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1 **Engineered Three-Dimensional Cardiac Fibrotic Tissue to Study Fibrotic Remodeling**

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41
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44

45 **Abstract**

46 Upon myocardial injury, activated cardiac fibroblasts (myofibroblasts (MyoFs)) play an essential
47 role in adverse cardiac remodeling, which in the long term cause cardiac fibrosis. As a result,
48 there is an increased risk of cardiac death due to arrhythmias and heart failure. However, the
49 abilities to study this process is complicated, as cardiac fibroblasts usually activate
50 spontaneously into cardiac MyoFs when cultured on two-dimensional (2D) culture plates. Here,
51 we present a simplified three-dimensional (3D) hydrogel platform of contractile cardiac tissue,
52 stimulated by transforming growth factor- β 1 (TGF- β 1), to recapitulate a cardiac fibrogenic
53 environment. We hypothesized that the quiescent state of cardiac fibroblasts can be controlled by
54 mimicking the mechanical stiffness of native heart tissue. To test this hypothesis, we created an
55 *in vitro* 3D cell culture model consisting of primary cardiomyocytes and cardiac fibroblasts
56 encapsulated within mechanically engineered gelatin methacryloyl (GelMA) hydrogel. We then
57 characterized the metabolic activity, structure, and contractility of the engineered heart tissue
58 constructs. Treatment with a beta-adrenergic agonist (isoprenaline) increased beating frequency
59 in the engineered cardiac tissues, demonstrating physiologic-like behavior of the constructs.
60 Subsequently, quiescent cardiac fibroblasts within the constructs were activated by the
61 exogenous addition of TGF- β 1. The expression of fibrotic protein markers (collagen I,
62 fibronectin, α -smooth muscle actin (α -SMA)) and the functional changes (eg. proliferation,
63 arrhythmogenicity) of the fibrotic-like tissues were analyzed to validate the model. This 3D

64 culture model of cardiomyocytes and cardiac fibroblasts exhibited physiological functions of
65 cardiac tissue and enabled controlled activation of MyoFs, thus demonstrating the usability of
66 this platform as a 3D culture model to study cardiac fibrotic remodeling. Furthermore, this
67 platform may be used as a more pathophysiologic-like culture model to study the effects of new
68 therapeutic agents.

69

70 **1. Introduction**

71

72 Cardiovascular diseases (CVDs), such as ischemic heart disease and hypertension, have
73 remained in the top 10 major causes of death worldwide.¹ Myocardial infarction (MI), which is
74 responsible for more than 50% of the deaths attributable to CVDs, results in a significant loss of
75 cardiomyocytes.^{2,3} This loss results in the initiation of a reparative wound healing process, which
76 is characterized by an initial inflammatory phase and followed by the proliferation and activation
77 of quiescent cardiac fibroblasts into cardiac myofibroblasts (MyoFs).⁴ Cardiac fibrosis results
78 from the excessive synthesis and accumulation of extracellular matrix (ECM) components (eg.
79 collagen, fibronectin), and is caused by the persistent activation and proliferation of both cardiac
80 fibroblasts and MyoF (**Fig 1**).³⁻⁹ In addition, cardiac MyoFs (hallmarked by the expression of α -
81 smooth muscle actin (α -SMA)) (**Fig 1**) have contractile properties, which can lead to a sustained
82 contractile stress that is exerted on the infarcted area.^{10,11} In the short term outlook, these pro-
83 fibrotic processes can be beneficial for cardiac function as it prevents dilatation and rupturing of
84 the ventricular wall.¹² However, prolonged and excessive activity of MyoFs results in excessive
85 fibrosis and tissue stiffening, which ultimately impairs cardiac function, increases the risk of
86 arrhythmia, and leads to the progression of end-stage heart failure.^{6,10,13,14}

87 Although there are many identified biochemical (eg. transforming growth factor- β 1
88 (TGF- β 1), angiotensin II, endothelin-1, platelet derived growth factor),^{4,8,15} mechanobiological,
89 (eg. tissue stiffness, mechanical strain, and hemodynamic stress)^{16,17} and cellular processes (eg.
90 cardiac fibroblasts migration, MyoF activation)⁷ that play a role in cardiac fibrosis, there are a
91 limited number of therapies available that effectively target fibrosis associated heart disease.^{8,10,18}
92 An important cause of the limited development of improved and more specific therapies against
93 cardiac fibrosis is the lack of biomimetic *in vitro* platforms to investigate the fibrogenic
94 remodeling after cardiac injury.⁴ A suitable *in vitro* model would preferably maintain cardiac
95 fibroblasts in a quiescent state and enable the integration of more physiological factors, such as
96 contractile tissue activity, cell-cell, cell-ECM, and paracrine and hormonal interactions. Thus
97 there exists a need for a novel, *in vitro* model system to study the pathological changes in
98 biomimetic and *in vivo*-like conditions. These systems could not only be used for studying
99 fibrotic changes in heart tissue, but they can potentially contribute to the development of more
100 physiologically relevant assay systems for drug screening.¹⁹

101 During the last decade, tissue engineering strategies have shown promise in designing
102 biomimetic *in vitro* models of cardiac tissue through the use of cardiac cells encapsulated in
103 three-dimensional (3D) hydrogel-based ECM.²⁰⁻²² For instance, the use of a gelatin methacryloyl
104 (GelMA)-based hydrogel in creating a functional and contractile cardiac tissue was demonstrated
105 by the successful encapsulation of cardiomyocytes and cardiac fibroblasts in a mechanically
106 tunable hydrogel.^{20,23} Additionally, different natural (eg. collagen, hyaluronic acid) and
107 synthetic-derived (eg. polyethylene glycol (PEG)) hydrogel culture models have been developed
108 to control and direct the activation of cardiac fibroblasts and fibroblast-like cells into MyoFs.²⁴⁻²⁶
109 However, there are still remaining challenges in engineering cardiac-like tissues to study MyoF

110 activation and the associated fibrotic remodeling. To date, most of the model systems that have
111 been used to study this, were based on either 2D²⁴ or mono-cultures of cardiac fibroblasts^{24,26}.
112 Similarly, the previously engineered cardiac-like tissues have not been used to study the
113 pathological remodeling that occurs during cardiac fibrosis. Consequently, some of the crucial
114 factors that need to be incorporated within *in vitro* culture platforms are different cardiac cells in
115 an *in vivo* like 3D microenvironment, which can be stimulated (externally) to exhibit a fibrosis
116 phenotype.

117 In the present study, we developed a 3D hydrogel-platform, composed of cardiomyocytes
118 and cardiac fibroblasts, which are used to engineer a physiologically relevant *in vitro* platform to
119 control the activation of cardiac fibroblasts towards MyoF. We hypothesized that by
120 mechanically tuning the stiffness of the hydrogels, a native-like ECM environment can be
121 created to enhance the quiescent state of cardiac fibroblasts, and the functional behavior of
122 engineered cardiac tissues. In addition, the physiological properties of these *in vitro* cardiac
123 tissues were characterized and the pro-fibrotic consequences of a TGF- β 1 induced activation of
124 cardiac fibroblasts were observed. We believe that this disease model of myocardial fibrosis may
125 be a suitable *in vitro* model to study bio-mechanistic processes of cardiac fibrosis. Moreover, this
126 platform could contribute to the development of better biomimetic pre-clinical drug screening
127 platforms.

128 **2. Materials and methods**

129 Synthesis of GelMA

130 GelMA was synthesized as described in a previous protocol.²³ Briefly, type A gelatin (10%
131 (w/v)) from porcine skin (Sigma-Aldrich) was added to Dulbecco's phosphate buffer saline

132 (DPBS; Gibco). This mixture was then stirred and heated at 50 °C for 1h to obtain a clear gelatin
133 solution. Subsequently, 1.25% (v/v) or 8% (v/v) methacrylic anhydride (Sigma-Aldrich) was
134 added dropwise to synthesize middle- (MM) and high-degree methacryloyl modification (HM)
135 GelMA. The solution was stirred and remained on a hot plate for 1h (middle methacryloyl
136 modification) or 2h (high methacryloyl modification), after which, DPBS was added to stop the
137 reaction. Following this, the GelMA solution was dialyzed (molecular weight cut off: 12 – 14
138 kDa) with deionized water for 10 days at 40 °C to remove any salts and unreacted methacrylate
139 anhydride. Finally, the GelMA solution was filtered (0.2 μm), frozen (-80 °C), and lyophilized
140 for 5 days to obtain GelMA foam. The foam was stored at room temperature until further
141 experimental use.

142

143 Preparation of Hydrogel Constructs

144 GelMA pre-polymer solutions were prepared by dissolving 5%, 7%, and 10% (w/v) MM and
145 HM GelMA in DPBS containing 0.25% (w/v) photoinitiator (PI; Igracure 2595 Sigma). The
146 solutions were briefly vortexed, and placed in an oven at 80 °C for 15 min to obtain pre-polymer
147 solutions of GelMA. To prepare disc-shaped hydrogels, 12 μL of the pre-polymer solution was
148 pipetted between two 600 μm tall spacers and covered with a 3-(trimethoxysilyl) propyl
149 methacrylate (TMSPMA) treated glass-slide (**Supplemental Fig 1**). This pre-polymer solution
150 was placed into a customized UV-chamber and exposed to UV light (800 mW, 360-480 nm) for
151 20 s, resulting in the creation of 600 μm tall hydrogel discs (**Supplemental Fig 1**). After this, the
152 GelMA hydrogels were removed manually from the glass slide and utilized for further
153 experiments.

154

155 Cardiac Cell Isolation and Culture

156 Primary ventricular cardiomyocytes and cardiac fibroblasts were isolated from two-day-old
157 neonatal Sprague Dawley rats. These procedures were based on a previously well-defined
158 protocol approved by the Institution's Committee on Animal Care.²⁷ Briefly, the hearts of
159 neonatal pups were surgically removed from the thoracic cavity after euthanasia. Upon removing
160 the atria, the ventricular tissues were cut into multiple small pieces and incubated overnight (at 4
161 °C) on a shaker in a 0.05% (w/v) trypsin solution prepared in Hank's Balanced Salt Solution
162 (HBSS, Gibco, USA). The heart tissues were subjected to four collagenase type II (LS004176,
163 Worthington, Lakewood, NJ) digestions (10 minutes, 37 °C, 80 rpm) to further digest the heart
164 tissues. The cell suspension was then collected, centrifuged (1000 rpm) for 5 min, and pre-plated
165 for 1 h to enrich the cardiomyocytes for immediate experimental use. The attached cardiac
166 fibroblasts were cultured for a maximum of three passages for future experimental use. The
167 cardiac fibroblasts were cultured in Dulbecco's Modified Eagle Medium (DMEM, Gibco USA)
168 with 10% fetal bovine serum (FBS; Gibco, USA) and 1% penicillin/streptomycin (P/S; Gibco,
169 USA).

170

171 Engineering Cell-Laden Hydrogel Constructs

172 To fabricate cell-laden hydrogel constructs, a pre-polymer solution was prepared with minor
173 modifications to the described protocol above. In brief, 5%, 7%, and 10% (w/v) MM and HM
174 GelMA was dissolved in DMEM containing 0.25% PI, 50% FBS, 1% P/S, and 2% (w/v) L-
175 glutamine (Gibco USA). Pre-polymer solutions were removed from the 80 °C oven, and placed

176 in a water bath at 37 °C until cell encapsulation. Cultured cardiac fibroblasts (passage 1-3) were
177 trypsinized and mixed at a 1:1 ratio with the freshly isolated cardiomyocytes to obtain a final
178 concentration of 25×10^6 cells/mL. The cells were centrifuged at 1200 rpm for 5 min, and the
179 pellet was resuspended in the GelMA pre-polymer solution. Gels were created following the
180 preparation of hydrogel constructs (above) and placed in culture medium containing DMEM,
181 supplemented with 10% FBS, 1% P/S, and 2% L-glutamine. In some conditions, media was
182 additionally supplemented with TGF- β 1 at a concentration of 2 ng/mL (100-21C, PeproTech,
183 USA) Medium was replaced consistently every 24 h throughout all experimental conditions.

184

185 Characterization of Hydrogels and Engineered Cardiac Tissues

186 Hydrogels were fabricated according to the described methods above to determine the
187 compressive modulus of the constructed (cell-laden) hydrogels. After fabrication, non-cell-laden
188 hydrogels were detached from the glass slide and allowed to swell overnight in DPBS at 4 °C.
189 Engineered cardiac tissues, however, were cultured in normal and TGF- β 1 containing medium
190 for 14 days before mechanical testing (n=5). Hydrogels were cut with a 5 mm biopsy punch, and
191 excess liquid was removed from the hydrogel. Gels were compressed with a uniaxial tensile
192 loading machine (Instron, 5542, USA) at a rate of 1mm/min with a 10 N cell load capacity. The
193 compressive modulus was calculated as the slope from 0-15% strain (n=4).

194 Scanning electron microscopy (SEM; Zeiss Ultra 55 SEM; Carl Zeiss, Thornwood, NY,
195 USA) was performed to characterize the hydrogel porosity. Cell-laden hydrogels were fixed at
196 day 1 and day 14 in 4% (v/v) paraformaldehyde (PF, 15700, Electron Microscopy Sciences,
197 Hatfield, PA) for 30 min at room temperature. Following fixation, the cell-laden hydrogels were
198 washed with DPBS and incubated at 4 °C overnight. Hydrated hydrogels were placed in liquid

199 nitrogen for 20 min and stored at -80 °C overnight. After freezing, the hydrogels were
200 lyophilized for 2 days to obtain a porous and foam-like GelMA hydrogel. The foams were
201 broken in half and coated with Pt/Pd to allow for cross-sectional imaging by SEM.
202 Quantification of the pore-size was performed by measuring pore-size diameter (n=150) from
203 SEM images (n=3) made from 5% HM, 7% MM, and 10% MM GelMA foams.

204 Hydrogel degradation was assessed by fabricating hydrogels and subjecting them to collagenase-
205 induced degradation. Hydrogels were fabricated and allowed to swell in DPBS overnight at 4 °C.
206 Hydrogels were then placed in a 0.5 U/mL collagenase type II solution (in DPBS) at 37 °C.
207 Excess liquid was removed, and the hydrogels were weighed before and after incubation with
208 collagenase. The weight loss percent was determined after 0.5, 1, 3 and 6 h (n=3).

209

210 Characterization of Cell Spreading and Cell Viability

211 Cell spreading within the 3D engineered cardiac tissues was determined by visualizing the
212 organization of F-actin fibers within the cells. The cell-laden hydrogels were fixed with 4% PF
213 solution for 30 min. Subsequently, the 3D encapsulated cells were permeabilized with 0.1% X-
214 100 Triton (Sigma-Aldrich) for 40 min at room temperature. This was followed by 45 min
215 incubation with Alexa Fluor 488 Phalloidin (Invitrogen) with a 1:40 dilution in DPBS. Cell
216 nuclei were counterstained with 4',6-diamidino-2-phenyl indole dihydrochloride (DAPI; Vector
217 Laboratories) for 20 min at room temperature. Hydrogels were then washed three times in DPBS
218 for 5 min. 3D imaging was performed by confocal microscopy (Leica SP5 X MP, Germany) to
219 visualize the fluorescently stained F-actin fibers and to determine the degree of cell spreading
220 within the hydrogels. Z-stack (100 µm each) images were taken of each hydrogel per condition

221 and four areas (400 μm x 400 μm) were selected for further quantification of cell spreading.
222 Fractional area coverage by F-actin was determined within the four selected windows using
223 ImageJ software.

224 Cell viability was examined with a Live/Dead fluorescent labeling kit (Invitrogen) on day 1
225 of culture according to the manufacturer's protocol. Hydrogels were first washed with DPBS
226 followed by an incubation with calcein-AM (0.5 $\mu\text{L}/\text{mL}$) and ethidium homodimer-1 (2 $\mu\text{L}/\text{mL}$)
227 in DPBS for 15 min at 37 $^{\circ}\text{C}$. After washing with DPBS, fluorescent images were taken from 4
228 selected areas using an inverted microscope (Nikon TE 2000-U, Nikon instruments Inc., USA).
229 To quantify viability, images were taken at 4 different focal planes within the hydrogel by
230 adjusting the height of the objective manually. Three cell-laden hydrogels were used to
231 determine the cell viability in each condition, and ImageJ software was used to quantify the
232 number of viable cells. Data depicted represents the percentage of live cells within the
233 engineered constructs. Cell metabolic activity was assessed throughout culture with PrestoBlue®
234 Cell Viability Reagent (PB; Life Technologies). The cell-laden hydrogels were incubated with
235 PrestoBlue for 2 h at 37 $^{\circ}\text{C}$ in a 1:10 dilution in normal culture medium (n=4). The fluorescence
236 was determined (560 nm – 590 nm) using a fluorescence reader (Synergy HT-Reader, BioTek,
237 Winooski, VT). The data was normalized to hydrogel samples without encapsulated cells. The
238 data represent the normalized fluorescence absorbance at day 1, 5, 10, and 14 of culture.

239

240 Cell Proliferation Analysis

241 Click-iT Plus EdU Alexa Fluor® 488 Imaging Kit (Life Technologies) was used to specifically
242 and quantifiably assess the number of proliferating cells within 3D cardiac tissues. Proliferating
243 cells were labeled following the manufacturer's guidelines. Briefly, cell-encapsulated hydrogels

244 (n=3) were incubated with 10×10^{-6} M EdU in normal culture medium at 37 °C. After 24 h of
245 incubation, the samples were fixed with 4% PF (30 min) and permeabilized with 0.1% X-100
246 Triton (40 min) at room temperature. The samples were blocked with 3% (w/v) bovine serum
247 albumin (BSA; Sigma-Aldrich) solution and subsequently incubated for 30 min with the Click-iT
248 solution at room temperature. Additionally, cell-specific proliferation was assessed by
249 immunostaining with vimentin, a mesenchymal cell specific marker. Subsequently, the samples
250 were washed twice with DPBS and counterstained for 20 min with DAPI at room temperature.
251 3D z-stack (50 μ m each) images were taken by confocal microscopy. To quantify proliferation,
252 fluorescence images were taken with an inverted microscope at 3 different focal planes within
253 the hydrogel by adjusting the height of the objective manually (n=3). ImageJ software was used
254 to count the number of EdU positive cells. Positive control for EdU labeling was determined by
255 staining cardiac fibroblasts cultured for 24 h in normal, and TGF- β 1 supplemented culture
256 medium (n=3). Fluorescence images were taken from each sample by using an inverted
257 microscope (n=10). The percentage of proliferating cells was calculated by counting the EdU
258 labeled cells using ImageJ software. Cell proliferation was calculated by dividing the EdU
259 positive cells by the total number of DAPI positive cells.

260

261 Immunofluorescence Staining for Cardiac (fibrosis) Specific Markers

262 The 3D engineered cardiac tissues were immunostained for cardiac tissue (sarcomeric α -actinin,
263 connexin-43) and cardiac fibrosis (α -SMA, collagen type I, fibronectin, and matrix-
264 metalloproteinase-2 (MMP-2)) markers. Samples were fixed with a 4% PF solution for 30 min,
265 followed by three washing steps (5 min each) with DPBS. Subsequently, the cell-hydrogels were

266 permeabilized by incubation with 0.1% X-100 triton for 45 min, after which the samples were
267 washed with DPBS and blocked for 30 min with a 10% goat serum solution in DPBS. After
268 blocking, the hydrogels were incubated with a monoclonal mouse anti-sarcomeric α -actinin
269 (Abcam, catalogue #9465), polyclonal rabbit anti-connexin-43 (Abcam, catalogue #11370),
270 monoclonal rabbit anti-vimentin (Abcam, catalogue #92547), monoclonal rabbit anti- α -SMA
271 (Abcam, catalogue #32575), polyclonal rabbit anti-collagen I (Abcam, catalogue #292),
272 polyclonal rabbit anti-fibronectin (Abcam, catalogue #23751), or a polyclonal rabbit anti-
273 MMP-2 (Abcam, catalogue #37150) for 16 h at 4 °C. After incubation with the primary antibody
274 (diluted 1:200 in 10% goat serum), the samples were washed three times (10 min each) in DPBS
275 at room temperature. The secondary antibodies (goat anti-rabbit Alexa Fluor 594 or goat anti-
276 mouse Alexa Fluor 488 (Abcam)) were diluted 1:200 in 10% goat serum, followed by incubation
277 with the samples for 2 h at room temperature. The nuclei were counterstained with DAPI.
278 Immunofluorescence double staining was performed by incubating two primary antibodies (eg.
279 sarcomeric α -actinin and connexin-43) simultaneously. After washing with DPBS three times,
280 the secondary antibodies were incubated separately for 2 h each. 3D confocal z-stack images
281 (150 μ m each) were taken and processed with ImageJ software.

282

283 Characterization of the Beating Behavior of Engineered Cardiac Tissue

284 The beating behavior of 3D engineered cardiac tissues was characterized quantitatively by using
285 a temperature controlled chamber (at 37 °C) and real time video recording with a camera (Sony
286 XCD-X710) attached to an inverted optical microscope. Videos of the beating constructs (n=3)
287 were recorded every day from day 3 of culture onwards. The beating pattern and frequency of

288 the constructs was determined by a custom written MATLAB program.²⁸ The single cell beating
289 characteristics of the engineered tissues, cultured in growth medium supplemented with TGF- β 1,
290 were also assessed with a modified custom written MATLAB program.

291

292 GelMA Hydrogel Contraction Assay

293 A GelMA hydrogel contraction assay was performed to assess the contractile manifestation of
294 MyoFs inside the engineered fibrotic-like cardiac tissues. The GelMA hydrogel contraction assay
295 was performed in a similar manner as a previously described collagen contraction assay.²⁴
296 Briefly, 3D cell-laden hydrogels were fabricated as described above and were cultured in TGF-
297 β 1 containing culture medium for 14 days according to the protocol. After 14 days, the culture
298 medium was aspirated and optical images were taken, followed by a quantitative analysis of the
299 gel diameters using ImageJ software. Depicted data represents mean \pm SD of gel diameter in
300 each condition (n=5).

301

302 Real-Time Polymerase Chain Reaction for Expression of Cardiac Fibrosis Markers

303 Cell-laden hydrogels were used to examine the expression of cardiac fibrosis markers. First, 3D
304 cardiac tissues were mechanically disrupted and total RNA was extracted from all samples using
305 TRIzol reagent (Life Technologies) and total RNA yield was measured with a NanoDrop
306 (Thermo Scientific). 1 μ g of total RNA from each sample was reverse transcribed according to
307 the manufacturer's instructions using the QuantiTect $\text{\textcircled{R}}$ Reverse Transcription kit (Qiagen). All
308 RT-PCR was performed using the iTaqTM Universal SYBR $\text{\textcircled{R}}$ Green supermix (Bio-Rad, USA).
309 The 20 μ L volume reaction component included 10 μ L supermix, 1 μ L of primer mix (5 μ M

310 forward/reverse primer), 100 ng template and nuclease free water (variable). Predesigned
311 KiCqStart[®] SYBR[®] Green primers (Sigma-aldrich) were obtained for the following target genes:
312 Collagen1A1 (catalogue #KSPQ12012G), Fibronectin (catalogue #KSPQ12012G), α -SMA
313 (catalogue #KSPQ12012G), and MMP-2 (catalogue #KSPQ12012G). Relative expressions were
314 calculated using $\Delta\Delta C_t$ method and normalized to glyceraldehyde-3-phosphate dehydrogenase
315 (GAPDH) gene expression.

316

317 Statistical Analysis

318 The quantitative results on all sample conditions were plotted by mean \pm standard deviation
319 (error bars). To perform statistical analysis, a student's t-test or one-way ANOVA was used. For
320 multiple comparisons, we used a Tukey's test. Graphpad Prism (v.6, GraphPad, USA) software
321 was used to perform all statistical analyses and results were considered to be significantly
322 different with a $p < 0.05$.

323 **3. Results**

324 3.1 Engineering and Characterization of GelMA Scaffolds

325 The elastic moduli of a healthy neonatal rat heart ranges from 4 to ~ 11 kPa.²⁹ In this study, we
326 encapsulated cells from neonatal rat hearts in a GelMA-based hydrogel to engineer 3D
327 myocardial tissues *in vitro*. The hydrogels showed an increased stiffness with increasing
328 methacryloyl modification degree and macromer concentration (**Fig 2A**). As such, 10% HM-
329 GelMA hydrogel exhibited the highest mechanical stiffness (25.76 ± 6.07 kPa) compared to all
330 other hydrogel conditions ($p < 0.05$). However, 7% HM-GelMA hydrogel (12.97 ± 2.12 kPa)

331 showed a significantly higher compressive modulus than 7% MM-GelMA hydrogel (4.48 ± 0.76
332 kPa) ($p < 0.05$) and a significantly lower compressive modulus when compared to 10% HM-
333 GelMA hydrogel. This verifies that the mechanical stiffness can be tuned by varying the
334 methacryloyl modification degree and macromer concentration of GelMA independently.²³ The
335 following three hydrogels 5% HM-GelMA (9.76 ± 4.48 kPa), 7% MM-GelMA (4.48 ± 0.76
336 kPa), and 10% MM-GelMA (7.25 ± 1.38 kPa), exhibited a mechanical stiffness that was in the
337 range of native neonatal rat hearts (**Fig 2A**) and were therefore further characterized.

338 To access the effect of the GelMA macromer concentration and methacryloyl modification
339 degree on the morphology of hydrogel, all SEM samples were prepared by same cryogenic
340 treatment. SEM images indicated that all three selected hydrogels showed highly microporous
341 structure (**Fig 2**).³⁰ There is a significant decrease in porosity with an increased macromer
342 concentration. In addition, a significantly lower porosity was observed in the 5% HM-GelMA
343 hydrogel as compared to the 7% and 10% MM-GelMA hydrogels (**Fig 2B**). This indicates an
344 inverse relationship between porosity and degree of methacryloyl modification and macromer
345 concentration. Although there was a significant decrease in the porosity of 5% HM-GelMA
346 hydrogel, no significant increase was observed in the mechanical properties of 5% HM-GelMA
347 hydrogel when compared to 7% and 10% MM-GelMA hydrogels (**Fig 2A**).

348 In native cardiac tissue, MMPs are excreted and activated by cells to induce and promote the
349 cleavage of ECM components.³¹ These proteins play an important role in the maintenance and
350 remodeling of the heart ECM. To assess the presence of physiological binding substrates for an
351 MMP-mediated degradation of the GelMA-based scaffold, a degradation assay was performed
352 with collagenase type II (also known as MMP-8). The results revealed significantly faster
353 degradation – described as percentage of weight loss – of 5% HM-GelMA hydrogel after 3 and 6

354 h when compared to 10% MM-GelMA hydrogel ($p < 0.05$). Furthermore, a complete degradation
355 of all three hydrogels was observed after 15 h of incubation with MMP-8 (**Fig 2C**). These results
356 confirm the existence of MMP substrates in the GelMA hydrogel but also indicate the
357 opportunity for use of GelMA based scaffolds for engineered physiological heart ECM tissue.

358

359 3.2 GelMA Hydrogel Characteristics Affect Cell Spreading but not Viability

360 To determine what hydrogel condition enabled the best cellular spreading and networking, we
361 investigated the spreading of encapsulated cells inside the three selected GelMA hydrogels. To
362 visualize this, fluorescent confocal z-stack images were taken after F-actin (cytoskeletal fiber)
363 staining of the cell-laden hydrogels (**Fig 3A**). After 10 days of culture, the fluorescence images
364 demonstrated that the majority of the cells inside 5% HM-GelMA and 10% MM-GelMA
365 hydrogels had limited spreading, as the cells retained a round shape. Interestingly, the expression
366 of F-actin fibers throughout the 7% MM-GelMA hydrogel clearly demonstrated an increase in
367 cellular spreading and networking (**Fig 3A**). A higher cellular spreading was expected due to a
368 lower methacryloyl modification degree and macromer concentration, thereby allowing for an
369 increased degradation and spreading throughout the ECM by the cells. Therefore, a higher
370 cellular spreading and networking in 7% MM-GelMA hydrogel could be attributed to the lower
371 macromer concentration and degree of methacryloyl modification as compared to 10% MM-
372 GelMA and 5% HM-GelMA hydrogels, respectively. Additionally, a quantitative analysis of the
373 area covered by F-actin fibers confirmed a significantly higher percentage of fractional coverage
374 in 7% MM-GelMA hydrogel (74.98 ± 17.70 %) compared to 5% HM-GelMA (22.71 ± 4.66 %)
375 and 10% MM-GelMA (42.68 ± 8.98 %) hydrogels at day 10 of culture ($p < 0.05$) (**Fig 3B**).

376 The viability of encapsulated cells in the different conditions was also assessed after one day of
377 culture. This time point was chosen to evaluate the survival of cells following UV exposure
378 during the fabrication of the constructs. **Figure 3C** depicts the quantitative analysis of the
379 percentage of live cells at day 1 of culture. Across all experimental conditions, the percentage of
380 live cells was higher than 84% and was not significantly different between groups.

381

382 3.3 Functional Properties of the 3D Engineered Cardiac Tissues

383 For further experiments, we selected the 7% MM-GelMA hydrogel for having the best spreading
384 and networking features for the cells. Additionally, we studied the viability of cells in this
385 condition for 2 weeks of culture. The results revealed that the engineered cardiac tissues
386 remained viable throughout a culture period of 14 days (**Fig 4A**). Compared to day 1, there was a
387 significant increase in the metabolic activity after 14 days (**Fig 4A**) ($p < 0.05$). Given the fact
388 that cardiomyocytes have limited proliferative capacity²⁰, we believe that cardiac fibroblasts
389 were responsible for the increase in metabolic activity.

390 On day 14, we also assessed the phenotype of the encapsulated cardiomyocytes and cardiac
391 fibroblasts by immunostaining with sarcomeric α -actinin and vimentin. As depicted in **Figure**
392 **4B**, a confocal z-stack image of an immunostained cell-laden GelMA hydrogel displayed both
393 sarcomeric α -actinin and vimentin positive cells. In addition, we investigated the expression of
394 connexin-43, a gap junction protein that is important for electrical coupling of cardiomyocytes
395 and is typically found in the cardiac tissue.³² From the confocal image in **Figure 4C**, it is clear
396 that cardiomyocytes demonstrated the expression of both connexin-43 and sarcomeric α -actinin.
397 The expression of gap-junctions and functional electro-mechanical coupling was also confirmed

398 by the observation of spontaneous, synchronous, and cardiac tissue-like contraction of the
399 engineered 3D cardiac tissues (**Supplemental Video 1**). Spontaneous individual cell beating
400 activity began after 2 days of culture; however, videos of the beating were recorded and analyzed
401 from when synchronous and tissue-like contraction began (**Fig 4D**). The engineered cardiac
402 tissues maintained synchronous and tissue-like contraction for as long as 18 days of culture
403 (**Supplemental Video 2**). Additionally, quantitative analysis of synchronous beats per minute
404 (BPM) revealed that the constructs reached a maximum of 48 (± 19.75) BPM on day 10
405 (**Supplemental Video 3**) and a minimum of 13 (± 5.57) BPM on day 18 of culture. This variable
406 beating behavior is consistent with previously reported studies on engineered GelMA-based
407 cardiac tissues *in vitro*.^{20,33,34} The decrease in beating rate with increasing culture time was also
408 shown in previous studies, in which rat neonatal cardiomyocytes were cultured on hydrogel-
409 based tissue-engineered models.^{27,35} In one of these studies, it was hypothesized that this
410 decrease might be attributable to the phenotypical transition of fetal cardiomyocytes towards the
411 neonatal stage.²⁷ We further went on to analyze the beating pattern of the beating constructs. As
412 depicted in **Figure 4E**, the cardiac tissues showed a stable and regular beating pattern throughout
413 culture days 4, 8, 12, and 16.

414 To assess physiological functionality of the engineered cardiac tissues, we also investigated the
415 effect of a beta-adrenergic drug, isoproterenol (isoprenaline), on the beating behavior on day 6.
416 Videos of the beating samples were recorded before and after 45 min incubation with 1 μ M
417 isoproterenol. An analysis of the videos revealed that the cardiac tissues exhibited physiological
418 behavior in response to the drugs (**Supplemental Video 4 and 5**). **Figure 4F** and **4G** show the
419 synchronous beating pattern (and amplitude) and beating frequency (in beats/min), respectively,
420 of the cardiac tissues before and after the administration of the drug. In the presence of

421 isoproterenol, the cardiac tissues developed a significant increase in their spontaneous beating
422 frequency (**Fig 4F**). Furthermore, an increase in the amplitude of all the samples was observed
423 after exposure to the drug (**Fig 4G**).

424

425 3.4 TGF- β 1 Induces Proliferation of Cardiac Fibroblasts

426 TGF- β 1 is a well characterized protein in the pathophysiology of cardiac fibrosis, and is a potent
427 stimulator of cardiac fibroblast proliferation during the course of this disease.¹⁵ In order to assess
428 the effect of TGF- β 1 on cell proliferation within our engineered 3D cardiac tissues, we analyzed
429 cell proliferation by EdU labeling. Fluorescent z-stack images of the EdU labeled cardiac tissues
430 were taken on day 1, 7, and 14 of culture (**Fig 5A**). To identify the cellular phenotype of
431 proliferating cells, we stained the cardiac fibroblasts by positive immunostaining for vimentin
432 (**Fig 5A**). From **Figure 5A**, it is clear that the EdU labeled cells (green) were also positively
433 stained by vimentin (red), thus confirming that the increase in proliferation across the two culture
434 conditions was attributed to the proliferation of cardiac fibroblasts. Furthermore, a quantitative
435 analysis of the percentage of EdU stained cells, revealed that the TGF- β 1 treated samples
436 showed a significantly higher number of EdU positive cells compared to the non-treated samples
437 on day 1 and 7 (**Fig 5B**) ($P < 0.05$). However, there was no significant difference on day 14
438 between the percentage of EdU positive cells in TGF- β 1 treated (41.63 ± 11.31 %) and non-
439 treated samples (42.45 ± 10.11 %). Additionally, the percentage of EdU labeled cells in the non-
440 treated samples had a significant increase by day 14 (42.45 ± 10.11 %) compared to day 1
441 (16.74 ± 3.70 %) (**Fig 5B**) ($p < 0.05$). A positive control analysis of the proliferation was also
442 obtained by the EdU labeling of TGF- β 1 treated and non-treated cardiac fibroblasts after 24h of

443 2D-seeding in a well-plate (**Supplemental Fig 2**). As expected, the results were consistent with
444 the results from the engineered cardiac tissues.

445

446 3.6 Characterization of the Expression of Cardiac Fibrosis Markers

447 Quiescent cardiac fibroblasts spontaneously differentiate into activated MyoFs when cultured in
448 conventional 2D tissue culture polystyrene (TCPS) plates (**Supplemental Fig 3A**).
449 Consequently, no significant difference was found in the expression of fibrotic markers after
450 incubation of these cells with TGF- β 1 for 24h (**Supplemental Fig 3A, B**). This is thought to be
451 partly due to the higher mechanical stiffness (GPa range) of TCPS compared to native and even
452 fibrotic myocardium.^{16,36} To investigate the activation of cardiac fibroblasts in a 3D beating
453 heart-like environment and stiffness, we analyzed the expression of a specific MyoF protein
454 marker, α -SMA after 14 days of culture (**Fig 6A**). As depicted in **Figure 6A**, cells in the cell-
455 laden hydrogels showed a minimal expression of α -SMA, indicating that cardiac fibroblasts
456 within the cardiac tissues remained in a quiescent state when cultured in normal culture medium.
457 However, engineered 3D cardiac tissues cultured in the presence of TGF- β 1, exhibited a more
458 MyoFs-like phenotype after 14 days, as can be seen in **Figure 6A**. Furthermore, we analyzed the
459 expression of other cardiac fibrosis markers by positive immunofluorescence staining of
460 collagen-I, fibronectin, and MMP-2. Overall, confocal z-stack images clearly demonstrated an
461 increased expression of fibronectin and collagen-I inside the engineered 3D cardiac tissues
462 cultured in TGF- β 1 supplemented medium. However, the expression MMP-2 was not clearly
463 increased in the TGF- β 1 treated samples. In addition to qualitative protein expression analysis,
464 we also investigated the mRNA expression of α -SMA, collagen-I, fibronectin, and MMP-2 by

465 quantitative real time polymerase chain reaction (RT-PCR) to confirm the upregulation of
466 fibrotic protein expression in the TGF- β 1 stimulated samples (**Fig 6B**). The expression levels of
467 α -SMA, collagen-I, and fibronectin were shown to be elevated for TGF- β 1 stimulated samples
468 when compared to normal growth medium. The greatest increase was observed in the fibronectin
469 samples, where the mRNA expression was $3.84 (\pm 2.33)$ ($p < 0.05$) times higher in the TGF- β 1
470 treated samples as compared to the control group. Additionally, mRNA expression levels of α -
471 SMA and collagen-I were shown to be $2.51 (\pm 1.21)$ ($p < 0.05$) and $3.48 (\pm 1.55)$ ($p < 0.05$)
472 times higher, respectively, in TGF- β 1 cultured samples as compared to control medium.
473 However, as demonstrated by the protein expression analysis (**Fig 6A**), the expression of MMP-2
474 was not significantly changed in TGF- β 1 stimulated samples (1.01 ± 0.53). Altogether, these
475 results indicated that a quiescent cardiac fibroblast phenotype could be effectively cultured in a
476 mechanically tuned GelMA-based cardiac tissue construct. Moreover, these results demonstrated
477 that the phenotypic state of cardiac fibroblasts can be directed by designing a 3D cardiac tissue
478 with a physiological co-culture of cells and an *in vivo*-like dynamic contraction.

479

480 3.5 Analysis of the TGF- β 1 induced Pro-Fibrotic Changes

481 During fibrotic remodeling there is a higher risk of arrhythmogenicity as a result of an increased
482 ECM deposition and an altered electrical-coupling between cardiomyocytes and MyoFs.^{13,37} In
483 addition, human fibrotic cardiac tissue is hallmarked by an increased mechanical stiffness (~30-
484 70 kPa) when compared to healthy myocardial tissue (~10 kPa).^{38,39} Furthermore, MyoFs can
485 generate a higher contractile force than cardiac fibroblasts and thereby induce contraction and
486 scarring of the cardiac tissue. We hypothesized that the engineered cardiac tissues would exhibit

487 some of these fibrotic characteristics when the fibrotic response is simulated. To assess the pro-
488 fibrotic changes, we cultured the samples in normal culture medium and added TGF- β 1 for 14
489 days. All TGF- β 1 stimulated cardiac tissues showed spontaneous individual cell beating, from
490 day 7 to day 13. Video analysis of the beating behavior revealed that there was a non-
491 synchronous and irregular contraction of all the tissues when stimulated with TGF- β 1 (**Fig 7A**
492 and **Supplemental Video 6**). These results correlated well with previous reports that MyoFs
493 induce changes in the beating behavior of cardiomyocytes.^{13,37}

494 Compared with normal culture medium, we also observed a significant decrease of ~17 % in the
495 average diameter of the GelMA-based cardiac tissues, thus indicating an increase in the MyoF-
496 mediated contraction of the hydrogel (**Fig 7 B, C**) ($p < 0.05$). In addition, SEM images of both
497 culture conditions showed that cardiac tissues from TGF- β 1 stimulated samples clearly had a
498 more fibrous and fibrillar structure than cardiac tissues in the control group (**Fig 7D**). This is
499 thought to be mainly attributable to the elevated deposition of ECM components (eg. collagen-I,
500 fibronectin) in the fibrotic-like tissues. On day 14, the mechanical stiffness of TGF- β 1 treated
501 engineered 3D cardiac tissues was compared to cardiac tissues cultured in normal culture
502 medium. These measurements revealed an increase in the mechanical stiffness of TGF- β 1
503 stimulated encapsulated cells in GelMA hydrogels (**Fig 7E**).

504 Our findings highlight the opportunity to use these GelMA-based engineered 3D cardiac tissue
505 constructs to identify fibrotic changes, and to study the pathophysiological cells and factors that
506 play a role in cardiac remodeling and myocardial fibrosis.

507

508 **Discussion**

509 For many decades, conventional TCPS plates have successfully contributed to a better
510 understanding of fibrogenesis in the context of cardiovascular diseases. However, these models
511 lack the *in vivo* like presence of tissue-level properties (such as cell-ECM interactions).
512 Furthermore, cardiac fibroblasts cultured in a 2D TCPS plate, spontaneously activate into MyoFs,
513 thereby complicating the ability to study phenotypical changes of these cells during disease
514 development. In this work, we designed a simplified 3D *in vitro* model of cardiac fibrosis by
515 tuning the mechanical ECM of engineered cardiac tissues followed by stimulating these
516 hydrogel-based tissues with TGF- β 1; an established and potent mediator in fibrotic remodeling.¹⁵
517 During cardiac fibrosis, there is an increased synthesis and deposition of ECM components,
518 including collagen (mainly type I and III), laminin, fibronectin, and elastin.^{4,40} During this
519 process, the main effector cells are activated MyoFs and proliferating cardiac fibroblasts. In
520 addition, there is an increased (early phase remodeling) and decreased (late phase remodeling)
521 production and activation of MMPs, which play a role in the degradation of the ECM.^{31,41} This is
522 of importance since it can enable and facilitate the migration of cells (eg. cardiac fibroblasts) to
523 the area of injury at the early phases of wound healing.⁴² The elevated expression of pro-fibrotic
524 genes such as collagen-I, fibronectin, and α -SMA in our stimulated cardiac tissue was consistent
525 with the pathological changes that occur during cardiac fibrosis. Similarly, we observed an
526 activation of quiescent cardiac fibroblasts by positively immunostaining α -SMA; a widely used
527 marker of MyoFs.⁴³ Immunostaining images further revealed that the TGF- β 1 stimulated tissues
528 also increased the expression of collagen-I and fibronectin. Our results also indicated that there
529 was an induced proliferation of cells in the TGF- β 1 treated cardiac tissues, as identified by EdU
530 labeling. It is well established that TGF- β 1 is an inducer of cardiac fibroblast activation during
531 pathological fibrotic remodeling.⁴⁴ In addition, TGF- β 1 is a proliferative stimulator of cardiac

532 fibroblasts.^{44,45} Therefore, we investigated whether the proliferating cells are cardiac fibroblasts
533 or cardiomyocytes by simultaneously labeling with EdU and immunostaining specifically for
534 cardiac fibroblasts. Our results showed that the proliferating cells were cardiac fibroblasts rather
535 than cardiomyocytes, which have a low proliferative capacity.⁴⁶

536 This disease model also showed that it could recapitulate functional properties of MyoFs and
537 fibrotic cardiac tissue. TGF- β 1 induced stimulation of the functional cardiac tissues resulted in
538 an asynchronous and irregular beating behavior. Furthermore, the increased conversion of
539 cardiac fibroblasts into MyoFs, resulted in a higher contraction and shrinkage of the 3D GelMA
540 hydrogels. This however, could also be a result of degradation of GelMA hydrogel due to the
541 production of various MMPs in the TGF- β 1 stimulated tissues.

542 GelMA hydrogels have been a widely used scaffolding biomaterial in the past few years for
543 applications ranging from tissue engineering³³, organs-on-a-chip,⁴⁷ and *in vitro* disease models.²⁵
544 Most of these studies take advantage of the easy fabrication methods, high and easy accessibility,
545 and high biocompatibility of GelMA hydrogels.²³ Moreover, GelMA hydrogels can be tuned
546 mechanically to obtain a physiologically relevant stiffness.^{23,48} This can be obtained by varying
547 the macromer concentration, the degree of methacryloyl modification, the concentration of
548 photoinitiator, the UV intensity, or the UV exposure time.²³ Although GelMA hydrogel is a
549 semi-synthetic, photocrosslinkable scaffolding material, it provides relevant biomimetic cues of
550 the native ECM, such as RGD-binding peptides and MMP degradable sites.⁴⁹ In addition, gelatin
551 is a denatured form of collagen, which is the main constituent of native heart tissue, and
552 therefore exhibits comparable biocompatible and biomimetic properties to collagen. However, a
553 limitation of this *in vitro* system is the lack of other ECM components that are present in the
554 native heart (eg. collagens, glycosaminoglycans, laminin, and elastin). Interestingly, previous

555 studies have shown that it is possible to engineer functional cardiac tissue by using naturally
556 derived decellularized ECM from native heart tissues.^{50,51} In a recent study, Visser *et al.* have
557 combined the fabrication methods of GelMA molecules to engineer crosslinkable hydrogels
558 derived from various native tissues.⁵² Thus indicating that in future work, it could be possible to
559 combine the advantages of GelMA hydrogels with the native properties of a decellularized heart
560 ECM.

561
562 Mechanotransduction is a process in which cells sense their mechanical microenvironment and
563 transmit the physical stresses of their surroundings to biochemical signals that result in various
564 cellular outputs (eg. differentiation, proliferation).⁵³⁻⁵⁵ These mechano-physical signals are
565 mainly mediated through cell-ECM and cell-cell connections, which are converted to cellular
566 signaling pathways by integrins, focal adhesions, and cadherins.^{4,53,56} Cardiac fibroblasts and
567 cardiomyocytes can sense their extracellular microenvironment by attaching to their ECM with
568 focal adhesions.⁵⁷ Increasing ECM stiffness or ECM-induced strain, can affect both
569 cardiomyocytes and cardiac fibroblasts in a way that alters contractile function and stimulates
570 MyoFs activation.^{16,58} Consequently, the mechanical stiffness of the extracellular
571 microenvironment of cardiac tissue is an important factor in both normal physiology and cardiac
572 fibrosis. In fact, in a previous study by Engler *et al.* it was reported that cardiomyocytes cultured
573 on stiff, fibrotic-like (34 kPa) substrates, showed a decreased beating activity and lacked the
574 development of well-striated sarcomere structures.³⁹ In addition, results from other recent studies
575 have demonstrated that the mechanical stiffness of the matrix or substrate can facilitate the
576 transition of cardiac fibroblasts into MyoFs.^{24,26} Zhao and colleagues engineered PEG-substrates
577 with varying degrees of stiffness to study the migration, proliferation, and activation of quiescent

578 cardiac fibroblasts in a 2D *in vitro* model system.²⁴ This culture platform suggested that the
579 quiescent state of cardiac fibroblasts could be maintained and directed by mechanically tuning a
580 hydrogel-based substrate. These results highlight the importance of the tunability of matrix
581 stiffness for the engineering of functional and contractile cardiac tissues, while simultaneously
582 maintaining the quiescent-like phenotype of cardiac fibroblasts. In this study, we used 7% MM-
583 GelMA hydrogel (4.48 ± 0.76 kPa) to engineer a physiological stiffness in the same range of
584 native neonatal rat hearts (4-11 kPa).²⁹ We showed that neonatal rat cardiac fibroblasts remained
585 in a quiescent-state while co-cultured with cardiomyocytes in a mechanically tunable 3D
586 hydrogel model.

587 Cardiac muscle is a syncytium in which a network of cardiomyocytes and cardiac fibroblasts are
588 connected to each other electrically and mechanically. In order to engineer a physiologic-like
589 heart tissue *in vitro*, it is essential for cardiomyocytes and cardiac fibroblasts to interact with
590 each other through both direct, and indirect cell-cell and cell-ECM interactions. Here, we
591 hypothesized that the spreading and networking of cells over time would lead to the development
592 of a system which better mimics native cardiac tissue. Consequently, 7% MM-GelMA hydrogel
593 was selected as a 3D substrate for culturing cardiac fibroblasts and cardiomyocytes. The
594 expression of cardiac differentiation markers (sarcomeric α -actinin, connexin-43) demonstrated
595 a well-developed electrical coupling and contractile apparatus inside these hydrogel conditions.
596 Although the cardiomyocytes had an isotropic orientation inside GelMA hydrogel, there was a
597 clear elongation and a well-defined sarcomeric structure visible. In native cardiac tissue however,
598 the cardiomyocytes are elongated and aligned in anisotropic layers of muscle tissue.⁵⁹ This is
599 important for the anisotropic propagation of electrical signals, which plays a critical role in
600 synchronous and rhythmic cardiac contraction.^{60,61} One limitation of our engineered cardiac

601 tissue was the lack of cellular anisotropic alignment, which could potentially lead to impaired
602 electrical pulse propagation and contraction in comparison to the native heart muscle. In future
603 studies, we could overcome this through the use of engineering strategies by applying
604 topographical and electrical cues to the tissues to enhance the elongation and alignment of
605 cells.^{62,63}

606 In this work, we used neonatal rat cardiomyocytes and cardiac fibroblasts to model an *in vitro*
607 platform of myocardial fibrosis. Currently, the use of rats for *in vitro* studies of heart disease is
608 the standard in both academia and the pharmaceutical and biotechnology industry. Therefore,
609 this indicates that our model may be a suitable and pre-clinically relevant *in vitro* platform for
610 studies of pathophysiology and drug screening applications. However, we recognize the
611 simplicity of this *in vitro* platform, as it lacks the incorporation of other dynamic (eg. blood flow)
612 and physiological factors (eg. blood vessels, inflammatory cells), which are present in the native
613 heart. Recently, microfluidic organs-on-a-chip have emerged and demonstrated the possibility of
614 incorporating several dynamic factors (such as fluid flow and mechanical stretch) within
615 microengineered cardiac tissues.^{64,65} Further development of microfluidic techniques and tissue
616 engineering strategies could therefore aid in the creation of a more physiologically relevant
617 cardiac tissue in the future.

618 **Conclusion**

619 In this study, we engineered a simplified but functional and physiologic-like heart tissue to study
620 cardiac fibrosis in a 3D GelMA-based hydrogel platform. By tuning the mechanical stiffness of
621 GelMA hydrogels, we were able to create a well-defined and beating network of cardiomyocytes
622 and quiescent cardiac fibroblasts. Subsequently, we showed that we were able to stimulate the

623 activation of cardiac fibroblasts into MyoFs by adding TGF- β 1 to the culture medium.
624 Furthermore, our results demonstrated that the engineered fibrotic tissues presented electrical and
625 mechanical alterations that are comparable to fibrotic heart tissues. In conclusion, our study
626 presents a physiologic-like *in vitro* model of cardiac fibrosis that could enhance our
627 understanding of this disease, while increasing the potential of these systems to be used for pre-
628 clinical drug screenings.

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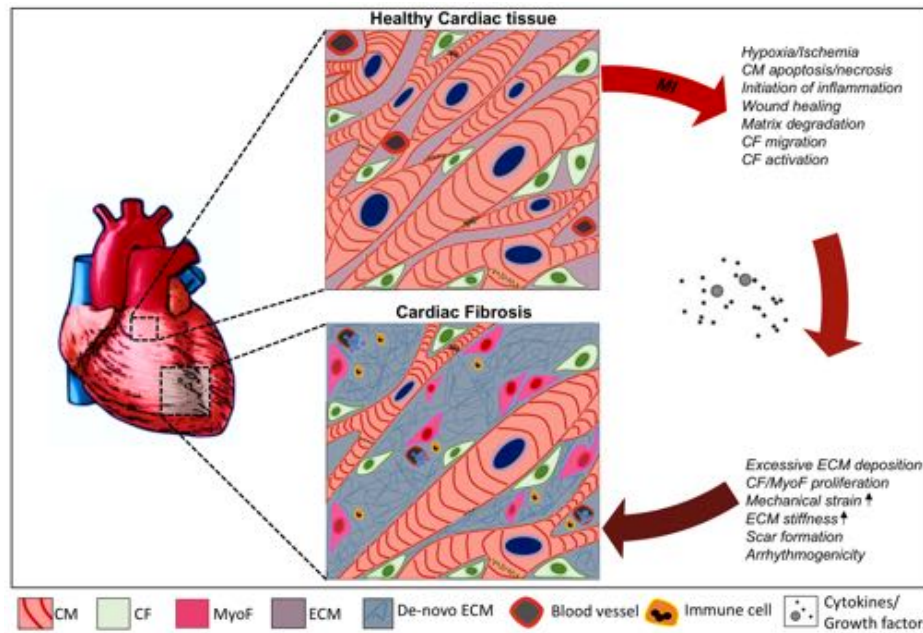
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649 **Figure 1. Schematic illustration of the pathophysiological changes during fibrotic cardiac**

650 **remodeling.** Healthy myocardial tissue consists of a network of cardiomyocytes (CM) and

651 quiescent cardiac fibroblasts (CF) that are interspersed within the extracellular matrix (ECM).

652 After myocardial injury (eg. myocardial infarct (MI)), CMs die and a reparative inflammatory

653 and wound healing process is initiated by the release of various cytokines and growth factors

654 (such as transforming growth factor-β1, angiotensin-II etc.). This results in the activation of

655 cardiac fibroblasts into cardiac myofibroblasts (MyoF). These cells and other resident cardiac

656 fibroblasts are responsible for an excessive and prolonged synthesis and deposition of de-novo

657 ECM proteins (eg. collagen-I, fibronectin, laminin). This results in scarring of the heart tissue

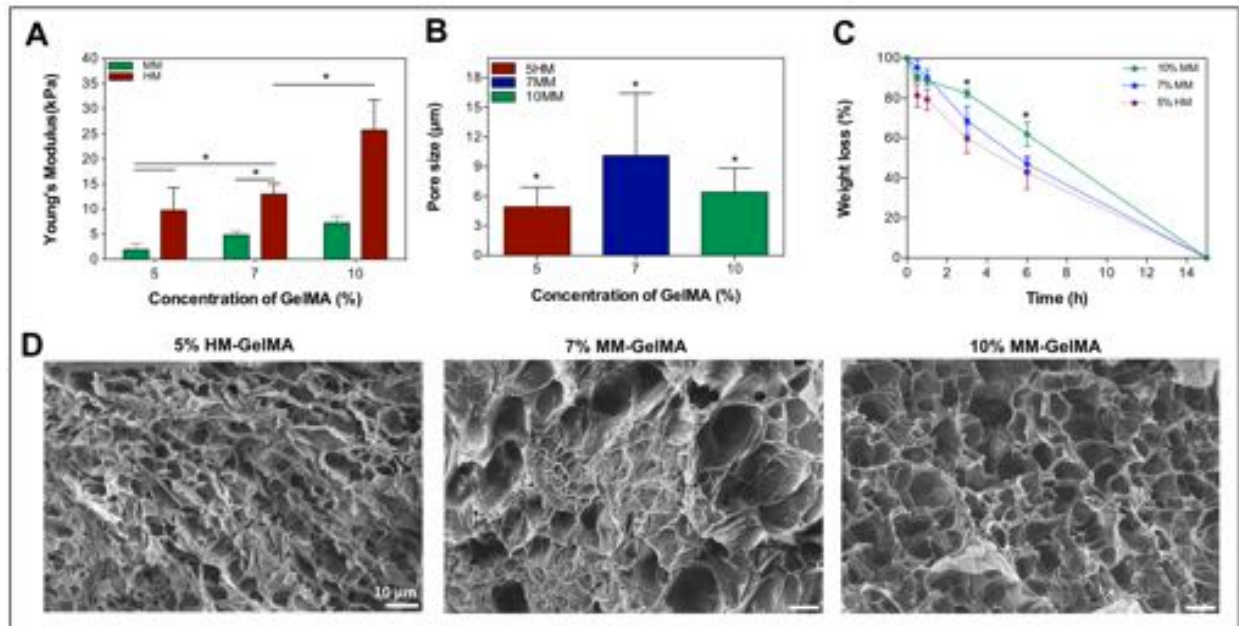
658 and leads to the deterioration of ventricular function followed by diastolic and systolic

659 ventricular dysfunction and may eventually lead to life threatening arrhythmogenicity or heart

660 failure.

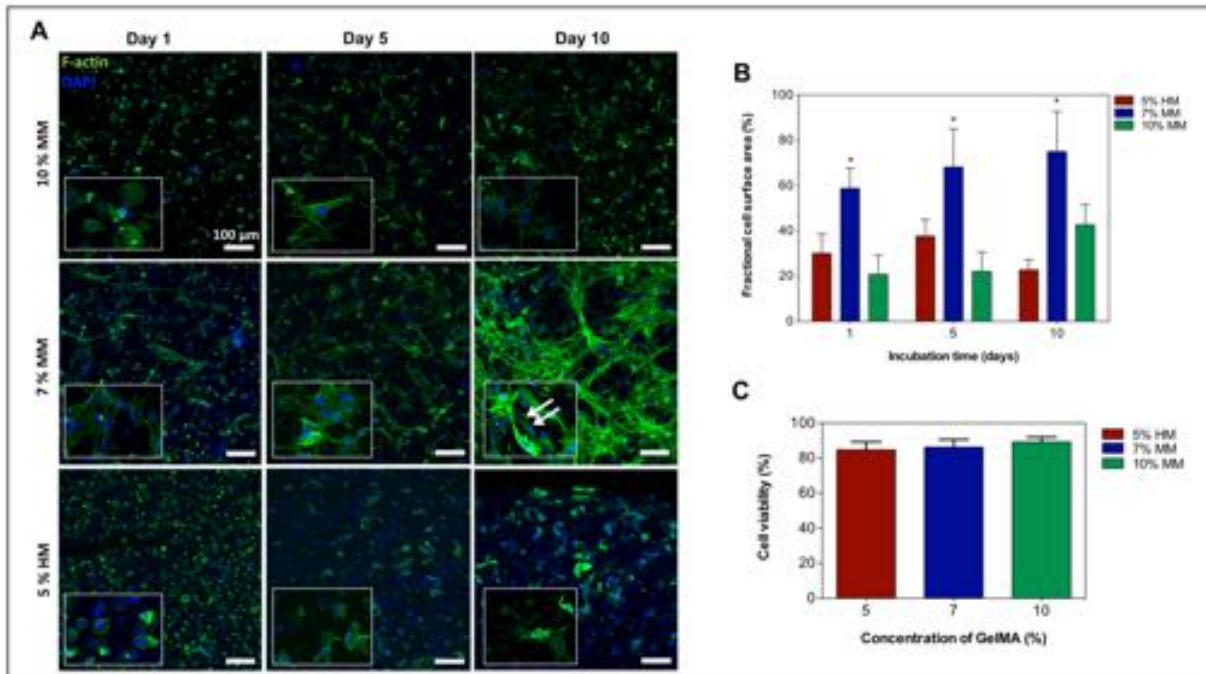
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664 **Figure 2. Mechanical, porosity and degradation properties of GelMA hydrogels.** A) pre
665 modulus of GelMA-hydrogels varies with different macromer concentration and degree of
666 methacryloyl modification. B) Pore size analysis of GelMA hydrogels. C) Degradation of
667 GelMA hydrogels with various macromer concentration and methacryloyl modification degree in
668 the presence of collagenase. D) Cross-Sectional scanning electron microscopy images of 5%
669 HM-GelMA, 7% MM-GelMA, and 10% MM-GelMA hydrogels reveal different porosity. Data
670 depict Mean \pm Standard deviation. * $p < 0.05$



671 **Figure 3. Viability and spreading characteristics of cardiac fibroblasts and cardiomyocytes**
 672 **encapsulated in mechanically tuned GelMA hydrogels.** A) Representative fluorescence
 673 images of encapsulated cardiac fibroblasts and cardiomyocytes within various GelMA hydrogels
 674 at day 1, 5, and 10 of culture. B) Representation of a quantitative analysis of fractional F-actin
 675 coverage within selected windows of 400 μm x 400 μm . C) Quantitative analysis of the viability
 676 of cardiomyocytes and cardiac fibroblasts within various GelMA hydrogels conditions on day 1
 677 of culture. Data depict Mean \pm Standard deviation. * $p < 0.05$

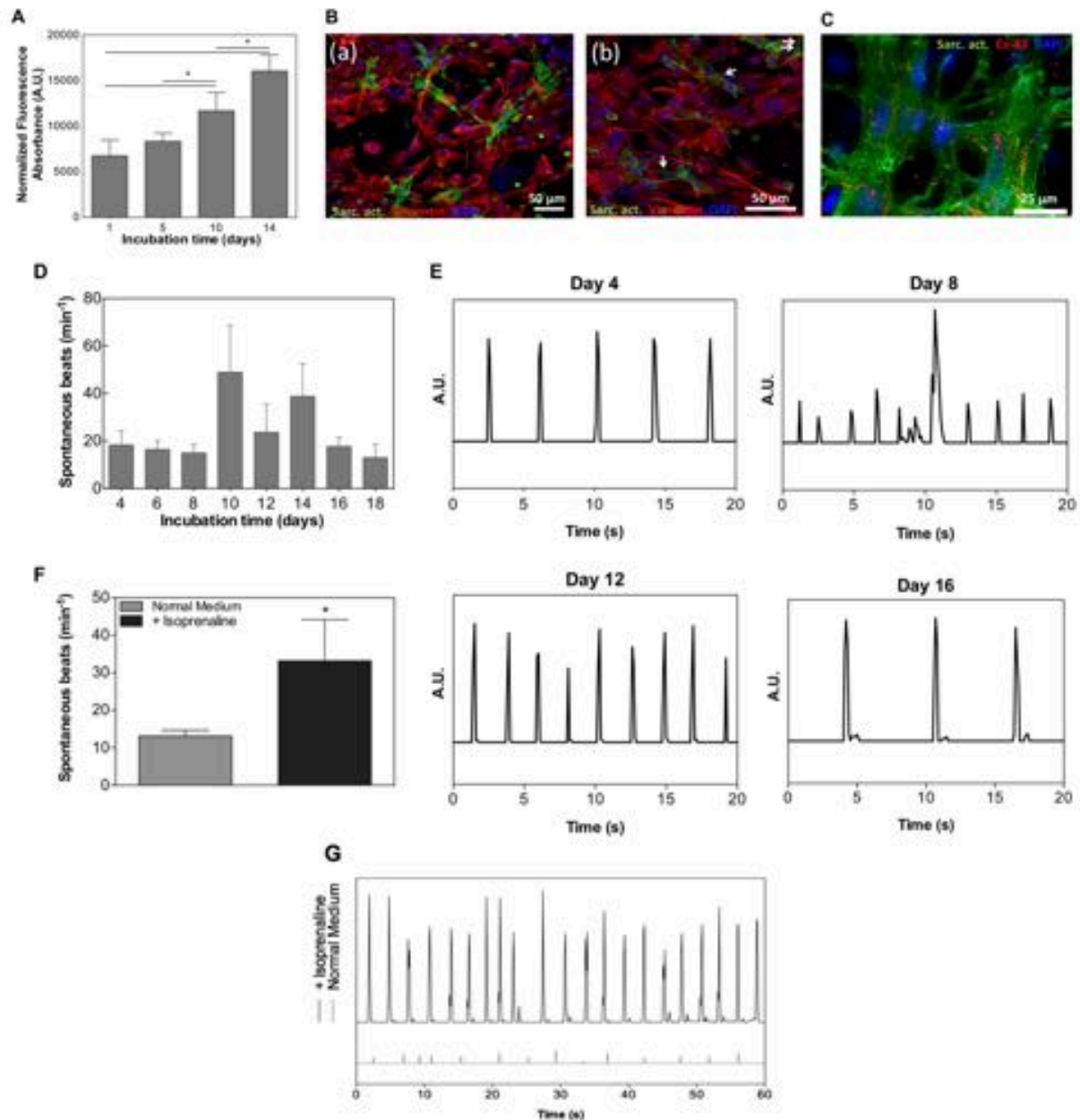
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 684 **Figure 4. Functional Properties of 3D engineered cardiac tissues.** A) Quantitative analysis of
 685 cellular metabolic activity throughout 14 days of culture. B) Representative fluorescence images
 686 of immunostained cardiomyocytes (α -sarcomeric actin = green) and cardiac fibroblasts (vimentin
 687 = red) on day 14 of culture (a). Higher magnification images of immunostained cardiomyocytes
 688 and cardiac fibroblasts showed sarcomeric cross-striations (white arrows) (b). C) Fluorescent
 689 image showing well developed sarcomeric striations (= green) and the expression of a gap-

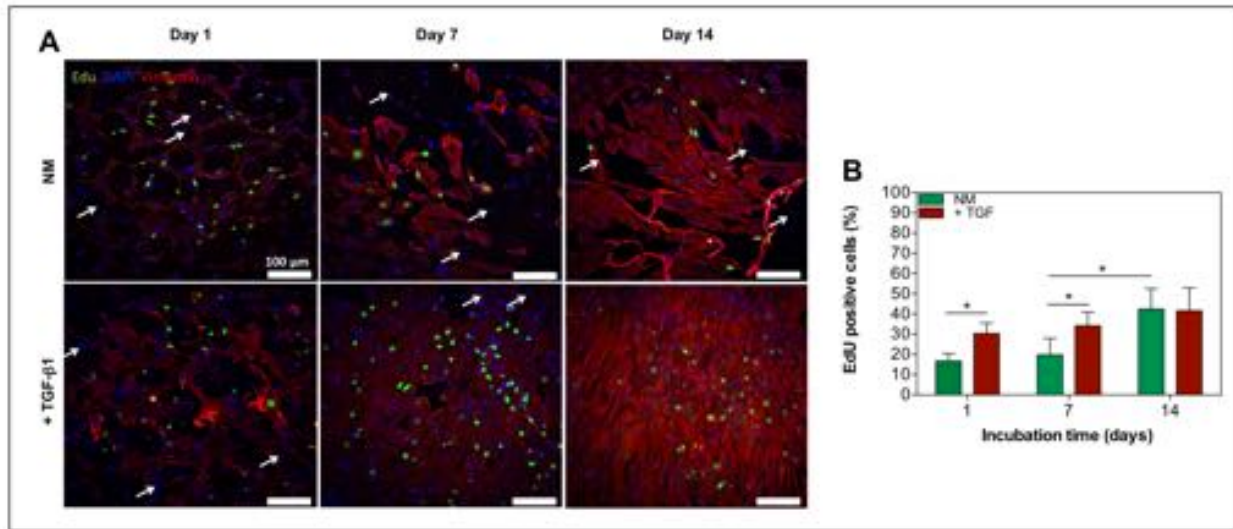
690 junctional protein, connexin-43 (= red). **D)** Quantitative representation of the spontaneous beats
691 per minute of 3D engineered cardiac tissues from day 4 up until a maximum of 18 days of
692 culture. **E)** Representative beating pattern of the cardiac tissues at day 4, 8, 12, and 16 of culture.
693 **F)** Quantitative analysis of the spontaneous beats per minute of cardiac tissues (n=3) in the
694 absence and presence of 1 μ M isoproterenol (isoprenaline). **G)** Representation of the beating
695 pattern of cardiac tissues in the absence and presence of isoproterenol at day 6. Data depict Mean
696 \pm Standard deviation. * p <0.05

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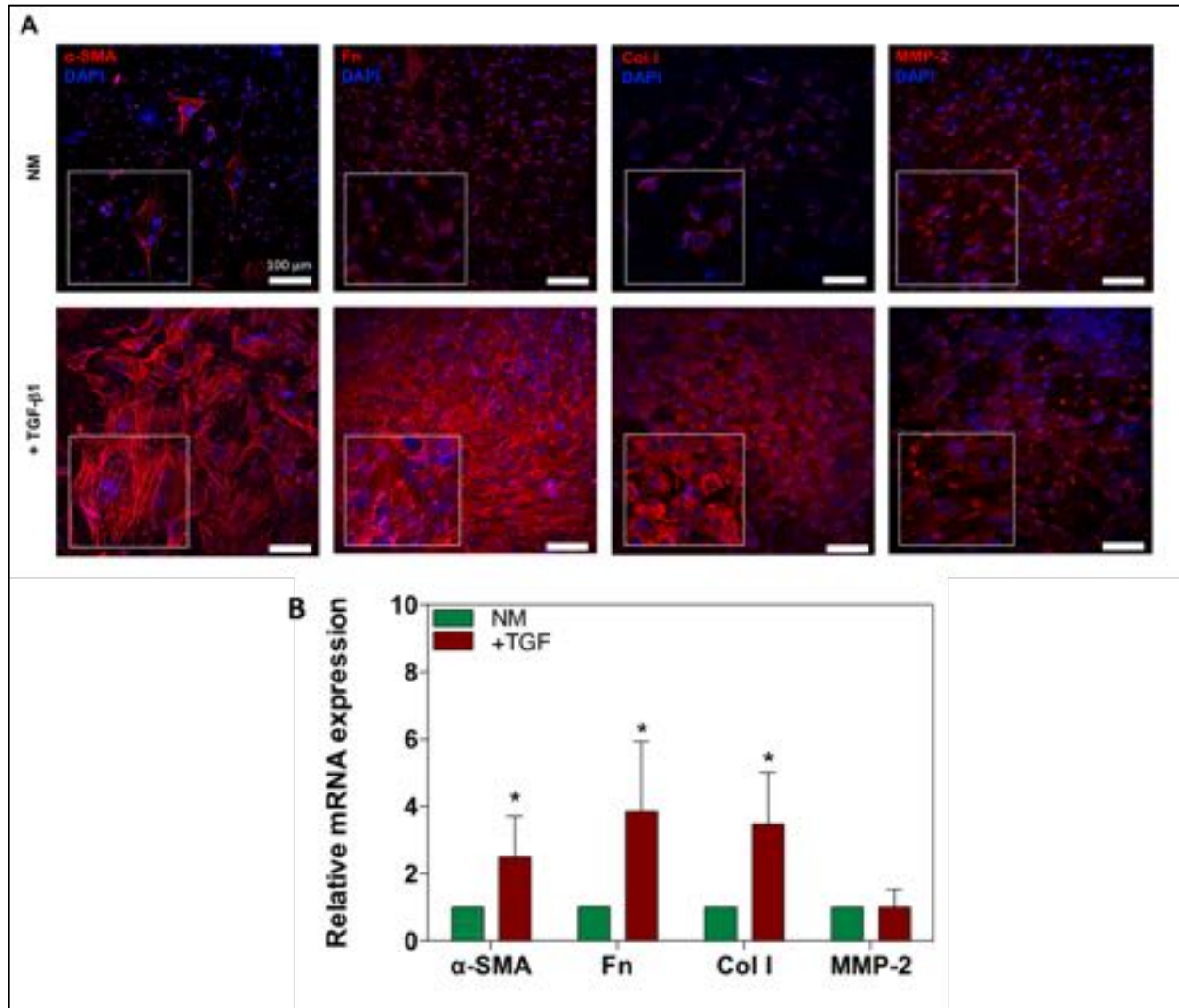
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700
 701 **Figure 5. The exogenous addition of TGF-β1 affects proliferation of cardiac fibroblasts in**
 702 **3D engineered cardiac tissues. A)** Confocal images of immunofluorescence staining of a
 703 cardiac fibroblast marker, vimentin (= red), and EdU click-iT labeling (= green) of 3D
 704 engineered cardiac tissues with and without the addition of TGF-β1 at day 1, 7, and 14.
 705 Cardiomyocytes were not stained and showed no positive EdU labeling (white arrows) **B)**
 706 Representative quantification of proliferating cells inside 3D cardiac tissues as determined by the
 707 percentage of EdU positive cells at day 1, 7, and 14 of culture (n=3). Data depict Mean ±
 708 Standard deviation. * $p < 0.05$

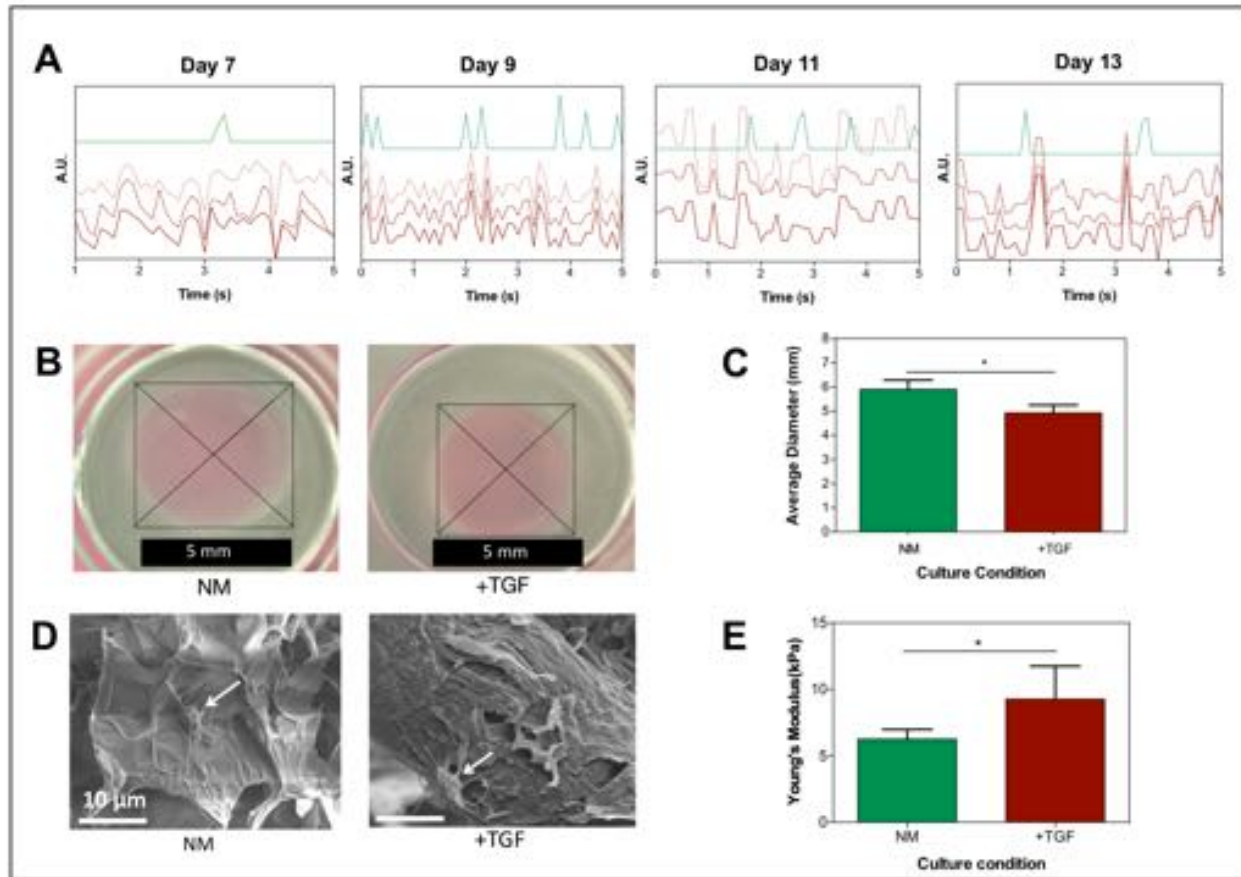
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712 **Figure 6. Increased expression of fibrotic makers and increased differentiation of quiescent**
 713 **cardiac fibroblasts into MyoFs by TGF- β 1.** A) Confocal images of immunofluorescence
 714 stained markers of cardiac fibrosis and MyoF differentiation; α -SMA, collagen-I (Col I),
 715 fibronectin (Fn), and MMP-2 after 14 days of culture. B) Data representing RT-PCR of mRNA
 716 expression of α -SMA, collagen-I, fibronectin, and MMP-2 in normal culture medium (NM)
 717 compared to NM + TGF- β 1 after 14 days of culture. Data depict fold-change \pm standard
 718 deviation. * $p < 0.05$.



719

720 **Figure 7. TGF- β 1 induces pro-fibrotic changes, such as increased contractility of hydrogels,**
 721 **increased mechanical stiffness, and asynchronous beating, in 3D engineered cardiac tissues.**

722 **A)** Beating patterns of 3D engineered cardiac tissues cultured in NM (green) and NM + TGF- β 1
 723 (red) at day 7, 9, 11, and 13. The three red lines (solid and 3dotted) represent three independently
 724 areas of beating within the same area of view. **B)** Hydrogel contraction test. Optical images of
 725 TGF- β 1 treated and non-treated hydrogels with encapsulated cardiomyocytes and cardiac
 726 fibroblasts. **C)** Quantitative analysis of the contraction test of cardiomyocytes/cardiac fibroblast
 727 encapsulated GelMA hydrogels in NM compared to NM + TGF- β 1. **D)** Representative scanning
 728 electron microscopy images of cell (white arrows)-encapsulated GelMA hydrogels cultured in
 729 NM and NM + TGF- β 1 on day 14 of culture. **E)** Mechanical stiffness of the 3D cardiac tissues in

730 the two different culture conditions (NM and NM+TGF- β 1) at day 14 of culture. Data depict

731 Mean \pm Standard deviation. * p <0.05

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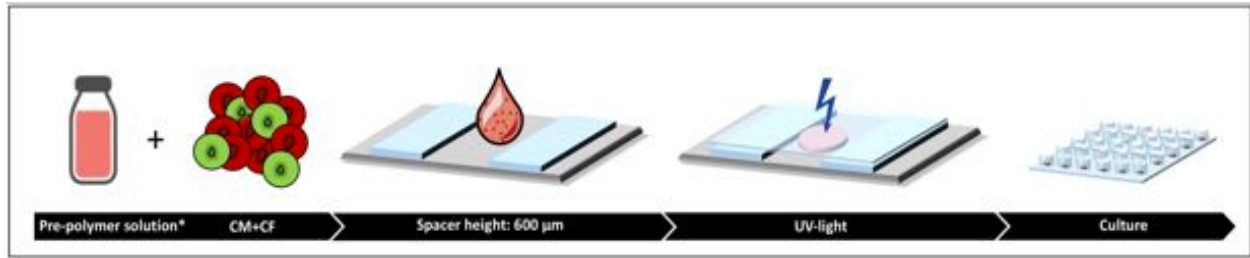
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746 **Supplemental information**



747

748 **Supplemental figure 1. Encapsulation of cardiomyocytes and cardiac fibroblasts within**

749 **GelMA hydrogels.** Primary neonatal rat cardiomyocytes and cardiac fibroblasts are isolated and

750 resuspended in a GelMA-based pre-polymer solution. Twelve microliters of the cell-laden pre-

751 polymer solution is pipetted between two spacers with 600 μm height. A sterile TMSPMA

752 treated glass slide is placed on top of this cell-laden GelMA solution and subsequently cross-

753 linked by UV-exposure for 20 s. This was followed by the separation of the hydrogels from the

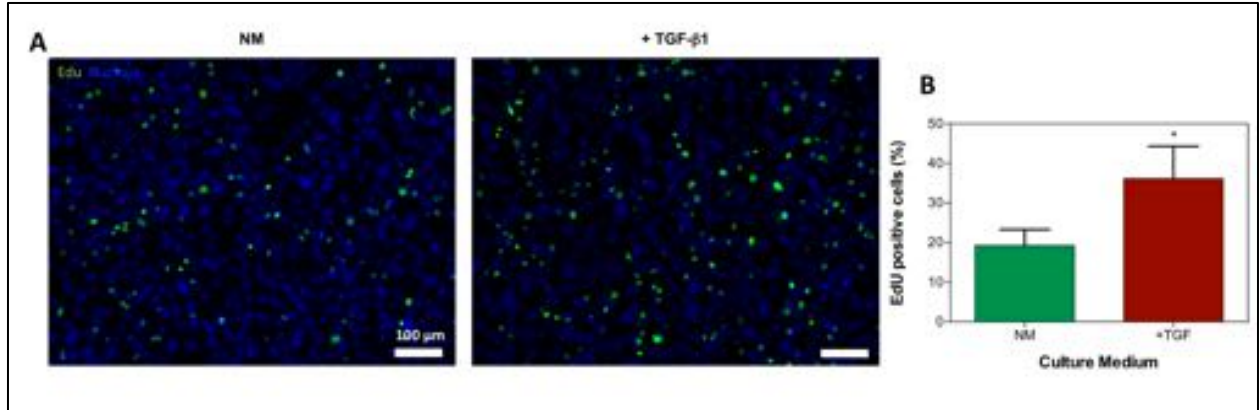
754 glass slide by a thin coverslip to culture the Cell-laden GelMA solution in a 48-well plate.

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760 **Supplemental figure 2. EdU labeling of cardiac fibroblasts cultured on TCPS. A)**

761 Fluorescence images of positive EdU labeled cardiac fibroblasts with and without TGF-β1, after

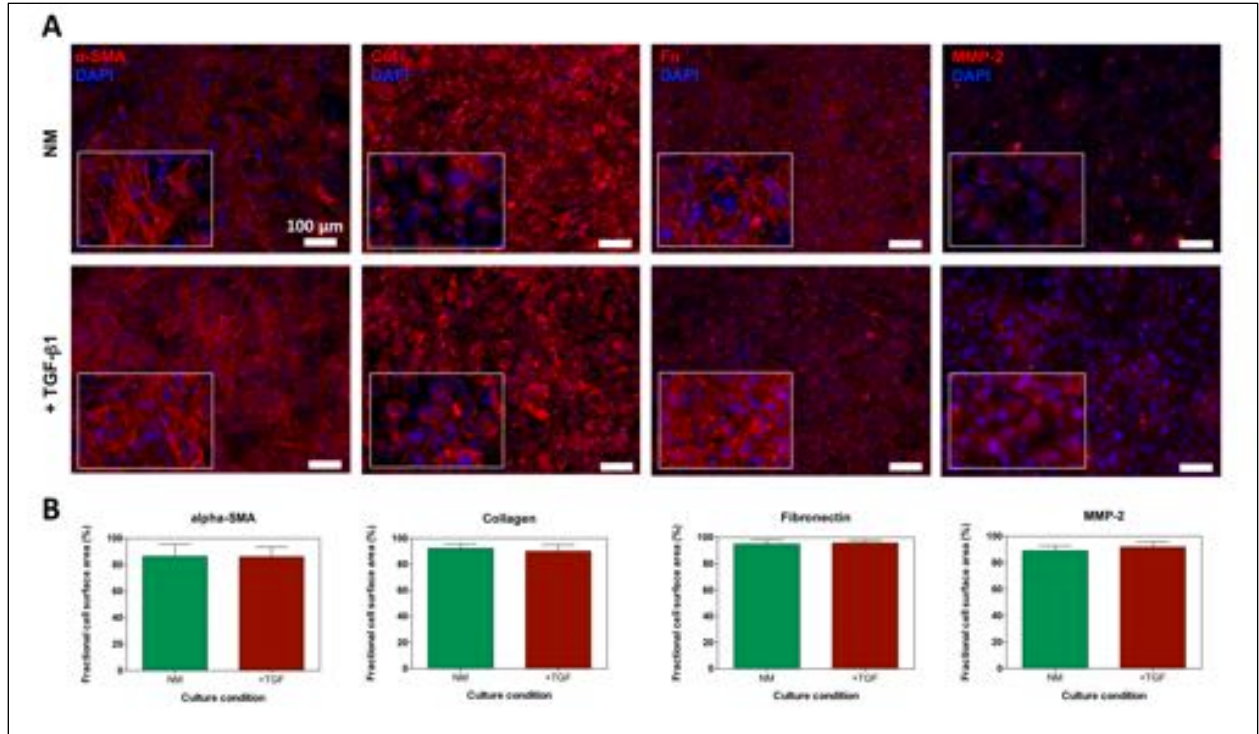
762 24h of culture. **B)** Representative quantification of proliferating cardiac fibroblasts in both the

763 presence (+TGF-β1) and absence (NM) of TGF-β1 after 24h of culture. Data depict Mean ±

764 Standard deviation. * $p < 0.05$

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768 **Supplemental figure 3. Expression profiles of cardiac fibrosis markers in cardiac**
 769 **fibroblasts cultured on TCPS. A).** Fluorescence images of cardiac fibrosis markers; alpha-
 770 smooth muscle actin (α -SMA), collagen-I (col I), fibronectin (Fn), and matrix metalloproteinase-
 771 2 (MMP-2) in the presence and absence of TGF- β 1, after 24h of culture. **B)** Representation of a
 772 quantitative analysis of fractional cell surface area. Data depict Mean \pm Standard deviation.

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