Mechanosensitive Osteogenesis on Native Cellulose Scaffolds for Bone Tissue Engineering.

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Abstract

In recent years, plant-derived cellulosic biomaterials have become a popular way to create scaffolds for a variety of tissue engineering applications. Moreover, such scaffolds possess similar physical properties (porosity, stiffness) that resemble bone tissues and have been explored as potential biomaterials for tissue engineering applications. Here, plant-derived cellulose scaffolds were seeded with MC3T3-E1 pre-osteoblast cells. Moreover, to assess the potential these biomaterials we also applied cyclic hydrostatic pressure (HP) to the cells and scaffolds over time to mimic a bone-like environment more closely. After one week of proliferation, cell-seeded scaffolds were exposed to HP up to 270 KPa at a frequency of 1Hz, once per day, for up to two weeks. Scaffolds were incubated in osteogenic inducing media (OM) or regular culture media (CM). The effect of cyclic HP combined with OM on cell-seeded scaffolds resulted in an increase of differentiated cells. This corresponded with an upregulation of alkaline phosphatase activity and scaffold mineralization. The results reveal that in vitro, the mechanosensitive pathways which regulate osteogenesis appear to be functional on novel plant-derived cellulosic biomaterials.

Introduction

Upon injury or break, bone has the ability to self-renew. However, large defects created by either injury or disease may require graft placement to avoid non-union or malunion of the bone tissue (Andrzejowski and Giannoudis, 2019). Such grafts can be derived directly from the patient (autologous grafts), and is considered the "gold standard" in regenerative orthopedics (Campana et al., 2014; Parikh, 2002; Sakkas et al., 2017; Wang and Yeung, 2017). However, limited size grafts, donor site morbidity and infections, cost and post-operative pain at both donor and receiver site has led to the development of alternative approaches (Parikh, 2002; Wang and Yeung, 2017): cadaver donors (allograft), animal sources (xenograft), or artificially derived (alloplastic). Such alternatives all have their own benefits and drawbacks, the later however provides a potential alternative with lower risk of transmitted diseases and infections , as well as overcoming the size limitation barrier (Parikh, 2002; Wang and Yeung, 2017) .

Alloplastic grafts are also considered a more ethical alternative to allografts and xenografts (Fernández et al., 2015). Physical properties are key parameters for graft development, such as pore size, interconnectivity and elasticity (Campana et al., 2014; Gao et al., 2019; Nukavarapu et al., 2015). Fine tuning of these parameters leads to better mechanical support and stability of the implant.

Long bones are highly dynamic structural tissues. A whole spectrum of forces are acting on different areas of this part of the skeletal system. For instance, the pressure found in the femur head in human adults can reach 5 MPa during normal locomotion, and can reach up to 18 MPa for other activities (Morrell et al., 2005). On a microscopic level, these forces are transmitted to the osteocytes through Wnt/β-catenin mechano-sensing pathways in the lacuna-canaliculi network (Bonewald and Johnson, 2008). Force-regulated mechanisms lead to formation and removal of bone tissue through bone remodeling processes (Bonewald and Johnson, 2008) and the pressure inside the lacuna-canaliculi network is around 280 kPa (Zhang et al., 1998). Bioreactors have also been developed to apply stresses to osteoblast cells to replicate the native bone environment via uniaxial compression/tension, biaxial compression/tension, shear-stress, etc (Brunelli et al., 2019; Pörtner et al., 2005). Hydrostatic pressure (HP) stimulation on cultured cells has also been achieved by compressing the gas phase above an incompressible media (Gardinier et al., 2009; Henstock et al., 2013; Liu et al., 2010; Reinwald et al., 2015; Reinwald and El Haj, 2018; Stavenschi et al., 2018; Zhao et al., 2015). Three-dimensional (3D) culturing of the cells is also critical for better representing *in vivo* conditions. With a specific tissue-oriented scaffold structure and appropriate applied mechanical stimuli one can potentially predict the performance of a biomaterial scaffold prior to *in vivo* animal studies.

A variety of 3D biomaterials have been utilized support the growth and proliferation of cells in environments that mimic bone tissues. These include hydroxyapatite, tricalcium phosphate, bioceramics or bioactive glass (Wang and Yeung, 2017). These materials are osteoconductive, can promote osteointegration and provide structural support at the implant site (Wang and Yeung, 2017). However, they show poor osteoconduction and little osteogenesic response (Wang and Yeung, 2017). Polymer biomaterials such as poly(glycolic acid) (PGA), poly(lactic acid) (PLA) and poly(ε-caprolactone) (PCL) are also biocompatible, possess tunable degradation rates and can be chemically modify to change the surface chemistry (Wei et al., 2020). However, *in vivo* degradation creates acidic bi-products which can lead to an inflammatory response and decrease the efficiency bone repair (Wei et al., 2020). Finally, naturally occurring polymers such as collagen, chitosan and silk are also common (Wei et al., 2020). However, due to their inherent mechanical properties and structural stability, these materials are often utilized as coipmosites with additional polymers and coatings in BTE applications (Di Martino et al., 2005; Lee and Volpicelli, 2017; Wei et al., 2020) . More recently, plant-derived decellularized cellulose scaffolds have been shown to be effective in BTE applications (Hickey and Pelling, 2019; Lee et al., 2019; Torgbo and Sukyai, 2018).

Previous studies by our group and others have shown that cellulose based scaffolds derived from plants can be used as tissue engineering scaffolds (Gershlak et al., 2017; Hickey et al., 2018; Lee et al., 2019; Modulevsky et al., 2016, 2014). These biomaterials are often sourced from plants with a microstructure that closely mimics the tissue to be replicated (Hickey et al., 2018). Successful experiments *in vitro* and *in vivo* showed that these biomaterials are biocompatible and support angiogenesis (Hickey et al., 2018; Modulevsky et al., 2016, 2014). Due to their pore size interconnectivity, other groups have successfully differentiated human pluripotent stem cells into bone-like tissues within scaffolds derived from either decellularized mushrooms (Balasundari et al., 2012) or apple tissues (Lee et al., 2019). The in vivo performance of the apple-derived scaffolds were further examined by implanting disk-shaped

98   scaffolds in rat cranial defects (Lee et al., 2019). The findings demonstrate partial bone
99   regeneration within the implant, type 1 collagen deposition and blood vessel formation (Lee et
100 al., 2019). However, to date, the mechanobiology of cells cultured on plant-derived scaffolds
101 has not been examined. It remains poorly understood how mechanical signal transduction
102 pathways are, or are not, impacted when cultured on unconventional plant-derived cellulosic
103 biomaterials. Here, to further examine the potential role plant-based biomaterials can play in
104 BTE applications, we explore how mechanical stimulation impacts the differentiation of pre-
105 osteoblasts when cultured on plant cellulose scaffolds. In a custom-built bioreactor, we apply
106 cyclic HP stimulation to differentiating osteoblasts and examine the regulation of key markers of
107 osteogenesis and mineralization. The results reveal that application of HP, in combination with
108 osteogenic inducing media, leads to enhanced differentiation. This work provides further
109 evidence that plant-derived cellulose scaffolds support osteogenesis and have potential
110 applications in BTE.
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Materials and Methods

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Scaffold fabrication

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117 Samples were prepared following established protocols (Hickey et al., 2018; Modulevsky et al.,
118 2016, 2014). MacIntosh apples (Canada Fancy) were cut with a mandolin slicer to 1 mm-thick
119 slices. A biopsy punch was used to create 5 mm-diameter disks in the hypanthium tissue of the
120 apple slices. The disks were decellularized in a 0.1% sodium dodecyl sulfate solution (SDS,
121 Fisher Scientific, Fair Lawn, NJ) for two days. Then, the decellularized disks were washed in
122 deionized water, before incubation in 100 mM $CaCl_2$ for two days. The samples were sterilized
123 with 70% ethanol, washed in deionized water, and placed in a 96-well culture plate prior to cell
124 seeding.
125
126 MC3T3-E1 Subclone 4 cells (ATCC® CRL-2593™, Manassas, VA) were cultured and
127 maintained in a humidified atmosphere of 95% air and 5% CO_2, at 37°C. The cells were cultured
128 in Minimum Essential Medium (ThermoFisher, Waltham, MA), supplemented with 10% Fetal
129 Bovine Serum (Hyclone Laboratories Inc., Logan, UT) and 1% Penicillin/Streptomycin (Hyclone
130 Laboratories Inc). Scaffolds were placed in 96-well plates. Prior to cell seeding, scaffolds were
131 immersed in culture media and incubated in a humidified atmosphere of 95% air and 5% CO_2, at
132 37°C, for 30 min. The culture media was completely aspirated from the wells. Cells were
133 suspended and a 30 µL drop containing 5×10^4 cells, was pipetted on each scaffold. The cells
134 were left to adhere on the scaffolds for 2 hours before adding 200 µL of culture media to the
135 culture wells. Culture media was then changed every 3-4 days for 1 week. Cell seeded scaffolds
136 were then either incubated in osteogenic media (OM) by adding 50 µg/mL of ascorbic acid and
137 10 mM β-glycerophosphate to the culture media or incubated in culture media (CM) for 2 weeks,
138 with or without the application of HP.
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Cyclic hydrostatic pressure stimulation

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142 Cyclic HP was applied by modulating the pressure in the gas phase above the culture wells in a
143 custom-build pressure chamber (Figure 1, A). The humidified, 95% air and 5% CO_2 incubator
144 atmosphere was compressed using an air compressor and injected in the chamber. A Particle
145 Photon microcontroller (Particle Industries, San Francisco, CA) was used to control the
146 frequency of the applied pressure remotely via a custom-made cellphone application through
147 the Blynk IoT platform (Blynk, New York, NY)). Cyclic HP was applied 1 hour per day, for up to 2

148 weeks (Figure 1, B) at a frequency 1Hz, oscillating between 0 and 280 kPa with respect to
149 ambient pressure. Pressure was monitored using a pressure transducer. The samples were
150 removed from the pressure chamber after each cycle and kept at ambient pressure between the
151 stimulation phases.

153 Cell-seeded scaffolds were either stimulated with cyclic HP with and without the presence of
154 OM, leading to four experimental conditions (Figure 1, B): Cyclic HP in regular culture media
155 (CM-HP), cyclic HP in osteogenic culture media (OM-HP), non-stimulated in osteogenic media
156 (OM-CTRL) and non-stimulated in regular culture media (CM-CTRL). The OM-CTRL and CM-
157 CTRL conditions were in a humidified, 5% CO2 incubator at 37°C.

159 **Scaffold imaging**

161 After 1 week or 2 weeks, scaffolds were washed with PBS and fixed with 10% neutral buffered
162 formalin for 10 min. Scaffolds were washed with PBS and incubated in a 0.01% Congo Red
163 staining solution (Sigma-Aldrich, St. Louis, MO) for 20 min at room temperature. Scaffolds were
164 washed with PBS. Cell nuclei were stained with 1:1000 Hoechst (ThermoFisher, Waltham, MA)
165 for 30 min in the dark. Samples were washed with PBS and stored in wash buffer solution (5%
166 FBS in PBS) prior to imaging. The cell-seeded surface of the scaffolds was imaged with a laser
167 scanning confocal microscope (Nikon Ti-E A1-R) equipped with a 10X objective. Maximum
168 intensity projections were used for cell counting with ImageJ software (Schindelin et al., 2012).
169 Cells were counted on a 1.3 by 1.3 mm^2 area (N=3 per experimental conditions with 3 randomly
170 selected area per scaffold).

172 **Alkaline phosphatase activity assay**

174 Alkaline phosphatase (ALP) activity in media was measured using an ALP assay kit (BioAssay
175 Systems, Hayward, CA). Working solution was prepared with 5 mM magnesium acetate and 10
176 mM p-nitrophenyl phosphate (pNPP) in assay buffer, following manufacturer's protocol. 150 μL
177 of working solution was pipetted in 96-well plate. 200 μL of calibrator solution and 200 μL of
178 dH$_2$O were pipetted in separated well, in the same 96-well plate. At 1 week and 2 weeks, 20 μL
179 of incubation media was pipetted into the working solution's well. All wells were read at 405 nm
180 for 10 minutes, every 30 seconds. ALP activity was calculated by taking the slope of the 405 nm
181 readings vs time, calibrated with the calibrator solution and dH$_2$O. Wells were read in triplicates
182 (N=3 per experimental conditions).

184 **Alizarin red S staining and mineral deposit quantification**

186 Samples were fixed with 10% neutral buffered formalin for 10 min, after 1 week or 2 weeks.
187 Calcium quantification was performed using previously published protocol (Gregory et al.,
188 2004). Samples were transferred to a 24-well plate and carefully washed with deionized water
189 and incubated in 1 mL of 40mM (pH=4.1) alizarin red s (ARS, Sigma-Aldrich) solution for 20
190 minutes at room temperature, with light agitation. The samples were washed with deionized
191 water and placed in 15 mL tubes filled with 10 mL dH$_2$O. The tubes were placed on a rotary
192 shaker at 120 rpm for 60 min and dH$_2$O was replaced every 15 min. Thereafter, samples were
193 incubated in 800 μL of 10% acetic acid on an orbital shaker at 60 rpm for 30 min. The eluted
194 ARS/acetic acid solution was pipettes out of the well and transferred to 1.5 mL centrifuge tubes.
195 Tubes were centrifuged at 17 x 10^4 g for 15 min. 500 μL of supernatants were transferred to
196 new centrifuge tube and 200 μL of 10% ammonium hydroxide was pipetted into the tubes.
197 Finally, 150 μL of the solution was pipetted into a 96-well plate and the absorption at 405nm

198 was read using a plate reader (BioTek Instruments, Winooski, VT). Wells were read in triplicates
199 (N=3 per experimental conditions).
200
201 **Statistical analysis**
202
203 Values reported in this manuscript are the average value ± standard error of the mean (SEM).
204 Statistical significance was determined using one-way ANOVA and post hoc Tukey test. A value
205 of $p < 0.05$ was considered to be statistically significant.
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207
208 # Results
209 **Scaffold imaging and cell counting**
210
211 Mechanical stimulation of the cultured scaffolds was carried as described in the Methods
212 (Figure 1). The application of HP significantly increases the density of cells (Figure 2) after 1
213 week in OM compared to the static condition ($p=10^{-5}$), but the increase was not significant after
214 2 weeks (p=0.07). Conversely, in CM a non-significant increase in the density of cells was
215 observed with applied HP after 1 week (p=0.21) and 2 weeks (p=0.92). Importantly, we also
216 observed a significant increase when incubated in OM compared to CM after 1 week of HP
217 stimulation (p=0.02). After 2 weeks of HP stimulation, samples cultured in OM exhibited a
218 similar density to samples cultured in CM (p=0.23). The results indicate that cell density
219 increases more rapidly in the first week of stimulation in OM compared to CM media but that by
220 two weeks the cell densities become equal. No significant difference was observed in the static
221 cases after 1 or 2 weeks (p=0.99 in both cases).
222
223 **Alkaline phosphatase activity assay**
224
225 The stimulation with cyclic HP significantly increased the ALP activity (Figure 3) in scaffolds
226 incubated in OM after 1 and 2 weeks compared to static condition ($p=4\times10^{-8}$ in both cases). A
227 similar effect was observed in CM after 1 and 2 weeks (p=0.03 and $p=5\times10^{-8}$ respectively).
228 However, the incubation in OM significantly increased ALP activity when HP is applied
229 compared to incubation in CM, after 1 week ($p<10^{-8}$) but was not significantly different after 2
230 weeks (p=0.99). Consistent with the cell density data the HP-driven increases in ALP activity
231 are only observed during the first week and equalize by the second week of culture. In the
232 absence of HP, the choice of incubation media did not significantly change the ALP activity after
233 1 or 2 weeks (p=0.25 and p=0.08 respectively).
234
235 **Alizarin red S staining and mineral deposit quantification**
236
237 The application of cyclic HP significantly increased mineral deposition (Figure 4) for samples
238 incubated in OM compared to static condition after 1 week and 2 weeks ($p=2\times10^{-7}$ and $p=2\times10^{-8}$
239 respectively. Similarly in samples cultured in CM, cyclic HP significantly increased mineral
240 deposition after 1 week and 2 weeks ($p=1\times10^{-6}$ and $p=2\times10^{-8}$ respectively). Moreover, the
241 incubation in OM significantly increased mineral deposition when HP is applied compared to
242 incubation in CM, after 1 week ($p=2\times10^{-4}$), but was not significant after 2 weeks (p=0.99). These
243 results are again consistent with the findings from assays of cell density and ALP activity. Under
244 static conditions mineralization still occurred in OM as expected and was significantly increased
245 compared to CM after 1 week ($p=10^{-3}$), but was not significant after 2 weeks (p=0.75).
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Discussion

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253 Cells utilize a variety of mechanisms to sense and respond to a variety of mechanical stimuli
254 (Vining and Mooney, 2017). Mechanical stimuli are known to affect cell differentiation, tissue
255 regeneration, cytokines and protein expression and proliferation (Martino et al., 2018; Vining
256 and Mooney, 2017). Thus, adequate representation of physical environment is critical in implant
257 design (Campana et al., 2014; Gao et al., 2019; Nukavarapu et al., 2015; Parikh, 2002; Wang
258 and Yeung, 2017). Bones are subjected to constant mechanical stresses and adapt through
259 remodeling process (Bonewald and Johnson, 2008). *In vivo*, HP stimulates bone cells and
260 impacts cell differentiation, maker expression and mineralisation (Henstock et al., 2013; Huang
261 and Ogawa, 2012; Reinwald and El Haj, 2018). Plant-derived cellulose scaffolds are an
262 emerging biomaterial in BTE (Lee et al., 2019), and therefore it is of interest to understand their
263 performance under the mechanical conditions that are found *in vivo* (Gao et al., 2019;
264 Nukavarapu et al., 2015). To our knowledge, cyclic HP effect of pre-osteoblasts seeded on
265 plant-derived scaffolds have not yet been studied. Cellulose biomaterials derived from plant
266 tissues have shown promising results *in vitro* and *in vivo* for targeted tissue engineering (Hickey
267 et al., 2018; Modulevsky et al., 2016, 2014) and have been used to host osteoblastic
268 differentiation (Lee et al., 2019). In addition, apple-derived cellulose scaffolds exhibit similar
269 morphological characteristics to trabecular bone and were previously used for BTE applications
270 (Lee et al., 2019).

271

272 In this study, we replicated the mechanical stimuli present during human locomotion by
273 measuring the impact of differentiation markers in cellulose scaffolds seeded with pre-osteoblast
274 cells. External pressure was applied on the scaffolds in similar magnitude of the lacuna-
275 canaliculi network with a frequency mimicking human locomotion (Henstock et al., 2013; Zhang
276 et al., 1998). After proliferation, our scaffolds were either cultured in standard culture media, or
277 in osteogenic-inducing differentiation media, with or without high pressure stimulation. Other
278 groups utilizing similar cell lines (Gardinier et al., 2014, 2009), bone marrow skeletal stem cells
279 (Reinwald and El Haj, 2018; Stavenschi et al., 2018; Zhao et al., 2015) or ex-vivo chick femur
280 (Henstock et al., 2013) have also studied the effects if cyclic HP on either 2D surfaces,
281 biomaterial meshes or ex vivo bones. In general, our results show a sensitivity ot HP during the
282 early differentiation of the cells during the first week of stimulation. This sensitivity largely
283 subsides during the second week of culture.

284

285 Cell counting revealed that the application of HP enhances MC3T3-E1 proliferation when
286 cultured in OM or CM. Consistent with our work, are the results from previous studies in which
287 metabolic activity has also been shown to be upregulated by mechanical stimulation in
288 comparison to non-stimulated samples (Reinwald and El Haj, 2018). Moreover, it was shown
289 that the application of HP accelerates cell proliferation through upregulated cell cycle initiation
290 (Zhao et al., 2015). Similarly, reports have shown that physical stimulation of MC3T3-E1 cells
291 induced expression of paracrine factors that leads enhancement of cell proliferation (Stavenschi
292 et al., 2018). Importantly, when cultured in OM, cell density increases more rapidly with HP
293 compared to CM during the first week of culture, but the cell densities equalize by the end of the
294 second week. This data is consistent with previous reports of a time-dependent increase in cell
295 number when cultured in OM (Quarles et al., 1992). Similarly, no significant difference between
296 incubation of MC3T3-E1 cells in similar OM after 2 weeks (Hong et al., 2010). Our findings

corroborate these studies and further suggest that the application of HP influences the replication rate at early stages of stimulation for samples cultured in OM.

Finally, ALP is an important enzyme expressed in the early stages of osteoblastic differentiation (Golub and Boesze-Battaglia, 2007). Our results indicate that the application of cyclic HP significantly increases ALP activity, compared to the static case. These findings are consistent with other studies on more conventional scaffolds (Reinwald and El Haj, 2018). For example, a significant increase in ALP activity was also reported after the incubation of scaffolds in osteogenic-inducing differentiation media, similarly to reports on 2D culture systems (Hong et al., 2010; Quarles et al., 1992). The application of HP significantly increased the mineral content in the scaffolds after 1 week and 2 weeks of stimulation, in both types of incubation media. Other groups have shown that a cyclic 300 kPa pressure at 2 Hz frequency on human BMSCs promoted significant mineral deposition (Stavenschi et al., 2018). The increase in mineral deposition also noted in ex vivo bone samples, with similar HP force application (Henstock et al., 2013). Furthermore, the incubation in OM increased the mineral content in the scaffolds, which is consistent with other studies (Hong et al., 2010; Quarles et al., 1992). Along with ALP expression, mineral content expression further confirms the ongoing differentiation of MC3T3-E1 onto osteoblast, either by applied HP, chemically (induction in OM) or a combination of both.

Conclusion

Plant-derived scaffolds have demonstrated their efficiency for tissue engineering applications (Fontana et al., 2017; Gershlak et al., 2017; Hickey et al., 2018; Lee et al., 2019; Modulevsky et al., 2016, 2014). The novelty, widespread availability, ease of use makes an interesting alternative to autografts, xenografts and synthetics implants. However, it has remined poorly understood how mechanosensitive pathways in bone precursor cells are impacted by being cultured in plant-derived cellulosic biomaterials. Previous studies have not included these phenomena. In this study, mechanical stimulation of cell-seeded scaffolds and the subsequent cellular response allowed us to understand the implants mechanosensitive behaviors in these novel scaffolds. As bone tissues are highly dynamic environments, relevant forces should be considered in the implant design. Importantly, this work provides the early evidence that bone-precursor cells appear to possess intact mechanosensing and mechanotransduction pathways when cultured on plant-derived scaffolds in a mechanically active environment. The results obtained here are consistent with data on traditional scaffold choices. The results reveal that application of cyclic HP, in combination with OM, leads to an increase in the number of cells, ALP activity and mineralization over time. These results combined with past *in vitro* and *in vivo* studies using apple-derived scaffold biomaterials demonstrate their potential in BTE applications.

Acknowledgements

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FIGURE LEGENDS

Figure 1: (A) Cyclic hydrostatic pressure device schematics. Hydrostatic pressure was applied by modulating the pressure in the gas phase above the culture wells in a custom-build pressure chamber. Air from incubator atmosphere was compressed using a compressor and injected in the pressure chamber using solenoid valves. (B) Experimental conditions. After 1 week of proliferation, cyclic hydrostatic pressure stimulation was applied during 1 hour per day, for up to 2 weeks at a frequency 1Hz, oscillating between 0 and 280 kPa with respect to ambient pressure. The samples were removed from the pressure chamber after each cycle and kept at ambient pressure between the stimulation phases.

Figure 2: (A) Representative confocal laser scanning microscope image showing seeded cells scaffolds (scale bar = 100 µm – applies to all). The scaffolds were stained for cellulose (red) and for cell nuclei (blue). (B) Cellular density after 1 week or 2 weeks of stimulation. Statistical significance (* indicates $p<0.05$) was determined using a one-way ANOVA and Tukey post-hoc tests. Data are presented as means ± S.E.M. of three replicate samples per condition, with three areas per sample.

Figure 3: Alkaline phosphatase (ALP) activity after 1 week or 2 weeks of stimulation. Statistical significance (* indicates $p<0.05$) was determined using a one-way ANOVA and Tukey post-hoc tests. Data are presented as means ± S.E.M. of three replicate samples per condition.

Figure 4: Mineral deposit quantification with Alizarin Red S (ARS) staining after 1 week or 2 weeks of stimulation. Statistical significance (* indicates $p<0.05$) was determined using a one-way ANOVA and Tukey post-hoc tests. Data are presented as means ± S.E.M. of three replicate samples per condition.
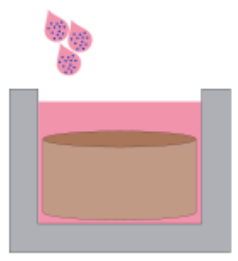
Figure 1

A

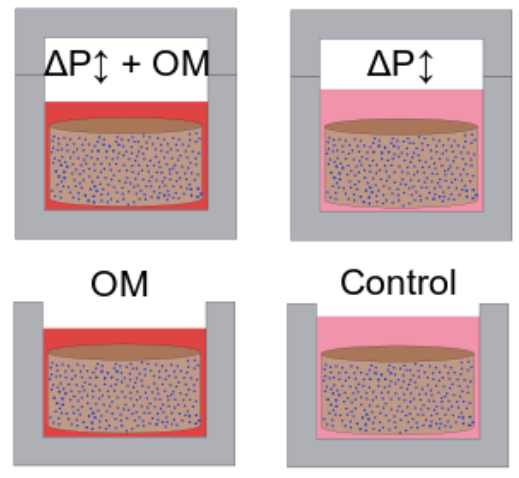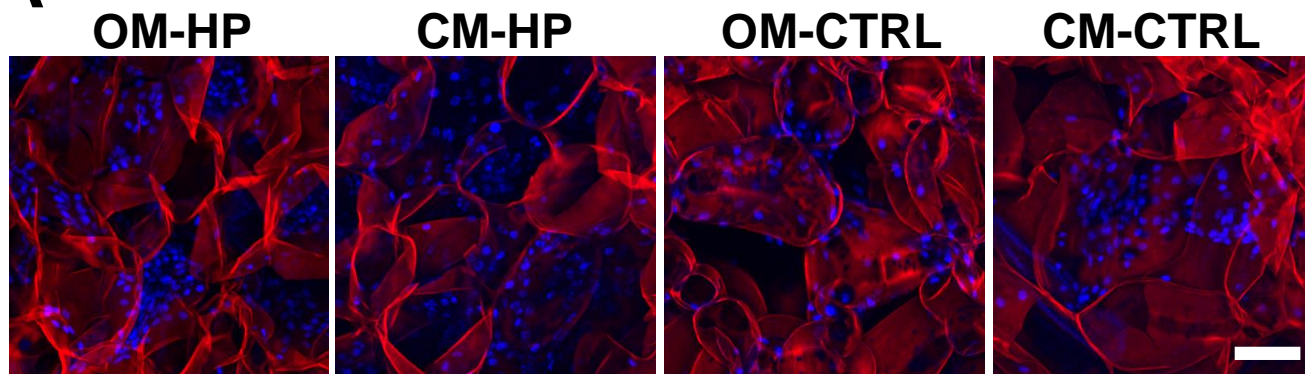
B

Proliferation

Stimulation

$\Delta P\updownarrow$ + OM

$\Delta P\updownarrow$

OM

Control
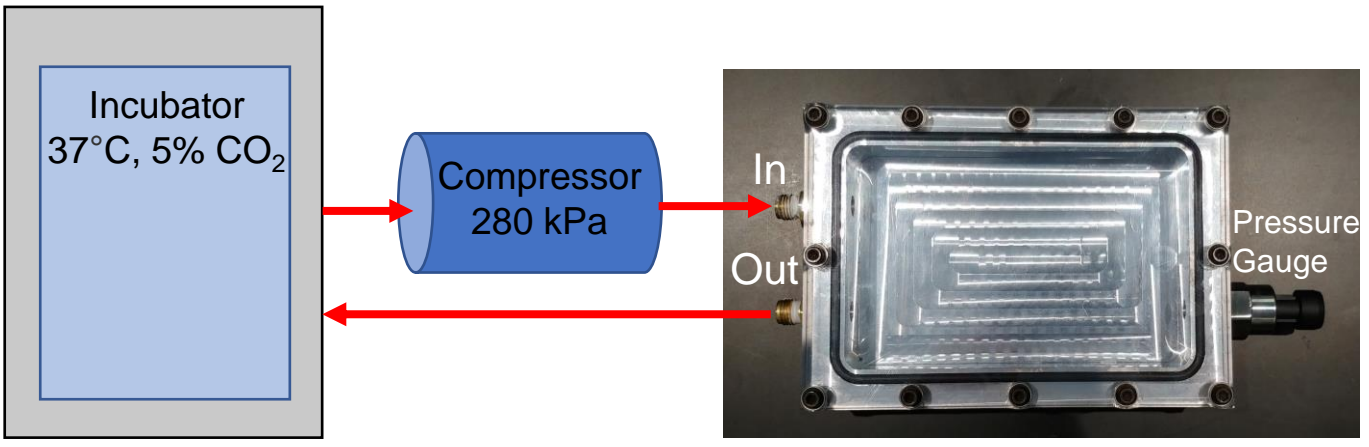
= Culture media
= Osteogenic media

Figure 2

A

OM-HP   CM-HP   OM-CTRL   CM-CTRL

B

Figure 3

Figure 4