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1 **Dynamic actuation enhances transport and extends therapeutic lifespan in an implantable**
2 **drug delivery platform**

3
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1 **Abstract**

2 Fibrous capsule (FC) formation, secondary to the foreign body response (FBR), impedes molecular
3 transport and is detrimental to the long-term efficacy of implantable drug delivery devices,
4 especially when tunable, temporal control is necessary. We report the development of an
5 implantable mechanotherapeutic drug delivery platform to mitigate and overcome this host
6 immune response using two distinct, yet synergistic soft robotic strategies. Firstly, daily
7 intermittent actuation (cycling at 1 Hz for 5 minutes every 12 hours) preserves long-term, rapid
8 delivery of a model drug (insulin) over 8 weeks of implantation, by mediating local
9 immunomodulation of the cellular FBR and inducing multiphasic temporal FC changes. Secondly,
10 actuation-mediated rapid release of therapy can enhance mass transport and therapeutic effect with
11 tunable, temporal control. In a step towards clinical translation, we utilise a minimally invasive
12 percutaneous approach to implant a scaled-up device in a human cadaveric model. Our soft
13 actuable platform has potential clinical utility for a variety of indications where transport is
14 affected by fibrosis, such as the management of type 1 diabetes.

1 **Introduction**

2 Our immune system has evolved to acquire a robust defence mechanism against foreign body
3 invasion. In the presence of a “foreign object”, neutrophil infiltration initiates a cascade of
4 inflammatory and wound healing processes, which precipitates the formation of a dense,
5 encapsulating fibrous capsule (FC)^{1,2}. The foreign body response (FBR) minimises exposure to
6 potential toxins and is often advantageous; for example, soldiers with bullet wounds rarely develop
7 clinical symptoms of lead poisoning^{3,4}.

8 This protective response, however, is detrimental to the long-term durability of implantable
9 biomedical devices such as breast implants^{5,6}, heart valves⁷, and pacemakers⁸. These devices have
10 transformed modern patient care, but the immune infiltration and fibrotic response can negate
11 device function over time, necessitating painful revision or replacement surgery. This fibrous
12 barrier is particularly deleterious for biosensors, such as continuous glucose monitors, and
13 controlled drug release devices, such as insulin pumps, which rely on interactive communication
14 with their local tissue environment⁹⁻¹¹. In such cases, the formation of a hypopermeable capsule
15 can impede transport of molecules, both to¹² and from^{13,14} the implant, and lead to therapy failure.

16 One pertinent example is the management of type 1 diabetes, a chronic disease affecting
17 18 million people worldwide, with an annual economic burden of greater than \$90 billion USD¹⁵.
18 Successful implementation and clinical adoption of an artificial pancreas combining continuous
19 glucose monitoring with the rapid, responsive, release of insulin (or glucagon) would vastly
20 improve outcomes and quality of life for this patient population. The development of a fully
21 automated closed loop insulin delivery system would reduce user burden, remove the need for
22 multiple daily injections, and increase time spent in the optimal blood glucose range, which is
23 imperative for the prevention of long-term diabetic complications. Unfortunately, current efforts

1 at developing such a device have been hindered by the dynamic and unpredictable FBR, leading
2 to glucose sensing inaccuracy, inhibition of insulin release, and gradual loss of functionality in the
3 weeks to months following implantation^{4,16-18}. Looking towards the future, living implants
4 containing stem cell derived pancreatic β -cells represent a potential cure for diabetes. However,
5 the attenuation of oxygen and molecular transport due to the FC barrier still constitutes a major
6 hurdle to successful clinical translation of these implants^{9,10,19,20}. It is evident that a method to (i)
7 mitigate the FBR or (ii) improve transport across the FC could transform the management of this
8 pervasive disease. Furthermore, such a method could have broader implications for a range of
9 diseases and device-based treatments affected by the FBR.

10 Conventional strategies to mitigate the FBR have focused on changing the attributes of the
11 implant material itself, such as its size, shape, topography, and surface coating²¹⁻²⁹, or involved
12 the concomitant delivery of FBR modifying drugs, such as steroidal anti-inflammatory, anti-
13 fibrotic, and anti-proliferative agents³⁰. While these strategies have shown promise, they have not
14 succeeded in completely disarming the FBR and possess several limitations. Firstly, materials are
15 designed in a pre-defined manner and generally target only one component or timepoint of an
16 immune response that is multifaceted and temporally dynamic. Secondly, the use of FBR
17 modifying therapeutics presents safety concerns due to untargeted adverse effects and local
18 toxicity³¹⁻³³. Sustained, systemic delivery of therapeutics such as non-steroidal anti-
19 inflammatories is associated with a range of toxicities in the liver, kidneys, heart, and
20 gastrointestinal tract³³. Local targeted delivery can reduce off-target effects but may still adversely
21 affect the underlying tissue or interfere with the mechanism of action of the implantable device.
22 For example, local delivery of dexamethasone can mitigate the FBR, but not without suppressing
23 underlying tissue regeneration³⁴. Furthermore, the effect of long-term immunosuppression on the

1 behaviour and secretome of cell-based therapeutics is unclear. Finally, a local depot of drug is
2 finite, often lasting 1-2 months, while many needs are lifelong³³. Thus, depending on drug
3 pharmacology and clinical context, the immune and fibrotic response may rebound once the
4 residual effects of drug inhibition dissipate. A long-term, drug-free method that can modulate and
5 adapt to the FBR over time would, therefore, be highly desirable to address these limitations.

6 Dynamically altering the local biomechanical environment at the implant site is one such
7 promising, yet underexplored, drug-free approach³⁵. Cells in our body are exquisitely sensitive to
8 their mechanical environment, with loading playing a pivotal role in cell functions such as
9 differentiation³⁶, proliferation³⁷, and migration³⁸. Historically, studies have observed
10 biomechanical stress as a pro-fibrotic or regenerative stimulus¹, demonstrating that application of
11 stretch³⁹⁻⁴¹, fluid flow^{42,43}, or compression⁴⁴ to cells can lead to increased deposition of
12 collagenous matrix. Accordingly, many anti-FBR strategies have focused on minimising the
13 mechanical mismatch, interfacial stress, and movement between the implant and local tissue. Our
14 research seeks to challenge this *status quo* and reveals the potential for a dynamic
15 mechanotherapeutic that uses low-magnitude, atraumatic, tissue strain and convective flow as a
16 defence mechanism against the invading cellular FBR. Interestingly, some studies have observed
17 that small magnitude, dynamic loading has anti-inflammatory and pro-regenerative effects.
18 Previous work applying dynamic loading to tissue has used daily mechanical, pneumatic, or
19 magnetic stimuli, either internally or externally, to apply cyclic loads, inducing strains ranging
20 from 4-50%. with each cycle lasting between 1 second and 10 minutes⁴⁵⁻⁵¹. These studies have
21 demonstrated beneficial effects in terms of vascularisation^{45,46,51}, functional tissue
22 regeneration^{47,50}, and anti-inflammatory gene expression⁴⁸. These prior works on mechanical

1 loading have indicated the presence of a therapeutic threshold, beyond which tissue damage and
2 inflammation occurs^{48,50}.

3 Preceding work from our group demonstrated a fibrosis attenuating effect elicited by a
4 dynamic soft reservoir following acute implantation³⁵. Here, we build on this work, and introduce
5 a soft transport augmenting reservoir (STAR) which can persistently mitigate the dynamic FBR
6 and maintain long-term, rapid, molecular communication with its tissue environment using two
7 distinct and synergistic soft robotic actuation strategies: intermittent actuation (IA) and actuation-
8 mediated rapid release (RR). Importantly, we shed light on the mechanistic underpinnings of IA
9 and reveal an immunomodulatory effect in the acute phase of implantation, with a significant
10 reduction in neutrophil infiltration at the pericapsular site, followed by multiphasic temporal
11 capsular changes with chronic implantation. Lastly, in a step towards clinical translation, we
12 demonstrate minimally invasive percutaneous delivery of a human-scale STAR device.

13

14 **Results**

15 **Design of a soft transport augmenting reservoir (STAR).** Our lab previously demonstrated the
16 fibrotic attenuating potential of a dynamic device during the initial stages of the FBR (2 weeks)³⁵.
17 Based on this foundational work, we propose that application of intermittent, cyclical, low
18 amplitude actuation can act as an oscillating shield against the invading, multiphasic FBR, induce
19 local immunomodulatory effects, and create a favourable environment for the rapid long-term
20 transport of macromolecular drug therapy (Figure 1a).

21 To test this hypothesis, we first designed a reservoir suitable for long-term tissue
22 implantation and the precise repeatable delivery of both drug and actuation therapy. Figure 1b
23 shows the multi-layered composition of STAR, with a low-profile design that minimises the

1 presence of sharp angles or edges which may exacerbate the FBR^{52,53}. A therapeutic chamber lies
2 in direct contact with underlying tissue and is separated by a membrane with an array of 10 µm
3 pores (Supplementary Figure 1). A connected indwelling catheter line allows for delivery of drug
4 therapy with temporal control (Figure 1b,c). Superimposed on the therapeutic chamber is an
5 actuation chamber that can be pressurised to elicit controlled oscillation of the porous, tissue-
6 contacting membrane (Figure 1c,d; Supplementary Movie 1). Imbalances between the mechanical
7 properties of the implant and the surrounding tissue are also known to exacerbate FC formation,
8 with stiffer implants eliciting a heightened immune response⁵³. For this reason, STAR was
9 manufactured from thermoplastic polyurethane (TPU) with an elastic modulus of ~15 MPa (Figure
10 1d), similar to that of extracellular matrix^{54,55}. STAR can easily be scaled between animal models
11 using 3D printed moulds and a simple thermoforming/heat-sealing process (Supplementary
12 Figure 2).

13 As part of device design and optimisation, we performed finite element (FE) simulations
14 to understand the biomechanical changes mediated by actuation, particularly the relationships
15 between membrane deflection, convective flow, and tissue strain (Figure 1e,f; Supplementary
16 Figure 3). Based on recently reported results⁵⁰, we designed our soft robotic actuation strategy to
17 induce tissue strain that would fall within the atraumatic range (<40%), and hypothesise that this
18 regimen would mitigate the FBR by creating convective flow disruptive to the cellular immune
19 response.

20

21 **Insulin Transport test (ITT): A longitudinal, *in vivo* method to study the effect of the FBR**
22 **on therapy transport.** Following STAR design and manufacture, we next developed a method to
23 longitudinally monitor the detrimental and progressive effect of the FBR on therapy transport

1 (Figure 2a). Insulin was chosen as our model macromolecular drug to allow for a real time, dose-
2 dependent measurement of functional response as insulin crosses the FC and enters the
3 bloodstream.

4 First, we implanted static STAR devices (without IA) on the subcutaneous dorsal aspect of
5 C57BL/6 mice (Supplementary Figure 4). Next, we injected short-acting human insulin into the
6 device and monitored diffusion-based release across the FC into the systemic circulation *via* serial
7 blood glucose measurements at day 3 (baseline, BL), 2 weeks, and 3 weeks post implantation
8 (Figure 2b).

9 The functional efficacy of an equivalent dose of insulin decreased with implantation time
10 and with progression of the FBR, as indicated by the maximum blood glucose (BG) drop
11 (Figure 2c) and the area under the BG curve (AUC; Supplementary Figure 5a,b). To corroborate
12 these results, we analysed FC thickness longitudinally using 2D micro-computed tomography
13 (μ CT; Figure 2d) and related it to these functional results. As expected, thickness of the capsule
14 increased with time (Figure 2e). Importantly, we observed an inverse linear relationship ($r =$
15 -0.929) between FC thickness and insulin efficacy metrics (Figure 2f, Supplementary Figure 5c).

16 In a final validation step, we examined the effect of FC thickness on therapy release using
17 a multiphysics computational diffusion model. Our simulations corroborate our experimental
18 results, also indicating that increasing FC thickness has a pronounced effect on drug transport
19 (Figure 2g), introducing a time lag for the desired therapeutic concentration to cross the capsule
20 and elicit a functional effect (Figure 2h).

21 In summary, this data demonstrates the development and validation of a pre-clinical model
22 that can detect real-time changes in FC formation *via* its effect on macromolecular transport and
23 track these changes over time.

1
2 **Dynamic intermittent actuation (IA) extends therapeutic lifespan of STAR.** Following device
3 and *in vivo* model development, we designed an 8-week longitudinal preclinical study to test the
4 ability of STAR to modulate the FBR and improve macromolecular delivery across the formed
5 FC.

6 We implanted STAR devices (without drug) on the dorsal subcutaneous aspect of 3 groups
7 of mice (Figure 3a). In two experimental groups, we performed STAR-enabled IA with cyclic
8 pressure input of 2 psi at 1 Hz for 5 min every 12 hours using a custom-made pneumatic control
9 system (Supplementary Figure 6). One group (8W IA) was intermittently actuated for the total
10 study duration of 8 weeks, while the second group (3W IA) received 3 weeks of IA followed by
11 no actuation for the remainder of the study. A third group which did not receive IA served as the
12 control. We then injected short-acting human insulin (2 IU/kg) into the device at various time
13 points post-implantation: 2, 3, 4, 5, and 8 weeks as well as day 3, which served as a baseline (BL).
14 We monitored passive, diffusion-based transport across the formed FC and into the bloodstream
15 *via* serial blood glucose measurements at these timepoints.

16 At baseline, insulin administration produced a similar drop in blood glucose in both IA and
17 control groups (Figure 3b, Supplementary Note 1). Over 8 weeks, the blood glucose curves
18 separate, with decreased insulin responsiveness in the control and 3W IA groups, compared to the
19 8W IA group. Impressively, the 8W IA group maintained its rapid drop in blood glucose over the
20 entire study duration (Figure 3b). Despite chronic implantation and FC development, there was no
21 statistical difference in the maximum blood glucose drop at 3 days (BL; $72.5 \pm 2.2\%$) and 8 weeks
22 ($68.3 \pm 3.4\%$) post-implantation in the 8W IA group (Figure 3c, Supplementary Note 1). In
23 contrast, the control group achieved only a $20.9 \pm 4.3\%$ maximum drop in blood glucose at the 8-
24 week timepoint, reflecting nearly complete loss of drug delivery functionality due to implant

1 isolation by the FC. The 3W IA group had a similar loss of function, with a maximum blood
2 glucose drop that was not significantly better than control at the end of the study (see
3 Supplementary Note 1 for details).

4 The mean time to achieve a physiologically relevant response (30% drop in blood glucose)
5 was preserved at less than 30 min in the 8W IA group over the entire study duration (Figure 3d,
6 Supplementary Note 1), with all mice achieving this 30% drop within 48 min at 8 weeks
7 (Figure 3e). In contrast, the mean time to effect for the control and 3W IA groups progressively
8 increased with implantation time. The mean time to therapeutic effect increased to > 65 min in the
9 3W IA group (with 2 out of 5 mice not responding) by week 8, and was not detectable within the
10 120 min experimental timeframe in the control group (Figure 3d,e, Supplementary Note 1).
11 Notably, 8W IA devices were able to achieve a therapeutic blood glucose drop twice as fast as
12 control devices at 4 weeks (mean time to 30% drop: 27.43 ± 4.48 min vs 73.55 ± 14.85 min) and
13 four times as fast at 8 weeks (26.33 ± 6.16 min vs >120 min).

14 When time and magnitude were integrated by calculating AUC (Supplementary Figure 5a),
15 8W IA produced a robust treatment effect with significant benefits in drug delivery at all time
16 points in comparison to the control group (Figure 3f, Supplementary Note 1). Stopping actuation
17 at 3 weeks in the 3W IA group led to a worsening of insulin response, with an AUC progression
18 that paralleled the control group (Figure 3g). Though there appears to be a trend towards better
19 functional effect in the 3W IA compared to control at the later time points, this was not statistically
20 significant, which implies that continued mechanical dosing will be needed for robust long-term
21 beneficial effects on transport (Figure 3f,g). Overall, this data suggests that IA can mitigate the
22 FBR and extend the therapeutic lifespan of implantable drug delivery devices.

23

1 **Multiphasic temporal effects of intermittent actuation.** Following the completion of our pre-
2 clinical study, we next set out to analyse differences in FC composition at evolving timepoints to
3 better understand the multiphasic cellular changes and key drivers of enhanced drug transport
4 caused by IA (Figure 4a).

5 First, we investigated the initial, acute phase of the inflammatory FBR. Using
6 immunofluorescent Ly-6G⁺ staining, we examined the pericapsular region for the presence of
7 neutrophils, the first responders of the immune defence. We found that IA significantly reduces
8 the presence of neutrophils at day 5 in comparison to the control (Figure 4b,c). This result indicates
9 that the application of IA can mediate a localised immunomodulatory effect.

10 We next assessed the activation of matrix producing cells into a myofibroblast phenotype,
11 a key contractile cell in fibrosis progression. The IA group exhibited a significant reduction in
12 α SMA expression when compared to the control at 2 weeks (Figure 4d,e). Despite observing
13 differences in individual cell populations (neutrophils, myofibroblasts), we did not detect
14 differences in overall cell number at equivalent timepoints (Figure 4f,g).

15 Next, we investigated the macroscale capsular changes responsible for improved therapy
16 transport. We examined evolving capsule thickness with longer periods of STAR implantation. IA
17 mitigated capsule growth in the first 2 weeks following implantation, with a significant reduction
18 in thickness observed w.r.t. the control at 2 weeks (Figure 4h,i). This result aligns with the
19 enhanced blood glucose responsiveness of the IA group at early timepoints (Figure 3) and suggests
20 that FC thickness is an important contributor to initial improvements in macromolecular transport.

21 By 8 weeks, however, capsule thickness had equalized between the IA and control groups
22 (Figure 4i). This result suggests that additional mechanisms are responsible for the sustained
23 improvement in functional response to insulin in the IA group at later timepoints (Figure 3). To

1 investigate this further, we examined capsule vascularity, density, and the maturity of the collagen
2 fibres. However, we did not observe differences that would account for improvements in
3 macromolecular transport and functional effect (Supplementary Figure 7, Supplementary Note 1).
4 By optical coherence analysis of polarised light microscopy images, we found that collagen fibres
5 exhibited higher alignment in the IA group compared to control at week 8 (Figure 4j,k). IA
6 appeared to increase collagen alignment over time from 2 weeks to 8 weeks, whereas there was no
7 temporal change in alignment observed in the control group (Figure 4k). We posit that the lower
8 degree of alignment, and therefore greater degree of fibre entanglement, in the control group
9 creates steric hindrance which potentially slows or immobilises transport of macromolecules
10 through the collagenous matrix^{56,57}.

11 Finally, we examined the ability of IA to protect against cellular invasion and blockage of
12 the porous membrane of STAR. Scanning electron microscopy demonstrated clear differences in
13 cellular infiltration between the control and IA group at the 8-week timepoint (Figure 4l). This
14 effect could be attributed to convective flow generated by STAR upon actuation (Supplementary
15 Figure 3g). These capsular analyses reveal the pleiotropic role of IA in modulating the FBR and
16 highlight cellular and structural changes that lead to improved transport of macromolecular
17 therapy.

18
19 **On demand, actuation-mediated rapid release of drug using STAR.** In addition to intermittent
20 immunomodulatory actuation, we demonstrate another soft robotic actuation-based mechanism of
21 augmenting drug transport. Actuation-mediated rapid release (RR) of a drug loaded STAR device
22 consists of a few (~1–5) cycles of on-demand actuation at the same magnitude as IA (2 psi, 1 Hz).
23 This strategy can accelerate mass transport of drug from the device reservoir (Figure 5a,

1 Supplementary Movie 2) into surrounding tissue (Figure 5b, Supplementary Movie 3). Using this on
2 demand, convective flow-based approach, we investigated if RR could overcome a diffusion-
3 limiting FC barrier by inducing higher concentration and pressure gradients (Figure 5c,d).

4 First, we developed a multiphysics computational model comparing passive diffusion-
5 based transport to RR (Figure 5d–f). RR enhances drug transport across the capsule and thus higher
6 concentrations can reach the therapeutic target in a temporally controlled manner (Figure 5d,e).
7 Furthermore, multiple actuation cycles can increase trans-capsular transport in comparison to a
8 single cycle, and thus dosing can be adapted to the specific clinical scenario (Figure 5f). Péclet
9 number (Pe) calculations⁵⁸ estimate $Pe = 2.35$ for passive diffusion and $Pe = 70.18$ for actuation-
10 mediated RR, suggesting that for a given dose of drug, the time required for passive drug delivery
11 through a diffusion dominated process far exceeds that of actuation-mediated drug delivery, which
12 is convection dominated.

13 To substantiate these simulations, we next investigated the utility of RR *in vivo*, following
14 long-term implantation and development of a FC. We implanted two STAR devices in a Sprague
15 Dawley rat model to evaluate the spatial distribution of drug with and without RR. On day 24
16 following implantation, we monitored the distribution area of a fluorescent small molecule drug
17 analogue (Genhance 750) using an *in vivo* imaging system (IVIS) (Figure 5g). While passive
18 diffusion of Genhance was slow, RR led to a sharp increase (~7 fold) in drug distribution, despite
19 presence of a FC (Figure 5h).

20 In a final example, we demonstrated enhanced mass transport and downstream functional
21 effect using RR in our ITT model at 2 weeks after STAR implantation (Figure 5i). Passive
22 diffusion of insulin led to a drop in blood glucose over 120 min in all animals. At this point, food
23 was given to one group to allow recovery of blood glucose levels towards baseline. At 150 min,

1 this group was subjected to 5 cycles of actuation (with the same parameters of 2 psi at 1 Hz). No
2 additional insulin was administered after the initial dose given at the start of the ITT. Despite a
3 reduced insulin concentration gradient across the device and attenuated insulin sensitivity in the
4 post-prandial animals, actuation-mediated RR led to a significant reduction in blood glucose levels
5 over 15 min due to augmented release of insulin from STAR (Figure 5i).

6 These results support the development of an on-demand actuation-based method to
7 enhance transport across a diffusion limiting FC.

8

9 **Minimally invasive surgical implantation and vision for clinical translation.** We demonstrate
10 in a human cadaver model that the soft and foldable characteristics of STAR lend themselves to
11 minimally invasive implantation, establishing feasibility of clinical translation. Our choice of
12 material (TPU) enables scalability to clinically relevant dimensions (80 mm × 120 mm) and
13 integration of additional elements such as deployment and adhesive channels, without changing
14 the manufacturing process (Supplementary Figure 8). We designed a deployment system and
15 surgical plan (Supplementary Figure 9) to allow for minimally invasive implantation of STAR to
16 an intermuscular delivery site through a 1 cm incision. The deployment system consists of a
17 delivery sheath, space creating balloon, and delivery cartridge containing STAR. We selected the
18 *transversus abdominis* plane, lying in the anterior abdominal wall between the internal oblique
19 and *transversus abdominis* muscles, as the implant site (Figure 6a). This potential space is well
20 vascularized and is an established tissue plane that is frequently accessed by healthcare providers,
21 e.g., for the delivery of analgesia during abdominal surgery^{59,60}.

22 Guided by ultrasound imaging, we first accessed the intermuscular *transversus abdominis*
23 space with an 18-gauge needle in the anterior abdominal wall and used hydro-dissection to separate

1 the tissue plane between the internal oblique and *transversus abdominis* muscles (Figure 6b). Next,
2 we utilized the Seldinger technique to exchange the needle over a wire⁶¹ and verify correct tissue
3 placement with ultrasound guidance (Figure 6c). We then made a 1 cm skin incision to facilitate
4 device delivery and used a commercially available dilator set to expand the space and
5 accommodate positioning of our bespoke delivery sheath (Figure 6d). Finally, we completed
6 separation of the tissue planes with a space creating balloon, which could be visualized with
7 ultrasound during filling. The scaled-up STAR device was then preloaded into the delivery
8 cartridge, advanced easily through the sheath to the submuscular plane (Figure 6e) and deployed.
9 Pressurisation of the deployment channel with echogenic contrast enabled opening of the device
10 under ultrasound visualisation (Figure 6f,g). Postprocedural dissection of the tissue indicated that
11 the device was successfully delivered to the appropriate space.

12

13 **Discussion**

14

15 We present STAR, an implantable platform which can evade and overcome the diffusional barrier
16 of the FC to achieve long-term enhanced therapy transport using two synergistic soft robotic
17 strategies: (1) intermittent immunomodulatory actuation, and (2) actuation-mediated rapid release.

18 Prior to testing these mechanotherapeutic strategies, we developed a robust *in vivo* method
19 (ITT) of detecting the effect of the FBR on macromolecular transport of insulin and monitoring
20 this outcome over time (Figure 2). The ITT has several advantageous features which aid
21 technology development. Firstly, its measurement in real time allows for agile feedback and
22 iterative development. Secondly, the method allows for precise, quantitative assessment of the
23 intervention using clinically relevant parameters including time to effect, maximum effect, or an
24 integration of both time and magnitude *via* AUC. Finally, the repeated, non-invasive

1 measurements enable longitudinal studies of complex, multiphasic phenomena with the ability to
2 track individual animals and treatment groups over time.

3 IA is the first element in STAR's armamentarium. By inducing strain at the tissue-
4 contacting membrane and perturbing peri-device fluid flow, STAR acts as an oscillating
5 mechanical shield against the invading cellular FBR. These localised mechanical effects create a
6 favourable environment for the long-term transport of macromolecules. IA is able to preserve the
7 functional effect of STAR over 8 weeks at the same level as that seen immediately following
8 implantation, i.e., prior to formation of a significant inhibitory FC (Figure 3c). In sharp contrast,
9 the insulin responsiveness of the control group decreased with longer implantation time, until near
10 complete FC isolation and implant failure. The extension of therapeutic effect using a simple
11 5 min, twice daily actuation regimen represents an attractive and innovative strategy for mitigating
12 the FBR.

13 By combining ITT blood glucose results with a range of *ex vivo* capsular analytical
14 techniques, we were able to unravel the multiphasic, temporal effects of IA on cell infiltration and
15 capsule formation. When examining the inflammatory response in the IA and control groups, we
16 detected significant differences in distinct cell populations at times of expected peak infiltration,
17 despite no difference in total pericapsular cell number (Figure 4).

18 We demonstrated that IA produces a localised immunomodulatory effect by clearing
19 neutrophils from the pericapsular site (Figure 4b,c). The infiltration of neutrophils is an important
20 first step in the FBR, initiating and propagating the inflammatory process, which causes the
21 subsequent recruitment of cell populations known to develop the FC (e.g., macrophages). Early
22 modulation of the neutrophil response may have important long-term consequences on the FBR.
23 Indeed, we observed a trend towards better functional effect in the 3W IA group compared to

1 control even after stopping actuation. Though this effect did not reach statistical significance, there
2 may be some long-term benefits in therapy transport even with initial periods of actuation;
3 however, it is clear that maximal anti-inflammatory benefit comes with continued actuation
4 (Figure 3f). A recent study by Seo et al.⁵⁰ corroborates the mechanosensitive nature of neutrophil
5 cell populations following dynamic loading for skeletal muscle regeneration. The authors
6 postulated that mechanical flushing of chemoattractants was responsible for the reported decrease
7 in neutrophils⁵⁰. In this context, our work motivates further study on the effect of IA on pro-
8 inflammatory chemoattractant gradients such as interleukin 1, 6 and 8 versus direct mechanical
9 effects on cell attachment, orientation, and function.

10 We noted a significant decrease in myofibroblast cell number with IA compared to control
11 (Figure 4d,e). The activation of matrix-producing cells into a myofibroblast phenotype,
12 characterised by α SMA expression, is a critical step in fibrosis progression. Increased expression
13 leads to heightened contractile activity, formation of stress fibres, and synthesis of extra-cellular
14 matrix. Furthermore, activated fibrogenic cells can produce cytokines responsible for additional
15 cell recruitment and propagation of the deleterious fibrotic response⁴⁴. It is becoming increasingly
16 evident that stiffness precedes, or is an important contributor to fibrosis⁶². Thus, reducing
17 myofibroblast expression and its effect on matrix stiffness may be a key strategy to modifying this
18 self-perpetuating fibrotic effect.

19 In addition to multiphasic cellular changes with time, we also observed multiphasic
20 differences in macro-capsular architectural evolution, with distinct differences between the IA and
21 control groups. At early timepoints (2 weeks) following implantation, we observed differences in
22 capsule thickness between groups (Figure 4h,i), which aligns well with improved diffusion-based
23 transport at these timepoints (Figure 3b,f). FC thickness equalises between groups after 8 weeks

1 of implantation, indicating that other mechanisms are responsible for the improved transport at
2 later timepoints. We rejected several relevant hypotheses for improved transport in the IA group
3 including vascularity, capsule density, and collagen maturity (Supplementary Figure 7). However,
4 we noted differences in collagen architecture and cellular infiltration into the device reservoir
5 which may contribute to late differences in transport and functional effects between the groups
6 (Figure 4j–l). Further study will be necessary to fully elucidate and understand the multiple
7 mechanisms at play. Note that 2 out of 6 mice in the control group had to be withdrawn from the
8 study at later time points due to self-inflicted damage to their dorsal tissue and subcutaneous
9 implant. Interestingly, this did not occur in any IA group mice. Capsular contracture mediated by
10 excessive FBR is a painful condition⁶³ that could rationalise this group difference, and future work
11 could investigate this observation further.

12 In addition to IA, STAR possesses a second transport-augmenting strategy of RR.
13 Actuation of a drug loaded STAR can induce higher concentration and pressure gradients^{32,33}, and
14 thus improve drug transport across a formed FC with temporal control (Figure 5). RR can be
15 particularly advantageous for potent drugs where accurate dosing and a rapid time to functional
16 effect is important, or macro-drugs such as proteins, where convective flow enhances diffusion-
17 based flux, principally governed by molecular weight and concentration gradient^{34,35}. Convection-
18 enhanced delivery has been successfully used to improve distribution of chemotherapy in deep
19 brain tumour targets, albeit with modest clinical outcomes^{27,28,33,64}.

20 We envision several clinical use scenarios for IA and RR technologies demonstrated in this
21 work. RR could be used independently of IA, such as for the rapid, on-demand delivery of
22 therapeutics in response to a clinical emergency. Some relevant examples include the delivery of

1 adrenaline for the treatment of anaphylaxis or glucagon for the treatment of hypoglycaemic coma.
2 In both cases, impedance of delivery due to FC formation would have grave consequences.

3 An optimal FC mitigating strategy could also combine both IA and RR. For example, short
4 daily bursts of IA (without drug) could attenuate the FBR to extend device lifespan and improve
5 performance in long-term implantation. A temporally controlled and modular RR actuation regime
6 could then be used to make rapid, precise dosage adjustments in accordance with the patient-
7 specific clinical scenario and/or FBR severity. IA and RR are complementary strategies that
8 elegantly make use of a single device design and pump, making STAR well suited to a variety of
9 clinical scenarios.

10 The management of type 1 diabetes is one relevant clinical area where STAR could provide
11 synergistic benefit. For example, IA could be applied to extend the lifespan of an artificial
12 pancreas⁶⁵, preventing unnecessary FBR-mediated blockages, linked hyperglycaemic events, and
13 ultimately simplifying the dosing regimen and patient experience. In synergy, actuation-mediated
14 RR could make rapid insulin adjustments and maintain blood glucose levels in the narrow window
15 necessary to prevent long-term complications⁶⁶. Looking further into the future, application of
16 STAR could enable translation of next-generation bioartificial technologies utilising human-
17 derived insulin-producing islet cells by modifying the transport-limiting FC, which has been a
18 major barrier to the viability of cell-based therapeutics^{19,67-69}. Considering the need for daily IA
19 for long-term efficacy, it is likely that embodiments will require a wearable pump, similar to those
20 used in existing insulin pumps⁷⁰.

21 STAR's soft material endows it with biocompatibility advantages over rigid implantable
22 drug delivery systems⁷¹, and lends itself to minimally invasive catheter implantation. In a step
23 towards clinical translation, we scaled-up the STAR device and developed a bespoke delivery tool

1 as well as a minimally invasive procedure that is congruent with conventional interventional
2 radiology techniques (Figure 6). We demonstrated delivery of STAR to a clinically accessible
3 intermuscular space in the anterior abdominal wall in a human cadaver model. This approach
4 allowed for a short procedure time (<20 min) for implantation of a human scale device through a
5 modified sheath using ultrasound guidance in the hands of an experienced interventional
6 radiologist. Additional design features demonstrated correct deployment and adhesive delivery to
7 maintain device position in the tissue plane. Thus, STAR can be quickly implanted by
8 interventionalists in an outpatient setting under local anaesthesia using an established imaging
9 modality.

10 Though we have demonstrated robust preclinical results in a long-term mouse model, there
11 are several limitations and barriers to clinical translation. Our findings from implantation in the
12 dorsal subcutaneous space of mice may not directly predict similar results in humans at different
13 anatomical locations (e.g., abdominal intermuscular space), and further work is needed to
14 understand how these anatomic and microenvironmental differences impact the effects of STAR.
15 Although rodent models have been extensively used to study the FBR^{9,27}, rodents have been shown
16 to have different tissue collagen content in the subcutaneous space surrounding an implant and
17 different metabolites in the interstitial fluid at implant interfaces, as compared to humans⁷².
18 Moreover, differences in rodent skin, fur, and behaviours may subject implanted devices to
19 different biomechanical forces than in humans⁵². Encouragingly, thus far, we have observed
20 similar FBR mitigating effects with an IA regimen that is agnostic of species and device design³⁵.
21 In addition to this, the presence of conserved inflammatory pathways in FC formation across
22 species⁷³ suggest that STAR may have a similar benefit in prolonging implant lifespan in humans.

1 Although there have been a number of prior studies that examine the effect of mechanical
2 loading on inflammation and tissue regeneration, there is significant heterogeneity with regards to
3 actuation methods, regimens, resulting deformations, target tissues, and animal models used ⁴⁵⁻⁵¹.
4 Only a few studies have attempted to address the effect of varying tissue strain^{48,50,51} and loading
5 frequency⁵¹; therefore, significant work is needed to define the optimal loading parameters that
6 maximise the anti-inflammatory effects of mechanical actuation, which may differ with the type
7 of tissue and mechanical stimulus.

8 We can draw six conclusions from this study: (1) The ITT represents a robust, longitudinal
9 method to monitor macromolecular therapy transport across a developing FC *in vivo*. (2) The FBR
10 can negate insulin transport over time from a static STAR device until complete implant isolation
11 and therapy failure. (3) Intermittent actuation can preserve therapy transport at baseline levels and
12 extend the therapeutic lifespan of STAR, even with long-term implantation. (4) IA can mediate
13 immunomodulatory changes in the neutrophil inflammatory response and elicit downstream
14 multiphasic temporal changes in cellular infiltration and capsule formation. (5) Actuation-
15 mediated RR of a drug loaded STAR device can synergistically enhance mass transport and
16 therapeutic effect with tunable, temporal control, despite the presence of a FC. (6) Minimally
17 invasive catheter implantation of STAR was possible in a human cadaver model, showing clinical
18 translatability of our approach.

19 In summary, the STAR platform represents a new mechanotherapeutic approach to both
20 mitigate and overcome the FBR, extending the lifespan and efficacy of implantable drug delivery
21 devices. It holds vast clinical utility for a variety of indications where transport is affected by
22 fibrosis, such as the management of type 1 diabetes.

23

1 **Methods**

2 **Device manufacture for pre-clinical mouse model.** Positive two-channel, positive one-channel,
3 and corresponding negative moulds were 3D printed using VeroBlue resin (Stratasys Objet30)
4 (Supplementary Figure 2a). Thermoplastic urethane (TPU; 0.3 mm, XGD0385, QING GEN) was
5 vacuum thermoformed (Yescom Dental) over the positive two-channel mould (Supplementary
6 Figure 2b). This process was then repeated with the positive one-channel using a thinner TPU
7 (0.076 mm, HTM-8001-M, polyether, American Polyfilm) (Supplementary Figure 2b). Pores with
8 10 μm diameter were laser cut in a TPU membrane (0.076 mm, HTM-8001-M, polyether,
9 American Polyfilm) using a UV Nanosecond laser (National Centre for Laser Applications,
10 National University of Ireland Galway).

11 The thermoformed and laser cut membranes were assembled in a negative mould
12 (Supplementary Figure 2c). Mandrels of outer diameter 0.21 mm were inserted into the channels
13 to retain patency. The assembly was heat sealed together using a heat transfer machine (330QXAI,
14 PowerPress). The mandrel was removed and a TPU catheter tubing (0.037" \times 0.023"; MRE037,
15 Micro-Renathane, Braintree Scientific) was inserted and heat sealed to the device using heat shrink
16 tubing. The final assembled devices measured 15 mm (width) \times 18 mm (length) \times 2 mm (height)
17 and consisted of two chambers – the larger measuring 12 \times 6 mm and the smaller measuring 3
18 \times 12 mm (Supplementary Figure 3a, b).

19 **Device manufacture for pre-clinical rat model.** Rat scale devices were produced as previously
20 described³⁵. Final devices measured 12 mm in length with the semi-circular reservoir measuring
21 3.9 mm in height and 3.5 mm in diameter, with variable lengths of 3Fr TPU catheter tubing.

22 **Device manufacture for human scale prototype.** Human scale devices measuring 120 \times 80 mm
23 were produced as previously described (Supplementary Figure 8)⁷⁴. An additional deployment

1 channel and actuation chamber were included to allow for minimally invasive delivery through a
2 sheath and dynamic actuation following implantation (Supplementary Figure 9).

3 **Electropneumatic actuation and control system.** A custom-made electropneumatic system to
4 deliver actuation to the implanted device was developed as described in Supplementary Figure 6.
5 The system consisted of pre-programmed electrical signalling to control pneumatic power sources.
6 Pneumatic components included a positive pressure and vacuum generator, a pressure regulator,
7 and electropneumatic (solenoid) valves. A programmable microcontroller board (Arduino Uno)
8 along with a power source were used to establish an open loop control of the pneumatic power.
9 Positive pressure was guided through an electropneumatic pressure regulator (ITV1030; SMC
10 Inc.) which was controlled *via* the microcontroller board to adjust the precise actuation pressure.
11 Actuation of the implanted device was then achieved by alternating the positive pressure for device
12 expansion and negative pressure for device deflation. The delivery of this pneumatic actuation
13 pattern was ensured by two electropneumatic solenoid valves (NVKF333; SMC Inc.) for positive
14 and negative pressure which were controlled using the same microcontroller and two MOSFETs
15 coupled to the electrical supply power (Supplementary Figure 6a). A manifold was used to actuate
16 multiple devices in separate animals simultaneously, ensuring that the set pressure level was
17 consistently achieved on all the manifold channels (Supplementary Figure 6b,c).

18 **STAR membrane deflection characterisation.** STAR devices were manufactured and placed in
19 a custom-made 3D printed holder (Objet30 Prime, Stratasys). Devices were pneumatically inflated
20 from 1-9 psi using the electropneumatic actuation and control system described above. Images of
21 membrane deflection were captured using a digital camera (Nikon DSLR) and tripod positioned
22 in the side view. Deflection magnitude was subsequently analysed using ImageJ. Based on this
23 strategy, a bi-chambered configuration was selected to investigate the effect of two distinct

1 deflection magnitudes (0.58 and 1.3 mm) in our pre-clinical mouse model. It should be noted that
2 the lower deflection magnitude was closely matched to our previous work.

3 **STAR membrane pore characterisation.** 5 mm × 5 mm pieces of laser-cut porous TPU
4 membranes were characterised by scanning electron microscopy (SEM) using a Hitachi S2400
5 microscope operating at 20 kV electron accelerating potential in backscatter electron imaging
6 mode and a sample working distance between 8–10 mm. After imaging, pore diameters were
7 measured from the images using the Hough circle transform function in Fiji 2.0.0 (ImageJ)
8 (Supplementary Figure 1).

9 **Computational modelling.** Fluid-structure interaction (FSI) simulations using smoothed particle
10 hydrodynamic (SPH) method were conducted to investigate the peri-implant fluid flow and the
11 dynamics of drug transport under active delivery. All FSI simulations were created using
12 Abaqus/Explicit 2018 (Dassault Systèmes, Vélizy-Villacoublay, France). The device was
13 modelled as a 3D surface geometry and meshed with 14,636 four-node shell elements (Abaqus
14 node type S4R). A Dirichlet boundary condition, where nodal displacement in all directions were
15 fixed to zero, was applied on the edge of the bottom porous membrane to prevent rigid body
16 motion. A pressure loading which linearly ramped up to 2 psi in 500 ms and ramped down to 0 psi
17 in the next 500 ms was applied to the internal surface of the outer and middle membranes. The
18 membranes were modelled using an Ogden 3rd order hyperelastic material with parameters,
19 $\mu_1 = -8.31$ MPa, $\mu_2 = -0.36$ MPa, $\mu_3 = 17.89$ MPa, $\alpha_1 = 0.46$, $\alpha_2 = 3.62$, $\alpha_3 = -3.10$. The fluid
20 domains, drug and outer fluid, were meshed with linear tetrahedral elements and each element was
21 converted to an SPH particle located at its centroid. The drug domain and the outer fluid domain
22 contained 107,406 and 174,516 particles, respectively. The particles were assigned with the

1 following properties: density of $9.96\text{E-}7 \text{ kg/mm}^3$, bulk modulus of 2.094 GPa and dynamic
2 viscosity of $3.56\text{E-}8 \text{ MPa-s}$.

3 A structural finite element (FE) model was constructed to investigate the deformation of
4 the tissue underneath the device. The device was modelled with same geometry and material model
5 as the FSI simulations. To model the skin constraint when the device is implanted subcutaneously,
6 a dome shape shell structure with a linear elastic property ($E = 1 \text{ MPa}$) was added on top of the
7 device. The tissue was also modelled as a linear elastic material ($E = 15 \text{ kPa}$). In terms of boundary
8 conditions, the edge of the skin and the bottom face of the tissue were both fixed in all directions.
9 Tie constraints were used to model the suture attachments between the device and the tissue. A
10 general contact with friction coefficient of 0.5 was applied throughout the entire model. The inner
11 chamber of the device was subjected to different linearly increasing pressure loadings (1 psi, 2 psi
12 and 3 psi). From the FE simulations, strain contour plots and downward displacement contour
13 plots of the tissue were extracted. The average strain and deflection were calculated at the interface
14 between the device and the tissue.

15 Mass transport simulations were conducted using COMSOL Multiphysics software version
16 5.6 (COMSOL, Burlington, MA). The 2D model contains three domains: the drug reservoir, the
17 FC, and the outer fluid domain representing the body. The FC is a thin layer surrounding the device
18 with thickness ranging from $50 \text{ }\mu\text{m}$ to $200 \text{ }\mu\text{m}$. The outer fluid domain is a rectangular region with
19 height and width of 5 mm and 20 mm, respectively. Mass transport was modelled using the
20 transient diffusion-convection equation in the “Transport of Diluted Species” module in
21 COMSOL. Diffusivity in the reservoir and outer fluid domain was given as $855 \text{ }\mu\text{m}^2/\text{s}$ and the
22 diffusivity in the FC was given as $50 \text{ }\mu\text{m}^2/\text{s}$ ⁷⁵. Initial concentration of 1 mol/mm^3 was applied in
23 the drug reservoir domain. Fluid velocity between pure fluid domain and porous media was

1 calculated using the Brinkman Equations in COMSOL. The drug reservoir and the outer fluid were
2 modelled as pure fluid domain with density of 997 kg/m^3 and viscosity of $8.9\text{E-}4 \text{ Pa}\cdot\text{s}$. The FC
3 was modelled as porous media with permeability of $8.9\text{E-}16 \text{ m}^2$ and porosity of 0.8. Pressure inlet
4 of 2 psi ramped up and down within 1 s was applied on the boundary of the reservoir when device
5 actuation was triggered; and zero outlet pressure was applied on the boundary of the outer fluid
6 domain. The model showed the transient changes of drug distribution profile within 30 min under
7 diffusion and convection. In all simulations, both local concentration and regionally accumulated
8 concentration were extracted over time as quantitative outputs.

9 **Péclet number calculations.** The Péclet number for mass transfer, for a characteristic length L , is
10 defined as $Pe_L = uL/D$, where u is the local flow velocity and D is the diffusivity⁵⁸. For
11 representative calculations, $L = 1 \text{ mm}$ was assumed. This assumption is based on experimental and
12 computational data showing that the maximum device membrane deflection is $\sim 1.5 \text{ mm}$ and the
13 estimated tissue deflection is $\sim 0.73 \text{ mm}$ (Supplementary Fig. 3). A diffusivity of $D = 855 \text{ }\mu\text{m}^2/\text{s}$,
14 which is the same value used in the COMSOL Multiphysics models above, was used⁷⁵. For the
15 passive diffusion scenario, a reasonable estimate for u is the velocity of interstitial fluid, which has
16 been reported widely to be in the range $0.1\text{-}2 \text{ }\mu\text{m/s}$ ⁷⁶. A flow velocity of 0.06 mm/s adjacent to
17 the porous membrane immediately following actuation was calculated from the COMSOL
18 simulations above.

19 **Pre-clinical studies.** Animal procedures were reviewed and approved according to ethical
20 regulations by the Institutional Animal Care and Use Committee at Massachusetts Institute of
21 Technology. Animals were housed in a facility with 12 hour on/off light cycle, at $20\text{-}22^\circ\text{C}$ with a
22 relative humidity ranging between 30-70%. Animals were singly housed with standard bedding

1 and food for the duration of the study. All devices were sterilised using ethylene oxide before
2 implantation.

3 **Mouse pre-clinical surgery:** Male C57BL/6 mice (25-30 g) were placed under anaesthesia using
4 inhalable isoflurane (1-3%). A single dose of sustained-release buprenorphine (Bup-SR, 1 mg/kg)
5 was administered subcutaneously to control pain. A STAR device was implanted subcutaneously
6 in the mouse as depicted in Supplementary Figure 4. To prepare the surgical site, the hair on the
7 back of the mouse was removed using a clipper and topical depilatory cream, and the sites was
8 sterilised with three washes of Povidone-iodine and 70% ethanol. Medial dorsal incisions were
9 made at the base of the neck and 1 cm from the tail (Supplementary Figure 4a). A blunt dissection
10 was made at the incision sites, and a curved haemostat was used to tunnel subcutaneously from
11 the superior to the inferior sites. A transcutaneous self-sealing port available from Instech
12 laboratories (VABM2B/22R22) was connected to the dorsal end of the therapy and actuation
13 catheter of each STAR device. The 15 × 18 × 2 mm STAR device (Supplementary Figure 3a,b)
14 was then inserted under the skin *via* the superior incision site and tunnelled inferiorly into position.
15 The device was secured to the underlying fascia with one suture at either side (7-0 monofilament).
16 The port was then inserted under the skin at the superior incision site (Supplementary Figure 4b).
17 The skin at each incision site was then closed with interrupted sutures (5-0 Maxon monofilament)
18 (Supplementary Figure 4c), and the animal was allowed to recover on a heated pad. To replete
19 intraoperative fluid losses, 0.2 mL of warm saline was administered subcutaneously.

20 **Insulin transport test (ITT):** A kinetic measure of insulin release from STAR devices and
21 subsequent effect on blood glucose levels was measured through an ITT. The mouse was weighed
22 and a solution containing a dose of 1 IU/kg/150 µL was prepared from a stock solution of Humulin
23 R U-100 short acting human insulin. Animals were fasted for 4 hours prior to the start of the ITT,

1 and were kept in a clean cage without food and bedding for the duration of the test. An initial blood
2 glucose measurement was taken to establish a baseline. An insulin preparation at a total dose of 2
3 IU/kg was then administered into the device *via* the transcutaneous self-sealing port available from
4 Instech laboratories (VABM2B/22R22) at time = 0 min. A PNP3M connected to a 1mL Luer Lock
5 disposable syringe (BD) was used to administer the dose (Supplementary Figure 4e). Following
6 administration, blood was sampled from the lateral tail vein of the mouse and serial blood glucose
7 measurements were performed over 120 min using a Bayer Contour Next Blood Glucose
8 Monitoring System at time = 15, 30, 45, 60, 75, 90, and 120 min. The animal was restrained using
9 a commercial restrainer (TV-RED 150-STD, Braintree Scientific). The tail was warmed using a
10 HotHand warmer for 10 seconds prior to blood sampling. The area was then disinfected with a
11 Kimwipe soaked in 70% ethanol. Finally, venepuncture was performed using a 27-gauge needle
12 (BD) and the measurement was recorded.

13 ***Intermittent actuation:*** Intermittent actuation was performed by connecting a custom-made
14 electropneumatic actuation and control system (described above) to the self-sealing transcutaneous
15 actuation port using a PNP3M connector (Instech) (Supplementary Figure 4d). The device was
16 then cyclically actuated at a controlled input pressure of 2 psi at 1 Hz for 5 min every 12 hours as
17 previously described^{35,47}. No drug was present in the device during IA throughout the study.

18 ***Long-term mouse preclinical study:*** Devices were implanted into male C57BL/6 mice as
19 described above. ITTs as described above were performed in all mice at 2, 3, 4, 5, and 8 weeks
20 following device implantation, after which point animals were euthanized by CO₂. Six devices
21 were static controls, and ten devices were dynamically actuated for 5 min every 12 hours. The
22 actuation group was split at three weeks post-actuation, with one group (n=5) stopping actuation

1 and remaining passive thereafter, and another group (n=5) continuing dynamic actuation for the
2 entire 8- week study period (Figure 3a).

3 ***Actuation-mediated rapid release of insulin:*** At 2 weeks following implantation, serial blood
4 glucose measurements were performed over 120 min after insulin injection as described above.
5 Food was given to the RR group at 120 min to allow recovery of blood glucose levels. Actuation-
6 mediated RR was then performed by actuating the STAR device at 150 min. Note that no additional
7 insulin was administered after the initial dose given at time = 0 min. The STAR actuation reservoir
8 was connected to a custom-made pneumatic control unit, *via* the transcutaneous self-sealing access
9 port, using a PNP3M connected (Instech) and pneumatically activated for 5 cycles of 2 psi cyclical
10 pressure at 1 Hz. The duration of activation and evacuation were equivalent. Blood glucose levels
11 were measured by tail vein sampling at four additional 15 min increments in 4 mice per group.

12 ***Rat pre-clinical study:*** Two female Sprague Dawley rats (250-300g) were placed under
13 anaesthesia using inhalable isoflurane (1-3%). Bup-SR (1 mg/kg) was administered
14 subcutaneously to control pain. To prepare the surgical site, the hair on the back of the rats was
15 removed, and the sites was sterilized with three washes of Betadine and 70% ethanol. A superior
16 incision was made at the base of the neck for the port, and two inferior incisions were made 9 cm
17 from the original incision along the back of the rat and 1 cm lateral of the spine. A blunt dissection
18 was made at all incisions, and a pair of forceps was used to tunnel subcutaneously from the anterior
19 to the posterior sites. A transcutaneous self-sealing port available from Instech laboratories
20 (VABM2B/22R22) was connected to the dorsal end of the therapy and actuation catheter of each
21 STAR device. The ports were placed in position at the base of the neck, and the devices were
22 tunnelled posteriorly into position. Each port was secured to the underlying fascia using at least
23 one interrupted suture (5-0 monofilament). Each STAR device was secured to the underlying

1 fascia with one suture at either side (7-0 monofilament). The skin was closed with interrupted
2 sutures (5-0 monofilament), and the animal was then allowed to recover on a heated pad. Warm
3 saline (300 μ L) was administered subcutaneously to mitigate fluid loss and dehydration caused by
4 surgery. Animals were euthanized by CO₂.

5 ***In vivo imaging:*** On day 24 of the rat preclinical study, release of a small molecule fluorescent
6 drug analogue, Genhance 750 (Perkin Elmer), was assessed using IVIS Spectrum *in vivo* imaging
7 system (Perkin Elmer). An excitation filter of 745 nm and an emission filter of 800 nm was used
8 to acquire the images. First, the animals were anesthetized using inhalable isoflurane. The hair was
9 removed above the subcutaneous STAR reservoir, and the circumference of each reservoir was
10 marked on the skin, to aid identification of the region of interest (ROI). After preparation, a control
11 image was acquired. 35 μ L of the Genhance drug analogue was then injected *via* the
12 transcutaneous port and refill line into each STAR reservoir using a syringe pump (Harvard
13 Apparatus). The refill line and reservoir were first cleared by applying a vacuum. Images were
14 acquired every 3 min in a sequence. After 14 min following injection of Genhance into STAR,
15 imaging was paused, and the intervention reservoir was pneumatically activated. Imaging was then
16 restarted. Using a consistent image threshold, a custom region of interest (ROI) was used to
17 delineate the area of diffusion in each image (Living Image 4.5.4, Perkin Elmer). The initial area
18 of diffusion at $t = 0$ following injection was subtracted from all subsequent readings for analysis.

19 ***Photoacoustic ultrasound:*** Methylene blue solution (1 mg/mL) was loaded into the STAR
20 reservoir, and the actuation line of the device was filled with water using a three-way valve
21 (Qosina), as air could cause interference with the ultrasound signal. The device was implanted
22 subcutaneously in one Sprague Dawley rat just after euthanasia. Ultrasound and photoacoustic
23 imaging (PAI) were performed using the Vevo LAZR-X Photoacoustic and micro-ultrasound

1 Imaging System (FUJIFILM VisualSonics, Toronto, Canada) using a MX550 transducer at a
2 frequency of 40 MHz, and a resolution of 40 μm . The narrow Vevo Optical Fiber, composed of
3 high efficiency fused silica, and surrounded by a MX550 Fiber Jacket was used to perform
4 Multispectral PAI imaging.

5 The probe was moved across the skin, and the subcutaneous device, using a 3D stepper
6 motor to obtain a 3D data set. Using Nanostepper mode, each slice of the 3D scan consisted of
7 images taken at 680, 730, 750, 800, 850, and 900 nm wavelengths. In addition, a Spectro mode
8 acquisition was performed at a single slice, where images were taken between 680 and 970 nm in
9 5 nm increments. Images were rendered using VevoLAB 3.2.0 software (FUJIFILM VisualSonics,
10 Toronto, Canada).

11 **Human cadaveric studies.** The protocol for this study was approved by the NUI Galway Research
12 Ethics Committee. All cadaveric material was bequeathed to the Medical School, National
13 University of Ireland Galway, for further advancement of medical knowledge. Informed consent
14 was obtained from next of kin as part of the bequeathment process. This is covered by legislation
15 governing the practice of Anatomy in the Republic of Ireland (Medical Practitioners Act 2007).
16 Whole adult male human cadavers were fixed with embalming fluid containing 21% methanol,
17 21% glycerine, 5.6% phenol, and 3.1% formaldehyde. A Seldinger technique was used to access
18 the *transversus abdominis* plane under ultrasound guidance⁶¹. The delivery system consisted of a
19 delivery sheath, a space-creating balloon, and the STAR delivery cartridge (Supplementary Figure
20 9). The delivery sheath and STAR delivery cartridge were directly 3D printed (Form 2, Formlabs,
21 Somerville, MA). The space creating balloon consisted of a 3D printed shaft connected to a TPU
22 balloon. A 4-14 MHz linear ultrasound transducer (Clarius) was used to visualize the muscular
23 layers of the anterior abdominal wall in the cadaver. An 18-gauge needle was advanced into the

1 transversus abdominis plane and physiological saline as injected to separate the muscle planes with
2 hydro-dissection. The needle was exchanged over a 5 Fr, 10 cm sheath, and a 1 cm skin incision
3 was made using a scalpel. An 0.035” Amplatz super stiff wire (Boston Scientific) was then
4 advanced into the space, and the serial dilation was performed using an Amplatz-type renal dilator
5 set (Boston Scientific), followed by the advancement of the custom-made delivery sheath into the
6 tissue plane. The space-creating balloon was used to fully separate the tissue planes, and then
7 exchanged for the delivery cartridge. STAR was then deployed using the cartridge into the
8 intermuscular space. The cartridge and delivery sheath were removed, and reservoir filling was
9 demonstrated with infusion of an ultrasound contrast agent (SonoVue) and visualized under
10 ultrasound. Proper deployment of the device and positioning was confirmed with dissection.

11 **Histology and immunohistochemistry.** After euthanizing the animals with CO₂, each device and
12 the immediate surrounding tissue were extracted. Tissues were fixed for 24 hours using 10%
13 formalin (pH 7.4). The tissue was then washed and stored in PBS. Fixed tissue samples were
14 transected in half, oriented, and embedded in paraffin wax blocks for histological and
15 immunohistochemical analyses. Each block was assigned a code for randomization and blinding
16 purposes. Sections of 7 µm were cut, deparaffinized in xylene, and rehydrated through a series of
17 graded alcohols. For assessment of FC collagen maturity and arrangement, sections were stained
18 in 0.1% Fast Green and then in 0.1% Sirius red in saturated picric acid. Slides were then dehydrated
19 through graded alcohols and cleared in two changes of xylene. The slides are cover slipped using
20 DPX mounting medium and left to dry. Slides were imaged using Ocular 2.0 Imaging Software on
21 an Olympus BX4 polarised light microscope (Mason Technology Ltd. Dublin, Ireland) at 20x
22 magnification.

1 To determine total cell number per area, samples were stained with haematoxylin and eosin
2 following widely established protocols⁷⁷. Ocular 2.0 Imaging Software on an Olympus microscope
3 (Mason Technology Ltd. Dublin, Ireland) at 40x magnification. For immunohistochemical
4 analysis, primary antibodies of CD31 (1:200; Ab182981, Abcam) and α SMA (1:500; ab7817,
5 Abcam) were incubated for 1 hour at 37°C. Secondary antibodies of Alexa Fluor 594 goat anti-
6 mouse immunoglobulin G (IgG; 1:200 Thermo Fisher Scientific), Alexa Fluor 594 goat anti-rabbit
7 IgG (1:200; Thermo Fisher Scientific), and Alexa Fluor 488 goat anti-mouse IgG (1:200; Thermo
8 Fisher Scientific) were incubated for 1 hour at room temperature, respectively. Primary antibody
9 Ly-6G (1:100; Biolegend 127602) was incubated for 1 hour at 37°C after Tris-EDTA (pH 9)
10 antigen retrieval. A ready probes mouse-on-mouse (Invitrogen, R37621) blocking solution was
11 performed to block endogenous binding. Secondary antibody Alexa Fluor 488 goat anti-rat 488
12 (1:100; Thermo Fisher Scientific) was incubated for 1 hour at room temperature. Sections were
13 stained with Hoechst and cover slipped using fluoromount. Immunofluorescence-stained slides
14 were observed using a spinning disc inverted confocal microscope (CSU22, Yokagawa) combined
15 with Andor iQ 2.3 software. For blood vessel analysis, CD31 (1:200; Ab182981, Abcam) was
16 incubated for 1 hour at 37°C after sodium citrate antigen retrieval (pH 7.2). A Dako EnVision+
17 System–HRP (DAB) secondary and counterstain (haematoxylin) was applied following
18 manufacturer instructions.

19 ***Analysis of total cell number:*** Five random fields of view were acquired from two sections using
20 light microscopy. Images were cropped to ensure only the capsule was included in the analysis.
21 Sections were converted to 8-bit and nuclei were manually thresholded from background tissue.
22 Particles were outlined and analysed using ImageJ (Fiji version 2.0.0) software.

1 ***Analysis of myofibroblasts and neutrophils:*** Ten random fields of view were acquired from two
2 sections using confocal microscopy. The volume fraction of α SMA+ cells and Ly-6G+ cells within
3 the FC were estimated by an unbiased stereological counting technique using ImageJ (Fiji version
4 2.0.0) software. The same unbiased stereological method was used to count both myofibroblasts
5 and neutrophils. A random offset stereological square grid (10,000 cm^2) was superimposed onto
6 the images to provide test points. To calculate area fraction, intersections falling on positively
7 stained cells were counted and expressed as a ratio of total grid intersections within the FC. To
8 estimate the relative volume of myofibroblast cells per FC and assess whether the presence of
9 myofibroblasts was perturbed by actuation, the volume fraction of α SMA+ cells was normalized
10 to a volume defined by FC thickness multiplied by unit area using the following equation: Relative
11 Volume of α SMA+ cells (mm^3) = Volume Fraction of α SMA+ cells / FC Thickness (mm) \times 1 mm
12 \times 1 mm.

13 ***Numerical density of blood vessels per unit area:*** Quantification of vascularization of the FC was
14 performed using a previously reported technique^{5,78,79}. A systemic random sampling strategy was
15 used. From each tissue section, ten non-overlapping images were taken of the FC in the area of
16 interest. Number of blood vessels per area (N_a) were calculated using an unbiased counting frame
17 (grid size = 2000 cm^2). The numbers of points which coincided with blood vessels were counted.
18 The volume fractions of blood vessels were then calculated by expressing the proportion of points
19 hitting blood vessels as a fraction of the total number of points observed in the tissue. The
20 forbidden line rule was followed to ensure blood vessels were only counted once. The application
21 of this counting rule generates an unbiased estimate of the number of blood vessels per unit area
22 ($N_a = C_N \times C_{\text{pts}} \times A$; taking C_N as cumulative number of blood vessels counted and C_{pts} as
23 cumulative number of points in the area of interest).

1 **Analysis of collagen orientation:** Quantification of the collagen content was performed using a
2 previously reported technique^{5,35,80-82}. In ImageJ, six ROIs were manually selected for each image
3 obtained by polarised light microscopy to completely capture all collagen fibres imaged while
4 minimizing noise from background artifact. Using the OrientationJ 2.0.5 plugin, the coherency of
5 each ROI was calculated. Each device sample generated 60 total datapoints (6 ROIs per section,
6 10 sections per sample) (Figure 4k).

7 **Micro-computed tomography analysis.** Following μ CT imaging of implanted devices, FC
8 thickness was quantified using Materialise MIMICS Research 18.0.0.525 and Materialise 3-matic
9 Research 10.0.0.212 software. DICOM files generated from μ CT scans were imported to
10 MIMICS, and a threshold mask applied to the field of view allowing the FC to be visually
11 identified by a change in signal intensity superior to the fascia layer. The threshold region was
12 cropped to include only the region of the FC, and manual segmentation of the FC in the sagittal
13 view was performed. This was repeated on every five DICOM slices, with interpolation between
14 slices performed to mask the FC in the intermediate regions. Upon completion of the masking
15 process, a rectangular section was isolated from the centre of the mask. An STL file of this
16 rectangular section was exported for thickness analysis using 3-matic Research. In 3-matic, the fix
17 wizard was used to fix errors in the STL file and a smoothing factor of 0.8 was applied, following
18 which the model was remeshed using the auto-remesh feature. A wall thickness analysis was
19 generated, and data was exported for statistical analyses.

20 Wall thickness of the isolated FC under both the small and large chambers was measured
21 (Supplementary Figure 10a,b), which demonstrated no difference. Similarly, thickness
22 measurements were taken at device edges which did not differ significantly from thickness values
23 underneath device chambers (Supplementary Figure 10c,d).

1 **Density analysis:** The density of tissue in the fibrotic capsule was measured using the ‘Density in
2 Rectangle’ function in Materialise MIMICS Research 18.0.0.525 on μ CT scans of explanted
3 STAR devices. Three radiodensity measurements of rectangular sections (0.01 mm^2) of fibrotic
4 capsule were measured under each chamber, at five locations across the span of the STAR device.
5 Background radiodensity measurements were obtained from non-sample regions of the scan and
6 were subtracted from the average radiodensity reading for the FC under each chamber.

7 **Scanning electron microscopy.** Following μ CT imaging, devices with associated surrounding
8 tissue were processed for SEM. Tissue samples were first trimmed of excess tissue and hair was
9 removed to prevent unwanted interaction with the electron beam. Samples were first dehydrated
10 through graded alcohols (70%, 95%, 100%, 100%) before being placed into a LEICA EM CPD300
11 critical point dryer for 3 hours. Dried samples were then mounted onto aluminium stubs with
12 carbon adhesive tabs and gold sputter-coated using a Quorum Q150R ES plus. Samples were
13 observed and imaged at 40x magnification 15kV on a HITACHI S-2600N Scanning Electron
14 Microscope.

15 **Statistical analysis.** Unpaired, one-tailed, two-sample *t*-tests were used to assess the differences
16 in blood glucose levels and FC thicknesses between groups. Unpaired, two-tailed, two-sample *t*-
17 tests were used to assess the histological, immunohistochemical, vascularity, and radiodensity
18 differences between the control and IA groups. Before performing *t*-tests, equality of variances
19 was verified between groups using Levene’s test. The analyses were performed in OriginPro 2018b
20 (OriginLab Corp.) and the same software was used to generate all plots. Data are represented as
21 mean \pm standard error of mean. Between-group differences were evaluated at a significance level
22 of 95% ($\alpha = 0.05$). The Bonferroni correction was applied for multiple comparisons by dividing

1 α by m , where m is the number of comparisons. p was deemed significant at $p < \alpha/m$. Exact p -
2 values for all statistical comparisons are presented in Supplementary Note 1.

3

4 **Data Availability**

5 All data supporting the findings of this study are available within the article and the Supplementary
6 Information. Data is available from the corresponding author(s) upon request.

7

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27

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10

11 **Author Contributions**

12 W.W., D.G., S.X.W., G.P.D., E.B.D., and E.T.R. designed the study. W.W., D.G., S.X.W.,
13 N.A.W., R.E.L., R.B., S.T.R., D.S., R.O'C., D.S.M., K.L.M., C.E.V., M.A.H., J.O'D., and A.S.R.
14 performed the experiments. Y.F. performed computational modelling. W.W., D.G., S.X.W., Y.F.,
15 N.A.W., R.E.L., R.B., L.T., D.A.D.-L, R.W., and E.B.D. analysed and reviewed the data. W.W.,
16 D.G., S.X.W., Y.F., N.A.W., R.E.L., S.T.R., G.P.D., E.B.D., and E.T.R. wrote the paper. All
17 authors reviewed and edited the paper.

18

19 **Competing Interests**

20 W.W., S.T.R., K.L.M., C.E.V., G.P.D., E.B.D., and E.T.R. are inventors on a pending patent
21 application related to the device described here. The other authors declare no competing interests.

22 **Figure Captions**

23

24 **Figure 1 | Design of soft transport augmenting reservoir (STAR). a,** Proposed mechanism of
25 STAR: intermittent actuation shields against the foreign body response, creating a favourable

1 environment for the rapid long-term transport of macromolecular drug therapy. **b**, Exploded view
2 showing the different layers comprising STAR. **c**, Deflection of the actuation and porous layers
3 during actuation. **d**, A prototype of STAR showing the deflection of the porous layer during an
4 actuation cycle. Scale bar is 5 mm. **e**, FE model showing peri-implant fluid velocity of convective
5 flow during actuation. **f**, FE model estimating maximum principal tissue strain induced by
6 actuation.

7
8 **Figure 2 | Development of a pre-clinical model to monitor the effect of FBR on therapy**
9 **transport longitudinally.** **a**, Schematic demonstrating detrimental effect of fibrous capsule (FC)
10 formation on therapy delivery with time. **b**, Blood glucose (BG) response to human insulin
11 delivered *via* STAR, measured over 120 min at baseline (day 3), week 2 and week 3. $n = 5$ mice
12 at each time point. **c**, Temporal evolution of the maximum BG % drop (denoting functional effect),
13 calculated from **b**. **d**, Representative 2D μ CT slice of STAR with fibrous encapsulation. Scale bar
14 is 1 mm. **e**, Average FC thickness encapsulating STAR at baseline (day 3), week 2 and week 3
15 following implantation. $n = 3$ mice at baseline and 2 weeks, 5 mice at 3 weeks. Data are means \pm
16 standard error of mean. **f**, Relationship between FC thickness and maximum effect of insulin
17 measured by reduction in blood glucose level. **g**, COMSOL Multiphysics simulations showing
18 spatial drug diffusion through FCs of varying thicknesses. **h**, Temporal evolution of drug release
19 percentage for varying FC thicknesses.

20
21 **Figure 3 | Intermittent actuation (IA) improves long-term macromolecule delivery.** **a**,
22 Preclinical study timeline used to evaluate the effect of IA on insulin transport through a fibrous
23 capsule. **b**, Blood glucose (BG) response to human insulin, measured over 120 min at day 3
24 (baseline), week 3 and week 8. **c**, Maximum BG% change at the 8-week timepoint. **d**, Time to
25 achieve a 30% drop in BG level, measured longitudinally. **e**, Cumulative incidence curves
26 demonstrating probability of achieving a 30% BG drop over 120 min for all groups at both baseline
27 (3 days) and 8-week timepoints. **f**, Area under BG % curve (AUC) denoting overall functional
28 effect mapped over study duration. Statistical comparisons w.r.t. control group. **g**, Change in AUC
29 from 3-week timepoint. Data are means \pm standard error of mean; $*p < 0.05$, $**p < 0.01$,
30 $***p < 0.001$. See Supplementary Note 1 for detailed statistical analyses. † Baseline study was

1 performed post-hoc with separate mice. # Control mice removed from study at intermediate time
2 points due to self-inflicted device damage with subsequent decrease in *n*.

3
4 **Figure 4 | Multiphasic temporal effects of intermittent actuation (IA).** **a**, Timeline of
5 multiphasic cellular and fibrous capsule (FC) changes induced by IA. **b**, Representative fluorescent
6 images of the FC stained with Ly-6G+ (green) and DAPI (blue). Scale bars are 20 μm . **c**,
7 Quantification of neutrophils present within FC +/- IA at day 3 and 5. **d**, Representative fluorescent
8 images of the FC stained with α -SMA (green) and CD31 (red). Scale bars are 50 μm . **e**,
9 Quantification of myofibroblasts present within FC +/- IA at 2 weeks. **f**, Representative histologic
10 images of the FC stained with haematoxylin and eosin. Scale bars are 20 μm . **g**, Quantification of
11 total cells/capsular area +/- IA at day 3, day 5, and 2 weeks. **h**, Representative topographical
12 reconstructions of μCT images showing the differences in FC thickness +/- IA at 2 weeks. **i**,
13 Average FC thickness of the control and actuated groups at day 3, day 5, 2 weeks, and 8 weeks
14 with two measurements taken per animal. **j**, Representative polarised light microscopy images of
15 the FC obtained after picrosirius red staining at 8 weeks. Scale bars are 100 μm . **k**, Quantification
16 of the FC collagen fibre orientation by optical coherency with 60 ROIs per animal. **l**,
17 Representative SEM images demonstrating reduced cellular invasion with actuation at the 8-week
18 timepoint. Scale bars are 500 μm . *n* = 2–6 animals per group; data are means \pm standard error of
19 mean; **p* < 0.05, ***p* < 0.01, ****p* < 0.001, *****p* < 0.0001. See Supplementary Note 1 for
20 detailed statistical analyses.

21
22 **Figure 5 | On demand, actuation-mediated rapid release (RR) of drug using STAR.** **a**, RR
23 enables convective flow of a model drug, methylene blue, from the therapy reservoir of STAR.
24 Scale bar is 5 mm. **b**, Photoacoustic images showing subcutaneously implanted STAR in a rat
25 model: RR enables convective flow of drug analogue (red) from the therapy reservoir into the
26 surrounding tissue pocket. **c**, Schematic showing actuation-mediated RR overcoming fibrous
27 encapsulation. **d**, Snapshots from COMSOL Multiphysics simulation showing the rate-limiting
28 diffusion barrier created by a FC and the ability to improve transport using RR. **e**, Concentration
29 of drug outside the FC comparing passive diffusion alone to RR at 200 seconds. **f**, Concentration
30 of drug outside the FC for a thin (100 μm) or thick (200 μm) FC with 1 or 5 RR actuation cycles.
31 **g**, *In vivo* images of rat model with two STAR devices implanted. Fluorescence shows the

1 distribution of drug analogue Genhance 750. Red arrow indicates the device after undergoing RR
2 actuation. **h**, Temporal evolution of the drug diffusion area of Genhance 750 in passive (control)
3 and RR actuated STAR, quantified by fluorescent IVIS imaging. **i**, Blood glucose response to
4 insulin in control (passive diffusion only) and in RR actuated (at $t = 150$ min) STAR devices, at 2
5 weeks following implantation. $n = 4$ animals per group; data represents means \pm standard error of
6 mean. p -value calculated from paired one-tailed t -test.

7 **Figure 6 | Minimally invasive surgical implantation in a human cadaver model.** **a**, Location
8 of the *transversus abdominis* plane in the anterior abdominal wall. **b**, Ultrasound guided needle
9 access to the desired tissue intermuscular plane in the anterior abdominal wall and hydro-dissection
10 to generate a potential space (EOM: External Oblique Muscle, IOM: Internal Oblique Muscle,
11 TAM: Transversus Abdominus Muscle). **c**, Seldinger technique was used to get needle access to
12 the *transversus abdominis* plane and a 5 Fr sheath was exchanged over a guide wire to maintain
13 durable access to the tissue plane. **d**, A commercially available dilator set is used to expand the
14 space to accommodate positioning of the deployment sheath. **e**, STAR advancement through
15 sheath into tissue space. **f**, **g**, An echogenic contrast agent was used to inflate the deployment
16 channel to ensure complete unfolding of the STAR device within the plane using ultrasound
17 guidance.