**Experimental**

(1) **NRVM and hiPS-CM preparation and seeding**

**NRVM primary cell harvest:** Neonatal Ventricular myocytes were isolated from day 2 neonate Sprague Dawley rats following a procedure approved by the Harvard University Animal Care and Use Committee. In short, after sacrificing the animals, the ventricles were removed and incubated ~12 hours in cold (4°C) 0.1% (w/v) trypsin (USB Corp., Cleveland, OH). The tissue was subsequently exposed to serial treatments (2 minutes each) of 0.1% (w/v) warm (37ºC) collagenase type II (Worthington Biochemical, Lakewood, NJ) solution. Isolated NRVMs were then seeded onto the engineered substrates at desired densities and maintained in culture medium based on Medium 199 (Life Technologies) supplemented with 10% (v/v) heat-inactivated fetal bovine serum (FBS), 20 mM glucose, 2 mM L-glutamine, 10 mM HEPES, 1.5 µL vitamin B-12, and 50 U/ml penicillin. After 48 hours the media was switched to lower FBS concentration of 2%.

**hiPS-CM preparation:** hiPS-CMs and accompanying media were acquired commercially (Cor4U; Axiogenesis, Cologne, Germany), shipped in cryovials containing 1 million viable hiPS-CMs each, and stored in liquid nitrogen upon arrival. hiPS-CMs were cultured according to manufacturer’s instructions. Briefly, for each cryovial, 3 wells of a 6-well tissue culture plate were coated with 0.01 µg/mL fibronectin (FN) (BD Biosciences, Bedford, MA) for 3 hours in a 37ºC incubator. Cryovials of Cor.4U hiPS-CMs were quickly thawed in a 37ºC water bath, resuspended in 9 mL of complete culture medium provided by the manufacturer supplemented with 4.5 µL of 10 mg/mL puromycin, to eliminate un-differentiated stem cells from the culture, and plated on the FN-coated tissue culture plates at 37ºC, 5% CO₂. 24 hours after plating, cell cultures were washed once with warm PBS, followed by additional 48 hours of culture in complete culture medium. Cells were then dissociated with 0.25% trypsin-EDTA (Life Technologies) and seeded onto devices at a density of 220 k/cm². Prior to seeding, devices were exposed to UVO for 8 minutes and incubated with 50 µg/mL fibronectin (Human natural fibronectin, BD Biosciences) for 1 hour. The seeded devices were incubated at 37ºC, 5% CO₂. Media was changed 24 hours after seeding and then every other day throughout the culture period.
(2) Determining Ink rheological properties

TPU ink: To study the changes in viscosity of the TPU ink that occur upon printing and evaporation of the volatile solvent THF, a series of inks were made, and their viscosities determined by a flow sweep using a TA Discovery DHR-3 rheometer equipped with a 40 mm stainless steel cone plate attachment and a solvent trap. A Peltier plate was used to hold sample temperature at 22°C for all tests. The inks had a constant net TPU content while the amount THF solvent was decreased from initial (1:4 v:v) DMF:THF (0% loss) to pure DMF (100% loss), thereby increasing TPU concentration from 15 wt% to 47 wt%. For all inks a newtonian behavior was observed as evidenced by a linear relationship between shear stress and shear rate. The viscosity reported is the slope of the best fit line.

CB:TPU ink: To study the changes in viscosity of the CB:TPU ink that occur upon printing and evaporation of the volatile solvent THF, a series of inks were made, and their viscosities determined by a flow sweep using a TA Discovery DHR-3 rheometer with a 40 mm stainless steel cone plate attachment and a solvent trap. The inks had a constant net TPU and CB content while the amount THF solvent was decreased from initial (1:4 v:v) DMF:THF (0% loss) to pure DMF (100% loss), thereby increasing TPU concentration from 15 wt% to 47 wt% and CB: from 5 wt% to 16 wt%. For all inks the viscosity decreased significantly with increasing shear rate, and best fit either a power law model or a Hershel Buckley model. To recapitulate ink behavior after ejection from the nozzle, 10^2 (1/s) shear rate was used for determination of all CB:TPU viscosities. This was the lowest shear rate that could be reliably applied and measured with the rheometer.

Ag:PA ink: To study the shear-thinning properties of the Ag:PA ink, a TA Instruments AR2000ex rheometer was used to perform an oscillation amplitude sweep of the ink. The rheometer was outfitted with a 60 mm 2.005 degree stainless steel cone plate attachment and Peltier plate, the temperature was held at 22°C, and the ink was given 3 minutes to equilibrate after sample loading. An oscillation amplitude sweep was run at an angular frequency of 1.0rad/s with a strain percentage ranging from 5x10^{-3}% to 500% with 10 points per decade. A solvent trap was used to prevent the evaporation of pentanol during the test.

Soft PDMS ink: To study the shear-thinning properties of the soft PDMS ink, a TA Instruments AR2000ex rheometer was used to perform an oscillation amplitude sweep of the ink. The rheometer was outfitted with a 60 mm 2.005 degree stainless steel cone plate attachment and Peltier plate, the temperature was held at 22°C, and the ink was given 3 minutes to equilibrate after sample loading. An oscillation amplitude sweep was run at an angular frequency of 1.0rad/s with a strain percentage ranging from 5x10^{-3}% to 500% with 10 points per decade.

(3) Determining stress-strain behavior of printed and cured features of cantilever

TPU features: The mechanical properties of TPU features in cured state were determined using an Instron uniaxial tensile test setup. A strain rate of 0.33 mm/s was applied, equivalent to 1 %/s. Linear fit to data in range up to 1%, gave an averaged Young’s modulus of 1.62 ± 0.04 MPa (n=3). Test samples were shaped according to the base area of ASTM D412 Die C and fabricated by casting the TPU ink into a mold and curing overnight at 100°C. TPU ink was prepared with 40 wt% total solids content to facilitate casting.
**CB:TPU features**: The mechanical properties of CB:TPU features in cured state were determined using an Instron uniaxial tensile test setup. The strain rate was 0.33 mm/s, equivalent to 1 %/s. A strain rate of 0.33 mm/s was applied, equivalent to 1 %/s. Linear fit to data in range up to 1%, gave an averaged Young’s modulus of 8.77 ± 0.95 MPa (n=3). Test samples were shaped according to the base area of ASTM D412 Die C and fabricated by doctor blading the CB:TPU ink into a laser cut adhesive stencil onto a Teflon coated aluminum substrate. The cast films were allowed to air dry for 4 hours at room temperature and subsequently cured overnight at 100°C. CB:TPU ink was prepared with 40 wt% total solids content to facilitate casting.

**Cantilever soft PDMS features**: The mechanical properties of soft PDMS features in cured state were determined using an Instron uniaxial tensile test setup. A strain rate of 0.165 mm/s was applied, equivalent to 0.5 %/s. Linear fit to data in range up to 1%, gave an averaged Young’s modulus of 1.280 ± 0.085 MPa (n=3). Test samples were shaped according to the base area of ASTM D412 Die C and fabricated by 3D printing using a 0.41 mm nozzle. The tool path was generated using the slicing algorithm Euclid provided by Voxel8, Inc. The tool path was set such that the infill direction was parallel to the longitudinal axis of the dogbone. Two external perimeters were printed to give the exterior of the parts a smooth finish with minimal stress concentrators. After printing, dog-bones were cured overnight at 100°C. Raw stress values were corrected according to total mass of sample to compensate for any trapped air incorporated between the filaments due to the nature of the 3D printing process. The theoretical density of 1.13 g/cm$^3$, reported by the supplier, was used for SE1700. The ink was prepared with a 1:25 curing-agent to base weight ratio.

(4) **Electrical property measurements**

**Resistivity of CB:TPU and Ag:PA in cured state**: To determine conductivity of CB:TPU and Ag:PA inks the cured state, a series of lines with constant cross-sectional area and varying length in the range of 10-60 mm were printed. The cross-sectional areas of the lines were determined by stylus profiling. The resistances of the lines were determined using a Keithley 3706a DMM. Resistivity was determined using linear fit to data to resistance x area vs. length. For CB:TPU resistivity of 1.19 $\Omega \cdot$cm was measured, while for Ag:PA the measured resistivity was 6.60x10$^{-5}$ $\Omega \cdot$cm.

**Strain-resistance relationship for cured CB:TPU**: The relationship between resistance and strain for CB:TPU wires in Instron uniaxial tensile test setup. Cast TPU dogbones with cast CB:TPU wires embedded within the center were applied as test substrates. The samples were subjected to triangular cyclic straining 1-3 Hz in the range 0.0125% to 1%, to mimic strain exerted by the laminar cardiac tissues strain. Gauge factor was determined for 0.1% range. Average gauge factor was found to be 2.51 ± 0.09 (n=4).

(5) **Tissue structural analysis**

**Immuno-staining and imaging**: Following culture, samples were fixed and stained at room temperature. Samples were first fixed with 4% PFA/PBS (v/v) solution for 15 minutes and then permeabailized with 0.05% Triton-X/PBS (v/v) solution for 10 minutes. Subsequently, samples
were incubated for 1 hour with a monoclonal sarcomeric α-actinin (clone EA-53; Sigma-Aldrich) primary antibody, washed three times in PBS, and finally counterstained with: Alexa Fluor 488-conjugated anti-mouse secondary antibody, Alexa Fluor 633-conjugated Phalloidin and DAPI (Invitrogen) and imaged using confocal microscopy.

(6) Alternate well and cover materials

Alternate material types: In an effort to reduce the possible absorption of hydrophobic drug due to the incorporation of PDMS into the device, two thermoplastic material alternatives were fabricated and tested. Clear acrylonitrile butadiene styrene (ABS), and natural colored polylactic acid (PLA) filament were purchased from E3D everyday filaments in 1.75 mm diameter and subsequently used for printing the wells and covers of the devices. The materials were secured to the device using a ~200 µm thick layer of printed SE1700 PDMS as an adhesive gasket material, thus significantly reducing the total volume of exposed PDMS in the device. PLA and ABS were chosen as replacement options as thermoplastics are not prone to volume absorption of hydrophobic drugs like PDMS (Berthier E, Young EWK, Beebe D, Lab on a Chip, 2012, 1224-1237). Equally important, both printed ABS and PLA are biocompatible (Bhattacharjee N, Urrios A, Kanga S, Folch A, Lab on a Chip, 2016, 1720 & Rosenzweig DH, Carelli E, Steffen T, et al. Int. J. Mol. Sci. 2015). Indeed, devices fabricated with both PLA and ABS wells did not affect cell viability and ability to generate contractile stresses.

Well and cover fabrication and curing: The wells and covers were modeled using Solidworks 2015, and exported as .STL files. Files were sliced and printed using a Voxel8 Developer’s Kit 3D printer and tool-pathing software. Settings for ABS printing: nozzle temperature: 235°C, bed temperature: 100°C, nozzle diameter: 0.35 mm, layer height: 0.19 mm, print speed: 30 mm/s. Settings for PLA printing: nozzle temperature: 210°C, bed temperature: 50°C, nozzle diameter: 0.35 mm, layer height: 0.19 mm, print speed: 30 mm/s. The printed wells and covers were aligned with the layer of printed SE 1700 adhesive that was printed onto the glass substrate of the device and compressed to form a thin watertight seal. The wells and covers were then permanently adhered to the substrate by a final curing step of 70°C for 24 hour.
Supplementary Figure S1 | TPU ink viscosity increases upon THF solvent evaporation.
Semilog plot of TPU ink viscosity as a function of THF solvent loss. The pristine ink contains a 1:4 ratio of DMF:THF (denoted as 0% solvent loss). Due to the substantial differences in solvent volatility, THF evaporates more quickly than DMF. Consequently, DMF fraction remains upon THF evaporation (pure DMF fraction denoted as 100% solvent loss). (n=3) error bars are st.dev.

Supplementary Figure S2 | CB:TPU ink viscosity increases upon THF solvent evaporation.
Semilog plot of CB:TPU ink viscosity as a function of THF solvent loss. The pristine ink contains a 1:4 ratio of DMF:THF (denoted as 0% solvent loss). Due to the substantial differences in solvent volatility, THF evaporates more quickly than DMF. DMF fraction thus remains upon THF evaporation (pure DMF fraction denoted as 100% solvent loss). (n=3) error bars are st.dev.
**Supplementary Figure S3 | Stiffness of printed and cured features of cantilever.**
Representative stress-strain curves for cured tensile specimens made from the: (a) TPU ink, (b) CB:TPU ink, and (c) soft PDMS ink.

**Supplementary Figure S4 | Electrical resistivity of printed features.**
(a) The electrical resistance of CB:TPU wires of varying length, (n=5) error bars are st.dev. Dotted line is linear fit to data. Slope of fitted line indicates electrical resistivity of 1.19 Ω-cm, equivalent to a conductivity of 0.841 S/cm (b) The electrical resistance of Ag:PA wires of varying length, (n=4) error bars are st. dev. Dotted line is linear fit to data. Slope of fitted line indicates resistivity of 6.60·10⁻⁵ Ω-cm, equivalent to a conductivity of 1.52·10⁴ S/cm.
Supplementary Figure S5 | Strain-dependent electrical resistance for CB:TPU strain sensors.

(a) Up to 0.1% strain, fast return to baseline and linear strain-resistance relationship is observed.

(b) At 1% strain, slow baseline return and significant increase from linear strain-resistance relationship is observed.
Supplementary Figure S6 | CB:TPU strain sensors tested at frequencies 1-3 Hz. (a) Left to Right: 0.02% strain exerted at 1, 2 and 3 Hz, (b) Left to right: 0.1% strain exerted at 1, 2 and 3 Hz. In these strain regimes relevant for the final device, negligible influence of the strain-rate (frequency) was observed on the sensor signal.
Supplementary Figure S7 | Determining gauge factor for CB:TPU strain sensor at 0.1% strain.

(a-b) Relative change in CB:TPU resistance upon triangular cyclic straining to 0.1% at 1 Hz.
(c-f) Relative resistance change vs. applied strain for 4 independent dog-bone samples. Dotted line indicates linear fit to part of strain cycles with increasing strain. CB:TPU gauges displayed limited hysteresis and conserved linearity during repeated strains. Average gauge factor is 2.51 ± 0.09.
Supplementary Figure S8 | Ag:PA ink rheology.
Log-log plot of the Ag:PA ink storage and loss moduli vs. oscillation strain %. At low strains/stresses, the storage modulus exceeds the loss modulus, indicating that the ink is behaving like a non-flowing solid. As the amplitude of strain increases, the ink exceeds its yield stress, and the loss modulus overtakes the storage modulus, indicating that the ink is behaving like a liquid that flows under high shear stresses.

Supplementary Figure S9 | Soft PDMS ink rheology.
Log-log plot of the soft PDMS ink storage and loss moduli vs. oscillation strain %. At low strains/stresses, the storage modulus exceeds the loss modulus, indicating that the ink is behaving like a non-flowing solid. As the amplitude of strain increases, the ink exceeds its yield stress, and the loss modulus overtakes the storage modulus, indicating that the ink is behaving like a liquid that flows under high shear stresses.
Supplementary Figure S10 | Devices employing thermoplastic ABS or PLA for wells and covers.

(a) Device with printed acrylonitrile butadiene styrene (ABS) wells and covers. Insert: example readout from cantilever in ABS device. (b) Device with printed poly(lactic acid) (PLA) wells and covers. Insert: example readout from cantilever in PLA device. ~200 µm thick printed PDMS gaskets below wells and covers ensured watertight seal. These approaches limit the amount of exposed and incorporated PDMS, thereby limiting absorption of hydrophobic drug molecules.
Supplementary Figure S11 | Anisotropic laminar cardiac tissue.
NRVM tissue fixed and stained at day 4 after seeding. Standard substrate of soft PDMS filaments spaced by 60 μm. Blue: DAPI nuclei stain, White: α-actinin sarcomere stain. Scale bar 25 μm.
Supplementary Figure S12 | Converting electrical and optical readout to tissue stress using mechanical model based on a modified Stoney’s equation.

(a) A linear conversion factor can be established between relative change in resistance, radius of curvature and tissue stress, by establishing analytical mechanical model. (b) Optical tracking is used to measure cantilever curvature during the tests as an external independent validation of the mechanical model. (c) To establish a sufficiently descriptive mechanical model, we determined local neutral axis in the two distinct regions of the cantilever; the wire region and the non-wire region. The relative placement of layers and mechanical properties were derived from stylus profilometer data and Instron mechanical tests.
Deriving the relationship between the contractile stress, cantilever curvature and the electrical readout

The printed instrumented MTF cantilever has a large width-to-thickness ratio (~100) and consists of two distinct strips: the wire-containing strip and the wire-free strip, (supplementary Fig. 12c). For this geometry, to make the mechanics modeling mathematically tractable, we assume that the two strips have the same curvature but different neutral surfaces when subjected to the tissue contraction. For each strip, cylindrical bending theory will be applied to solve for the position of the neutral surface to yield a final modified Stoney’s equation that relate the electrical and optical measurements to the contractile stress of the tissue.

For the entire derivation, the uncontracted tissue and non-deflected cantilever is assumed as reference state, thus strains will be stated relative to this state. As illustrated in supplementary S12, for the wire-containing strip, \( b \) denotes the distance from the bottom of the cantilever to the neutral surface and \( c \) denotes the curvature of the film. The normal strain \( \varepsilon_x \) in the longitudinal direction can then be written as:

\[
\varepsilon_x = c(b - z) \tag{1}
\]

The wire-containing strip is composed of 5 layers: TPU layer 1, Wire layer, TPU layer 2, SE1700 PDMS rectangular layer, SE1700 sinusoidal layer. The normal stress \( \sigma_x \) for the \( i \)-th layer is

\[
\sigma_x = \frac{E_i}{1-v_i^2} \varepsilon_x \tag{2}
\]

where \( E_i \) and \( v_i \) are Young’s modulus and Poisson ratio of the material of the \( i \)-th layer, respectively. Note that the uniaxial modulus \( \frac{E_i}{1-v_i^2} \) is used in this model, because the anisotropic contraction of the cell layer bends the plate into a cylindrical shape instead of a bowl or cap-like shape. For the theoretical case of a bowl-like curved plate induced by the isotropic contraction, the biaxial modulus \( \frac{E_i}{1-v_i} \) should be used. This approach notably ignores the cantilever being constrained in the transverse direction at the base. The top surface of the sinusoidal grooves is described by:

\[
z_{TG} = z_{BG} + a \left( 1 - \cos \left( \frac{2\pi}{\lambda} y \right) \right), \tag{3}
\]

where \( z_{BG} \) denotes the position of the valley of the grooves, \( a \) denotes the half-depth of the grooves, \( \lambda \) denotes the wavelength, \( y \) is the coordinate in the width direction. The middle plane of the tissue layer is:

\[
z_c = z_{TG} + h/2 \tag{4}
\]

where \( h \) is the thickness of the tissue layer. The mean strain in the tissue layer resulting upon bending can be determined using Equation (1) at the middle plane of the tissue layer.
\[ \varepsilon_{cx} = c(b - z_c) \]  

(5)

Since the tissue layer is much softer than the substrate layers (10 kPa <1 MPa), the bending resistance of the tissue is less than 1\% that of the beam. The bending resistance of the tissue can thus be ignored, and the potential energy of the MTF bending reduced to two parts; \( V_c \), which is the negative of the work done by the cell contraction, and the strain energy of bending in the plate \( V_b \). Consequently, the stress in the tissue can be reduced to an active uniform stress, equivalent to that exerted by a tissue on a flat substrate.

Denoting \( \bar{\sigma}_c \) the active contractile stress of the tissue, the negative of the work done by the cell contraction is calculated as a volume integral:

\[ V_c = \int_{tissue\ layer} \bar{\sigma}_c \varepsilon_{cx} \, dv \]  

(6)

For the sinusoidal tissue layer, \( dv = Lhds \), where the arc length \( ds = \sqrt{1 + \left( \frac{dz_c}{dy} \right)^2} \, dy \). Then, we have

\[ V_c = \bar{\sigma}_c cLh \int_0^W (b - z_c) \sqrt{1 + \left( \frac{dz_c}{dy} \right)^2} \, dy = \bar{\sigma}_c cLh \left( F_1 \bar{b} - F_2 \right) \]  

(7)

where \( F_1 = \int_0^W \sqrt{1 + \left( \frac{dz_c}{dy} \right)^2} \, dy \), \( F_2 = \int_0^W z_c \sqrt{1 + \left( \frac{dz_c}{dy} \right)^2} \, dy \), and \( \frac{dz_c}{dy} = \frac{a}{h} \sin \left( \frac{2\pi}{h} \right) \). \( W \) is the width of the wire-containing strip. Taking derivatives of \( V_c \) with respect to curvature \( c \) and position of neutral axis \( \bar{b} \), we have

\[ \frac{dv_c}{dc} = \bar{\sigma}_c cLh \left( F_1 \bar{b} - F_2 \right) \]  

(8)

\[ \frac{dv_c}{db} = \bar{\sigma}_c cLh F_1 \]  

(9)

The strain energy of bending \( V_b \) can be calculated via the volume integral of the elastic energy density,

\[ V_b = \int \frac{1}{2} \bar{\varepsilon}_x \varepsilon_x \, dv \]  

(10)

For the layers with rectangular cross sections, \( dv = LhW \, dz \), plugging Eq. (1) and (2) into the above equation we have

\[ V_{b,i} = LhW \int_{z_{B,i}}^{z_{T,i}} \frac{1}{2} \bar{E}_i c^2 (z - \bar{b})^2 \, dz \]  

(11)

where \( z_{B,i} \) and \( z_{T,i} \) are lower and upper positions of Layer \( i \), \( \bar{E}_i = \frac{E_i}{1 - \nu_i^2} \). The derivatives can be computed as
\[
\frac{\partial v_{b1}}{\partial b} = L c^2 [A_{1b} \tilde{b} - A_{21}] \tag{12}
\]

\[
\frac{\partial v_{b1}}{\partial c} = L c [A_3 - 2\tilde{b}A_2 + \tilde{b}^2 A_1] \tag{13}
\]

where

\[
A_{1b} = \bar{W} \bar{E}_i t_i
\]

\[
A_{2i} = \frac{1}{2} \bar{W} \bar{E}_i (Z_{T,1,i}^2 - Z_{B,1,i}^2)
\]

\[
A_{3i} = \frac{1}{3} \bar{W} \bar{E}_i (Z_{T,1,i}^3 - Z_{B,1,i}^3)
\]

For the sinusoidal layer, \(dv = L dz dy\), the strain energy of it is

\[
V_{BG} = L \int_0^W \left[ \int_{z_{BG}}^{z_{RG}} \frac{1}{2} \bar{E}_c c^2 (z - \bar{b})^2 \right] dz dy
\]

and the derivatives are

\[
\frac{\partial v_{BG}}{\partial b} = L c^2 [A_{1g} \tilde{b} - A_{2g}] \tag{14}
\]

\[
\frac{\partial v_{BG}}{\partial c} = L c [A_{3g} - 2\tilde{b}A_{2g} + \tilde{b}^2 A_{1g}] \tag{15}
\]

where

\[
A_{1g} = \bar{E}_i \int_0^W (z_{RG} - z_{BG}) dy
\]

\[
A_{2g} = \frac{1}{2} \bar{E}_i \int_0^W (Z_{T,G}^2 - Z_{B,G}^2) dy
\]

\[
A_{3g} = \frac{1}{3} \bar{E}_i \int_0^W (Z_{T,G}^3 - Z_{B,G}^3) dy
\]

The principle of the minimal potential energy yields,

\[
\frac{\partial v_{b}}{\partial b} + \frac{\partial v_{c}}{\partial b} = 0 \Rightarrow c [A_1 \tilde{b} - A_2] + \tilde{\sigma}_c h F_1 = 0 \tag{16}
\]

\[
\frac{\partial v_{b}}{\partial c} + \frac{\partial v_{c}}{\partial c} = 0 \Rightarrow c [A_3 - 2\tilde{b}A_2 + \tilde{b}^2 A_1] + \tilde{\sigma}_c h (F_1 \tilde{b} - F_2) = 0 \tag{17}
\]
where \( A_1 = A_{1G} + \sum_{i=1}^{4} A_{1i}, A_2 = A_{2G} + \sum_{i=1}^{4} A_{2i}, A_3 = A_{3G} + \sum_{i=1}^{4} A_{3i} \).

Rearranging (16) yields

\[
\delta_c = \frac{c}{F_i} \left[ A_2 - A_1 \bar{b} \right] \quad (18)
\]

Plugging (18) into (17) and simplifying the result, we have

\[
A_3 - \bar{b} A_2 - A_2 F_{21} + A_1 F_{21} \bar{b} = 0
\]

where \( F_{21} = F_2/F_1 \), solving it for \( \bar{b} \), we have

\[
\bar{b} = \frac{A_3 - A_2 F_{21}}{A_2 - A_1 F_{21}} \quad (19)
\]

Substituting Eq. (19) into Eq. (18) leads to the final modified Stoney equation for the cantilever including sinusoidal grooves:

\[
\delta_c = \frac{c}{F_i} \left[ A_2 - A_1 \frac{A_3 - A_2 F_{21}}{A_2 - A_1 F_{21}} \right] \quad (20)
\]

The average strain of the wire is

\[
\varepsilon_w = c \left[ \bar{b} - \frac{z_{T1} + z_{B1}}{2} \right] \quad (21)
\]

Therefore, the curvature \( c \) can be represented as

\[
c = \frac{\varepsilon_w}{(\bar{b} - (z_{T1} + z_{B1})/2)} \quad (22)
\]

Substituting Eq. (22) into (21) we have

\[
\delta_c = \frac{\varepsilon_w}{\left( \bar{b} - \frac{z_{T1} + z_{B1}}{2} \right) h F_i} \left[ A_2 - A_1 \frac{A_3 - A_2 F_{21}}{A_2 - A_1 F_{21}} \right] \quad (23)
\]

Here the strain of the wire can be computed from the electrical readout by the following equation

\[
\Delta R/R = GF \varepsilon_w \quad (24)
\]

Therefore finally we have the relation between the contractile stress and the electric readout

\[
\delta_c = \frac{\Delta R/R}{GF} \left( \bar{b} - \frac{z_{T1} + z_{B1}}{2} \right) h F_i \left[ A_2 - A_1 \frac{A_3 - A_2 F_{21}}{A_2 - A_1 F_{21}} \right] \quad (25)
\]
For the wire-free strip, the same procedure can be taken to derive the position of the neutral surface \( \hat{b} \), as well as the relationship between the curvature \( \hat{c} \) and the contractile stress \( \hat{\sigma}_c \) in the wire-free strip. Using this, we calculated the total work required for a given curvature of the full cantilever. To obtain the equivalent collected tissue stress, we assumed that the stress was uniform across the width of the tissue. This collected contractile stress \( \hat{\sigma}_c \) for the whole MTF can readily be shown to be equivalent to the geometric average of the apparent contractile stresses in the wire-containing strip and the wire-free strip,

\[
\sigma_c = \frac{\hat{\sigma}_c W + \hat{\sigma}_f \hat{W}}{\hat{W} + \hat{W}}
\]  

(26)

where \( \hat{W} \) is the width of the wire-free strip.

| Wire Region | Layer | top [µm] | bottom [µm] | \( E \) [MPa] |
|-------------|-------|----------|-------------|-------------|
| TPU base | 0 | 1.25 | 1.62 |
| CB TPU wire | 1.25 | 7.75 | 8.77 |
| TPU coat | 7.75 | 9.25 | 1.62 |
| PDMS grooves sheet | 9.25 | 12.25 | 1.28 |
| PDMS grooves sinusoidal | 12.25 | 30.25 | 1.28 |

| Non-wire Region | Layer | top [µm] | bottom [µm] | \( E \) [MPa] |
|-----------------|-------|----------|-------------|-------------|
| TPU base | 0 | 4.5 | 1.62 |
| PDMS grooves sheet | 4.5 | 7.5 | 1.28 |
| PDMS grooves sinusoidal | 7.5 | 25.5 | 1.28 |

**Supplementary Table S1 | Values applied in analytical mechanical model.**

Z-positions for each layer were obtained from stylus profiling data. For the CB:TPU wire partial integration in TPU bottom layer is observed. For grooved region, a sinusoidal wave with 60 µm wave length, 18 µm amplitude and a groove sheet thickness 3 µm, were applied. For approximating stress in control samples with 40 µm groove spacing, a sinusoidal wave of with 40 µm wavelength, 6 µm amplitude and a groove sheet thickness 17 µm, were applied. For approximating stress in control samples with 80 µm groove spacing, a sinusoidal wave of with 80 µm wavelength, 22 µm amplitude and a groove sheet thickness 2 µm, were applied. These values reflect variations occurring from changes in filament overlap. Instron test were used to measure Young’s modulus values. Poissons ratio of 0.5 was assumed as materials are elastic. Even minor variations in wire thickness have significant influence in the final signal. To account for these changes wire layer thickness was corrected using the resistance of a given cantilever wire loop, when comparing absolute stresses generated by different tissues.
Supplementary Figure S13 | Comparing optical and electrical readout to verify mechanical model.

To evaluate the robustness of the model, we performed synchronous optical tracking and electrical recordings of three independent cantilevers, seeded with NRVM anisotropic tissues. (a,c,e) optical tracking of cantilevers, Left: Diastole, Right: Systole (b,d,f) Comparison between stress calculated from optical tracking and synchronous gauge recordings. Variations are attributed to geometric variations in the curvature generated by the tissue and to variations in device printing from batch to batch.
Supplementary Figure S14 | Micro-filament spacing influence tissue stress and device readout. 
(a-d) Micro-filaments guiding self-assembly of laminar NRVM tissues. Increased filament spacing shifts neutral axis towards sensor wire. (e-h) Still images of cantilever deflection for cantilevers with micro-filaments spacing 40-100 µm (scale bar 1 mm). For 80 and 100 µm spacing, transverse deformation is observed. This deformation is caused partly by increasingly isotropic contraction, but is in particular influenced by the larger groove spacing leading to decreased transverse bending modulus. Thus, no transverse deformation is observed for 40 µm spacing, which has similar degree of cytoskeletal anisotropy as 100 µm samples. Notably, for all cases, cantilever transverse deformation remains constrained at base. (i-l) Examples of raw signals from strain gauges. For 80 and 100 µm spacing, notable sensor compression signal is seen, as the neutral axis shift towards wire. (m) Readout from strain gauges for cantilevers with micro-filaments spacing ranging from 40 to 100 µm. (n≥4) Error bars are S.E.M. (n) Approximate twitch stress for cantilevers with micro-filaments spacing ranging from 40 to 100 µm. (n≥4) Error bars are S.E.M. To establish signal-to-stress conversion factors for all cases, the mechanical model was modified based on profilometer data for 40 µm and 80 µm samples. For 100 µm samples, the conversion factor for 80 µm was applied, as the approximation of sinusoidal groove patterns no longer is valid.
Supplementary Figure S15 | Multi-well Recording setup.
*Top:* Cell-seeded device in recording holder. *Bottom:* Setup inside cell incubator.

Supplementary Figure S16 | Example force-frequency relationship of laminar NRVM tissue.
Example force-frequency relationship trace for laminar NRVM tissues. The NRVM tissues generally displayed negative force-frequency relationship.
Supplementary Figure 17 | Laminar NRVM tissues spontaneous drug dose response to verapamil recorded inside incubator.

(a) Example traces of stress generated by the NRVM tissue when exposing tissue to increased concentrations of verapamil. (b) Drug dose response curve for spontaneous beating frequency at increasing verapamil concentrations (n=13) error bars are S.E.M. Apparent EC$_{50}$ of 1.59 x10$^{-7}$ M
(c) Drug dose response curve for stress at increasing verapamil concentrations, (n=13) error bars are S.E.M. Stress normalized to initial stress prior to drug exposure. Apparent EC$_{50}$ 4.72x10$^{-7}$ M

Supplementary Figure 18 | Laminar NRVM tissues chronotropic drug dose response to isoproterenol recorded inside incubator.

(a) Example traces of increased an irregular spontaneous beating of NRVM tissues at increasing concentrations of isoproterenol in regular 199 media (b) Drug dose curve for peak spontaneous beating frequency at increasing isoproterenol concentrations, (n=10), error bars are S.E.M. Upon increasing dosage, the spontaneous beat rate of the tissue became increasingly irregular and tachycardia sequences of elevated beat rates are interspersed with sequences of reduced activity. Plotting the maximum observed beat rate as a function of dosage, we obtained an EC$_{50}$ value of 3.14x10$^{-9}$ M.
Supplementary Figure S19 | Structural maturation of hiPS-CM tissue in device during culture. Tissue fixed and stained at day 2 (top) and day 28 (bottom) after seeding. Standard substrate of soft PDMS filaments spaced by 60 µm. Blue: DAPI nuclei stain, White: α-actinin sarcomere stain. Scale bars 25 µm.
Supplementary Figure S20 | Force-frequency relationship of laminar hiPS-CM tissues.

(a) Example trace at day 8, note that for 1 Hz pacing spontaneous beating lead to higher actual beat rate
(b) Example trace at day 28. Slightly negative force-frequency relationships observed for day 28.
Supplementary Figure S21 | Thin and thicker laminar NRVM tissues on standard cantilever.

(a) Example orthogonal view of standard laminar tissue in groove. (b) Example orthogonal view of thicker laminar tissue in groove. Scale bars 10 µm. Blue: DAPI nuclei stain Green: Actin, Red: α-actinin. (c-d) Example still images of cantilever deflection for cantilevers with regular laminar tissue. (e-f) Example still images of regular cantilevers being overly bend due to the contraction of thicker laminar tissue.
Supplementary Figure S22 | Thick laminar NRVM tissue on cantilever surface in micro-pins.
NRVM Tissue fixed and stained at day 5. Blue: DAPI nuclei stain, White: α-actinin sarcomere stain, Red: Actin. (a) Z-projection, Scale bar 30 µm. (b) Example orthogonal view of thick tissue in groove. Tissue thickness on top of groove (4 µm) and in bottom of groove (15 µm) indicated.
Supplementary Figure S23 | Sarcomere organization of thicker laminar NRVM tissue.
NRVM Tissue fixed and stained at day 5. Left to right: Z-projection of top to bottom section of tissue. Blue: DAPI nuclei stain White: α-actinin. Scale bars 10 μm. Anisotropy is conserved throughout depth of thicker laminar tissue.

Supplementary Figure S24 | Example force-frequency relationship of thicker laminar NRVM tissues. NRVM-based thicker laminar tissues generally displayed negative force-frequency relationship. Notably peak systolic stress was largely unaffected by increasing pacing frequency.
List of Supplementary Movies

**Movie S1** | Timelapse movie (time x 40) of automation procedure for substrate height profiling and determination of the relative x-y-z positions of the four individually addressable nozzles on the multimaterial 3D printing stage.

**Movie S2** | Timelapse movie (time x 132) of automated printing of a cardiac microphysiological device on a 2 inch x 3 inch glass slide substrate in 7 steps.
1) dextran thin film sacrificial layer
2) TPU thin film cantilever base
3) CB:TPU strain sensor loop
4) TPU wire cover
5) PDMS micro-filaments
6) Ag:PA electrical leads and contacts
7) PDMS covers to insulate exposed wires and wells to contain cells and media

**Movie S3** | Close-up movie (real-time) of the individual printing steps in the fabrication of the cardiac microphysiological device.

**Movie S4-7** | Optical mapping system videos (time x 0.1) of the propagation of the derivative of tissue activation potential. Laminar NRVM tissues developed on substrates with soft PDMS filaments at 40 (S4), 60 (S5), 80 (S6) and 100 µm (S7) spacing. Tissues electrically paced at 2 Hz in top left corner. 4 mm x 4 mm field of view.

**Movie S8** | Microscope movie (real-time) of laminar NRVM tissues deflecting cantilever.

**Movie S9-11** | Microscope movie (real-time) of three spontaneously beating cantilevers with overlaid optical tracking of wire edge. Samples were used for independent verification of stress values obtained from mechanical model.

**Movie S12** | Microscope movie (real-time) of laminar NRVM tissue deflecting cantilever paced at 1-3 Hz.

**Movie S13** | Close-up movie (real-time) of the micro-pin and micro-well printing steps used for thicker tissue version of device.

**Movie S14** | Microscope movie (real-time) of thicker laminar NRVM tissue deflecting cantilever with micro-pins spontaneously.

**Movie S15** | Microscope movie (real-time) of thicker laminar NRVM tissue deflecting cantilever with micro-pins, paced at 1-3 Hz.
**Custom MATLAB scripts**

*Script to calculate conversion factors between resistance changes, radius of curvature and stress exerted by tissue. Based on the multilayer Stoney’s equation model derived above:*

```matlab
% length: milimeter
% elastic modulus: kPa
% Stress: kPa

%% input

gauge_factor = 2.51;

%% elastic modulus

Poisson_ratio = 0.5;   % Poisson ratio
TPU_E = 1.62e3;          % Young's modulus of TPU
wire_E = 8.77e3;         % 8.8e3;  % Young's modulus of CB wire
SE_E = 1.28e3;          % Young's modulus of PDMS

wire_width_each = 0.4; % width of each wire loop
total_wire_width = 2*wire_width_each;

non_wire_width = 2.2; % width of non-wire region

tissue_thickness_h = 4e-3;

rec_baseres = 125;

standardwirethick = 6.5e-3;
standardwireres = 125;

wirethick = (standardwireres/rec_baseres)*standardwirethick

%% parameters for: CB wire

zB1 = 1.25e-3     % z coordinates of the bottom of the CB wire
zT1 = (zB1) + wirethick     % z coordinates of the top of the CB wire

E_bar_1 = wire_E/(1-Poisson_ratio^2);

%% parameters for: TPU, excluding the wire region

zT2 = 4.5e-3;       % z coordinates of the top of the TPU
zB2 = 0.0;          % z coordinates of the bottom of the TPU

E_bar_2 = TPU_E/(1-Poisson_ratio^2);

%% parameters for: TPU, the region under the wires

zT3 = 1.25e-3;     % z coordinates of the top of the TPU
```
\[ z_{B3} = 0.0; \] % z coordinates of the bottom of the TPU

\[ E_{\text{bar}_3} = E_{\text{bar}_2}; \]

\text{% parameters for: TPU, the region above the wires}

\[ z_{T4} = (z_{T1} + 1.5e-3); \] % z coordinates of the top of the TPU

\[ z_{B4} = z_{T1}; \] % z coordinates of the bottom of the TPU

\[ E_{\text{bar}_4} = E_{\text{bar}_2}; \]

\text{% parameters for: SE sheet, excluding the wire region}

\[ z_{T5} = 7.5e-3; \] % z coordinates of the top of the SE sheet

\[ z_{B5} = 4.5e-3; \] % z coordinates of the bottom of the SE sheet

\[ E_{\text{bar}_5} = \frac{SE_E}{1 - \text{Poisson ratio}^2}; \]

\text{% parameters for: SE sheet, region above the wires}

\[ z_{T6} = (z_{T4} + 3e-3); \] % z coordinates of the top of the SE sheet

\[ z_{B6} = z_{T4}; \] % z coordinates of the bottom of the SE sheet

\[ E_{\text{bar}_6} = E_{\text{bar}_5}; \]

\text{% parameters for SE grooves, non-wire region}

\[ \text{wave\_length} = 60e-3; \] % wave length of grooves

\[ z_{BG\_1} = z_{T5}; \] % z coordinate of bottom of grooves

\[ \text{half\_depth\_a} = 9e-3; \] % half depth of groove

\[ E_{\text{bar\_G}} = E_{\text{bar}_5}; \]

\[ \text{N\_wave\_1} = \text{non\_wire\_width}/\text{wave\_length}; \]

\text{% parameters for SE grooves, above the wires}

\[ z_{BG\_2} = z_{T6}; \] % z coordinate of bottom of grooves in the wire region

\[ \text{N\_wave\_2} = \text{total\_wire\_width}/\text{wave\_length}; \]

\text{% grooves in non-wire region}

\[ \text{N\_Gaussian} = 10; \]

\[ [\text{Gauss\_point}, \text{weight\_coef}] = \text{lgwt}(\text{N\_Gaussian}, 0, \text{wave\_length}); \]

\[ z_{TG\_point} = z_{BG\_1} + \text{half\_depth\_a} \times (1 - \cos(2\pi \times \text{Gauss\_point}/\text{wave\_length})); \]

\[ A1\_G\_n = E_{\text{bar\_G}} \times \text{sum}((z_{TG\_point} - z_{BG\_1}) \times \text{weight\_coef}) \times \text{N\_wave\_1}; \]
A2G_n = 0.5*E_bar_G.*sum((zTG_point.^2-zBG_1.^2).*weight_coef,1)*N_wave_1;
A3G_n = 1/3*E_bar_G.*sum((zTG_point.^3-zBG_1.^3).*weight_coef,1)*N_wave_1;

zc_y = half_depth_a^2*pi/wave_length*sin(2*pi/wave_length*Gauss_point);
F1_n = sum(sqrt(1+zc_y.*zc_y).*weight_coef,1)*N_wave_1;
F2_n = sum((zTG_point+tissue_thickness_h/2).*sqrt(1+zc_y.*zc_y).*weight_coef,1)*N_wave_1;

%% grooves in the wire region

[Gauss_point,weight_coef]=lgwt(N_Gaussian,0,wave_length);
zTG_point = zBG_2 + half_depth_a*(1-cos(2*pi*Gauss_point/wave_length));
A1G_w = E_bar_G.*sum((zTG_point-zBG_2).*weight_coef,1)*N_wave_2;
A2G_w = 0.5*E_bar_G.*sum((zTG_point.^2-zBG_2.^2).*weight_coef,1)*N_wave_2;
A3G_w = 1/3*E_bar_G.*sum((zTG_point.^3-zBG_2.^3).*weight_coef,1)*N_wave_2;

zc_y = half_depth_a^2*pi/wave_length*sin(2*pi/wave_length*Gauss_point);
F1_w = sum(sqrt(1+zc_y.*zc_y).*weight_coef,1)*N_wave_2;
F2_w = sum((zTG_point+tissue_thickness_h/2).*sqrt(1+zc_y.*zc_y).*weight_coef,1)*N_wave_2;

%% wire

A11 = total_wire_width.*E_bar_1.*(zT1-zB1);
A21 = 0.5*total_wire_width.*E_bar_1.*(zT1.^2-zB1.^2);
A31 = 1/3*total_wire_width.*E_bar_1.*(zT1.^3-zB1.^3);

% TPU

A12 = non_wire_width.*E_bar_2.*(zT2-zB2);
A22 = 0.5*non_wire_width.*E_bar_2.*(zT2.^2-zB2.^2);
A32 = 1/3*non_wire_width.*E_bar_2.*(zT2.^3-zB2.^3);

A13 = total_wire_width.*E_bar_3.*(zT3-zB3);
A23 = 0.5*total_wire_width.*E_bar_3.*(zT3.^2-zB3.^2);
A33 = 1/3*total_wire_width.*E_bar_3.*(zT3.^3-zB3.^3);

A14 = total_wire_width.*E_bar_4.*(zT4-zB4);
A24 = 0.5*total_wire_width.*E_bar_4.*(zT4.^2-zB4.^2);
A34 = 1/3*total_wire_width.*E_bar_4.*(zT4.^3-zB4.^3);

% SE sheet

A15 = non_wire_width.*E_bar_5.*(zT5-zB5);
A25 = 0.5*non_wire_width.*E_bar_5.*(zT5.^2-zB5.^2);
A35 = 1/3*non_wire_width.*E_bar_5.*(zT5.^3-zB5.^3);

A16 = total_wire_width.*E_bar_6.*(zT6-zB6);
A26 = 0.5*total_wire_width.*E_bar_6.*(zT6.^2-zB6.^2);
A36 = 1/3*total_wire_width.*E_bar_6.*(zT6.^3-zB6.^3);

%%

% wire region

A1_w = A1G_w + A11 + A13 + A14 + A16;
A2_w = A2G_w + A21 + A23 + A24 + A26;
A3_w = A3G_w + A31 + A33 + A34 + A36;
% non-wire region
A1_n = A1G_n + A12 + A15;
A2_n = A2G_n + A22 + A25;
A3_n = A3G_n + A32 + A35;

%% model 1: having separate neutral axis

neutral_position_b_w = (A3_w-A2_w*(F2_w/F1_w))/(A2_w-A1_w*(F2_w/F1_w))
neutral_position_b_n = (A3_n-A2_n*(F2_n/F1_n))/(A2_n-A1_n*(F2_n/F1_n))

electrical2curvature = 1/(gauge_factor*(neutral_position_b_w -(zT1+zB1)/2 )) % this factor times relative resistance change gives you curvature

curvature2stress_w = (A2_w-A1_w.*neutral_position_b_w)./tissue_thickness_h/F1_w ; % K1
curvature2stress_n = (A2_n-A1_n.*neutral_position_b_n)/tissue_thickness_h/F1_n ; % K1

curvature2stress_averaged = (curvature2stress_w*total_wire_width + curvature2stress_n*non_wire_width) /
(total_wire_width+non_wire_width)

%% model 2: coupled, but two neutral axis

% a11 = A1_w*F2_w - A2_w*F1_w;
% a12 = A1_n*F2_n - A2_n*F1_n;
% a21 = A1_w*F1_n;
% a22 = -A1_n*F1_w;
% b1  =  A2_w*F2_w - A3_w*F1_w + A2_n*F2_n - A3_n*F1_n;
% b2  = A2_w*F1_n - A2_n*F1_w;
% % A = [a11 a12; a21 a22]; b = [b1 b2];
% % x = A;
% % % b_w = x(1) % neutral axis of the wire region
% % b_n = x(2) % neutral axis of the non-wire region
% %
% % % curvature2stress = (A3_w - 2*b_w*A2_w + b_w*b_w*A1_w + A3_n - 2*b_n*A2_n + b_n*b_n*A1_n)/ ... 
% % (tissue_thickness_h*(F2_w-F1_w*b_w + F2_n-F1_n*b_n)); % this factor times curvature gives you stress

combined_conversionfactor = electrical2curvature*curvature2stress_averaged
Script to detect relative resistance change signal values and frequencies, from multichannel resistance readout:

This code accepts data in the form of two matrices:

1) A time matrix "timetable" with time values for each recording channel,
2) A resistance data matrix "datatable" with resistance values corresponding to a given timepoint, for each recording channel.

```
[mX,nX] = size(timetable);
tot_noofchannels = nX;
readingsperchannel = mX;

% Option: Choose a subset or all channels for analysis
specificchannels = (1:tot_noofchannels);
% specificchannels = [1,2,3];

% Option: Choose a temporal subset for analysis.
startpointstoignore = 0;
endpointstoignore = 0;
nopointsforanalysis = (readingsperchannel-startpointstoignore-endpointstoignore);

% Creating tables for calculated and corrected data

% Tables for raw and corrected data
timetablecorr = zeros((nopointsforanalysis),tot_noofchannels);
datatablecorr1 = zeros((nopointsforanalysis),tot_noofchannels);
datatablecorr2 = zeros((nopointsforanalysis),tot_noofchannels);
expfitdatatable = zeros((nopointsforanalysis),tot_noofchannels);
expfittable = zeros((nopointsforanalysis),tot_noofchannels);

% Tables for calculated values
peaknolist = zeros(tot_noofchannels,1);
peakbundleduration = zeros(tot_noofchannels,1);
beatratelist = zeros(tot_noofchannels,1);

peakloccell = cell(tot_noofchannels,1);
deltarescell = cell(tot_noofchannels,1);
baserescell = cell(tot_noofchannels,1);
reldeltarescell = cell(tot_noofchannels,1);

% Tables for calculated value means
mean_deltares = zeros(tot_noofchannels,1);
mean_baseres = zeros(tot_noofchannels,1);
mean_reldeltares = zeros(tot_noofchannels,1);

% Tables for calculated value medians
median_deltares = zeros(tot_noofchannels,1);
median_baseres = zeros(tot_noofchannels,1);
median_reldeltares = zeros(tot_noofchannels,1);

% Choose analysis approach
```
% fit 1 to assist initial baseline correction and facilitate peak detection
% Polynomial baseline is default. Choose degree of polynomial:
PolynomialDegree = 5;
% Option: Consider applying a median filter. Default is (1) for no filter:
MedianDegree = 1;

% fit 2 to calculate peak values and compare to local baseline value and calculate relative resistance changes.
% Choose:
% 1) Moving median between peaks, excluding compression dips before peak.
% In case of no plateau median of all negative values between peak. (Default)
% 2) Global median of values that are below zero after first baseline correction.
ChooserFit2 = 1;

% First correction and peak finding.
% Peak detection thresholds can be adjusted in code
% please note 60 Hz sampling is assumed

for n = specificchannels

    Yfilt = datatable((startpoints:endpoints),n);
    Y = medfilt1(Yfilt,MedianDegree);
    X = (0:length(Y)-1)./60;

    f1 = polyfit(X,Y,PolynomialDegree);
    Yest = polyval(f1,X);
    Ycorr = (Y-Yest);

    halfmaxlimit = (max(Ycorr)*0.5);
    [noisepks,noiselocs] = findpeaks(Ycorr);
    noiselimit = median(noisepks)*0.125;
    minpeakval = max([noiselimit halfmaxlimit]);
    [pks,locs] = findpeaks(Ycorr,'MINPEAKHEIGHT',minpeakval,'MINPEAKDISTANCE',20);
    peaknolist(n) = length(pks);
    peakloccell{n,1}= locs;

    if peaknolist(n) > 1
        peakbundleduration(n) = max(locs)-min(locs);
        beatratelist(n) = ((peaknolist(n)-1)/(peakbundleduration(n)/60));
    else
        beatratelist(n) = peaknolist(n)/((nopointsforanalysis)/60);
    end

    expfittable(:,n) = Y;
    expfitdatatable(:,n) = Yest;
    datatablecorr1(:,n) = Ycorr;
    timetablecorr(:,n) = X;
end

% Second correction. Choose zeroing procedure

if ChooserFit2 == 1;
    for n = specificchannels
        for n = 1 : noofchannels
peakno = peaknolist(n);
locsb = peakloccell{n,1};

Ycorr2 = datatablecorr1(:,n);
Yest2 = expfittable(:,n);
Ycorr3 = Ycorr2;

if (peakno>1)

    spacecorr = zeros(peakno,1);
deltas = zeros(peakno,1);
baseres = zeros(peakno,1);
reldeltas = zeros(peakno,1);

    for i = 1 : (peakno-1)
    space = Ycorr2(locsb(i):(locsb(i+1)-1));
    [pksCC,locsCC]= findpeaks(space);

    AA = find(diff(space)== 0,2,'first');
    BB = find(diff(space)== 0,2,'last');

    if (isempty(AA) == 0 && isempty(BB) == 0 && AA < BB);
    spacecorr(i) = median(space(AA:BB));
    newspace = space - spacecorr(i);
    deltares(i) = (Ycorr2(locsb(i))-spacecorr(i));
    baseres(i) = Yest2(locsb(i))+spacecorr(i);
    reldeltares(i) = deltares(i,1)/baseres(i,1);
    else
    spacecorr(i) = median(space(space < 0 ));
    newspace = space - spacecorr(i);
    deltares(i) = (Ycorr2(locsb(i))-spacecorr(i));
    baseres(i) = Yest2(locsb(i))+spacecorr(i);
    reldeltares(i) = deltares(i,1)/baseres(i,1);
    end
    Ycorr3(locsb(i):(locsb(i+1)-1))= newspace;
    end

    postspace = Ycorr2((locsb(peakno):length(Ycorr2)));
    newpostspace = postspace - spacecorr(peakno-1);
    Ycorr3((locsb(peakno):length(Ycorr2)))= newpostspace;

deltas(peakno) = (Ycorr2(locsb(peakno))- spacecorr(peakno-1));
baseres(peakno) = Yest2(locsb(peakno))+ spacecorr(peakno-1);
reldeltas(peakno) = deltas(peakno)/baseres(peakno-1);
datatablecorr2(:,n) = Ycorr3;

elseif (peakno == 1)

    spacecorr = median(Ycorr2((Ycorr2 <0)));
deltares = zeros(peakno,1);
baseres = zeros(peakno,1);
reldeltares = zeros(peakno,1);

deltares(peakno) = Ycorr2(locsb(peakno))- spacecorr;
baseres(peakno) = Yest2(locsb(peakno))+ spacecorr;
reldeltares(peakno) = deltares(peakno)/baseres(peakno);

Ycorr3 = Ycorr2 - spacecorr;
datatablecorr2(:,n) = Ycorr3;

elseif (peakno == 0)
    deltares = 0;
baseres = 0;
reldeltares = 0;
datatablecorr2(:,n) = Ycorr3;
end

deltarescell{n,1} = deltares;
baserescell{n,1} = baseres;
reldeltarescell{n,1} = reldeltares;

mean_deltares(n) = mean(deltares);
mean_baseres(n) = mean(baseres);
mean_reldeltares(n) = mean(reldeltares);

median_deltares(n) = median(deltares);
median_baseres(n) = median(baseres);
median_reldeltares(n) = median(reldeltares);

end
end

% 2) Global median of values below zero. Good for stacked signals
if Chooserfit2 == 2;
    for n = specificchannels
        peakno = peaknolist(n);
        locsb = peakloccell{n,1};

        Ycorr2 = datatablecorr1(:,n);
        Yest2 = expfittable(:,n);
        Ycorr3 = Ycorr2;

        spacecorr = median(Ycorr2((Ycorr2 < 0)));
% spacecorr = median(Ycorr2(200:300));

        if (peakno>0)
            deltares = zeros(peakno,1);
baseres = zeros(peakno,1);
reldeltares = zeros(peakno,1);
        end
    end
end
for i = 1 : (peakno-1)
    space = Ycorr2(locsb(i):(locsb(i+1)-1));
    newspace = space - spacecorr;
    deltares(i) = (Ycorr2(locsb(i))-spacecorr);
    baseres(i) = Yest2(locsb(i))+spacecorr;
    reldeltares(i) = deltares(i,1)/baseres(i,1);
    Ycorr3(locsb(i):(locsb(i+1)-1))= newspace;
end

postspace = Ycorr2((locsb(peakno):length(Ycorr2)));
newpostspace = postspace - spacecorr;
Ycorr3((locsb(peakno):length(Ycorr2))= newpostspace;

deltares(peakno) = (Ycorr2(locsb(peakno))- spacecorr);
baseres(peakno) = Yest2(locsb(peakno))+ spacecorr;
reldeltares(peakno) = deltares(peakno)/baseres(peakno);
datatablecorr2(:,n) = Ycorr3;
elseif (peakno == 0)
deltares = 0;
baseres = 0;
reldeltares = 0;
end

deltarescell{n,1} = deltares;
baserescell{n,1} = baseres;
reldeltarescell{n,1} = reldeltares;

mean_deltares(n) = mean(deltares);
mean_baseres(n) = mean(baseres);
mean_reldeltares(n) = mean(reldeltares);

median_deltares(n) = median(deltares);
median_baseres(n) = median(baseres);
median_reldeltares(n) = median(reldeltares);
end
end

To calculate stresses, use conversion factor from stoney's model on exported relative resistance change values.
For default plot, conversion factor calculated from standard gauge resistance of 125k used
Stress_conv_factor = 3.107528138998242e+004;

plotting loop
for n = specificchannels
locsc = peakloccell{n,1};
locstimes = timetablecorr(locsc,n);
basevals = baserescell{n,1};
peakvals = deltarescell{n,1};
twitchvals = (peakvals./basevals);

if peaknolist(n) > 1
    % % % data and detected peaks plot
    figure; plot(timetablecorr(:,n),expfitdatatable(:,n))
    hold on; plot(timetablecorr(:,n),expfittable(:,n),’r’)
    hold on; plot(locstimes,basevals,’b*’)
    hold on; plot(locstimes,(basevals+peakvals),’k^’,’markerfacecolor’,[1 0 0])
end

% % % approximate stress plot
figure; plot(timetablecorr(:,n),datatablecorr2(:,n)*(Stress_conv_factor/min(basevals)),’k’,’LineWidth’,1)

elseif peaknolist(n) == 0
    % % % data and first fit plot
    figure; plot(timetablecorr(:,n),expfitdatatable(:,n))
    hold on; plot(timetablecorr(:,n),expfittable(:,n),’r’)
end

end

% % % creating dataset arrays for saving and export

% expdata_valuetable = dataset(peaknolist, beatratelist, mean_reldeltares,mean_deltares, mean_baseres, median_reldeltares,
% median_deltares, median_baseres);
% expdata_traces = dataset(timetablecorr, expfitdatatable, expfittable,datatablecorr2);
% export(expdata_valuetable,’XLSFILE’,’name’)
%