Wearable healthcare devices monitor the condition of patients outside the hospital and hence increase treatment capacity and resources. At the same time, patients benefit from wireless data transmission and conformable device designs, speeding up the rehabilitation process and integration into normal daily activities. Yet, the successful implementation of personalized treatment creates an extensive engineering challenge as devices need to adapt to a vast number of different patients and treatment protocols. Herein, soft building blocks of mobile health (mHealth) devices that reversibly assemble through a magnetic click-on mechanism are introduced. Fabrication of reliable magnetic connectors with an inherent safeguard mechanism allows the realization of personalized wearable mHealth devices, independent of the (desired) measurement technique. Stretchable elastomer-based units combined with imperceptible electrodes are protected from overstretching by controlling the opening force of the magnetic connections. The stretchable devices retain both electrical and mechanical functionality for more than 10 000 opening cycles, on par with the standard for universal serial bus type C (USB C) connectors used in consumer electronics. A fully functional and autonomous pulse sensor wristband, assembled from the reliably connecting circuits, demonstrates the feasible implementation for mHealth devices. The plug-and-play modularity ensures the applicability independent of a patient’s needs, without sacrificing functionality and durability.

Medical monitoring is transforming from bulky stationary hospital machines into small-sized wearable devices for personalized diagnostics, with lifestyle products such as fitness-/activity trackers already on the market. Such gadgets are designed for checking specific body functions, with 24 h monitoring being most insightful, yet difficult to realize. Body temperature, heart rate, respiration rate, and blood pressure (BP) are the four main vital signs routinely measured by medical professionals, yet the needs of individuals often vary related to activities performed. Foot pressure distribution (FPD) measurements can disclose the origins of posture problems, whereas for long-term medical monitoring and surveillance, BP, electrocardiogram (ECG), or electroencephalogram (EEG) measurements (Figure 1A) are crucial. Therefore, mobile and user-friendly devices that can easily be handled by patients themselves are necessary. With rapid developments toward applications such as shapeable magnetoelectronics, on-skin blood flow sensors, and wearable sweat...
sensors, stretchable devices are being developed into a platform for configurable and wearable medical monitoring.\(^{[8–11]}\) A modular approach enables an informed patient to swap functional units with distinct input and output features based on current needs (Figure 1B). Reuse of complex, expensive modules, such as data processing and communication units, combined with the exchange of disposable components, such as sterile or biodegradable sensor patches, pose an excellent route for sustainable management of electronic devices and materials. Bao and co-workers, for example, exploit self-healing materials to achieve modularity of on-skin stretchable electronics, yet their approach still requires the design process to be done in advance, and does not offer plug-and-play modularity able to be performed by patients themselves.\(^{[12]}\) We here introduce safe magnetic connectors enabling the development of modular stretchable electronic appliances with an integrated safeguarding mechanism to protect the strain sensible parts from mechanical failure. Conventional modular electronics often use clamp or plug-and-socket connections. Yet, magnetic links are already well established in commercial or industrial applications, albeit limited to rigid devices. They are, for instance, used as power connectors in consumer laptops, being entirely rigid, as well as in the assembly of soft robots, microfluidic devices, and now also enter the field of medical electronic systems.\(^{[3,13,14]}\) Furthermore, the utilization of liquid metals with both electrical and magnetic properties, although still in early stages, poses a promising route of incorporating magnetic connections into stretchable devices.\(^{[15,16]}\) In this work, we demonstrate the feasibility of magnetic connectors to link individual functional units for mobile health (mHealth) applications both electrically and mechanically based on stretchable electronic circuits. In addition to medical applications, our connectors may also significantly simplify incorporation of sensors and data transmission units into soft robots, allowing smart stand-alone solutions.\(^{[17]}\) In our approach, we fabricate stretchable modules consisting of an elastomer matrix with embedded magnets and electronic circuitry rendered stretchable by wrinkling on top (Figure 1C). The magnets in the elastomer establish both mechanical and electrical connections between individual units with adjustable hold/release force (Figure 1D). This enables easy fabrication of diverse task-specific units, including power sources, sensors, and communication modules with an inherent safeguard against overstretching of sensible electronic parts. We illustrate the design process to achieve at least 10,000 opening and closing cycles, the fabrication process of the safe magnetic connectors, and a proof of concept with a modular mHealth pulse sensing wristband. The full device consists of three individual modules with distinct tasks. A custom-made pulse sensor exploits the optical reflection characteristics of blood, whereas a second unit hosts a microcontroller with an innate Bluetooth Low Energy (BLE) module acquiring sensor data and transferring it to any Bluetooth compatible device (e.g., mobile phone). A third module is implemented as power supply for the pulse and communication module (Figure 1C).

The use of magnetic over stiff plug-and-socket connectors allows frugal embedding into a stretchable matrix, maintaining the soft surface properties of a device. Moreover, they provide a zero-force mating capability and prevent misconnections. This design concept is demonstrated by two modules: one holding a sensor another one housing a communication unit. In both units, the islands that encapsulate rigid parts are fabricated from an elastomer with a higher Young’s modulus (polydimethylsiloxane (PDMS)) than the interconnecting bulk material (Ecoflex 00-30; Figure 2A). This rigid island approach with modulus graded elastomers reduces the stress on critical parts and hence avoids unfavorable movement of sensors.\(^{[18]}\) Finite element simulations and a prototype demonstrate the beneficial effects of this design (Figure 2B,C). The sensor/communication islands are

Figure 1. Body sensors and modular unit concept. A) A multitude of on-body sensors is available for monitoring body functions such as BP, electromyography (EMG), EEG, ECG, or FPD. B) Plug-and-play structure consisting of sensor, power source, communication, and signal processor unit. C) A fully functional demonstrator of a soft mHealth wristband for pulse monitoring, consisting of three units. The sensor unit for pulse measuring is shown in the background, the unit including microprocessor with integrated BLE module in the foreground and the power supply unit is mounted behind the BLE unit. D) Schematic of two modular units linked via the magnetic connector.
free from strain or stress under load and consequently are best suited for positioning rigid electronic components (Figure 2D, E).

Apart from ensuring the protection of the rigid modules, a continuous operation of electrodes is equally crucial. Magnetic connectors allow for the implementation of a reliable fail-safe for overstretch due to their inherently defined hold/release force, ensuring tight affixations of mechanical and electrical linkage. We integrate magnets into both ends of elastomer-based modules to enable both mechanical interconnection and electrical contact between two opposing imperceptible electrodes (Figure 2F). These electrodes are rendered stretchable by placing metallized ultrathin (1.4 μm) polyethylene terephthalate (PET) foils on prestretched elastomers.[19] Releasing the prestrain causes the electrodes to form out-of-plane wrinkles perpendicular to the stretching direction, thus creating a reversibly stretchable system up to the 75% prestrain (Figure 2G). The wavelength of the wrinkles is estimated with a buckling model (Equation (S1), Supporting Information) considering a stiff, thin-film layer on...

**Figure 2.** Design process of stretch-safe magnetic connectors. A) Finite element simulation of a stretchable unit comprising a RFDuino encapsulated with a stiffer elastomer placed in the middle of the communication module and of a pulse sensor module connected by magnets. B) Simulation results indicate no strain at the stiffer islands holding either the magnets or the electronic components. C) Demonstrator of the simulated module assembly. D) Stress simulation results of the assembly. E) Stress simulation indicates no significant amount of stress has to be taken up by the magnetic island connectors. F) Concept of stretch-safe magnetic connectors including imperceptible electrodes. The connector is designed to release when the pulling force reaches a critical point, set to be below the force required to impair the electrodes. G) Stretchable copper electrode (inset) on Ecoflex 00-30 substrate with embedded magnets at each side is clamped in its relaxed state to strong magnets hindering snapping. H) Stretching of the sample beyond the prestrain results in the wrinkles flattening in stretching direction. In addition, Poisson compression results in wrinkles perpendicular to the stretching direction (inset). I) Stretching of the sample far beyond its prestrain yields electrode rupture after plastic deformation. J) Design scheme for determining the required intermagnetic distance for reliable release of the magnetic connection at a defined sample strain. The stress–strain relation from a given elastomer and geometry (blue) gives insight into the maximum allowed tensile force for a maximum allowed strain. When the magnetic force equals the tensile force of the elastomer, the contact starts to release, allowing determination of the therefore required intermagnet distance (purple). The green shaded area marks the operational range for a sample designed to open at 30% strain, as for higher strain levels the magnetic force is exceeded by the elastic force. K) Tensile force and resistance measurement of a sample with a substrate prestrain of 50% being attached to a steel plate and stretched until the snapping release occurs. Photographs of the mechanism demonstrated by manual pulling are given as insets. L) Normalized resistance of samples being cyclically stretched to 50% without release and cyclic snapping cycles with release at 30% strain.
top of a thick, elastomer substrate. Stretching an imperceptible electrode beyond the innate prestrain results in rupture. Up to an elongation equal to the prestrain, the metal electrode unfolds and ultimately flattens out. In this regime, the tensile force is dominated by the elastic properties of the elastomeric substrate and the electrodes electrical resistance stays practically constant (Figure S1, Supporting Information). Along with the flattening of the electrode at increasing electrode strain, electrode stiffening occurs and transforms into the major contributor of the total tensile force (Equation (1))

$$F_{\text{total}} = F_{\text{elastomer}}(\varepsilon) + \Theta (\varepsilon - \varepsilon_{\text{pre}}) F_{\text{electrode}}(\varepsilon_{\text{electrode}})$$  \hspace{1cm} (1)

with

$$\varepsilon_{\text{electrode}} = (\varepsilon - \varepsilon_{\text{pre}})(1 + \varepsilon_{\text{pre}})^{-1}$$  \hspace{1cm} (2)

where $\varepsilon_{\text{pre}}$ is the prestrain of the elastomer, $\varepsilon$ the actual strain, $F_{\text{elastomer}}$ the force contribution of the elastomer substrate, $F_{\text{electrode}}$ the force contribution of the imperceptible electrode, the total tensile force $F_{\text{total}}$, and the Heaviside step function $\Theta$.

The elongation and plastic deformation of the PET film with deposited electrodes on top occurring beyond the prestrain is accompanied by macroscopic crack-induced electrode failures. Electrode resistance increases steeply as a result. In addition, wrinkles perpendicular to the stretching direction start to appear due to Poisson compression of the elastomer (Figure 2H). Ultimately, stretching the electrode over the PET necking region leads to the rupture of the electrode (Figure 2I). However, with our magnetic connections, we demonstrate a reliable safeguard mechanism, preventing overstretching of the electrodes and therefore paving the way toward the development of safer and longer operation of mHealth devices.

Considering the magnetic force and the stress–strain characteristics of the elastomer, the snapping point, i.e., the elongation at which the magnets release, is tailored to prevent the rupture of the imperceptible electrode (Figure 2J). The critical parameter for design considerations is the maximum strain that the imperceptible electrode endures. The corresponding force at which the connectors are to be released depends on the stress–strain relation of the substrate–electrode combination and is modeled according to Equation (2) by applying the gent model for $F_{\text{elastomer}}$ and using $\varepsilon > 80\%$ as prestrain. For a given set of magnets, the magnetic force between them is related to their intermagnet distance and was calculated using finite element modeling, as well as a simulation demonstrating the field distribution (Figure S3, Supporting Information), using a friction coefficient of 0.2 for PET.[21,22]

To ensure a reliable safeguarding mechanism, the magnetic interconnects should snap at a strain below the electrode prestrain. Therefore, the intermagnet distance is chosen in such a way that the magnetic force is higher than the elastic force up to the desired release strain but is not sufficient for higher strain levels, which leads to immediate release of the magnetic connections. The green-shaded area marks the region for a sample releasing at 50\% strain, and the required intermagnet distance being 1.3 mm. Dotted lines indicate the path to determine the required intermagnet distance (0.6 mm) to design a connection releasing at 50\% strain instead of 50\%.

The force contributions of both the PET–copper electrodes and the elastomer substrate are demonstrated in a snap-off test, with a sample exhibiting 50\% prestrain (Figure 2K). Up to this prestrain, the wrinkles are unfolded, whereas the tensile force (blue) is dominated by the elastomeric substrate. The electrical resistance (red) during that time is virtually constant. Between 50\% and 53\% strain, the electrode becomes fully extended resulting in a steep increase in the force due to the contribution of the PET substrate. The magnetic connectors start to slide apart, leading to a reduced contact area of the electrodes, with an accompanying increase in resistance. Ultimately the magnetic connector releases (Figure 2K, insets), the force vanishes to zero, whereas the resistance increases to infinity.

For the daily use of a connector in mHealth applications, reliable operation for thousands of stretch cycles is indispensable. Long-term properties of the electrodes and magnetic connectors are demonstrated by cyclic stretching without release of joints, and by repeated disconnect experiments. Therefore, a specimen (50% prestrain) is cycled between 0\% and 50\% strain, preventing an opening of the magnetic joints (Figure 2L). A change in electrode resistance of only 10\% after 10 000 stretching cycles confirms that the hard magnets have no influence on the long-term stability of the electrodes.[19] Electrode lifetime, simulating a typical life cycle of everyday applications, is investigated by measuring 10 000 connect–disconnect cycles of a magnetic joint, designed to open at 30\% strain. Herein, the resistance increases by 46\% after 5000 connect–disconnect cycles and by around 160\% after 10 000 cycles (Figure 2L), with the device still being operable with no critical failure. These stretchable connectors can operate up to 5 years with five closings and openings per day. The performed 10 000 connect–disconnect cycles correspond to the mechanical requirement of universal serial bus type C connectors, which render our magnetic connectors more than suitable for mHealth devices.[23] Future research directed at combining our technique with other sensor approaches, such as thin film electronics, combined with miniaturization of the connectors could lead to a new class of reusable, unobtrusive medical monitoring equipment, entering the daily life of users.

Based on the demonstrations of reliable operation and long lifetimes of our devices, we developed a mHealth pulse sensor wristband consisting of three modules with distinct tasks. A sensor unit features a custom-made pulse sensor circuit on a flexible printed circuit board gauging the optical reflection of blood (Figure 3A). Being in contact with the skin, it illuminates the upper skin layers with infrared light and measures the back-reflected light via a photodiode sensor. The acquired voltage signal contains a combination of a constant part, originating from artery, tissue, and veins and a pulsatile contribution, originating from the pulsating blood. The signal is amplified and the constant contribution and higher frequency noise is cut off by a two-stage active first-order band pass (−3 dB at 0.7/2.34 Hz, gain 10 000). The filtered and amplified signal is fed into a 10-bit analog/digital converter (ADC) embedded in an advanced RISC (reduced instruction set computing) machine (ARM) microcontroller. This control unit (RFduino RFD22301) also features a BLE module that transmits the acquired data in real time to a mobile phone, where it is plotted and logged. Wireless data transfer proves to be advantageous due to less restrictions during movement, with BLE providing the desired parameters required for mHealth applications.
Advantages over near-field communication (NFC) are its longer range, less power consumption, and smaller chip area, which is favorable for stretchable devices. Although recent advances in directly stretchable circuits are promising, the higher integration density provided by conventional integrated circuits (ICs) is still favorable. Both modules, sensor and control unit, are powered from a third module, housing a CR927 coin battery with a voltage of 3 V and a capacity of 30 mAh (Figure S2, Supporting Information). The Bluetooth chip inflicts a power consumption of 4–12 mA, depending on the desired range, and ~5 mA is needed for the sensor board. Therefore, the entire wristband with sensor and BLE module is powered with an off-the-shelf coin cell battery. The active elements of all three units (sensor circuit, microcontroller, and coin cell battery) and the magnets are embedded in a PDMS shell (E = 1.9 MPa) acting as rigid region compared with the soft Ecoflex straps (E = 55 kPa), which take up all the strain upon deformation. Electrical contact from the active devices to the stretchable electrodes was established with conductive epoxy. Once the units are connected and communication to the mobile phone is paired, the pulse sensor is read every 50 ms, with the data being transferred to the Android application executed on a mobile phone, displayed on a graph, and logged (Figure 3B). To test the pulse sensor under real conditions, a volunteer wears the wristband during a physically demanding workout. Before the workout, a rest pulse of 77 bpm is monitored, a 3 mm-thick PET covered with a thin PDMS adhesion layer). A 3 nm Cr adhesion layer was evaporated at a rate of 0.2 nm s⁻¹, and subsequently a 100 nm Cu layer at a rate of 3 nm s⁻¹. Laser cut shadow masks (120 μm polyimide[PI], DuPont) were used for electrode structuring. The elastomer (Ecoflex 00-30, Smooth-On) components were mixed in 1:1 ratio, degassed, and afterward injected into the mold. It was then cured for 24 h at 60 °C. After curing, the specimen was prestrained and the release mechanism. Our stretchable pulse sensor demonstrator operates in a plug-and-play manner, exhibiting the modularity and durability as well as the user-friendliness required for personalized, wearable mHealth applications.

Experimental Section

Fabrication of Imperceptible Electrodes: Thermally evaporated chromium–copper (Cr–Cu) metal bilayer structures on 1.4 μm-thick PET foils were used as imperceptible electrodes. To ease the handling, the 1.4 μm-thick PET foils were temporarily attached to a stiffer support prior to evaporation (125 μm-thick PET covered with a thin PDMS adhesion layer). A 3 nm Cr adhesion layer was evaporated at a rate of 0.2 nm s⁻¹, and subsequently a 100 nm Cu layer at a rate of 3 nm s⁻¹. Laser cut shadow masks (120 μm polyimide[PI], DuPont) were used for electrode structuring. The magnets were embedded into a cylindrical PDMS (Sylgard 184, 10:1, Dow Corning) 1 mm-thick shell using cast molding. The shells were precured for 30 min at 60 °C and afterward placed at their designated positions inside a mold onto a VHB 467 MP (3M) adhesive inlay. To allow for better bonding to the elastomer, the VHB adhesion inlay was poi treated with a primer (OS 1200, Dow Corning). After a 60 min drying process, the magnets were embedded into a cylindrical PDMS (Sylgard 184, 10:1, Dow Corning) 1 mm-thick shell using cast molding. Cast molds were prepared by fused deposition modeling with a Makerbot 2X printer using acrylonitrile
butadiene styrene (ABS) filament with a layer height of 0.15 mm. Again, VHB 467 MP was used as adhesive layer for the electrodes, prepared by laser cutting (Speedy 300, 100 W, Trotec) and treated with primer (1200 OS, Dow Corning, spin coated at 150 rpm for 30 s, 60 min drying). Sylgard 184 was prepared by weighing prepolymer and cross-linker in a 10:1 ratio into a polypropylene (PP) beaker, stirring with a sterile spatula for 5 min and subsequent degassing in a desiccator. Ecoflex 00-30 was prepared by weighing the two components in a 1:1 ratio, 3 min stirring with a sterile spatula and subsequent degassing.

Electronic Island Specimen Fabrication: Magnets, electrodes, and adhesion layer were manufactured, as described in the aforementioned section. The electronic components were casted into a Sylgard 184 island and cured for 30 min at 60 °C using mold casting. Before placing, the electrodes on the prestrained elastomers through holes were made into the electrodes, which enabled electric connection of the active devices to the imperceptible electrodes via conductive epoxy (CW2400, Chemtronics). The pulse sensor unit hosted a combination of infrared light-emitting diode (IR-LED) and photoelectric sensor (TCRT1000, Vishay). Driven by ≈5 mA, the IR-LED illuminated the tissue beneath it, whereas the photoelectric sensor picked up the pulse-modulated reflected light and was followed by a two-stage active first-order band-pass filter with an amplification of 101 for each stage. The cutoff boundaries (–3 dB point) for pulse detection were set to 0.7 and 2.34 Hz corresponding to a heart rate of 42 and 142 bpm. To allow bending of the sensor unit, it was fabricated onto a flexible printed circuit board (50 μm P, 35 μm Cu), structured by standard wet-etching technology and conventional soldering techniques. Electrical connection to the imperceptible electrodes was again established with conductive epoxy. The communication and control unit consisted of an RFDuino microcontroller (RFDuino RFD22301, RFDigital) with a 16 MHz ARM Cortex-M0 CPU, integrated BLE module, 10 bit ADC, and ready to use Arduino compatible bootloader. Programming was conducted with the Arduino platform. Once power is supplied to the microcontroller, it waits until the mobile phone sends a Bluetooth pairing request, followed by establishing the connection to the host. If the connection stands, the microcontroller samples the voltage signal from the pulse sensor board every 50 ms, and sends the acquired value to the mobile phone. Here, an Android app (programmed in Android Studio 7) took up the sensor readings, saved them into a logging file, and displayed the data in a progress plot. The power unit hosted a CR927 button cell with a specified capacity of 30 mAh. The battery was connected with conductive epoxy to flexible electrode flaps which were contacted to the imperceptible electrodes as described previously.

Simulation: Mechanical steady-state finite element simulations were conducted by using linear material models in Autodesk Inventor 2014. Simulations of magnetic fields were conducted with the commercial software COMSOL. With the relative permeabilities of copper, PET foil, and PDMS being close to that of air, the model had been simplified with a multimagnet system surrounded by an air space. The cylindric magnets with radius 100 mm (20 times that of magnet radius) had been used. Simulation and control unit consisted of an RFDuino microcontroller (RFDuino RFD22301, RFDigital) with a 16 MHz ARM Cortex-M0 CPU, integrated BLE module, 10 bit ADC, and ready to use Arduino compatible bootloader. Programming was conducted with the Arduino platform. Once power is supplied to the microcontroller, it waits until the mobile phone sends a Bluetooth pairing request, followed by establishing the connection to the host. If the connection stands, the microcontroller samples the voltage signal from the pulse sensor board every 50 ms, and sends the acquired value to the mobile phone. Here, an Android app (programmed in Android Studio 7) took up the sensor readings, saved them into a logging file, and displayed the data in a progress plot. The power unit hosted a CR927 button cell with a specified capacity of 30 mAh. The battery was connected with conductive epoxy to flexible electrode flaps which were contacted to the imperceptible electrodes as described previously.

Reliability Snap-Off Test: All tensile tests were conducted on a horizontal self-built linear stage driven by a stepper motor and a low clearance drive shaft. The position of the clamps was logged via the driven steps of the motor and the tensile force was acquired with a 50 N force gauge (Alluris FMI-2208S) connected to a National Instruments DAQ card. Electrical resistance was measured by four-wire gauging with a digital multimeter. A custom-made program was used to control the tensile tester and the measurement sequences and to acquire, display, and log all data. All tensile tests were performed with a constant elongation velocity of 1 mm s⁻¹.

Reliability Snap-Off Test: The linear stage was arranged vertically and a linear stage driven by a stepper motor and a low clearance drive shaft. The position of the clamps was logged via the driven steps of the motor and the tensile force was acquired with a 50 N force gauge (Alluris FMI-2208S) connected to a National Instruments DAQ card. Electrical resistance was measured by four-wire gauging with a digital multimeter. A custom-made program was used to control the tensile tester and the measurement sequences and to acquire, display, and log all data. All tensile tests were performed with a constant elongation velocity of 1 mm s⁻¹.

Supporting Information
Supporting Information is available from the Wiley Online Library or from the author.

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Conflict of Interest
The authors declare no conflict of interest.

Author Contributions
G.K., S.B., and M.K. conceived the research project; G.K. and D.D. manufactured the devices; G.K. and M.D. designed the experiments; G.K., M.D., and D.D. conducted electromechanical materials characterization; G.K., R.M., and C.M.S. conducted the mechanical simulations and modeling; G.K., D.D., D.W., and F.H. analyzed the data; G.M. conducted the magnetic field FEM simulation; M.K. gave input at all stages; G.K., D.D., D.W., F.H., M.K., and S.B. wrote the manuscript; all authors contributed to editing the manuscript; M.K. supervised the research.

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