DEVELOPMENT OF A 3D BIOPRINTED AIRWAY SMOOTH MUSCLE MODEL FOR MANIPULATING STRUCTURE AND MEASURING CONTRACTION

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ABSTRACT

The contractile function of airway smooth muscle (ASM) is inextricably linked to its mechanical properties and interaction with the surrounding mechanical environment. As tissue engineering approaches become more commonplace for studying lung biology, the inability to replicate realistic mechanical contexts for ASM will increasingly become a barrier to a fulsome understanding of lung health and disease. To address this knowledge gap, we describe the use of 3D bioprinting technology to generate a novel experimental model of ASM with a wide scope for modulating tissue mechanics.

Using a stiffness modifiable alginate-collagen-fibrinogen bioink, we demonstrate that modulating the stiffness of free-floating ASM 'bare rings' is unfeasible; bioink conditions favorable for muscle formation produce structures that rapidly collapse. However, the creation of novel 'sandwich' and 'spiderweb' designs that encapsulate the ASM bundle within stiff acellular load bearing frames successfully created variable elastic loads opposing tissue collapse and contraction. Sandwich and spiderweb constructs demonstrated realistic actin filament organisation, generated significant baseline tone, and responded appropriately to acetylcholine, potassium chloride and cytochalasin D. Importantly, the two designs feasibly simulate different mechanical contexts within the lung. Specifically, the sandwich was relatively compliant and subject to plastic deformation under high contractile loads, whereas the stiffer spiderweb was more robust and only deformed minimally after repeated maximal contractions.

Thus, our model represents a new paradigm for studying ASM contractile function in a realistic mechanical context. Moreover, it holds significant capacity to study the effects of ECM composition, multiple cell types and fibrosis on lung health and disease.

69 **INTRODUCTION**

70 Cell culture, laboratory animals, and *ex vivo* human tissue have been the foundation of biological research for
71 over 100 years. While each remains an essential scientific tool, it is becoming increasingly recognised that
72 these paradigms are not sufficient to fully understand both healthy and disease states. Biological and technical
73 issues limit the tissue- and organ-level relevance of cell culture, while ethical concerns mandate reductions in
74 the use of laboratory animals, and reliable supplies of human tissue remain scarce. Tissue engineering has
75 emerged as a compelling solution to these limitations, by promising to create a large supply of customised
76 human tissues from cultured cells to better understand basal biology and to simulate human health and
77 disease. In the context of the lung, many recent advancements have been made with 'organ-on-a-chip' and
78 bioreactor models that replicate the alveolar compartment, gas exchange, and cell-cell interactions in the
79 airway wall (11-13, 21, 26). Organoid models show extreme promise for modelling different aspects of lung
80 development and epithelial function (2, 4, 29, 31).
81
82 Despite these advances, major deficiencies persist in our ability to engineer a critically important lung
83 component: airway smooth muscle (ASM). ASM composes a large portion of the airway wall by volume, making
84 it a major inflammatory mediator (30). Moreover, muscle contraction is potently modulated by the mechanical
85 environment and accurately replicating these *in vivo* environments *in vitro* has proven to be challenging (7, 33).
86 Tension development and contraction of ASM also regulate vital aspects of lung physiology, and dysfunctional
87 ASM mechanics is a central player in the pathogenesis of lung diseases such as asthma (24). Key technical
88 difficulties arise in the ability to engineer realistic mechanical environments that can stand up to the rigors of
89 muscle contraction. Early attempts successfully used simple hydrogels to measure muscle contraction (3, 18,
90 20), although the mechanical context of these models is questionable. ASM microtissues manufactured on
91 silicone moulds (32) as well as electrospun scaffolds (22) have more realistically replicated many aspects of
92 functional ASM, although short lifespans in culture and narrow scope to modulate tissue mechanics remain
93 significant limitations.
94
95 The recent description of a 3D bioprinted experimental model of ASM (9) represents a novel approach for
96 creating functional ASM tissues. In this study, a commercial bioprinter (Aspect Biosystems' RX1™) and alginate-
97 collagen bioink (AGC-10) was used to create ASM muscle rings that responded appropriately to physiological
98 contractile and relaxant signals. However, similar to early hydrogel approaches, this model does not recreate a
99 realistic mechanical environment, necessitating further development to accurately depict different
100 environments within the lung. Alginate has been used in many tissue engineering applications to modulate
101 tissue stiffness (5, 19, 27, 35), and the RX1 bioprinter incorporates a multi-channel microfluidic printhead that

102 could feasibly deposit structural and muscle components separately, presenting several avenues for technical
103 and physiological advancement.
104
105 In the present study, we expanded on previous approaches by creating a stiffness-modifiable alginate bioink
106 and novel multi-material tissue structures. This enabled us to develop a 3D bioprinted model of ASM designed
107 specifically to interrogate the effect of tissue mechanics on muscle contraction. Critically, our approach
108 generated lifelike ASM that responded appropriately to contractile and relaxant signals and demonstrated that
109 different engineered structural properties can modulate the dynamics of ASM contraction. Finally, by testing
110 coronary artery smooth muscle (CSM) in parallel, we demonstrated that our model could be applicable to a
111 wide range of different organ contexts.

112 **METHODS**

113 **Cell Culture**

114 Senescence resistant human ASM cells (hTERT overexpression) were obtained as a generous gift from the
115 laboratory of Dr. Andrew Halayko (Department of Physiology & Pathophysiology, University of Manitoba). Cells
116 were propagated in feeder media consisting of DMEM/F12 (Gibco 11330-057) supplemented with 10% heat-
117 inactivated fetal bovine serum (Gibco 12483-020) and 1% penicillin-streptomycin (Gibco 15140-122), and
118 maintained at 37°C, 5% CO_2 and 85% humidity. To prepare for bioprinting, cells at approximately 90%
119 confluence were detached with TrypLE cell dissociation reagent (Gibco 12605-036), resuspended in a calcium-
120 free alginate diluent buffer (145 mM NaCl, 5 mM NaHCO3, pH 7.0), and gently centrifuged at 150×RCF for 5
121 minutes. The supernatant was aspirated, and cells resuspended into prepared bio-ink at a density of 2.5×10^7
122 cells/mL. Coronary Arterial Smooth Muscle cells (CSM; Applied Biological Materials T0557) were propagated
123 and prepared in the same manner, but were grown in flasks coated with rat tail collagen type I (Corning
124 354236) at 10 µg/mL in PBS.

125

126 **3D Designs**

127 Designs for 3D bioprinting and fused deposition modelling (FDM) 3D printing were created in Tinkercad
128 (https://www.tinkercad.com/) and exported as STL files (see Supplementary Files). Toolpaths for 3D bioprinting
129 were prepared from STL files in Aspect Studio software and saved as APJ files. 'Bare ring' designs were created
130 as a cylinder 8 or 10 mm in diameter and 0.6 mm high and printed with 0.04 mm layer thickness (for 15 layers),
131 0% infill, and one perimeter of cellular bioink (Fig 1A). 'Sandwich' designs encapsulated a bare ring within top
132 and bottom 13 mm diameter, 0.20 mm high cylinders as acellular layers (Fig 1B). The encapsulation layers were
133 set to a 0.04 mm layer height (for 5 layers), and 20% infill with rectilinear and concentric patterns in alternate
134 layers. 'Spiderweb' designs were created with additional mechanical supports designed to interface with a rigid
135 plastic tissue holder (Fig 1C). Briefly, tissue holders were designed with six internal posts and an outer diameter
136 set to fit securely inside a 24 mm ThinCert insert (Greiner 657638). We hand-coded bioprinter toolpaths to
137 generate a 3D design comprising a 10-layer 10 mm diameter cellular ring stacked between four layers of
138 acellular rectilinear infill from the sandwich design. This was further encapsulated by a 19 mm diameter
139 network of acellular material containing loops/voids positioned to engage with the tissue holder. Layer heights
140 were set such that the spiderweb construct was nominally 1.0 mm thick. To improve stability of constructs for
141 long-term live cell imaging, centring rings were designed to firmly hold the ThinCert inserts at the centre of a 6-
142 well microplate (Greiner 657160) without disrupting access to the basolateral chamber (see Supplementary
143 Files).

144

145 **FDM 3D Printing**

146 STL files were 3D printed using Ultimaker Cura 4 slicing software and an Ultimaker S3 3D printer (216932) fitted
147 with an AA 0.25 mm print core (200454), glass print surface (227639) and lid enclosure (Shop3D EnclosureS3).
148 Filament was stored in freshly regenerated silica gel between prints and kept within a PrintDry 2.0 filament
149 dryer at ≥65°C during printing to minimise moisture content and ensure print consistency. Tissue holders were
150 printed with Polycarbonate Transparent filament (Ultimaker 1640) using a modified version of the Cura Fine
151 (Experimental) profile (see Supplementary Files) and an adherence aid to prevent warping (Magigoo PC).
152 Printing the fine features of the tissue holder exceeded filament retraction limits, thus sacrificial cylinders were
153 interspersed between the tissue holders to consume filament and enable unlimited retractions without
154 filament damage (see Supplementary Files). Completed tissue holders were visually inspected, and those with
155 excessive stringing or obvious physical defects were discarded. Acceptable prints were washed several times
156 with distilled water then once with 70% EtOH to remove traces of the adherence aid, air dried, then passed
157 briefly through a Bunsen burner flame to remove minor surface stringing. Tissue holders were sanitized by
158 boiling in ddH_2O water for 10 minutes at full power in a 1000 W microwave oven, followed by soaking in 70%
159 EtOH under moderate vacuum for 30 minutes. They were then exposed under 70% EtOH to 400 mJ/cm^2 of UV-
160 C light per side in a Spectrolinker XL-1000 UV Crosslinker, before being transferred to a sterile field for air
161 drying and storage. ThinCert centring rings were printed with Ultrafuse PET Black filament (BASF Pet-0301a075)
162 using the Cura CPE Generic Fine profile with modifications (see Supplementary Files) and Magigoo Original
163 adherence aid. Centring rings were sanitised by the same process as for tissue holders, with the boiling step
164 omitted.

165

166 **Bioink Formulation**

167 All bioinks were comprised of alginate, collagen, and fibrinogen as appropriate to individual experiments. To
168 create stiff bioinks, we used an ultra-purified high guluronate low molecular weight alginate (NovaMatrix
169 PRONOVA SLG-20; referred to as SLG-20). Biocompatibility and soft bioinks were achieved with an RGD-
170 coupled high guluronate very low molecular weight alginate (NovaMatrix NOVATACH VLVG GRGDSP; referred
171 to as RGD-VLVG). Lyophilised alginates were diluted to working concentration of 3% w/v by adding alginate
172 diluent buffer and incubating at room temperature with gentle mixing until fully dissolved. Fibrinogen (Sigma
173 F8630-1G) was freshly prepared as a 20 mg/mL stock solution by carefully layering the powder on top of
174 alginate diluent pre-warmed to 37°C. The tube was gently swirled to mix and maintained at 37°C until fully
175 dissolved before the solution was sterilized by passing through a syringe filter (Sigma SLGV033RS), then kept on
176 ice. Collagen I was added to a pre-cooled 15 mL tube, before pre-calculated volumes of ice-cold 7.5% $NaHCO_3$,
177 1.5 M NaCl and ddH_2O were added dropwise while gently vortexing the tube to precisely neutralize the acid
178 from collagen and achieve salt concentrations of 145 mM NaCl, 5 mM $NaHCO_3$ within the final volume of the

179 completed bioink. The appropriate quantities of fibrinogen and alginate were then added dropwise with gentle
180 mixing on ice. A pH strip (5-8) was used to verify that the pH of the final mixture was in the 7.0 to 7.4 range.
181
182 **Bioprinter Setup and Printing Process**
183 An Aspect Biosystems RX1 bioprinter was set up according to the manufacturer's instructions. In brief, a DUO
184 (two bioink material channels) or QUAD (four bioink material channels) microfluidic printhead was mounted on
185 the bioprinter. Material tubes (Falcon 352096) containing bioinks and bottles containing alginate diluent buffer
186 and CAT-2 crosslinker (Aspect Biosystems 00000458) were sealed with pneumatic caps pre-threaded with
187 autoclaved PTFE microfluidic tubing (Diba Omnifit 008T16-080-20), before the material tubes were placed in an
188 ice filled tube holder. Tubing was connected to the corresponding material channels in the printhead, before
189 the system was primed to remove air and clear bubbles from the printhead and nozzle in the following order:
190 material 1 through material 4, buffer, buffer plus crosslinker, then buffer.
191
192 ThinCert inserts were pre-wetted with alginate diluent buffer and mounted on the RX1 24 mm vacuum chuck.
193 The porous membrane captured the printed material while allowing excess crosslinker and buffer to flow
194 through. Constructs were deposited layer by layer as per the specified APJ file, with pressures and speeds set
195 and constantly monitored to maintain high fidelity to the original design (Table 1 and Fig 1). All print runs were
196 completed within 30 minutes of adding cells to bio-inks to minimize variance arising from cell
197 sedimentation/clumping and premature precipitation/gelation of bioink components. For spiderweb
198 constructs, a sanitized tissue holder was manually applied immediately after printing to engage the posts with
199 loops/voids in the construct. ThinCert inserts containing the printed constructs were transferred to a 6-well
200 plate containing 4 mL of pre-warmed feeder media. Where the bioink contained fibrinogen, the feeder media
201 was supplemented with 1.25 U/mL thrombin (Sigma T4648-1KU) for 30 minutes to initiate fibrinogen cleavage
202 and polymerisation to fibrin, before constructs were switched to regular feeder media. For spiderweb designs,
203 a ThinCert Centring Ring was applied to prepare the constructs for post-print imaging (see below). Constructs
204 were maintained by changing media every two days. Prior to contraction assays, constructs were serum-
205 deprived for ≥24 hours by replacing feeder media with DMEM/F12 containing 0.5% FBS and 1% insulin-
206 transferrin-selenium (Gibco 41400-045).
207
208 **Imaging for Tissue Compaction and Contractile Function**
209 To assess compaction of bare rings and sandwich constructs, macroscopic images were captured daily using an
210 Olympus Tough TG-1 iHS digital camera mounted on a tripod. Camera height and focus were optimized using 6-
211 well plates with ThinCert inserts such that the entire plate was in the field of view. Videos of muscle responses
212 to contractile/relaxant agents and alginate decrosslinking were obtained using high-quality video mode (1080p

213 @ 30 fps). Videos were processed by ffmpegTool to create single image sequences. Spiderweb constructs were
214 imaged in a Cytation 5 MPV Cell Imaging Multimode Reader set at 37°C and 5% CO_2. Whole-well brightfield
215 images were obtained using the montage function of the Gen5 software and a 4× lens.
216
217 **Contraction Assays**
218 To determine contractile responses of sandwich and spiderweb constructs, time-lapse images representing
219 baseline and treatment conditions were captured. Drug treatments designed to elicit maximal responses were
220 prepared in low serum media; 100 µM acetylcholine (ACh; Sigma A6625), isotonic 80 mM KCl (Sigma, 746436-
221 500) and 10 µM cytochalasin D (Cyto D; Santa Cruz, sc-201442) as described previously (32). To initiate the
222 experiment, the media on constructs was replaced by the first drug treatment, adding one-third of the dose to
223 the ThinCert apical chamber and two-thirds to the basolateral chamber, taking care not to disturb the constructs.
224 After a 10-minute exposure window, treatment images were captured, before applying the next treatment in
225 the sequence. For single maximal contraction time course experiments in spiderweb constructs, a 20-minute KCl
226 exposure was followed by a 20-minute cytochalasin D exposure, with images captured continuously at the
227 maximum capture rate.
228
229 **Alginate Depolymerization Load Removal Assay**
230 To characterize shortening dynamics of the unloaded muscle bundle and determine the relative strength of the
231 mechanical load, calcium chelation was used to rapidly de-crosslink alginate (14, 34). Baseline images were
232 taken before media was replaced with a buffer containing 40 mM tri-sodium citrate (Sigma 111037) plus 15
233 mM ethylenediaminetetraacetic acid (EDTA; Sigma E5134-50G) pH 7.2. Videos (sandwiches) and montage
234 image sequences (spiderwebs) were collected for up to 10 minutes.
235
236 **Contraction Reproducibility**
237 To determine the robustness and stability of our designs, spiderweb constructs were subjected to five successive
238 contraction/relaxation cycles. After capturing baseline images, maximal contraction was achieved by 20-minute
239 exposures to 80 mM KCl, followed by a 30-minute washout period, with images captured at peak contraction
240 and relaxation. The final washout cycle was followed by an additional treatment with 10 mM cytochalasin D to
241 ensure complete tension ablation.
242
243 **Image Processing and Calculations**
244 The lumen area of bare ring, sandwich, and spiderweb constructs was quantified with Fiji Imaging software, using
245 the manual polygon selection feature. Where suitable, the Canny Edge Detector plugin was used to highlight
246 borders to simplify and expedite measurements. Changes in lumen area were calculated as a percentage of raw

247 pixel area relative to, or delta from, post-print area (for compaction assays) and pre-contraction baselines (for
248 contraction assays).
249
250 **Histology**
251 Constructs were fixed with 4% paraformaldehyde (PFA; Sigma P6148) in Hanks Balanced Salt Solution (HBSS;
252 Gibco 14065-056) for 15 min, permeabilized with 0.1% Triton-X 100 (X-100, Sigma-Aldrich) in 4% PFA/HBSS for
253 10 min, then washed 3 times with HBSS. Filamentous actin (f-actin) was stained using 1 U per construct of Alexa
254 Fluor 488 Phalloidin (Invitrogen A12379) in HBSS for 1 hour, before washing 3 times with HBSS, and
255 counterstained with 0.1 µg/mL Hoechst 33342 (Invitrogen H1399) in HBSS for up to 20 min. Constructs were
256 washed one final time to remove unbound stain before they were transferred to a glass bottomed 12 well plate
257 (Cellvis P12-1.5H-N). One-to-two drops of SlowFade Diamond Antifade Mountant (Invitrogen S36963) were
258 added, and an 18 mm round coverslip (VWR CA48382-041) placed on top. Fluorescent images were captured on
259 the Cytation 5; where necessary, images were uniformly adjusted for brightness and contrast only, ensuring that
260 no details were obscured.
261
262 **Data Analysis and Statistics**
263 All numerical data are presented as mean ± standard error. Each 'n' value represents results from a single
264 distinct bioprinted construct. Statistical tests were performed with the GraphPad Prism 6.07 software package,
265 with $p<0.05$ considered statistically significant. For all compaction time-course and contraction experiments,
266 independent samples 2-way ANOVA (factors: Stiffness, Time) was performed, with all-pairwise Tukey post-tests
267 performed for the Stiffness factor. Contractile reproducibility was assessed by independent samples 2-way
268 ANOVA (factors: Contraction/Relaxation, Cycles) with Tukey post-tests comparing contraction versus relaxation
269 within each cycle.

270 **RESULTS**

271 **Bare Ring Optimisation**

272 To begin formulating a stiffness-modifiable bioink suitable for use with ASM, we took inspiration from Aspect
273 Biosystems' previous work (9) and began producing bare ring designs containing ASM and CSM with variable
274 concentrations of SLG-20 alginate plus collagen. Rings with high concentrations of SLG-20 were mechanically
275 robust but exhibited no characteristics of functional muscle. F-actin staining revealed that cells within bare
276 rings were balled-up immediately after printing, as would be expected from the dissociation step during bioink
277 preparation (Fig 2A). However, the cells remained balled-up even after extended periods in culture (up to 7
278 days). As SLG-20 concentration was reduced to produce softer rings, or replaced in-part or completely by the
279 less-stiff RGD-VLVG, cells began to exhibit some signs of cell spreading and elongation. However, these less-stiff
280 alginate formulations resulted in rings that were extremely fragile, collapsed into a shapeless mass of tissue,
281 and began to exhibit tears in the structure (Fig 2B).

282

283 To ensure that this phenomenon was not specific to a particular cell type, several different ASM cell lines plus
284 CSM were trialled, each with similar findings (data not shown). Increasing collagen concentration also failed to
285 improve either cell spreading or structural integrity in any alginate formulation. This created a significant
286 impasse for optimisation of bare rings; stiff bioinks that facilitated structural integrity did not facilitate muscle
287 formation/maturation, and soft bioinks that supported muscle formation were unable to withstand long-term
288 cell culture conditions due to rapid loss of structural integrity.

289

290 **Sandwich Development**

291 Several approaches were trialled to achieve a delicate balance between cell spreading and structural stability in
292 constructs made with low stiffness alginates. Initial attempts to increase internal compressive loads by
293 modifying the aspect ratio of bare rings to make them wider and flatter yielded no benefits. Printing strategies
294 were then developed to exploit the multi-material capabilities of the RX1 bioprinter, by including acellular
295 components comprised of stiff alginate, intended to provide physical support. First, a heterogenous 'Janus'
296 fibre design was trialled where cellular and acellular materials were co-deposited into a single fibre, however
297 this proved extremely challenging to print and produced muscle tissue with little functionality. Second, an
298 'alternating ring' approach interspersed stiff acellular layers within the bare ring, yielding minor but insufficient
299 structural improvements (data not shown). Finally, several methods of fully encapsulating the muscle ring
300 within an acellular structure were tested and iterated, ultimately resulting in the sandwich design (Fig 1B). The
301 intent of this design was to reduce mechanical stress/burden on the muscle bundle, allowing the cellular bioink
302 to be formulated with low alginate concentrations that facilitated tissue remodelling, and the acellular bioink
303 could be made with higher and tunable stiffnesses to provide varying loads opposing ring collapse.

304

305 An extensive iterative design process was implemented to formulate optimal cellular and acellular bioinks.
306 These were based on SLG-20, RGD-VLVG, collagen I and fibrinogen (Table 2), using a single ASM cell line (D12)
307 and CSM cells to minimise variation, with success measured by qualitative tissue/print fidelity, cell spreading,
308 actin morphology, and tissue lifespan in culture. Notably, inclusion of SLG-20 in the cellular bioink at any
309 concentration completely inhibited muscle formation, and the ideal concentration of RGD-VLVG was relatively
310 narrow. Use of collagen and fibrinogen enhanced both muscle formation and technical aspects of printing,
311 although the exact concentrations of these components did not seem to be critical. Thus, the optimal cellular
312 bioink which offered a balance between practicality and tissue formation contained 0.375% RGD-VLVG, 1.0
313 mg/mL collagen, 5.0 mg/mL fibrinogen. The optimal acellular bioink contained 0.75-1.25% SLG-20, 1.0 mg/mL
314 collagen, 5.0 mg/mL fibrinogen.

315

316 Gross morphology of constructs printed with different stiffness acellular bioinks revealed that ASM sandwiches
317 (Fig 3A-B) have modest but significant unstimulated 'compaction' of lumen area over time ($p<0.0001$),
318 indicative of baseline tone development and tissue remodelling. This effect is stiffness modulated ($p<0.0001$)
319 with 0.75% alginate constructs experiencing significantly higher compaction. Consistent with previous results,
320 ASM bare rings compacted excessively, collapsed, and rapidly became non-functional. F-actin staining for ASM
321 (Fig 3C) tissues printed with optimal acellular and cellular bioinks demonstrated impressive cell
322 spreading/elongation and alignment, consistent with native muscle tissue. CSM sandwiches (Fig 3D-E) followed
323 a similar dynamic over time ($p<0.0001$), but we were unable to detect statistically significant differences in
324 compaction across the stiffness range trialled ($p=0.4693$). Collectively, these results show that the sandwich
325 design successfully stabilises tissue structure and facilitates lifelike muscle morphology, establishing it as a
326 suitable platform for the assessment of contractile function.

327

328 **Sandwich Contractile Function**

329 ASM sandwich constructs responded appropriately to classic contractile stimuli (Fig 4A; $p<0.0001$), exhibiting
330 mild contractions in response to ACh and a stronger response to KCl. Cytochalasin D reduced lumen area
331 consistent with cytoskeletal tension ablation, however lumen area failed to return to baseline levels.
332 Responses were stiffness modulated ($p=0.0438$), although this modest effect was not linear, with 1.0% acellular
333 sandwiches having a weaker contractile response than 0.75 and 1.25% constructs. CSM sandwich constructs
334 exhibited similar contractile trends with apparent reductions in lumen area in response to KCl and reversal by
335 cytochalasin D, but the treatment effects were not statistically significant over time due to extremely high
336 variability between tissues (Fig 4B; $p=0.1788$). There appeared to be a trend towards linear stiffness

modulation with 1.25% acellular constructs producing the highest mean contraction and weakest mean relaxation response, but the trend was not significant (p=0.2726).

The inability of cytochalasin D to completely reverse KCl contractions and expand lumen area above baseline levels was unexpected. This result presents two possibilities: 1) ASM and CSM in our model have little-to-no baseline contractile tone, or; 2) the sandwich can prevent ring collapse during tissue maturation but is insufficient to create an elastic load that can oppose strong or rapid contractions. To ascertain the relative degree of baseline tone versus strength of the load, ASM sandwich constructs were treated with a citrate/EDTA solution to chelate calcium and depolymerise the alginate, resulting in a sudden removal of the load opposing contraction. Figure 4C shows that load removal results in significant time (p<0.0001) and stiffness (p=0.0074) linked reductions in lumen area that far exceeds the magnitude of KCl contractions. These combined data indicate that the muscle exerts a significant degree of baseline contractile tone, and that incomplete recovery of lumen area with cytochalasin D is likely due to imperfect elasticity (i.e., plastic deformation) of the acellular structure.

Spiderweb Development

To reduce plastic deformation of sandwich constructs during contraction, we further iterated on the tissue design to improve the ability of the acellular structure to behave like an elastic spring. This could not be feasibly achieved with a bioprinted acellular structure alone (data not shown). Thus, we designed a FDM 3D printed polycarbonate tissue holder to sit tightly within a ThinCert insert to create a rigid, non-compliant load against which the construct could develop tension. Six posts were precisely positioned in the holder, and hand-coded bioprinter toolpaths were created comprising circular ‘loops’ around the post positions. A constant direction of rotation, careful optimisation of loop length/width, and tight control of bioprinting speed and pressures were found to be necessary to minimise slack without closing the loops. Five layers of looping acellular structure was coded to encapsulate two layers of sandwich infill above and below the muscle bundle, which was reduced from 15 to 10 layers to balance muscle force relative to load strength and increase longevity in culture. Tissue holders were manually placed on top of the bioprinted structure immediately after printing, completing the spiderweb design and preparing the constructs for automated imaging (Fig 5A-B).

Assessment of ASM spiderweb compaction at 4 days post printing (Fig 5C) revealed similar results to the ASM sandwich. There was significant reduction in lumen area over time (p<0.0001) and across the stiffness range (p=0.0036), although somewhat surprisingly the 1.0% acellular constructs showed modestly higher compaction.

370 **Spiderweb Contractile Function**

371 Spiderweb constructs responded predictably to KCl and cytochalasin D with significant contraction and
372 relaxation phases (Fig 5D; $p<0.0001$). However, in contrast to the sandwich design, spiderweb stiffness alone
373 was unable to modulate contractile responses ($p=0.7656$), lumen area reductions to KCl were of a smaller
374 magnitude, and lumen area increases to cytochalasin D rose above baseline levels. The small magnitude of
375 spiderweb contractions relative to sandwich contractions could be attributed to reduced contractility and/or
376 increased load. To clarify this question, alginate depolymerization experiments (Fig 5E) were performed
377 revealing a strikingly rapid reduction in lumen area ($p<0.0001$) that was stiffness modulated ($p=0.0126$), with
378 0.75% acellular constructs showing the strongest response. Importantly, the contraction appeared to be
379 considerably faster and more extensive than observed for sandwich constructs (Fig 4C). This suggests that the
380 spiderweb muscle exerts a high degree of baseline contractile tone, and that the acellular load provides
381 significant elasticity to oppose this force.

382

383 The dynamics of tension development in spiderweb designs was assessed by exposing a 0.75% acellular
384 construct to KCl and cytochalasin D for 20 minutes each (Fig 6A). The contraction phase resulted in a steady
385 reduction in lumen area that did not fully peak after 20 minutes. Cytochalasin D application resulted in a rapid
386 and extensive reversal of contraction with lumen area ending up larger than baseline levels, indicating the
387 spiderweb successfully generated a more effective spring than the sandwich.

388

389 During repeat contraction/washout cycles, lumen areas of 0.75% acellular spiderweb constructs progressively
390 decreased (Fig 6B). Analysis of each cycle (Fig 6C) revealed that the magnitude of contraction during the first
391 cycle was significantly larger than subsequent cycles ($p=0.0314$), and that contraction was significantly larger
392 than washout relaxation across the experimental time course ($p=0.0005$). A final treatment with cytochalasin D
393 after the last cycle resulted in significant additional relaxation, indicating a progressive increase in baseline
394 muscle tone. Since cytochalasin D could not fully return lumen area to baseline levels, this suggests that the
395 spiderweb construct is still subject to plastic deformation when exposed to excessive contractile force.

396 **DISCUSSION**

397 Creating physiologically relevant smooth muscle suitable for evaluating contraction remains a significant
398 challenge for tissue engineering. This is particularly important within the context of the lung, where structure,
399 function, and forces generated by ASM are inextricably linked. The recent description of a novel bioprinted
400 experimental model of ASM (9) raises intriguing possibilities to create customised tissues that replicate
401 different mechanical environments associated with development, health, and disease. In this study, we
402 describe the development of innovative multi-material tissue designs that were enabled by the RX1
403 bioprinter's ability to seamlessly switch between cellular and acellular bioinks within a single tissue engineered
404 construct. Combined with a stiffness-tunable alginate-collagen-fibrinogen bioink, this allowed us to create
405 widely different mechanical contexts for tissue engineered ASM. Importantly, we demonstrate that changes in
406 the mechanical environment significantly modulate ASM contractile function.
407

408 **Structural Modulation of Tissue Engineered Muscle**

409 Our study demonstrates the potent ability of tissue structure to drive a wide range of muscle function from the
410 same cells and bioink components. Importantly, we highlight the pitfalls of not considering mechanical factors
411 when engineering lung tissues that include a muscle component. Bare rings that represent muscle with an
412 inappropriate mechanical load opposing contraction rapidly collapse and become non-contractile, even as cell
413 viability and cellular morphology may remain qualitatively acceptable. Muscle formed in this manner may meet
414 many metrics of experimental success but has no *in situ* physiological equivalent and would likely provide
415 artifactual results for any experiment relating to contractile function.
416

417 The encapsulated sandwich and spiderweb designs represent a physiologically relevant mechanical load, and
418 direct parallels can be drawn to the structure and function of different lung tissues. The mild mechanical load
419 and resulting contractile dynamics offered by the sandwich design could be considered equivalent to a small
420 non-cartilaginous airway. In this context, the mild load allows a relatively high degree of muscle shortening
421 during contraction, and contraction/relaxation plastic deformation of the structure may simulate airway
422 closure. Further, the relatively slow rate of shortening observed when the mechanical load is removed by
423 depolymerising alginate with citrate/EDTA is comparable to the relatively low baseline tone seen in small
424 airway smooth muscle. In contrast, the more robust load provided by the spiderweb design appears more akin
425 to large cartilaginous airways. The magnitude of muscle shortening during KCl contractions is restricted by the
426 load, and the rapid shortening upon load removal is strikingly similar to the shortening that occurs during
427 dissection of highly tonic tracheal smooth muscle.
428

429 Within each tissue design, stiffness modulation of the structural component produced somewhat inconsistent
430 results. The predicted weaker loads produced by the 0.75% acellular bioink allow greater sandwich compaction
431 during ASM tissue formation, yet spiderweb compaction was significantly greater in 1.0% acellular ASM
432 constructs, and no statistically significant difference was observed in CSM. Load removal experiments exhibited
433 more consistency; in both sandwich and spiderweb tissues, 0.75% acellular constructs exhibited faster and
434 more extensive shortening during load removal, as expected from the reduced quantity/density of alginate
435 that needs to depolymerise. Contractile responses were also diverse in the different tissues and structures
436 trialled; ASM sandwiches had the smallest contraction in 1.0% acellular tissues, CSM sandwiches trended
437 towards larger contractions with increasing stiffness, and ASM spiderwebs contracted similarly irrespective of
438 acellular stiffness. A key confounding factor that possibly contributed to inconsistencies in contractile function
439 is that our results were normalised to a baseline lumen area recorded immediately prior to contraction
440 experiments. This is directly equivalent to standard protocols that set baseline length (6) or optimal passive
441 force (25) prior to *ex vivo* smooth muscle strip experiments. However, adopting this approach in our model
442 masks the effect of varying levels of muscle maturation and tissue compaction preceding evaluation of
443 contractile responses, which may represent different degrees of tissue remodelling versus baseline tone
444 development. Moving forward, it will be essential to image every construct immediately after printing and
445 before contraction experiments to account for this variable.
446
447 It is also possible that the effects of differing acellular load stiffness are below the sensitivity of instruments
448 used to capture contractile responses. Lumen area changes in spiderwebs are notably small and may be subject
449 to noise despite our rigorous imaging and measurement protocols. We expect that clearer patterns would
450 emerge with a wider range of acellular stiffnesses within each tissue design. In this study we used acellular SLG-
451 20 alginate concentrations from 0.75% to 1.25%, representing the lowest and highest bioink viscosities
452 printable with high fidelity and near-perfect reliability. Based on printing pressures (Table 1) this range covers
453 an approximate doubling of bioink viscosity and resultant fibre stiffness. However, to better highlight stiffness
454 modulation of contraction, an expanded concentration range of 0.5 to 1.5% SLG-20 is entirely feasible (albeit
455 with a lower success rate of producing high fidelity constructs) and would be expected to produce more than a
456 three-fold difference in fibre stiffness between the softest and stiffest constructs. Finally, this foundational
457 design study primarily considers muscle function in terms of physical shortening, which is a relatively narrow
458 view of the many potential factors modulated by tissue mechanics. Future studies will examine other
459 experimental outcomes with a wider dynamic range of measurements including contractile vs synthetic
460 phenotype, calcium handling, proliferation, and ECM remodelling.
461

462 **Contrasts to Other Tissue Engineering Approaches**

463 The bare ring, sandwich, and spiderweb designs developed in this study offer striking similarities and disparities
464 to comparable hydrogel models of tissue engineered ASM. Notably, a foundational 3D bioprinting paper by
465 Dickman *et al* used an alginate-collagen bioink (AGC-10) to create primary human ASM muscle rings that
466 compacted without collapsing [see Ref (9) Fig 2]. Our inability to replicate similar data may be related to
467 distinct properties of AGC-10, or attributable to use of ASM cells with significantly different mechanical and
468 contractile properties. Indeed, discussions with the authors revealed that muscle rings made with some ASM
469 cell samples did collapse, and these tissues were excluded from their experiments. Wide variation in primary
470 ASM cell samples is normal and expected, but such an approach may bias cell selection towards low-contractile
471 cells incapable of generating high amounts of force. This highlights a key benefit of our designs; the acellular
472 layers generate a significant load opposing cellular tension, allowing both weakly and highly contractile cells to
473 be used in identical conditions without fear of collapse.
474
475 The ASM microtissue model (32) provides an excellent point of comparison since the studies used the same
476 ASM cell line and key contractile experiments were performed with similar protocols. Indeed, several
477 physiological aspects of the sandwich and spiderweb constructs were remarkably similar to microtissues,
478 including actin morphology, the relative magnitude of ACh and KCl contractions, and time courses for maximal
479 contraction and relaxation. Yet, a marked point of difference between the microtissue model and this study lies
480 in the dynamics of multiple contraction and washout relaxation cycles in spiderweb constructs [see Ref (32) Fig
481 4E]. In both experimental models the first contraction was greater than subsequent contractions, but in
482 spiderweb designs each contraction was of greater magnitude than the subsequent washout relaxation,
483 resulting in a progressive decrease in lumen area with each cycle. The incomplete relaxation after the final
484 cytochalasin D treatment indicates that this phenomenon derives in-part from plastic deformation of the
485 spiderweb throughout the experiment. However, the large magnitude of cytochalasin D relaxation suggests
486 that the washout step alone is incapable of fully relaxing the muscle, and that active actinomyosin crossbridge
487 cycling and/or passive cytoskeletal tension accumulates with successive contractions.
488
489 The reason for this contrasting relaxation response is unclear, although key technical differences between the
490 models may contribute. Spiderweb muscle bundles are significantly larger than microtissues with a greater
491 diffusion barrier that potentially increases washout times. However, single-contraction time courses were
492 highly comparable between the models indicating this is not a major factor. The inclusion of supporting
493 fibroblasts in ASM microtissues is noteworthy, although it seems doubtful that fibroblasts would modulate
494 muscle relaxation in this manner. More likely, microtissues were cultured for shorter time periods than
495 spiderwebs (typically 3 days versus 5 days) and had less exposure to low serum media (2 hours versus 24

496 hours), each of which may significantly modulate muscle relaxation. Finally, the silicone cantilevers of the
497 microtissue model behave like perfect springs, with virtually no hysteresis nor plastic deformation, which may
498 pull on the tissues to facilitate lengthening during relaxation versus the less elastic spiderweb design.
499 Examining other features relating to relaxation, including the use of alternative relaxant agents (β_2 agonists,
500 adenylate cyclase activators, rho kinase inhibitors), tracking cytoplasmic calcium, and characterising
501 cytoskeletal morphology will be essential to develop a complete understanding of this behaviour.
502
503 In terms of practicalities, our bioprinted designs offer significant benefits over microtissues and other models
504 based on deformable silicone moulds. All aspects of the sandwich and spiderweb designs are far easier and less
505 resource intensive to produce than microtissue and other silicone substrates, tissue mechanical properties can
506 be altered 'on the fly' without changing silicone manufacturing processes, and the entire bioprinting process
507 requires only off-the-shelf equipment and reagents. The bioprinting process can be completed by a moderately
508 skilled operator and does not require laborious and highly skilled preparation of substrates including degassing
509 of wells, centrifugation, or substrate 'scraping' steps. Importantly, bioprinting may be more suitable to obtain
510 reliable results from rare or expensive cell samples. While large quantities of cells are required for each
511 bioprinting run ($\geq 5\times10^6$ cells), virtually all of the cells can be incorporated into constructs. Only 1×10^6 cells are
512 required for manufacturing microtissues, but it is estimated that less than 5×10^4 cells end up within constructs
513 due to manufacturing losses associated with centrifugation and scraping. Finally, bioprinted tissues last
514 significantly longer in culture than microtissues and automated measurement protocols are easier to establish,
515 making them suitable for a much wider range of experiments.
516
517 **Other Limitations and Opportunities**
518 The narrow range of alginate types and concentrations that generated functional smooth muscle tissue was
519 surprising. We had initially expected to create a bioink based on RGD-VLVG where muscle bundle stiffness
520 could be widely modulated by addition of stiffer alginates. However, alginate content needed to be kept
521 relatively low, and inclusion of higher molecular weight alginates – even at concentrations as a low as 0.1% –
522 completely inhibited tissue formation. This may be consistent with previous studies that show high native
523 alginate concentrations impact cell function due to poor diffusivity and porosity (27, 35), and that alginate
524 chain length specifically regulates cell spreading (19). In this context, a bioink comprised of ultra low molecular
525 weight alginate generated by alginate lyase treatment (5) may provide greater scope for modulating the
526 stiffness of the muscle bundle, since alginate concentration could likely be increased without inhibiting cell
527 spreading. This approach was not pursued in this study; the ability of the sandwich and spiderweb designs to
528 completely uncouple the cellular and acellular components provided significant freedom for construct
529 optimisation. Nevertheless, the inability to use alginate to stiffen the muscle bundle may reduce the capacity to

530 simulate some aspects of fibrosis, and alternative approaches such as matrix crosslinking (16, 23), inclusion of
531 fibroblasts (32), or treatment with mediators of fibrosis (9) may be more fruitful.
532
533 In contrast to the rigid limits on alginate composition, the wide range of ECM protein content achievable within
534 the bioinks was impressive. Inclusion of small amounts of collagen I significantly improved structural integrity
535 immediately after printing, likely due to collagen thermal gelling contributing to layer adhesion. Collagen I also
536 improved tissue formation versus RGD-alginate alone, consistent with its well-known role for improving
537 biocompatibility in cell culture and tissue engineering applications. Fibrinogen inclusion (with post-print
538 conversion to fibrin) was not inherently required to create contractile ASM, but was beneficial for improving
539 long-term structural integrity of constructs. This is expected from the excellent mechanical properties of
540 fibrin/collagen co-gels (17), the resistance of fibrin to degradation by proteases normally secreted by smooth
541 muscle (1, 10, 15), and the absence of plasminogen in our culture system. Intriguingly, the upper limit of
542 collagen and fibrinogen concentrations was restricted only by practicalities of stock solution concentrations
543 and the quantity that could be physically mixed into the bioink while still maintaining osmolarity and pH. This
544 raises the possibility that collagen and fibrinogen could be, in-part or completely, substituted by alternate ECM
545 proteins including decellularized ECM (8, 28) to test the effects of different ECM milieu on muscle function.
546 Usage of dECM may be particularly beneficial, as the inclusion of dECM in bioprinted intestinal smooth muscle
547 constructs was shown to improve contractile function (9). Moreover, the nature of the sandwich and
548 spiderweb designs mean that these ECM substitutions can be made freely without concern for interfering with
549 muscle loading, creating a powerful tool to study how ECM derangement interacts with tissue mechanics.
550
551 **Summary**
552 We have developed a physiologically relevant bioprinted model of ASM that enables extensive manipulation of
553 tissue mechanics. The combination of a stiffness-tunable bioink, partnered with a microfluidic-based multi-
554 material bioprinter than enables the novel sandwich and spiderweb designs, allows us to interrogate the effect
555 of lung structure on ASM contraction more readily than any other tissue engineered model. Future studies
556 examining a broader range of stiffness environments, alternative ECM components, and simulated fibrosis by
557 including fibroblasts or through ECM crosslinking, will provide valuable insights into lung development, health,
558 and disease.

SUPPLEMENTARY FILE LIST

Bare Ring.STL (STL file for 10 mm bare rings)

Sandwich.STL (STL file for sandwich design with 13 mm frame and 10 mm muscle ring)

UM PC for Tissue Holder.CURAPROFILE (printing profile for spiderweb tissue holder)

Tissue Holder Thincert v4 22.8.STL (STL file for spiderweb tissue holder)

Tissue Holder Retraction Count Buster.STL (STL file for sacrificial filament usage)

UMS3_Tissue Holder Thincert v4 22.8 (Cura Project file for tissue holder for Ultimaker S3)

BASF Ultrafuse PET for Centring Holder.CURAPROFILE (printing profile for Thincert Centring Ring)

Thincert Centring Holder v3.STL (STL file for Thincert centring ring)

UMS3_Thincert Centring Holder v3 (Cura 4 Project file for Thincert centring ring)

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AUTHOR CONTRIBUTIONS

Jeffery Osagie edited the manuscript, designed experiments, and performed data/statistical analysis, cell culture and contraction assays

Sanjana Syeda edited the manuscript, designed experiments, and performed data/statistical analysis, cell culture and contraction assays

Emily Turner-Brannen edited the manuscript, analysed data, and performed bioprinting, cell culture and contraction assays

Michelle Guimond performed 3D FDM design

Lumiere Parrenas tested 3D FDM designs, and performed cell culture and contraction assays

Ahsen Haroon tested 3D FDM designs, and performed cell culture and contraction assays

Philip Imasuen tested 3D FDM designs, and performed cell culture and contraction assays

Adrian R West conceived the study, wrote the manuscript, designed/supervised all aspects of experimentation, and performed data/statistical analysis, 3D design, FDM 3D printing and bioprinting

588 **TABLES**

589 **Table 1.**

590 Print pressures and speeds for select bioink formulations during bare ring and sandwich/spiderweb
591 optimisation and experimentation phases. Pressures were qualitatively monitored during printing to maintain
592 fidelity to the original 3D design and comparable deposition rates between different bioink viscosities.

593

Bare ring optimisation	**Print Pressure (mBar)**	**Print Speed (mm/sec)**
Cellular 0.25% RGD-VLVG	18-22	32
Cellular 0.375% RGD-VLVG	20-28	32
Cellular 0.5% RGD-VLVG	25-30	32
Cellular 0.75% RGD-VLVG	30-50	30
Cellular 0.75% SLG20	55-60	23
Cellular 1.0% SLG20	80-100	20
Optimised for sandwiches and spiderwebs	**Print Pressure (mBar)**	**Print Speed (mm/sec)**
Cellular, 0.375% RGD-VLVG	25-35	32
Acellular, 0.75% SLG20	80-100	23
Acellular, 1.0% SLG20	120-150	20
Acellular, 1.25% SLG20	160-200	18
Crosslinker	60-70	N/A
Buffer	100	N/A

594

595 **Table 2.**

596 Summary of bioink components and their usable concentration range within cellular (muscle) and acellular
597 (structural) components of 3D bioprinted sandwich and spiderweb constructs. For convenience, the acellular
598 bioink was typically made with the same collagen and fibrinogen content as the cellular bioink.

599

Cellular bioink component	**Optimal concentration (usable range)**	**Effect of lower than optimal concentrations**	**Effect of higher than optimal concentrations**
SLG20	<0.1%	N/A	Prevention of cell-cell contacts
RGD-VLVG	0.375% (0.25-0.50)	Poor morphology	Reduction of cell-cell contacts
Collagen I	1.0 mg/mL (0.5-2.0)	Unravelling of constructs immediately after printing	No benefit
Fibrinogen	5.0 mg/mL (1.0-20)	Reduced structural integrity, reduction of cell-cell contacts	No benefit
Acellular bioink component	**Optimal concentration (usable range)**	**Effect of lower than optimal concentrations**	**Effect of higher than optimal concentrations**
SLG20	0.75-1.25% (0.5-1.5%)	Low viscosity, decreased print fidelity, high print speeds	High viscosity, decreased print fidelity, low print speeds
RGD-VLVG	Not used	N/A	N/A
Collagen I	1.0 mg/mL (0.5-2.0)	Unravelling of constructs immediately after printing	No benefit
Fibrinogen	5.0 mg/mL (1.0-20)	Reduced structural integrity	Increased acellular stiffness

600

FIGURES

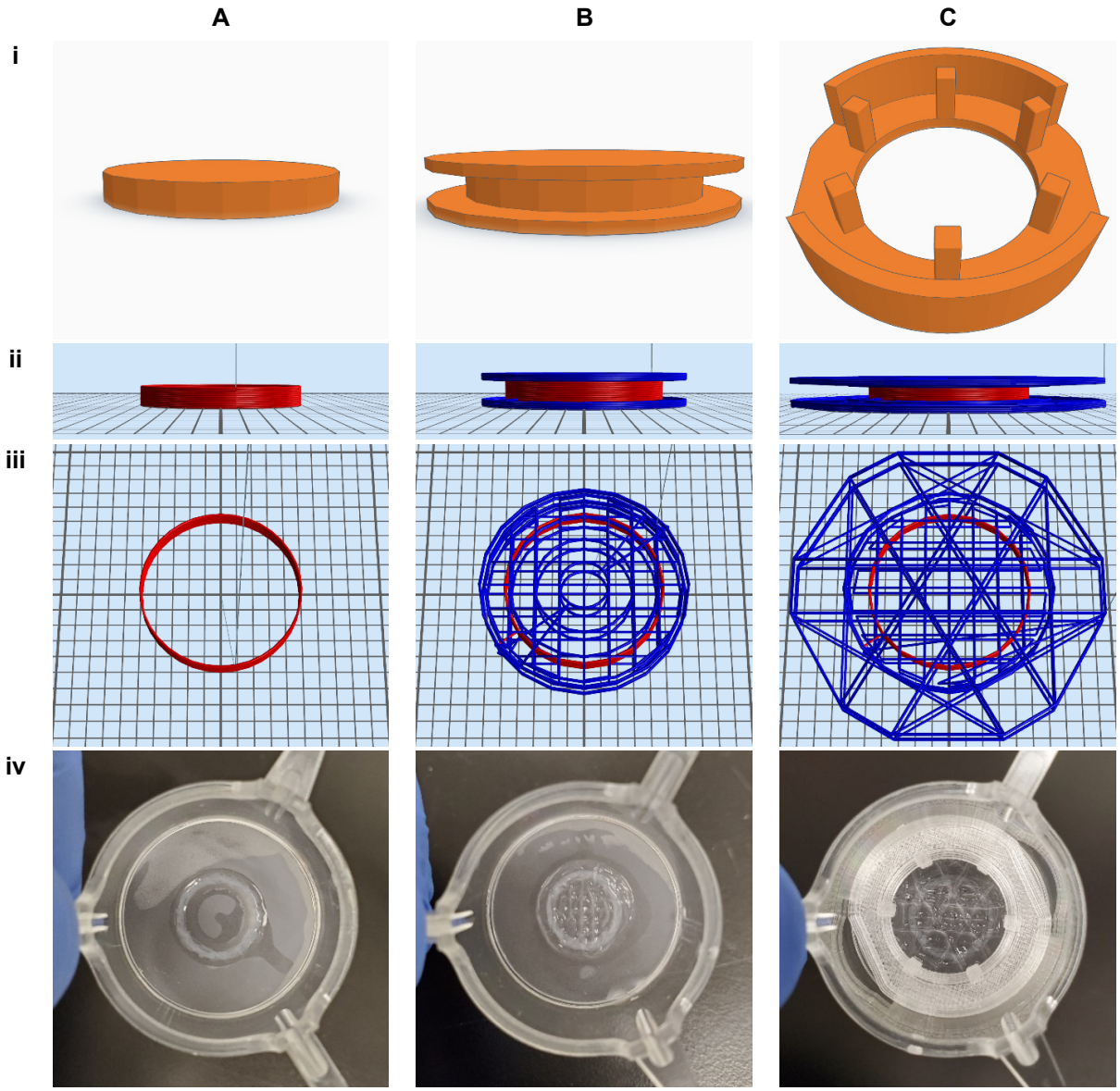

Figure 1. 3D STL designs and toolpaths for bioprinting and FDM printing.
Schematic of bare rings (A), sandwich (B), and spiderweb (C) designs showing STL files (i), bioprinter toolpaths (ii and iii) and photos of freshly printed constructs (iv). Bare rings consisted of a 10 mm diameter cylinder printed with cellular bioink (red) in a single circumference 15 layers high. The sandwich design encapsulated the bare cellular ring between two 13 mm dimeter cylinders, with alternating rectilinear and circular infill patterns of acellular bioink (blue) creating a spring-like load opposing tissue compaction and contraction. The spiderweb design began as a solid tissue holder; the posts were designed to engage with voids in a hand-coded toolpath for acellular bioink that further encapsulated and supported the cellular ring. Each of the three are printed with very high fidelity to the base design. The Z-height of ring and sandwich STL files, plus all toolpaths, are scaled to 200% for visual clarity.

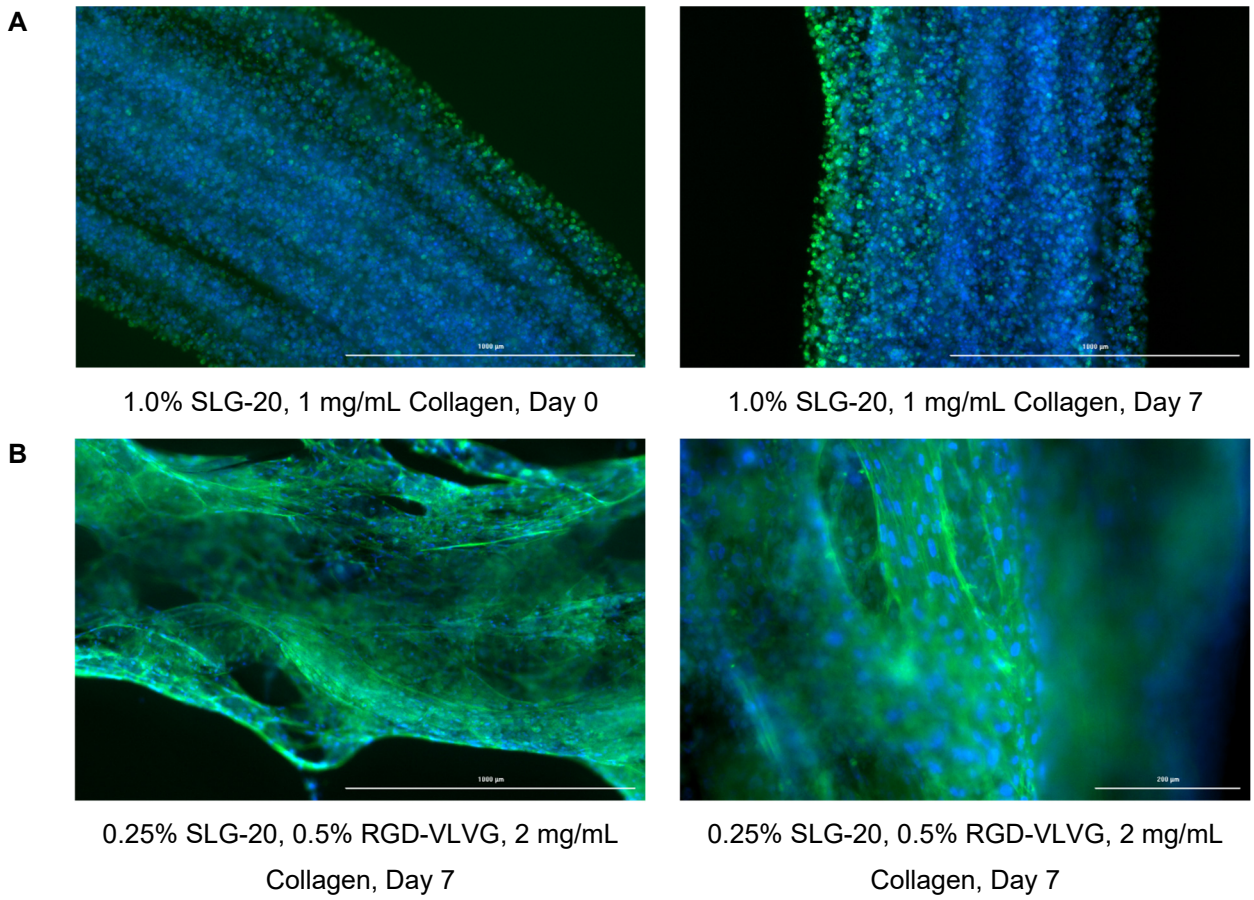

Figure 2. Histology of bare ASM rings with different bioink formulations.

Bare rings formulated with high concentrations of SLG-20 produced high fidelity and robust structures (A), however cells did not form functional muscle due to an absence of cell spreading, as evidenced by disorganised f-actin (green) and nuclei (blue) staining even after extended culture periods. Lower concentrations of SLG-20 and substitution of SLG-20 with RGD-VLVG allowed cells to spread within the construct (B). However, the macroscopic structure began to show tears in the fibre and collapses soon after printing. These results were consistent regardless of cell line/type and collagen concentration.

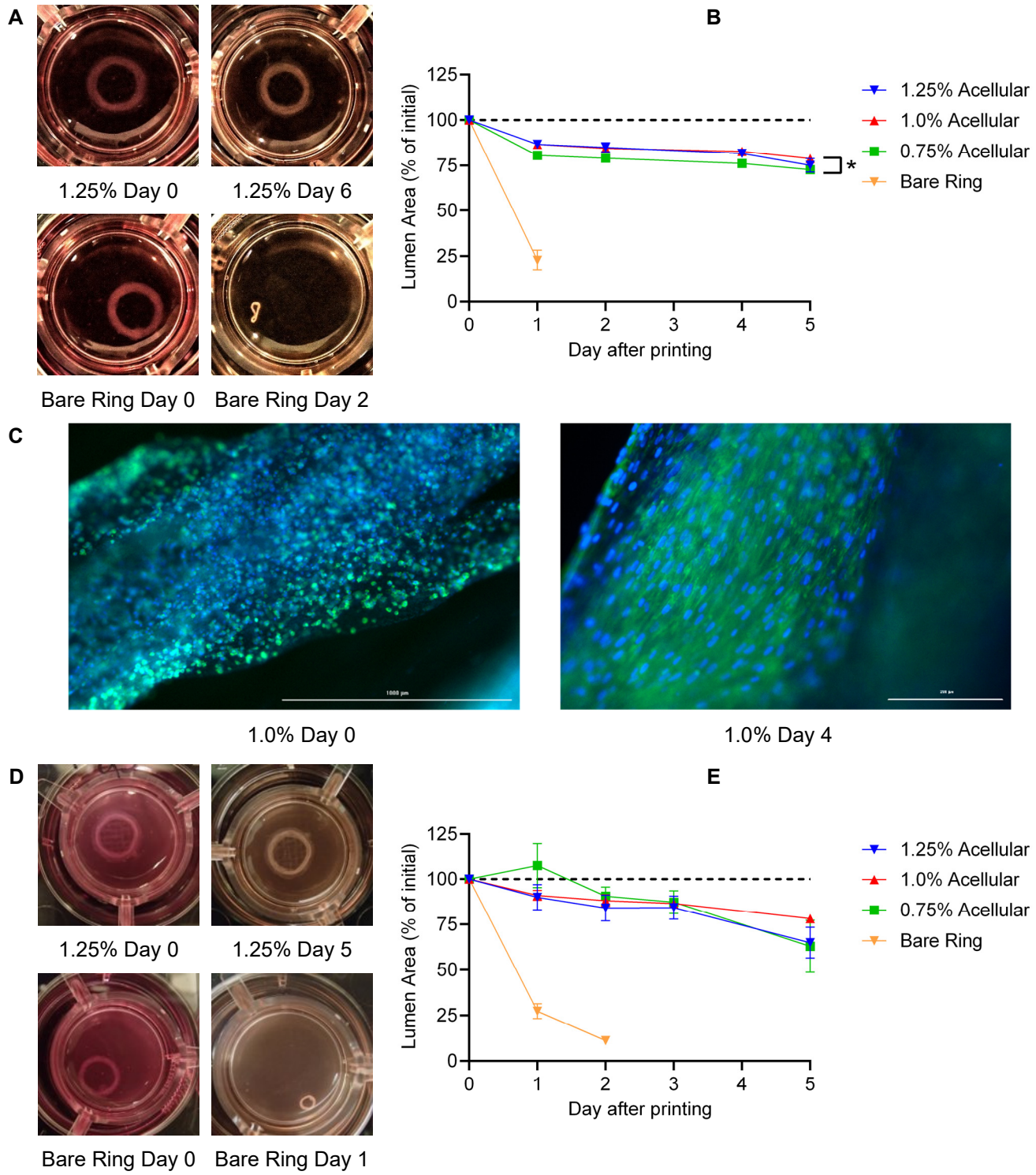

Figure 3. The sandwich design opposes muscle ring collapse and promotes muscle-like morphology. ASM sandwiches are qualitatively equivalent to bare rings immediately after printing, but the sandwich structure effectively prevents ring collapse (A). Compaction of ASM sandwiches is moderately but significantly regulated by times and acellular stiffness (B; interaction $p=0.5164$, time $p<0.0001$, stiffness $p<0.0001$, $n=12$). Histological staining of ASM sandwiches for f-actin (green) and nuclei (blue) reveals disorganised and rounded

625 cells immediately after printing, but cells rapidly take on an elongated morphology with f-actin alignment along
626 the length of the muscle bundle which is highly reminiscent of real muscle (C). CSM sandwiches are similarly
627 resistant to collapse over extended culture periods (D); they exhibit a highly similar time-based compaction,
628 although this is not modulated by acellular stiffness (E; interaction $p=0.5949$, time $p<0.0001$, stiffness
629 $p=0.4693$, $n=4$). Note that bare rings are shown on graphs for reference but are excluded from statistical
630 analysis. * represents a significant stiffness effect.

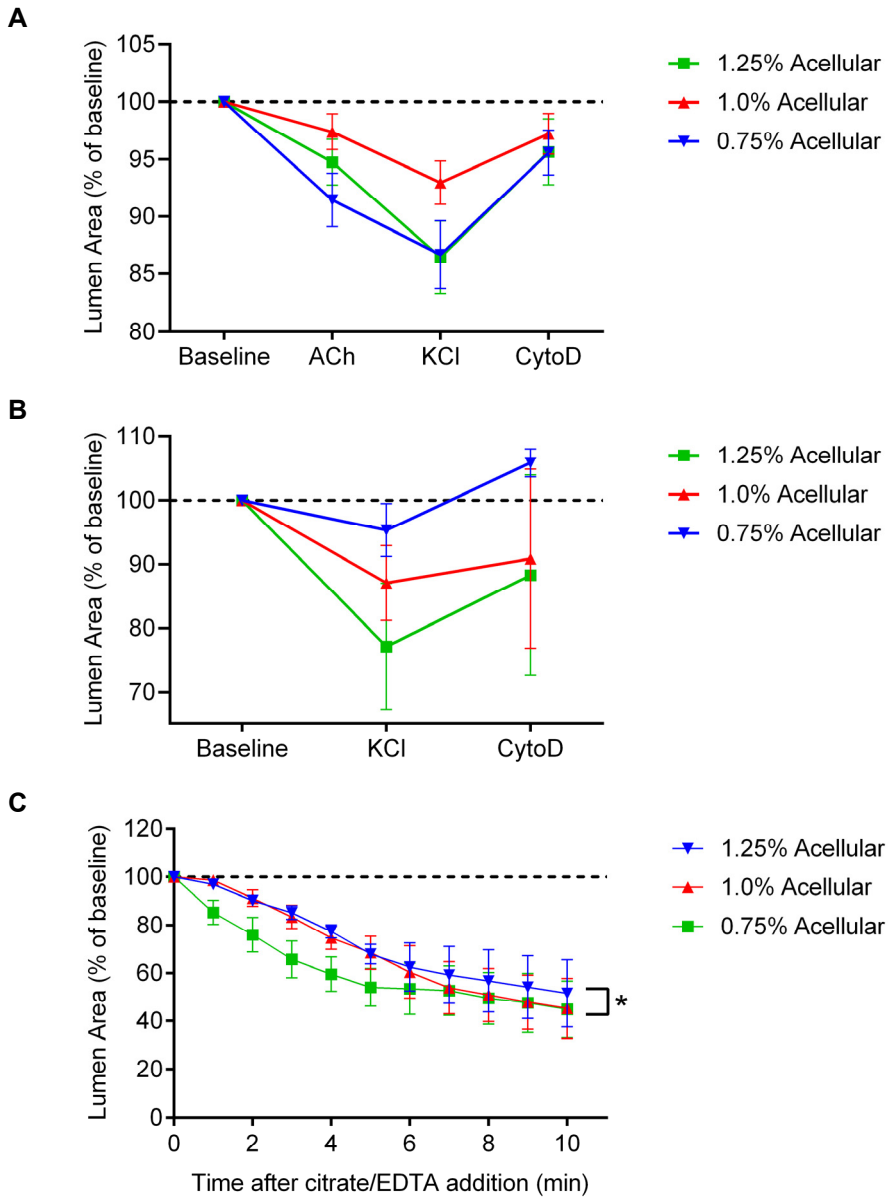

Figure 4. Contraction and relaxation of ASM and CSM sandwich designs.

ASM sandwich constructs exhibit a drug- and stiffness-dependent contractile response (A; interaction p=0.5524, drug p<0.0001, stiffness p=0.0438, n=10), with mild contraction responses to acetylcholine, a stronger response to KCl, but only partial reversal of contraction by cytochalasin D. Surprisingly, the 1.0% acellular constructs exhibited significantly weaker response, despite being the median of the stiffness range. CSM also appeared to have similar contractile responses, but no significant differences were detectable due to high variation between samples (B; interaction p=0.8110, drug p=0.1788, stiffness p=0.2726, n≥4). Chelation-depolymerization of the alginate load in ASM sandwich constructs resulted in time- and stiffness-dependent reductions in lumen area (C; interaction p=0.9981, time p<0.0001, stiffness p<0.0074, n≥2), indicating that the muscle has substantial baseline tone, and that contraction can cause plastic deformation of the sandwich. * represents a significant stiffness effect.

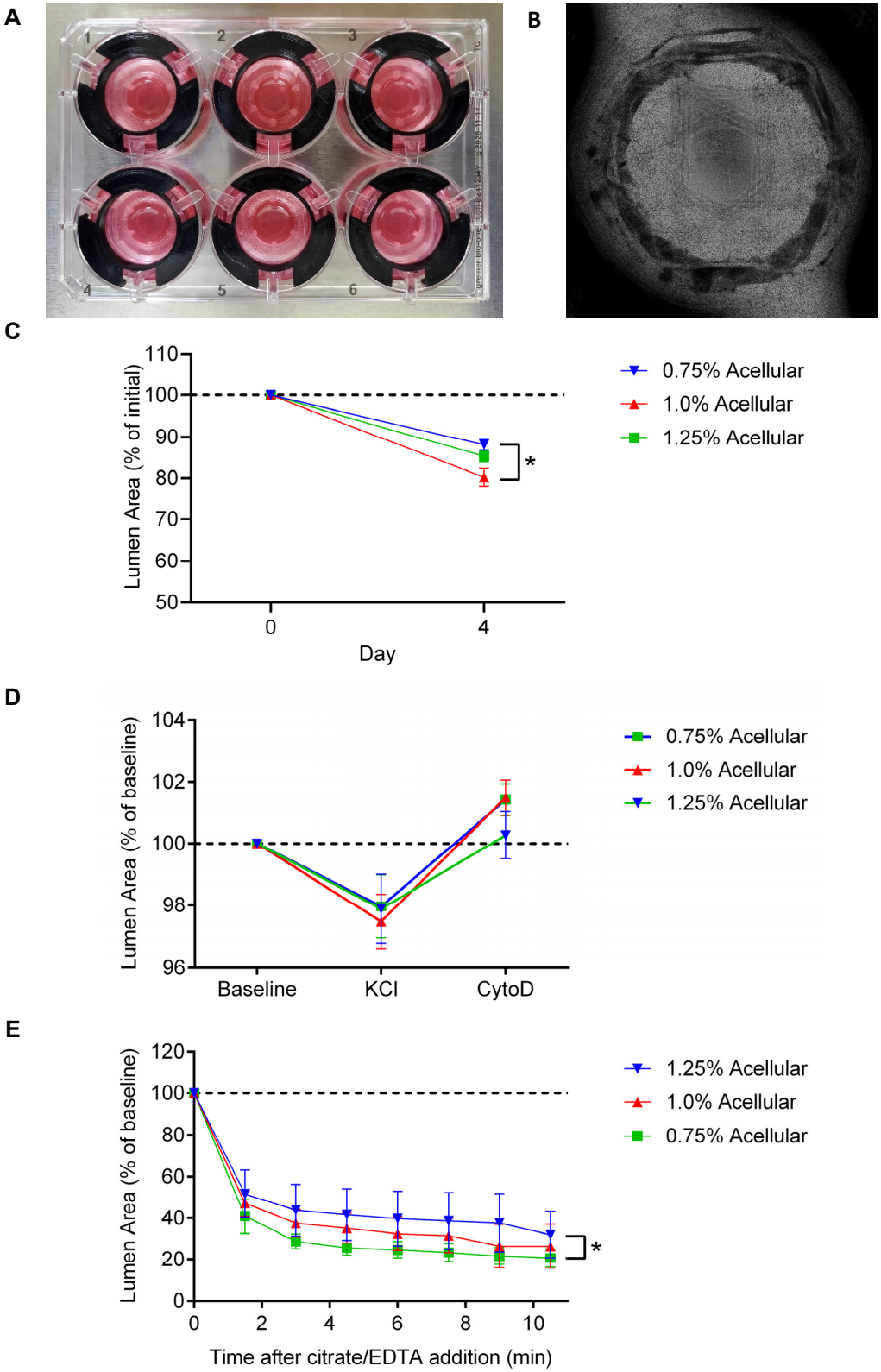

Figure 5. The spiderweb design presents a strong and elastic load opposing contraction. The ASM ring within spiderweb designs is fixed in place by ThinCert centring rings, the tissue holder, and the taut acellular structure (A), enabling automated capture of high-resolution montage images for accurate lumen area assessment (B). Compaction of ASM spiderweb constructs followed a similar trajectory to ASM and CSM

646 sandwich constructs, being modulated by both time and stiffness (C; interaction $p=0.0036$, day $p<0.0001$,
647 stiffness $p=0.0036$, $n\geq16$). The contractile response was drug-dependent (D; interaction $p=0.7925$, drug
648 $p<0.0001$, stiffness $p=0.7656$, $n\geq5$); modest reductions in lumen area were observed after KCl, while
649 cytochalasin D completely reversed the contraction and enlarged lumen area above baseline levels. The
650 chelation-depolymerization load removal resulted in a dramatic reduction in lumen area (E; interaction
651 $p>0.9999$, time $p<0.0001$, stiffness $p=0.0126$, $n\geq3$), indicating that the muscle in ASM spiderwebs exerts
652 significant baseline tension, and that the spiderweb provides a strong elastic load that opposes the generated
653 forces. * represents a significant stiffness effect.

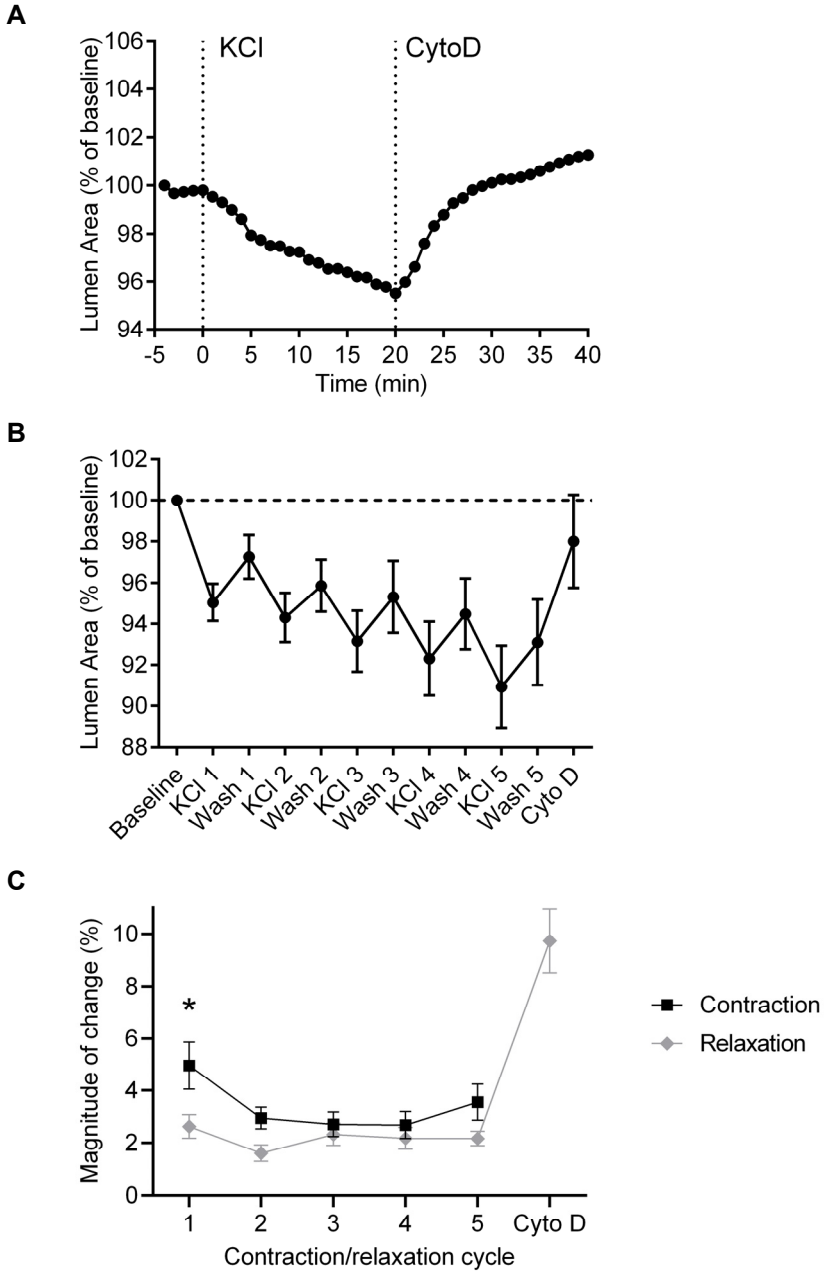

Figure 6. Dynamic properties of ASM spiderweb contractions.

Tracking a single contraction (A) reveals that a maximal dose of KCl results in a steady reduction in lumen area that does not peak after 20 minutes. Cytochalasin D results in lumen areas larger than baseline, entirely reversing the contraction plus highlighting baseline tone. Repeated contraction/relaxation cycles cause a progressive reduction in lumen area (B), resulting from each contraction being larger than the subsequent relaxation (C; interaction p=0.3465, cycle # p=0.0314, contract vs relax p=0.0005, n=10). Cytochalasin D can partly reverse the cumulative reduction in lumen area, indicating contributions from increasing muscle tension and plastic deformation of the construct. The final cytochalasin D treatment is shown for reference but is excluded from the statistical analysis. * represents a significant difference between contraction and relaxation.

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