements. (b) Pressure-dependent current response showing high sensitivity ($<300 \text{ kPa}$) and saturation at higher loads due to TPU matrix densification. (c) Creep test results demonstrating stable signal discrimination for 1 N load increments (1–15 N range). (d) Cyclic durability performance (250 cycles) highlighting initial stabilization and consistent piezoresistive response ($\text{CV} = 1.01 \%$), with dynamic loading enhancing sensitivity via rapid MWCNTs rearrangement.

initial cycles (0–25) as the composite microstructure reached equilibrium—MWCNTs networks established consistent conductive pathways, TPU viscoelastic relaxation minimized hysteresis, and electrode interfaces optimized through mechanical settling [33,34]. Subsequent cycles (50–250) showed stable operation with consistent 600–650 % current variation at 10 N loading with reproducible signal waveforms with minimal cycle-to-cycle variation ($CV = 1.01\%$ – see Fig. S3). The enhanced sensitivity observed under dynamic loading conditions is due to two synergistic effects: inertial forces that promote rapid MWCNTs rearrangement during compression and reduced viscoelastic energy dissipation in the TPU matrix at elevated force rates. Dynamic loads facilitate the reorientation of MWCNTs within the TPU matrix, altering the conductive network and contributing to the piezoresistive response of the sensor [35]. Additionally, at higher force rates, reduced energy dissipation allows more efficient energy transfer to the MWCNTs network, enhancing the electromechanical response of the sensor [36]. The 0.1-second sampling interval captures these rapid transitions without signal distortion, confirming the robustness of the acquisition methodology. This sustained sensitivity over hundreds of cycles indicates strong interfacial bonding — that is, robust physical and chemical interactions — between MWCNTs and TPU fibers. Such bonding facilitates effective load transfer and electrical connectivity, resulting in the observed cyclic stability, consistent with previous reports on MWCNT-enhanced TPU sensors [37].

2.3. Integration and characterization of the sensor within a soft ball

To evaluate real-world performance, we embedded the sensor in a 50 mm diameter spherical silicone ball (Ecoflex 00–50) (Fig. 3a, Supplementary Movie 1). Using a Mecmesin Multitest 5-xt system (100 N

load cell), we characterized the dynamic response of the sensor across clinically relevant loading/unloading speeds (Fig. 3b). The sensor reliably tracked motion profiles spanning therapeutic applications, including gentle rehabilitation (25 mm/min), coordination therapy (50 mm/min), and strength training (100 mm/min). At lower loading/unloading speeds, the compliant Ecoflex material enabled gradual force transfer, ensuring accurate signal reproduction. At 100 mm/min, however, we observed signal distortion and offset, suggesting the viscoelastic damping properties of the matrix limit force transmission rates. While switching to a softer elastomer (e.g., 00–30) might theoretically enhance high-speed response by reducing damping [38]. Excessive softness introduces a trade-off: it risks degrading geometric stability and measurement accuracy [39]. Thus, optimizing sensor performance requires a careful balance between material compliance (for dynamic response) and rigidity (for structural integrity) [40,41].

Uniaxial compression testing to 100 N (Fig. 3c) revealed characteristic non-linear deformation of the Ecoflex 00–50 spherical ball. Below 60 N, the sensor demonstrated a linear response ($R^2 = 0.97$) with approximately 120 % relative current change, reflecting effective force coupling through the compliant matrix. The pressure–response behavior of the sensor-integrated soft ball revealed a single linear sensitivity region. In the 0–30 kPa range, the sensor exhibited a sensitivity of 4.54 kPa^{-1} with a strong linearity ($R^2 = 0.96$), indicating reliable performance across a wide pressure range (see Table S1). Pressure values were estimated by dividing the applied force by the projected contact area of the 50 mm diameter ball, taking into account the embedded sensor location within the soft ball structure. The limit of detection (LOD) was calculated to be approximately 11.4 Pa, based on baseline noise analysis, indicating the minimum pressure change the sensor can reliably detect. Beyond this threshold, the system entered a stiffening regime

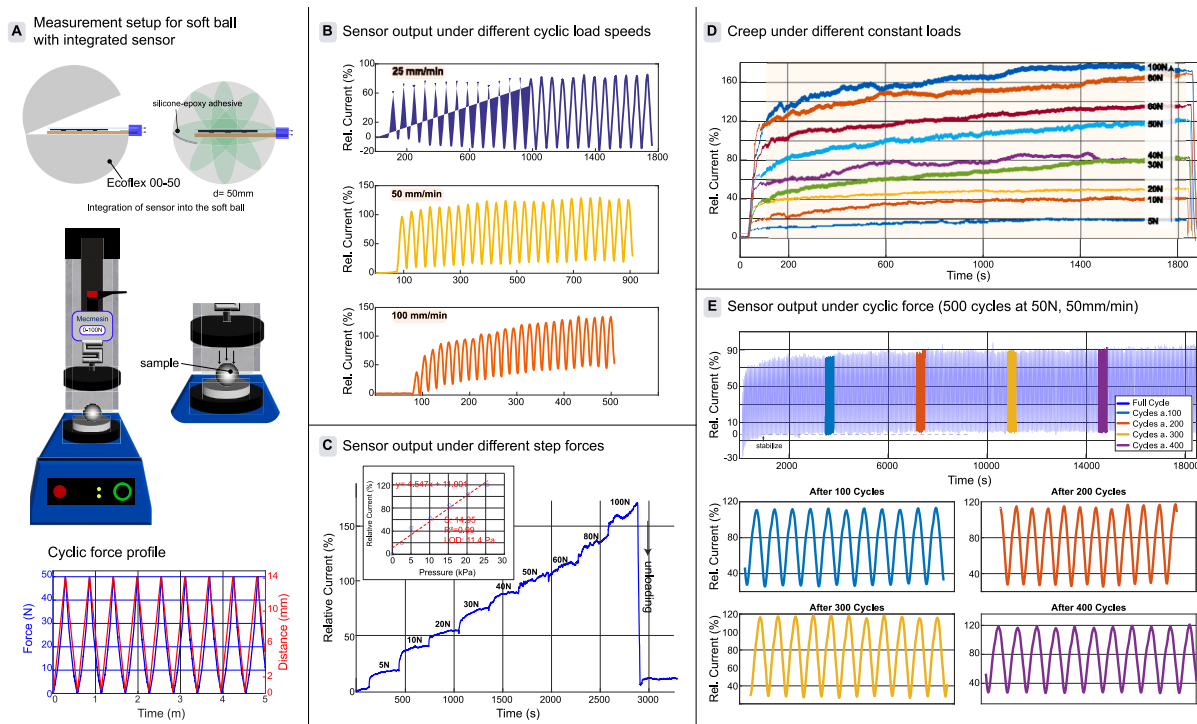


Fig. 3. Sensor integration and performance characterization within a soft Ecoflex 00–50 spherical ball. (a) Experimental setup showing the Mecmesin test system for dynamic compression of the sensor-embedded ball. (b) Motion tracking capability across therapeutic loading speeds (25–100 mm/min), revealing viscoelastic damping effects at higher speeds. (c) Uniaxial compression response (0–100 N) demonstrating linear sensitivity ($<60 \text{ N}$, $R^2 = 0.97$) and stiffening regime at higher loads. (d) Uniaxial compression response (0–100 N), showing a linear region up to 60 N with a sensitivity of 4.54 kPa^{-1} , a limit of detection (LOD) of 11.4 Pa, and a high linearity ($R^2 = 0.96$). Beyond 60 N, the response exhibits nonlinear behavior due to mechanical constraints of the encapsulation. (e) Cyclic durability (500 cycles at 50 N) with stable 80 % current variation ($CV = 2.55\%$), equivalent to 8 weeks of clinical use. Insets show reproducible waveforms at 100-cycle intervals. Bottom panel contrasts embedded vs. standalone sensor performance, highlighting the trade-off between force attenuation (30–40 % strain sensitivity) and mechanical protection enabled by spherical encapsulation.

where the elastomer approached its geometric compression limit—the reduced deformability of the material diminished force transmission efficiency to the embedded sensor [42,43]. This expected behavior for soft silicones manifests as decreasing sensitivity to additional pressure until reaching a stabilized state (> 80 N), where the ball achieves uniform compression [44]. The 60 N transition point thus represents the practical upper limit for proportional sensing, while the stabilized high-force regime remains useful for threshold detection applications.

Creep testing under sustained compression revealed clear time-dependent stability thresholds relevant to therapeutic applications. For forces below 30 N—which encompasses the majority of rehabilitation exercises—the sensor exhibited excellent stability, with signal drift constrained to $< 4\%$ over 30 min. In contrast, compressive loads in the 30–80 N range induced progressively larger signal deviations, reaching up to 14 % (Fig. S5b). These changes arise from two main mechanisms: (i) viscoelastic relaxation of the surrounding Ecoflex 00–50 matrix [45], and (ii) delayed elastic recovery and possible interfacial slippage between the embedded IDE-patterned electrospun TPU nanofiber film and the encapsulating silicone [46]. The integration of the thin-film sensor in the geometric center of the elastomer ball plays a critical role here: while it ensures symmetric load distribution and protection from external shear, it also introduces a compliant buffer that amplifies time-dependent viscoelastic effects under high, sustained loading. Importantly, such prolonged static compression (≥ 30 min) at high force levels far exceeds the loading cycles typically encountered in therapeutic protocols, which are dominated by intermittent, dynamic activities [47]. Therefore, the observed deviations, though measurable, remain clinically inconsequential for short-term strength training applications, and the 30 N transition can be considered a conservative threshold for precision monitoring during prolonged static exercises.

Cyclic testing at 50 N (speed of 50 mm/min) revealed stabilization behavior matching the standalone sensor characteristics (Fig. 3d). After an initial ~ 30 -cycle conditioning phase, the system maintained consistent response levels, demonstrating approximately 80 % relative current change at 50 N loading throughout 500 cycles. Periodic snapshots taken every 100 cycles (insets, Fig. 3d) confirm this stable behavior, showing reproducible signal waveforms with minimal CV of 2.55 % (Fig. S4). This endurance performance, equivalent to approximately 8 weeks of clinical rehabilitation sessions (assuming 3 daily sessions of 20 compression cycles), confirms the robustness of our system for therapeutic applications.

Integrating the standalone sensor within a soft ball fundamentally alters its mechanical coupling and response characteristics compared to standalone configurations. When embedded at the geometric center, applied forces distribute radially through 25 mm of elastomer, significantly attenuating force transmission to the sensor. Experimental data indicate the embedded system achieves approximately 30–40 % of the loading sensitivity observed in standalone sensors under equivalent loading, reflecting both the damping effect of the surrounding silicone and the transition from direct surface loading to volumetric compression mechanics. This reduced sensitivity emerges as an inherent trade-off of the embedded design [48,49]. While standalone sensors demonstrate exceptional responsiveness (600 % current change at 10 N loading), their unsupported $\sim 200\ \mu\text{m}$ architecture cannot withstand practical loading conditions. The spherical encapsulation enables robust operation up to 100 N by distributing stresses through the entire elastomer volume, with only a fraction of the applied surface stress reaching the central sensor region. While this naturally attenuates forces reaching the central sensor, the design maintains clinically sufficient sensitivity across the therapeutic 0–60 N range and demonstrates excellent durability, retaining 80 % response through 500 cycles. The performance characteristics reflect a deliberate optimization for rehabilitation applications, where durability and overload protection outweigh maximum sensitivity. The spherical geometry ensures progressive stiffening under compression, naturally limiting strain transmission at higher forces (> 60 N) while maintaining linear response in the

clinically relevant loading regime.

2.4. Battery-free hand grip strength monitoring system

To demonstrate practical rehabilitation applications, we developed a compact, battery-free hand grip strength monitoring system using near-field communication (NFC) technology (Fig. S1a, Supplementary Movie 2). The portable readout unit (25–35 mm, $< 1/5$ credit card size) centers on an SIC4341 NFC chip (Silicon Craft Technology PLC, Thailand), enabling wireless power harvesting and data transmission through standard smartphones (see Fig. S5). This preliminary implementation could potentially support various clinical applications, including post-stroke recovery monitoring, age-related muscle assessment, and neurological function evaluation through grip dynamics measurements, among others, while eliminating the need for wired connections or battery replacements.

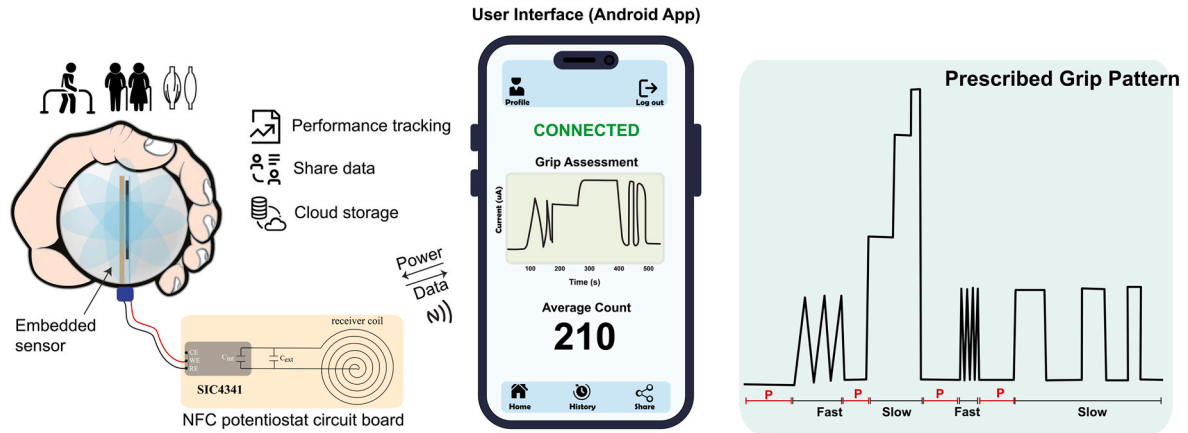
From the hand grip strength measurements (Fig. 4b), we observe that the soft ball effectively tracks the user-prescribed force path and consistently detects a higher strength output from the right hand of the user compared to the left. Notably, the relative current outputs during hand gripping reached up to $\sim 250\%$ for the right hand and $\sim 200\%$ for the left, indicating strong sensor responsiveness. These signal magnitudes are comparable to those observed during uniaxial mechanical bench tests, suggesting that the user exerted forces exceeding 100 N (~ 10 kgf). The similarity in output occurs despite fundamentally different force transmission mechanisms: in the controlled uniaxial compression setup, forces are applied vertically through the sphere's axis, resulting in a more uniform and directed stress distribution (see Fig. S8). For the hand gripping setup, however, the net force is distributed across approximately 47 cm² ($\sim 60\%$ of the surface) through multiple contact points of the fingers, generating complex multidirectional loading. Unlike uniaxial compression, which produces pure compressive stress, natural gripping induces a mix of mechanical actions: finger contact points generate local compression (increasing sensor output), while simultaneously creating tensile, bending, and twisting components in surrounding regions (decreasing output). This dynamic equilibrium of opposing stresses explains the reduced force transmission efficiency to the embedded sensor compared to the controlled bench tests. The spherical design compensates for these mechanical differences in three ways: (1) it transforms high-magnitude grip forces into moderate loads on the sensor through mechanical force distribution, (2) it offers robust sensor protection by dissipating peak stresses, and (3) it maintains measurement fidelity sufficient to detect inter-hand strength variations. While directly comparing the net force between gripping and compression remains challenging due to the complex multidirectional forces involved in hand gripping, the system reliably captures physiologically relevant strength measurements. This is evidenced by the consistent 200 % output difference observed between the left and right hands of the user, demonstrating the capability of the system to track relative strength variations despite the differing force transmission mechanics.

3. Conclusion

We have developed a scalable and portable grip strength monitoring system based on MWCNTs-decorated electrospun TPU sensors integrated into a spherical elastomeric ball. The platform successfully bridges laboratory-grade sensing capabilities with practical rehabilitation requirements through its unique mechanical design, which naturally distributes complex grip forces while protecting sensitive components. The NFC-enabled implementation demonstrates how battery-free operation can enable accessible strength assessment without compromising measurement quality.

While this work establishes a functional prototype for clinical grip monitoring, several key challenges remain for practical implementation.

A Measurement setup and mobile app interface for soft ball sensor



B Output of sensor-integrated soft ball via NFC-enabled android mobile app

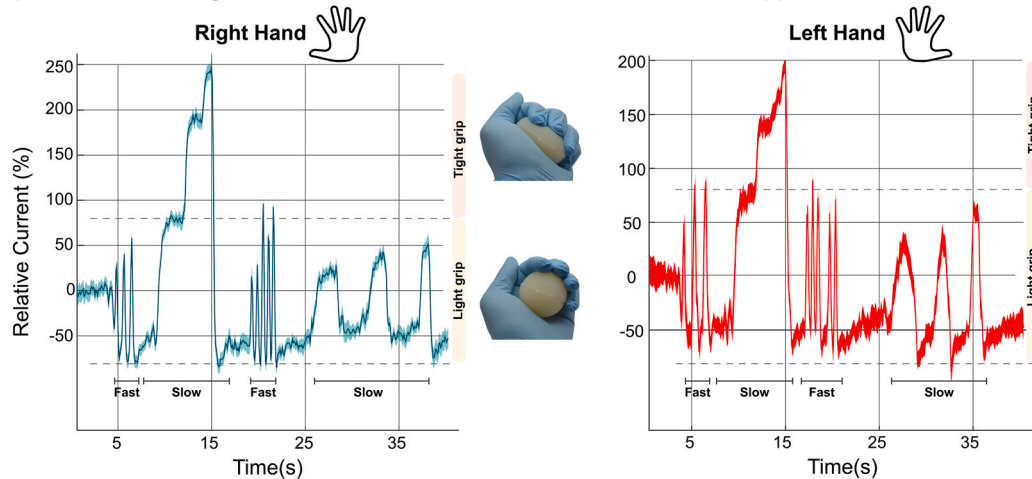


Fig. 4. Battery-free hand grip strength monitoring system and performance characterization. **(a)** Compact NFC-enabled readout unit (25–35 mm, <1/5 credit card size) utilizing an SIC4341 chip for wireless power harvesting and data transmission via a standard Android smartphone. **(b)** Hand grip strength measurements demonstrating accurate tracking of user-prescribed force profiles, consistent detection of left–right hand strength differences (~250 % current variation for the right hand, ~200 % for the left), and strong correlation with controlled uniaxial bench tests ($n = 3$).

- (i) First, the current battery-free design, while eliminating power constraints, still requires physical smartphone proximity for NFC coupling, limiting true wireless operation during exercises. Future iterations could explore hybrid NFC-BLE architectures to maintain power efficiency while enabling continuous monitoring.
- (ii) Second, our single-sensor vertical integration, though effective for basic strength assessment, cannot fully capture the multidimensional nature of hand grip forces. A distributed sensor array would better characterize grip patterns and detect asymmetries, particularly important for neurological rehabilitation.
- (iii) Third, while the spherical geometry provides excellent force distribution, it may not optimally fit all hand sizes or clinical scenarios. Since our sensor is scalable, it can be easily integrated into any soft device, accommodating diverse user populations, including those with hand deformities.
- (iv) A key limitation is quantifying absolute grip forces due to complex 3D contact mechanics. While the system detects relative strength differences well, natural gripping creates competing stresses (local compression increasing output versus surrounding tension/bending decreasing output) across $\sim 47 \text{ cm}^2$ ($\sim 60 \%$ surface area). This differs fundamentally from uniaxial bench

tests, requiring future Finite-element analysis [50] and experimental validation to establish precise force-output correlations.

Beyond these immediate challenges, the core technology presents numerous expansion opportunities. The same sensing principle could be adapted to flatter geometries for integration into rehabilitation equipment or daily objects. With appropriate modifications to the encapsulation materials and electrode configurations, the platform could evolve to monitor other biomechanical parameters like pinch strength or joint mobility. Most importantly, this work demonstrates how thoughtful material selection and mechanical design can overcome longstanding barriers in rehabilitation technology. By prioritizing scalable fabrication and practical usability alongside measurement accuracy, we establish a foundation for truly accessible strength monitoring that could transform both clinical and home-based rehabilitation practices. The remaining challenges represent not limitations of the core concept, but rather clear pathways for continued innovation in this critical healthcare technology domain.

4. Materials and methods

4.1. Materials

Carboxyl-functionalized multi-walled carbon nanotubes (MWCNTs, 92 % purity, outer diameter: 8–15 nm) were purchased from Molchem Technologies (London, UK). Thermoplastic polyurethane (TPU, Elastollan 1185A10, density: 1.12 g/cm³) was supplied by Biesterfeld, Germany, in granule form. N, N-Dimethylformamide (DMF, $\geq 99.8\%$) and sodium dodecyl sulfate (SDS, $\geq 99\%$) were obtained from Sigma-Aldrich and used without further purification.

4.2. Electrospinning unit and material characterization

Electrospun TPU fibers were fabricated using an Inovenso NS24XP electrospinning system (Istanbul, Turkey) under optimized parameters (applied voltage: 15 kV, collector speed: 700 rpm, feed rate: 1.2 mL/h, needle-to-collector distance: 20 cm). The morphology and dispersion of MWCNTs within the TPU matrix were analyzed via field-emission scanning electron microscopy (FE-SEM, Zeiss Supra 55 VP). Samples were sputter-coated with a 5 nm gold layer and imaged at an accelerating voltage of 10 kV and a working distance of 13 mm. Mechanical characterization was performed using a Mecmesin Multitest 5-xt system equipped with a 100 N load cell, with tensile tests conducted at a strain rate of 10 mm/min according to ASTM D638. Chronoamperometric measurements were performed using an Autolab PGSTAT204 potentiostat (Metrohm Autolab) in chronoamperometry mode (0.5 V applied potential, 0.1 s step duration), with data acquisition via Nova 2.1 software.

4.3. Fabrication of interdigitated electrodes

Interdigitated electrodes (IDE) were fabricated on a polyimide substrate (Kapton HN, thickness: 125 μm) using a piezoelectric inkjet printer (Fujifilm Dimatix DMP-2831) with a conductive silver nanoparticle ink (Sigma-Aldrich, sheet resistance: $<0.1\ \Omega/\text{sq}$, viscosity: 10–12 cP at 25 °C). The electrode pattern consisted of 3 interdigitated finger pairs with a finger width of 4 mm, spacing of 0.5 mm, and an overlap length of 3 mm, yielding a total active area of 0.25 cm². Post-printing, the electrodes were sintered at 120 °C for 30 min in a convection oven to achieve a bulk resistivity of $<5 \times 10^{-7}\ \Omega\cdot\text{m}$. Before printing, the Kapton substrate was cleaned with isopropanol and oxygen plasma-treated (Diener Plasma 30 W, 10 min) to enhance ink adhesion. Surface topography and electrode integrity of IDE were evaluated using an optical microscope (Brunel Microscopes Ltd).

4.4. Fabrication of spherical soft ball

Spherical soft elastomeric structures were fabricated using 3D-printed PLA molds (50 mm diameter) designed in Fusion 360 (Autodesk) and manufactured using a Bambu Lab A1 Mini printer (0.4 mm nozzle diameter, print temperature: 210 °C). The fabrication process began with casting Ecoflex 00–50 (Smooth-On, Bentley Advanced Materials) prepared in a stoichiometric 1:1 mass ratio (Part A: 32.5 g, Part B: 32.5 g) into one hemispherical mold cavity. After vacuum degassing (0.1 bar, 10 min) and partial curing (30 min at $23 \pm 2^\circ\text{C}$), the integrated sensor assembly was carefully positioned at the geometric center of the partially cured hemisphere. The structural assembly was completed by precisely aligning and joining the second molded hemisphere using Silpoxy® silicone-epoxy adhesive (Smooth-On Inc., PA, USA), with controlled pressure application to ensure uniform bonding interface formation. The composite structure underwent an optimized two-stage curing protocol: (1) thermal curing at 120 °C for 4 h to establish primary crosslinking networks, followed by (2) 24 h ambient conditioning ($23 \pm 2^\circ\text{C}$, $50 \pm 5\%$ RH) to achieve complete polymerization. The resulting spherical elastomer exhibited a Shore 00–50 hardness profile

(ASTM D2240).

4.5. NFC-enabled wireless data acquisition system

The battery-free wireless sensing platform integrates custom planar antenna design with advanced near-field communication instrumentation. A planar spiral antenna (15 mm diameter, 10 turns) was fabricated on FR4 substrate ($\epsilon_r = 4.3$, $\tan \delta = 0.025$, thickness: 1.6 mm), achieving 250 μm track width with 150 μm spacing (Fig. S7). Impedance characterization using a Keysight E4990A impedance analyzer confirmed the antenna exhibited an inductance of 1.32 μH with a quality factor (Q) exceeding 50 at the target frequency of 13.56 MHz. Resonance was achieved by pairing the SIC4341 NFC IC with a 54-pF tuning capacitor, complementing the inherent 50-pF capacitance of the ICs. The system architecture centers on the SIC4341 NFC Type 2 Tag IC, which provides integrated potentiostat functionality with two selectable current measurement ranges ($\pm 2.5\ \mu\text{A}$ and $\pm 20\ \mu\text{A}$) and adjustable bias voltage from $-0.8\ \text{V}$ to $+0.8\ \text{V}$. This configuration enables precise chronoamperometric measurements while operating entirely passively through harvested RF energy.

Within the soft elastomeric ball, the embedded sensing system employs a two-point measurement configuration implemented through a flexible polyimide-based printed circuit board. The working electrode (WE) and reference electrode (RE) connect to the functional sensor elements, while the counter electrode (CE) electrically couples to the reference electrode (RE), maintaining stable potential control during mechanical deformation of the spherical structure (see Fig. 4 and Fig. S5). Wireless communication operates through NFC Type 4 Tag emulation (ISO 14443-A standard) at 106 kbps data rate. The custom-developed Flutter-based Android mobile application performs three primary functions: real-time chronoamperometric data acquisition via the integrated SIC4341 potentiostat, onboard signal conditioning including baseline correction, and encrypted data transmission to a cloud-based MongoDB Atlas repository. The complete system achieves continuous operation at a maximum $\sim 5\ \text{cm}$ working distance from standard NFC readers while maintaining less than 1 % signal variation during dynamic deformation testing, demonstrating robust performance for clinical rehabilitation monitoring applications.

CRedit authorship contribution statement

Abdulkadir Sanli: Writing – review & editing, Writing – original draft, Visualization, Validation, Project administration, Methodology, Investigation, Funding acquisition, Formal analysis, Data curation, Conceptualization.

Declaration of competing interest

The author declares that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.matdes.2025.114257>.

Data availability

Data will be made available on request.

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