ABSTRACT

Title of Thesis: **STRONG AND STRETCHABLE GELS FOR ELECTROADHESION**
Uma Joy Kokilepersaud, Master of Science, 2022

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Electroadhesion (EA) was first shown between cationic and anionic gels: when the two are contacted under an electric field (10 V DC, for ~ 20 s), they adhere strongly and remain adhered after the field is removed. Recently, our lab demonstrated that EA can also be induced between a cationic gel and animal tissues. This suggests the intriguing possibility of using EA in surgery: i.e., for gels to be electroadhered over tears in tissues. However, current gels fail to properly seal tissue-tears because either they are not strong enough (they break when stretched over the torn tissue) or do not adhere well enough to the tissue. This project aims to create a cationic gel that is strong (i.e., has high elongation-at-break ε and tensile strength TS) and exhibits strong adhesion to tissue by EA (i.e., has high pull-off strength). Our base gel, made by polymerizing an acrylamide monomer (termed 'QDM'), is fortified by a second polymer. The optimal second polymer is shown to be gelatin, which forms physical gels upon cooling from a hot solution. QDM/gelatin gels are interpenetrating networks (IPNs) and exhibit ~ 20-30× higher ε and TS compared to QDM alone. We proceed to test if our strong gels can seal tears in chicken intestines. A tear in the tube is patched by the QDM/gelatin gel using EA, and water is then sent through the patched tube. Even for large tears (~ 3-4 mm width), the patched tube allows water to flow at a pressure exceeding 80 mm Hg (normal fluid pressure in the body). This suggests that EA could indeed be implemented in surgery using our strong QDM/gelatin gels.

STRONG AND STRETCHABLE GELS FOR ELECTROADHESION

By

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Thesis submitted to the Faculty of the Graduate School of the
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Dedication

This thesis is dedicated to my family and friends. Without their support I would not have made it this far.

To Mom, Dad, Mr. Tims and Kiran thank you for always encouraging me even when I wanted to give up. Without your support and knowledge none of this would be possible so thank you!

To Christian and Jen thank you for your help through so much tiring engineering coursework, your friendship means the world to me.

To Emily and Haleh thank you for being by my side for so many years at UMD, it has been difficult, but we made it! Thank you so much for all your guidance and friendship!

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Thank you to all my labmates past and present for helping me throughout the years: Hema, Nikhil, Faraz, Wenhao, Medha, Christine, Sam, Morine, all of your insight and friendship has been unmatched, thank you so much!

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Chapter 1: INTRODUCTION & OVERVIEW

The phenomenon of electroadhesion (EA) was first reported in gel-to-gel systems approximately a decade ago.[1,2] When a cationic and an anionic gel are brought into contact under an electric field (DC voltage of 10 V, applied for 10 to 20 s), they adhere strongly and persistently even after the field is removed (Figure 1.1A, 1.1B). If the adhered gels are then placed in the same DC field with a reversed polarity, the adhesion is reversed and the two gels can be detached (Figure 1.1C).

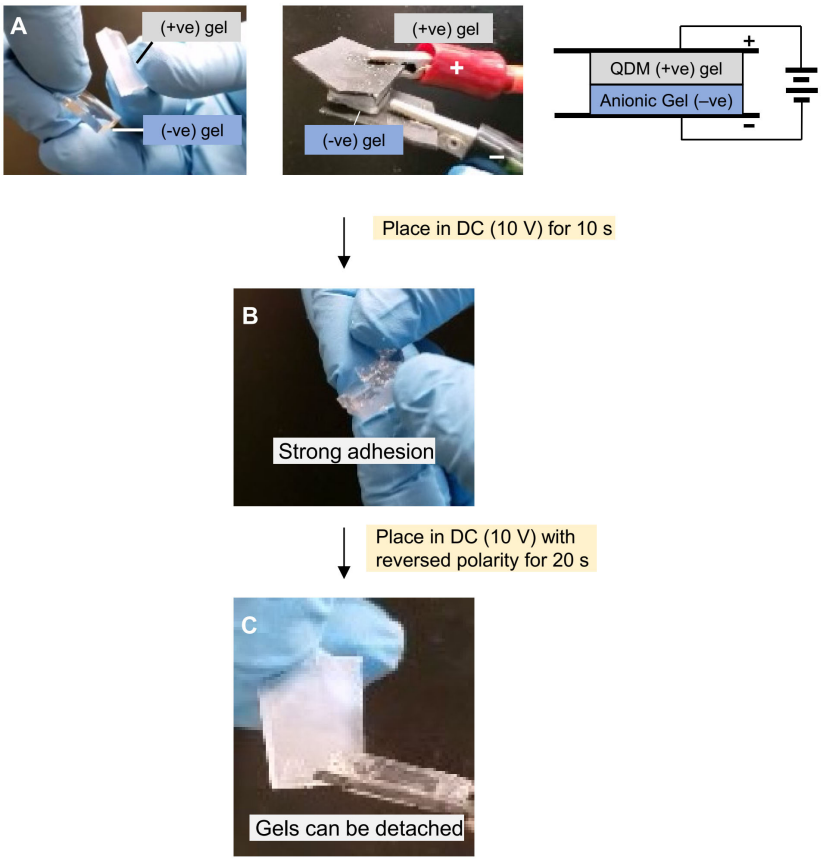

Figure 1.1. Electroadhesion of a cationic gel to an anionic gel. A DC voltage of 10 V applied for 10 s (A) causes adhesion (B). Reversing the polarity (C) reverses the adhesion.

The mechanism for EA is expected to involve electrophoresis of cationic and anionic chain segments across the gel-gel interface.[3-6] Note that electrophoresis refers to the movement of charged molecules or particles under an electric field. Most interestingly, the EA phenomenon goes beyond gel-to-gel systems. Recent studies from our lab have been the first to report EA between a cationic gel and animal tissue (Figure 1.2).[7]

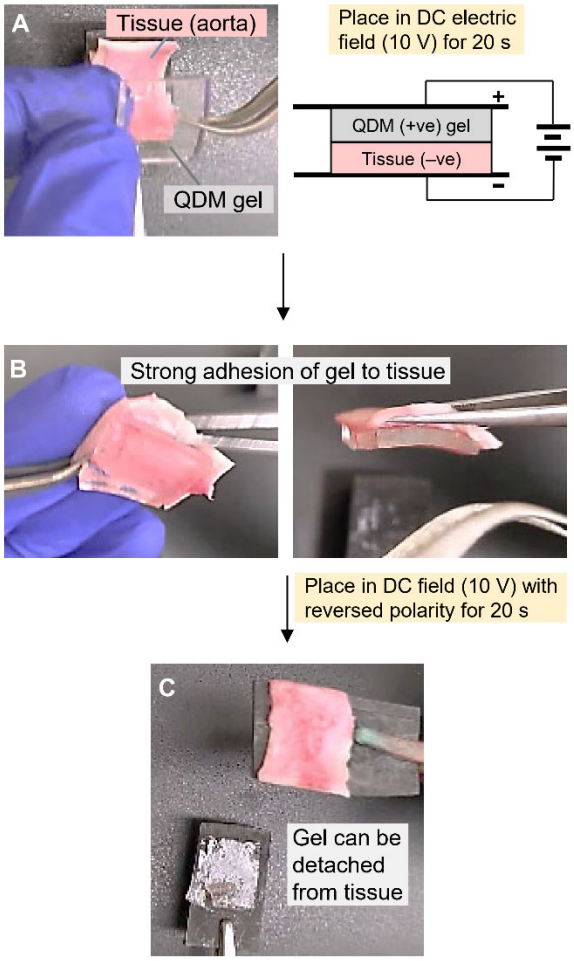

Figure 1.2. Electroadhesion of a cationic gel to animal tissue. (A) A cationic gel (QDM) is contacted with bovine aorta under a DC voltage of 10 V for 20 s, resulting in strong adhesion (B). Reversing the polarity (C) reverses the adhesion.

Figure 1.2 shows EA of a cationic gel (denoted by QDM) to a piece of bovine aorta. Similar EA was found to many bovine tissues. The QDM gel was made by free-radical polymerization of acrylamide monomers. Note that only cationic gels can be electroadhered, which implies that the tissues have anionic character. Gel-tissue EA presumably occurs by the same mechanism as gel-gel EA, i.e., by electrophoresis.

One obvious application of this novel gel-tissue adhesion is in surgery: i.e., the use of an electroadhered gel as a patch to seal cuts or tears in tissue. Currently, surgical repairs of torn tissue are done by suturing the torn edges together until the wound heals naturally.[8] However, using sutures to close a wound can be cumbersome and requires considerable skill on the part of a surgeon.[9,10] Thus, we would like to explore the use of EA as an alternative to suturing.

In this regard, we have performed pilot studies with chicken intestines as a prototype for a tubular tissue. Figure 1.3 shows the schematic of such an experiment. We take the intestinal tube and make a cut in the wall. Then we adhere a gel by EA over the cut. Thereafter, we flow water through the patched tube and measure the burst pressure at which the gel-patch fails. These experiments indicate that our current gels have some deficiencies. First, the typical QDM gel we use in EA experiments is rather brittle. Thus, it sometimes cracks even as it is bent and stretched around the curved surface of the tube. Moreover, when the QDM gel fails as a seal, the water stream breaks through the gel, indicating that the failure is because the gel is not strong enough. In our lab, we have experience with some other cationic gels, such as those made with the biopolymer chitosan. Some of these gels do have better strength than the QDM gel. However, when these stronger gels were tried as patches (using EA) in the same setup as Figure 1.3, they too

failed at relatively low burst pressures. In these cases, the failure mode was different: the water dislodged the patch as it leaked out, indicating that the failure was because of low gel-tissue adhesion.

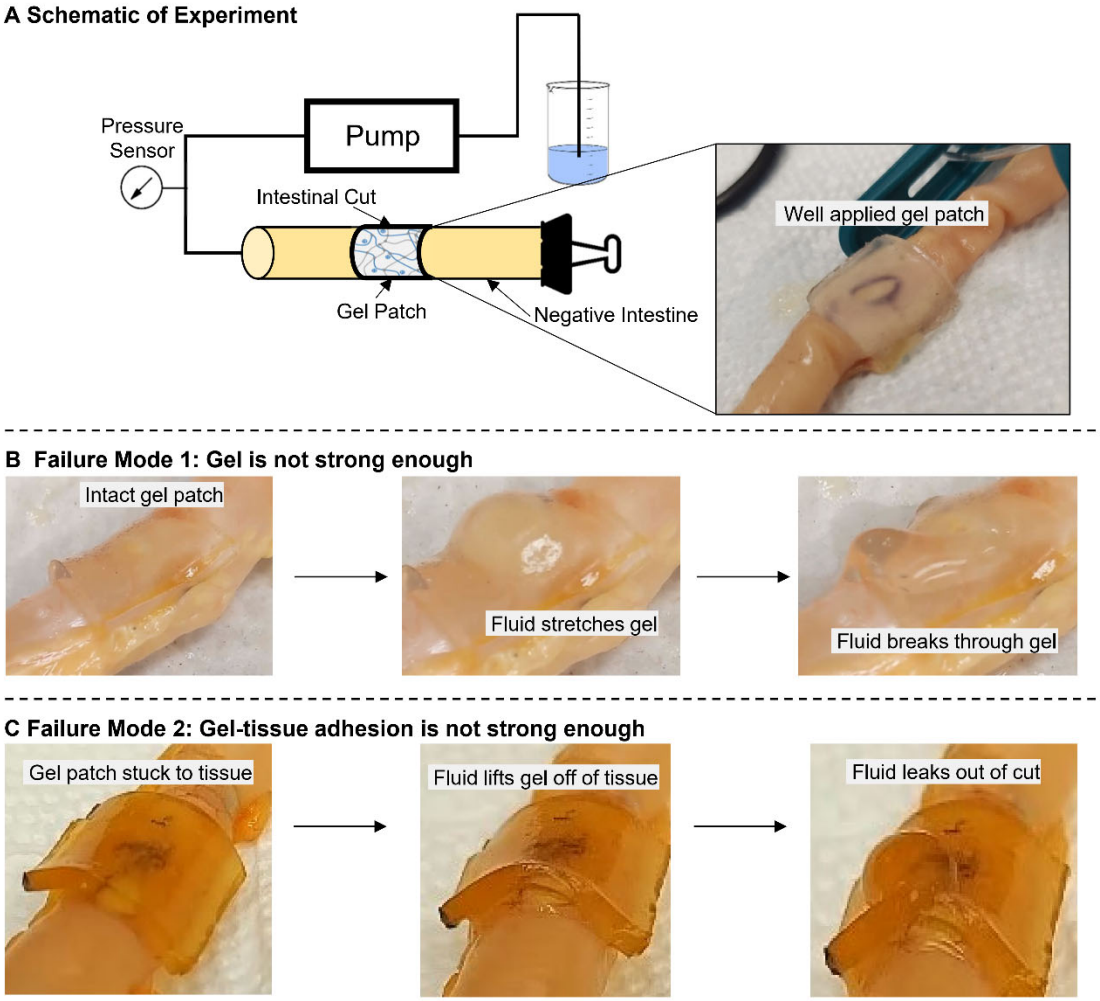

Figure 1.3. Electroadhesion of a gel over chicken intestine for sutureless surgery.
(A) Tear in intestine is closed by cationic gel patch using EA. (B) Sometimes the patch fails because the gel is too weak. (C) Sometimes the patch fails because gel-tissue adhesion is too weak.

The above experiments show that improvements are needed in the gels used for EA. The goal is for the gel-patches to remain affixed even when the water pressure in the tube is higher than the typical pressure *in vivo*. For reference, typical fluid pressure in the intestines is around 80

mm Hg (10665 Pa) while that in blood vessels ranges from 80 to 120 mm Hg (10665-16000 Pa).[11,12] For a gel-patch to withstand such pressures, the gel should both be **strong** as well as **highly adhesive to tissue**. This is the overall goal for this project.

In Chapter 3, we will describe our efforts to create and characterize such strong gels. Our strategy in this regard is adapted from the numerous studies published in the literature on improving the strength of gels.[13-15] The strategy is to introduce a second polymer network into the QDM gels, creating an 'interpenetrating network' (IPN). After much experimentation, we have found that an optimal second network that can reinforce QDM is that of gelatin. By this approach, we have been able to achieve QDM/gelatin gels that are considerably stronger than the native QDM gels while also exhibiting high adhesion to tissue. We have also tested these QDM/gelatin gels for their ability to withstand high burst-pressures in the chicken-intestine model (as shown in Figure 1.3). The results are promising, and indicate that our improved gels could pave the way for *in vivo* testing of electroadhesion in animal models.

Chapter 2: BACKGROUND

2.1. Hydrogels

Hydrogels are three-dimensional networks of polymer chains swollen with water.[16,17] In these networks, the chains are connected at crosslinking points or junctions. These crosslinks can be covalent (chemical) bonds or non-covalent (physical) bonds. Chemically crosslinked hydrogels are formed by free-radical polymerization of a monomer and crosslinker.[18] For example, a gel can be made by mixing the monomer acrylamide (AAm), the crosslinker N,N'-methylenebisacrylamide (Bis), the initiator ammonium persulfate (APS) and the accelerant tetramethylethylenediamine (TEMED) and allowing it to polymerize at room temperature. TEMED accelerates the decomposition of APS into free radicals.[19] These free radicals then initiate the polymerization reaction by binding to the AAm monomers, allowing the monomers to link into polymer chains. Bis molecules will connect to multiple AAm chains resulting in crosslinking of the chains. The final result will be a crosslinked network, i.e., a gel of AAm. To make the gel cationic, a cationic monomer such as quaternized dimethyl aminoethyl methacrylate (QDM) can be added to the recipe.[20] The QDM and AAm monomers will then copolymerize to form the chains in the network. Thus, the QDM/AAm molar ratio will dictate the charge density of the gel.

Another type of gel that we will investigate in this thesis are physically crosslinked gel networks such as gelatin. Gelatin gels are formed by dissolving gelatin powder in water at approximately 40°C and cooling the hot solution. Upon cooling, strands of gelatin, which is a denatured protein, bind to each other and form triple-helical segments through non-covalent interactions such as hydrogen bonding.[21-23] These segments act as crosslinks in the gelatin network.

The structure of a gelatin network is shown schematically in Figure 2.1. Note that the gel is thermoreversible, i.e., it reverts to a solution (sol) upon heating.

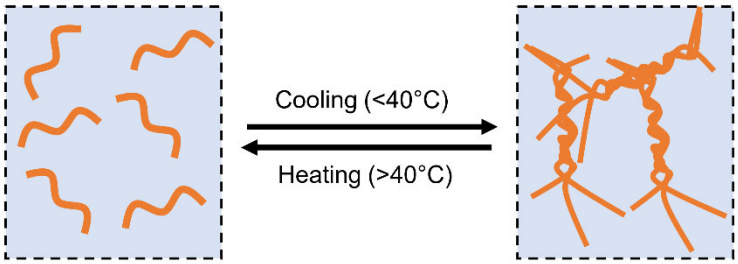

Figure 2.1. Schematic showing the structure of a gelatin gel. Chains of gelatin bind together at junctions where they form triple helices.

2.2. Strengthening Hydrogels

Covalently crosslinked gels such as those of AAm or QDM tend to have weak mechanical properties. If their crosslink density is low, they are too soft and floppy. Conversely, if they are highly crosslinked, the gels tend to be brittle and fragile, i.e., they break into pieces even if slightly deformed. There has been much research into finding ways to make gels stronger. A common strategy to make a covalent gel network stronger is to incorporate a second network within it. Such materials are called interpenetrating networks (IPNs) or double networks (DNs).[24-26] Classical IPN or DN gels often incorporate one chemical and one physical network.[27-29] This allows for a two-step polymerization process and a highly robust IPN.

One IPN explored in this thesis will be that of QDM/gelatin. To make this IPN, the two-step process outlined in Figure 2.2 will be used.[30] First, the monomers needed to make the QDM chemical gel (i.e., QDM, AAm, and Bis) are combined with gelatin in hot water. The gelatin at this stage exists as discrete chains and does not form a gel. Into this hot solution, the initiator (APS)

and accelerant (TEMED) are added. Chemical crosslinking is allowed to occur, leading to a QDM network in which the gelatin chains are incorporated by entanglement. Such a material is sometimes called a semi-interpenetrating network (SIPN). When this is cooled, the gelatin chains form triple-helical crosslinks and thereby a second network. This results in a full IPN, and we find that this IPN gel has much better mechanical properties compared to the QDM alone.

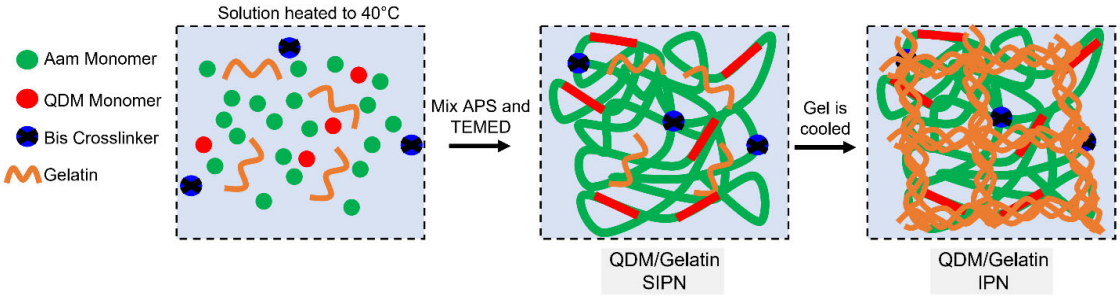

Figure 2.2. Schematics showing the synthesis of a QDM/gelatin interpenetrating network (IPN). Adapted from Ref 30.

2.3. Pull-Off Adhesion Testing

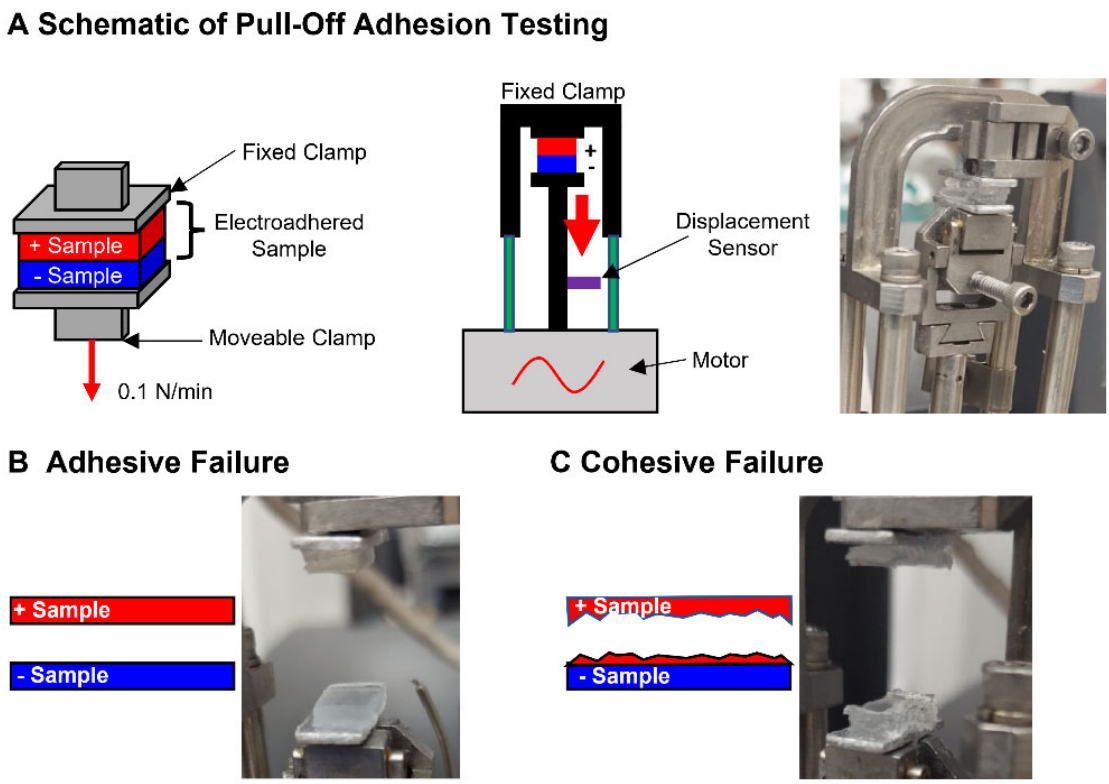

Figure 2.3. Pull-off adhesion testing and modes of sample failure. (A) Schematics of the test, which uses two clamps, one fixed and the other moving downwards at a constant rate. When the clamps detach, the failure can be adhesive (B), or cohesive (C).

In this thesis, we will aim to quantify the strength of adhesion between two gels or a gel and a tissue. For this, we will use a Dynamic Mechanical Analyser (DMA) and run pull-off tests (Figure 2.3).[31] The two materials we will adhere are denoted as "+ Sample" (to indicate the cationic gel) and "– Sample" (to indicate the anionic gel or tissue). The two samples are stuck on one side to the clamps of the DMA using a cyanoacrylate glue (Super Glue). On the other side of the samples, they are stuck either by pressing together (contact adhesion) or using electroadhesion (EA). Then, one clamp is moved downward by applying a force of 0.1 N/min. The other clamp

remains fixed. The stress required to detach the adhered pair is a measure of the pull-off adhesive strength. A higher stress indicates stronger adhesion between the samples.

When the samples detach, two failure modes can occur: adhesive and cohesive failure.[32] Adhesive failure results from failure at the interface between the two samples (Figure 2.3C). Both samples have "smooth" interfaces, indicating that the failure was because adhesion was not strong enough. In contrast, cohesive failure occurs when the adhesion is stronger than the tensile properties of either one of the samples (Figure 2.4D). This results in a "rough" interface as the adhesion still persists while the rupture occurs in one of the samples.

2.4. Tensile Testing

To assess the mechanical strength of our gels, we will test them under tension and shear. In tensile testing, the material is gripped on both ends by the apparatus in an instrument called an Instron. One end remains stationary, while the other end is pulled upwards, thereby stretching the gel (Figure 2.4A). The tensile stress can thus be plotted as a function of the elongation or tensile strain (Figure 2.4B) and the plot eventually ends at the point when the gel breaks. From this plot, several parameters can be extracted. First is the Young's Modulus Y (or elastic modulus), which is the slope in the linear region of the plot at low strains.[33,34] Next is the tensile strength (TS), which is the highest stress that the material can withstand before it breaks. This quantity is also sometimes called the ultimate tensile strength (UTS). In cases where the stress increases monotonically with strain, TS is the same as the stress-at-break.[35] In other cases, the stress can reach a peak at the TS and then decrease before the material breaks. A third quantity is the strain-at-break ε (or elongation at break), which is the maximum extent to which the material can be

deformed before it breaks. Stronger materials will have higher TS. In the case of our gels, we would like them to have both a high TS and a high ε, so that they are both strong and stretchable.

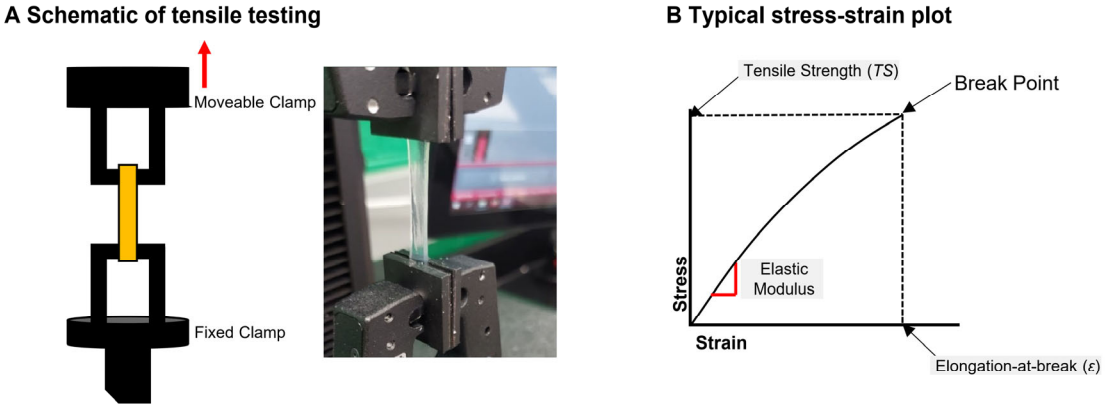

Figure 2.4. Tensile testing schematic and typical data. (A) The sample is stretched axially until it breaks, as shown in the schematic and photo. (B) Typical stress-strain plot for our gels, indicating the three key parameters we extract from the plot.

2.5. Rheology

One further technique used to quantify gel strength is rheology. Rheological studies on gels are usually conducted under dynamic oscillatory shear. Here the sample is deformed by a sinusoidal strain $\gamma = \gamma_0 \sin(\omega t)$, where ω and γ_0 are the frequency and the strain amplitude. The stress response will also be sinusoidal: $\sigma = \sigma_0 \sin(\omega t + \delta)$ where δ is the phase angle. This equation is rewritten in the following way:

$$\sigma = G'\gamma_0 \sin(\omega t) + G''\gamma_0 \cos(\omega t) \qquad (2.1)$$

where G' is the *storage* or *elastic* modulus and G'' is the *loss* or *viscous* modulus. These measurements must be done in the *linear viscoelastic* (LVE) *regime*. In this regime, G' and G'' are only functions of ω.

The rheological signature of a gel is that its elastic modulus G' will be nearly independent of ω, and moreover, G' will be higher than G''. This indicates the elastic response of the gel. The fact that G' is independent of ω implies that the network structure in the gel is permanent, i.e., it does not relax, regardless of the time scale. The value of G' quantifies the *stiffness* of the gel. Generally, the higher the density of crosslinks in a network, the higher the G', i.e., the stiffer the gel.

Chapter 3: STRONG AND STRETCHABLE GELS FOR ELECTROADHESION

3.1. Introduction

The goal of this study is to create gels with excellent mechanical properties that also adhere strongly to tissues by electroadhesion (EA). Our studies are motivated by the weak mechanical properties of the previous gel (QDM) that we employed in our studies using EA. QDM is a cationic gel, and we had previously confirmed that only cationic gels could adhere to tissues by EA (which showed that tissues are anionic). Thus, the gels we are looking for need to be strong *and* cationic.

In terms of mechanical properties, as discussed in Chapter 2, we will focus on the tensile strength TS and the tensile elongation at break ε. Both the TS and ε are low for the QDM gels, which are our control group in this study. Thus, we will seek to increase both these parameters in our gels. As for adhesion, we will measure the pull-off adhesion strength σ (see Chapter 2) for the cases of EA between our strong cationic gel and an anionic gel ($\sigma_{\text{Gel-Gel}}$ or σ_{GG}) or between the gel and tissue ($\sigma_{\text{Gel-Tissue}}$ or σ_{GT}). In the case of tissue, we will mostly work with chicken intestines.

The need to increase mechanical and adhesion strength is relevant for our intended application involving EA, which is to perform sutureless surgery to seal tears in animal tissue. The lab-scale model we will use in this regard is to make a cut in the wall of chicken intestines and patch this cut with a piece of our gel using EA. Then, we will examine how this gel-patch holds up to flow by measuring the burst pressure P_{burst} required to dislodge the patch. Gel strength (as measured by TS and ε) is critical to the success of our burst-pressure experiment. Our working hypothesis is that the stronger the gel (the higher the TS), the better the seal it will make over

tissue, i.e., the higher the P_{burst}. The gel should also be flexible and stretchable (high ε) as this will allow the gel-patch to be bent and folded around a tubular tissue like the intestine without breaking. The adhesion of the gel to the tissue (as measured by σ_{GT}) is also key to a successful burst-pressure experiment, and the stronger this adhesion, the higher we expect P_{burst} to be. We will put these hypotheses to test at the end of this study.

How do we increase the strength of QDM gels? Based on the literature, we hypothesized that the addition of a second water-soluble polymer to a QDM gel could be a simple way to make the gel stronger. Specifically, we focused on the well-known polysaccharides agarose, xanthan gum, alginate, and chitosan, as well as the protein gelatin. All these biopolymers are extensively used in biomedical applications.[36-38] Besides, when added to water, they are known to increase the viscosity (thicken) and form transient or permanent gels. Thus, we postulated that these second polymers could create an interpenetrating network (IPN), or at least a semi-IPN, which could result in improved mechanical properties. Our studies show that gelatin is the optimal second polymer that can be added to QDM to ensure higher gel strength while keeping strong adhesion to tissue. By further tweaking the composition, we have been able to achieve highly stretchy and robust gels for EA applications.

3.2. Experimental

Materials. The monomers acrylamide (AAm), sodium acrylate (SA), and quaternized dimethyl aminoethyl methacrylate (QDM), as well as the biopolymers xanthan gum from Xanthomonas campestris (XG), agarose, and gelatin from porcine skin were purchased from Sigma-Aldrich. The crosslinker N,N'-methylene-bis(acrylamide) (BIS), the initiator ammonium persulfate (APS) and the accelerant N,N,N',N'-tetramethylethylenediamine (TEMED), were also purchased from Sigma-Aldrich. Cyanoacrylate glue was purchased from Krazy Glue.

Gel Synthesis. To prepare our typical cationic QDM chemical gels 6 wt% Aam, 4 wt% QDM, 0.3 wt% BIS, 0.18 wt% APS, and 0.10 wt% TEMED were mixed in a solution of DI water. The aqueous solution was then poured into a glass petri dish to a thickness of 2 mm and left to cure under a flow of nitrogen gas for 2 hours.

To prepare anionic SA/Aam chemical gels 6.5 wt% Aam, 1.4 wt% SA, 0.3 wt% BIS, 0.18 wt% APS, and 0.10 wt% TEMED were mixed in a solution of DI water. The aqueous solution was then poured into a glass petri dish to a thickness of 2 mm and left to cure under a flow of nitrogen gas for 30 minutes.

To prepare cationic IPN gels, 4 wt% QDM, and varied weight percentages of AAm and BIS were mixed in a solution of DI water. The second polymer network was then added under vigorous mixing and heating. After the second network was dissolved, 0.18 wt% APS and 0.10 wt% TEMED was added to the solution and immediately transferred to a glass petri dish to a gel thickness of 2 mm. The gel was then left to cure under a flow of nitrogen gas for 2 hours.

Electroadhered Sample Preparation. To electroadhere our samples an Agilent, model E3612A DC power source was used. For gel-to-gel samples the voltage was set at 10 V. Graphite electrode slabs were cut to sizes of approximately 2 x 3 cm and were connected to the voltage source via alligator clips. The electrodes were placed in the proper configuration for electroadhesion as shown in Figures 1.1 and 1.2. Positive and negatively charged gel sample strips were cut to sizes of approximately 2 x 1 cm and contacted with the respective electrodes in the proper configuration for 10 seconds. For gel-to-tissue samples a similar method was utilized but with a 20 V setting and the samples remained under the electric field for 60 seconds.

Pull Off Adhesive Testing. The pull-off adhesive testing was done on a TA Instruments Q800 Dynamic Mechanical Analysis (DMA) testing apparatus. A sample of cationic gel and either anionic gel or tissue were electroadhered under the conditions specified above. The adhered pair was then super-glued to the DMA clamps with cyanoacrylate glue. The clamps were then attached to the device and a pull apart force of 0.1 N/min was used until failure.

Tensile Strength Testing. Tensile testing was done on a Instron 6800 Series testing device. Testing was done by cutting gel strips into 3 x 1 cm slabs before inserting them into the device grips. Once secured, the samples were then elongated at a rate of 10 mm/min until failure.

Rheological Testing. Rheological testing was done on a TA Instruments AR2000 rheometer at 25° C, using 20 mm parallel plate geometry. Gel samples were cut into 20 mm disks and put under oscillatory shear at 0.1% strain.

Burst Pressure Testing. Burst pressure testing was done using a Pharmacia-LKB peristaltic pump and a MadgeTech PRTemp pressure gauge. Chicken intestine was purchased from a local farm and cut to lengths of approximately 20 cm. The intestines were affixed to one end of the pump and clamped shut at the other end with a clip (see Figure 1.2). Pressure readings were obtained for water flow into the intestine before cut, during cut, and following cut repair. Final burst pressure was determined at the time of patch failure.

3.3. Results and Discussion

3.3.1 Second Polymer Comparisons

We begin by combining QDM with a second polymer, with the intention of creating an IPN or semi-IPN. The polymers we focus on are agarose (Agr, a nonionic polysaccharide), xanthan gum (XG, an anionic polysaccharide), and gelatin (Gln, a protein that is nearly neutral in charge at ambient pH). The parent QDM gel is our control and to it we add 2% of the above polymers prior to free-radical polymerization. The resultant gels are then tested under tension and shear. Figure 3.1A shows the tensile stress vs. strain curves for the gels. In all cases, the stress increases monotonically until the gel breaks. However, the curves are very different for the QDM (control) gel compared to the gels of QDM + a second polymer. For QDM, the tensile strength *TS* (i.e., the stress at break) is low (7 kPa) and the gel breaks when stretched up to its elongation at break ε of just 14%. This indicates that the gel is both weak and fragile. In comparison, adding 2% of Agr, XG, or Gln increases both *TS* and ε: note that the stress-strain curves are shifted upwards and to the right (i.e., to higher stresses and strains).

Figure 3.1B shows the *TS* and ε values extracted from the curves in Figure 3.1A. The QDM/Agr gel has the highest *TS* at 37 kPa, which is ~ 5× that of the control QDM gel. However, this gel is not very stretchable as it breaks at an ε of 20%, indicating that it is almost as fragile as QDM. On the other hand, the QDM/XG gel is less strong (*TS* of 27 kPa), but more stretchable (ε of 46%). Lastly, the QDM/Gln gel offers a compromise, with good strength (*TS* of 27 kPa) and stretchability (ε of 35%). To summarize, we do see marked improvement in the tensile properties upon adding 2% of various polymers to QDM gels.

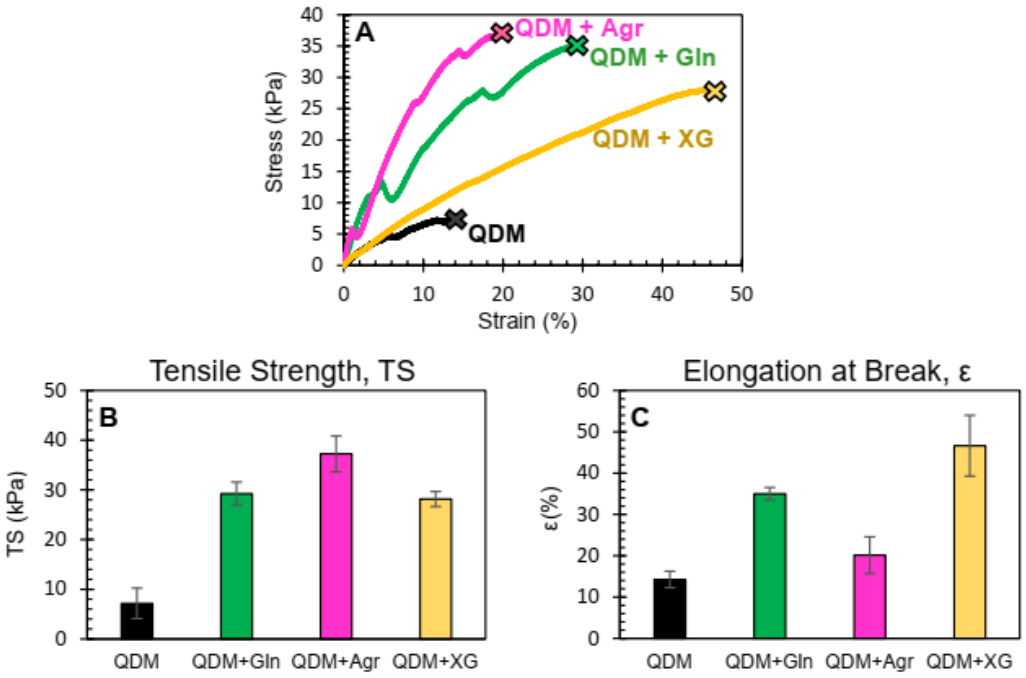

Figure 3.1. Tensile testing comparing the addition of several polymer networks to our original QDM gels. (A) Overall stress v. strain curves, (B) stress at break (*TS*), and (C) elongation at break (ε).

Next, we performed dynamic rheology (oscillatory shear) on these gels. Figure 3.2A shows a plot of the elastic modulus (G') and the viscous modulus (G'') as functions of frequency (ω). All the gels showed the elastic rheology expected of gels, with G' nearly independent of ω, and G' much higher than G''. The value of G' is a measure of the gel stiffness, and it increases when a second polymer is added to QDM (Figure 3.2B). This shows that the gels become stiffer due to the higher density of crosslinks in the IPN or semi-IPN. The highest G' is for QDM/Agr, and it is nearly 100-fold higher than for the QDM control.

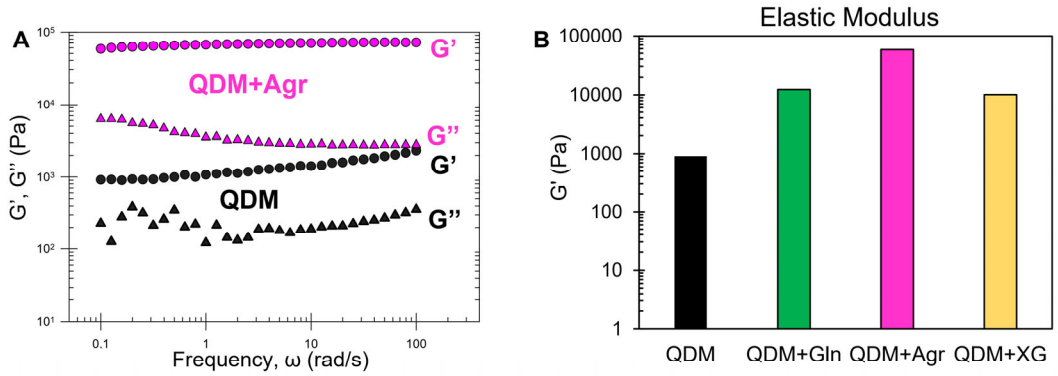

Figure 3.2. Typical rheological data and change in elastic modulus. (A) Typical rheology plot for our gels demonstrating a G' and G'' curve for a control (QDM) as well as a stiffer gel (QDM+Agr). (B) Comparing rheology data for all four gel types showed a higher G' when a second polymer was added.

The data thus far show that the addition of a second polymer makes QDM gels stronger. How does this polymer affect electroadhesion? To assess this, pull-off testing was done for gel-gel pairs where the cationic gel was the QDM/polymer hybrid and the anionic gel was sodium acrylate (SA). In all cases, the gel-gel pairs were contacted under a 10 V DC field for 20 s to induce adhesion. Thereafter, pull-off testing was performed, and the parameter plotted in Figure 3.3 is the pull-off adhesion strength σ_{GG}, which is the stress needed to separate the cationic gel from the anionic gel. σ_{GG} is the highest at 23 kPa for the control QDM-SA pair. When QDM is replaced with QDM/Agr or QDM/XG, σ_{GG} decreases dramatically to 2.0 and 1.1 kPa respectively. However, in the case of QDM/Gln, σ_{GG} is only slightly reduced to 19 kPa.

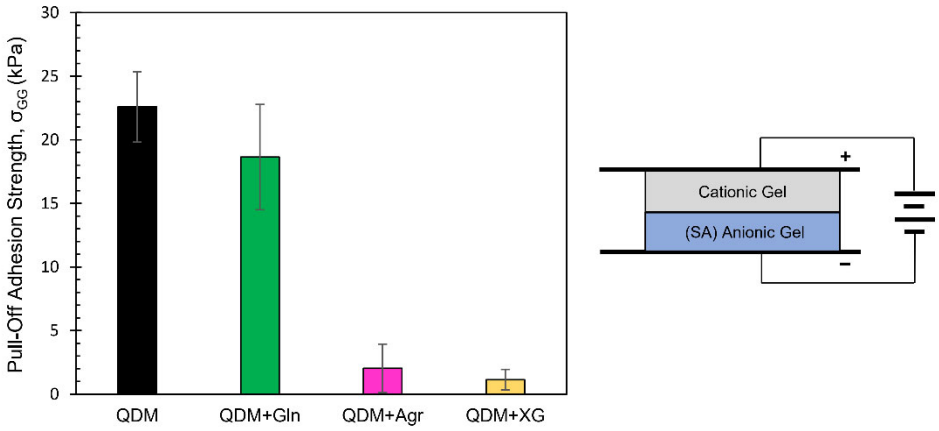

Figure 3.3. Comparing the electroadhesive strength of QDM gels with varied second polymer additions. Pull-off adhesion testing was done utilizing sodium acrylate (SA) anionic gel in an adhered pair with our cationic gels incorporating varied second polymers.

The sharp reduction in σ_{GG} in the case of QDM/XG could be because XG is an anionic polysaccharide (with carboxylate groups) that reduces the cationic character of the QDM gel.[39] If the gel is less cationic, its electroadhesion to the anionic SA gel is expected to be weaker. It is less clear why σ_{GG} is reduced in the case of QDM/Agr compared to QDM alone. Agarose (Agr) is the nonionic component of the anionic polysaccharide agar. It is possible that the Agr makes this gel too stiff (as seen in the rheology) and that this higher stiffness impairs the ability to adhere by EA. In the case of gelatin (Gln), its net charge at ambient pH is low and thus, we believe it does not adversely affect the EA.

3.3.2 Effects of Gelatin Concentration

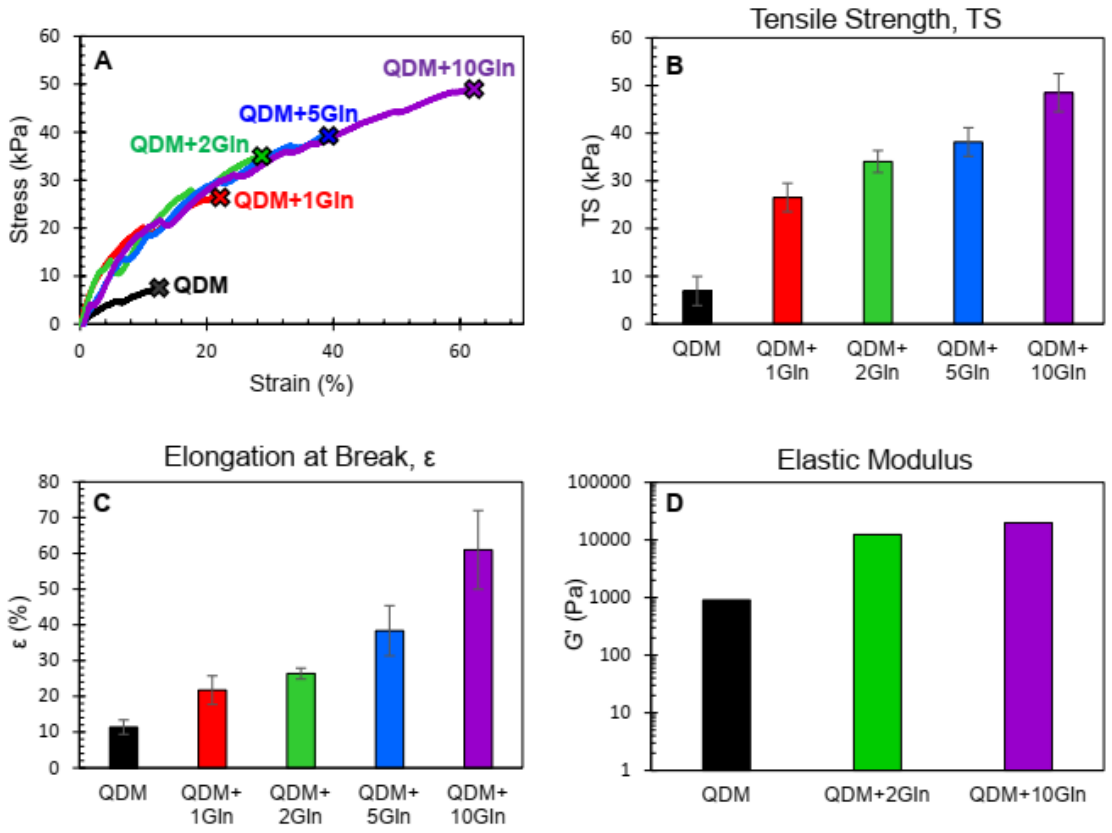

Figure 3.4. Tensile and rheological testing comparing the addition of varied gelatin concentrations to our original QDM gels. (A) Overall stress v. strain curves, (B) Tensile Strength (*TS*), (C) elongation at break (ε), and (D) G' (elastic modulus).

Our results so far have shown that the best secondary polymer to achieve a robust, EA-compatible cationic gel is gelatin (Gln). Therefore, we decided to study QDM/Gln gels in more detail, varying the Gln concentration. Figure 3.4A shows the tensile stress vs. strain curves for QDM (control) and QDM + 1, 2, 5, and 10% Gln. The curves systematically shift to higher stresses and strains as the Gln concentration is increased, indicating that Gln is indeed reinforcing the QDM network. From these plots, the tensile strength *TS* and the elongation at break ε are plotted for the various gels in Figures 3.4B and 3.4C, respectively. Both parameters steadily increase with Gln

concentration. The QDM + 10% Gln sample has a *TS* of 49 kPa (7× higher than QDM alone) and an ε of 61% (7× higher than QDM). We also performed dynamic rheology on these gels and the elastic modulus G' (Figure 3.4D) also increases substantially with Gln concentration. The QDM + 10% Gln gel has a G' of 20 kPa, which is 20× higher than the G' of QDM alone.

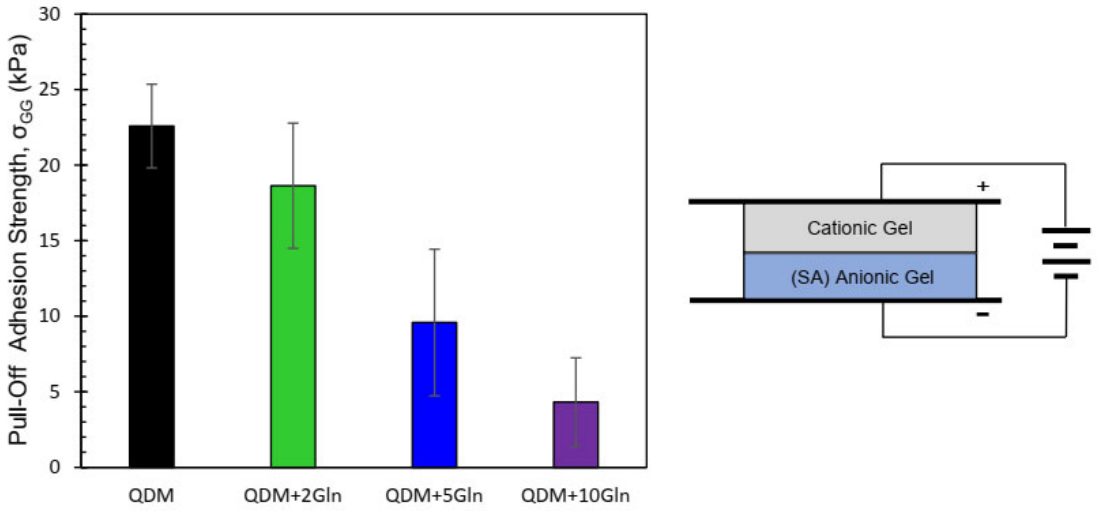

Figure 3.5. Comparing the electroadhesive strength of varied gelatin concentrations added to our QDM gels. Pull-off adhesion testing was done utilizing the anionic gel sodium acrylate (SA), as shown schematically, in an adhered pair with our cationic gels incorporating varied gelatin concentrations.

Next, electroadhesive strength was assessed for the above gels with different Gln concentrations. Once again, these experiments were done for gel-gel pairs where the cationic gel was QDM/Gln and the anionic gel was sodium acrylate (SA), with EA between the gels being induced by 10 V DC for 20 s. Figure 3.5 shows the pull-off adhesive strength σ_{GG}. As the Gln concentration increases, σ_{GG} decreases. For the QDM + 10% Gln sample, σ_{GG} is lowered to 4.3 kPa, which is 5× lower than the σ_{GG} for QDM alone. Thus, we see that in the case of Gln also, the

increase in strength comes at the price of lower adhesion via EA. We conclude that to increase gel strength without sacrificing adhesion, it is best to use a low amount of Gln, i.e., around 2%.

3.3.3 QDM/Gelatin Gel Optimization

Given the constraints based on gelatin concentration, we decided to look at other compositional variables to make our gels stronger. In this regard, the QDM gel itself is made by combining QDM (cationic monomer), AAm (acrylamide, nonionic monomer), and BIS (crosslinker). The wt% of QDM dictates the cationic nature of the gel, and we have kept this fixed at 4 wt%. The total monomer content (M = QDM+AAm) sets the density of polymer chains in the gel, and in our 'control' gel, this is 10 wt%. Finally, the BIS content sets the density of crosslinks, i.e., the "tightness" of the network, and in our control gel, this Crosslinker C = 0.3 wt%. Thus, the control gel has a M:C weight ratio of 10:0.3 = 33. When we add gelatin (2 wt%) to this gel, the resulting hybrid is labeled as Q33/G in Table 3.1.

We decided to examine two variations of the above recipe, with a view to make the QDM network 'looser'. Our rationale was that in an IPN, to achieve the best enhancement of mechanical properties, one network must be much looser than the other. Therefore, we increased the total monomer M and decreased the crosslinker C relative to the control. Whereas M:C was 33 in the control, we increased M:C to 253 in one case and to 800 in another case. These compositions are shown in Table 3.1. In all cases, we included 2% gelatin as the second network. Thus, the final

gels are designated as Q33/G, Q253/G, and Q800/G, respectively. Note that the original QDM gel by this notation would be simply Q33.

	% M	% C	Ratio M/C	Gel Code
Original QDM+2Gln	10	0.30	33	Q33/G
Variation 1	19	0.075	253	Q253/G
Variation 2	24	0.03	800	Q800/G

Table 3.1. Optimizing hybrid QDM/Gln IPN gels. Creating a looser QDM network by increasing the weight percentage of overall monomer M while decreasing the weight percentage of crosslinker C.

Figure 3.6A shows the tensile stress vs. strain curves for the gels in Table 3.1. The data show that loosening the QDM network has a dramatic effect on the mechanical properties of the gels. While Q33/G breaks at a relatively low stress and strain (27 kPa, 35%), Q253/G and Q800/G break at much higher values of these parameters. In the case of Q800/G, the tensile strength *TS* is increased to 135 kPa: note that this is 5× that of Q33/G and 20× that of the original QDM (i.e., Q33) gel. This gel can also be extended up to an ε of 400% before breaking, which is 11× that of Q33/G and 28× that of the original QDM. Thus, the combination of adding gelatin and loosening the QDM network is able to dramatically strengthen the original QDM gel. Dynamic rheological data for the elastic modulus G' (Figure 3.6D) shows that it is also increased for the looser networks. Here, there are opposing factors at play. While the crosslinker content is lower (fewer junctions in the network), the monomer content is much higher (more chains in the network).

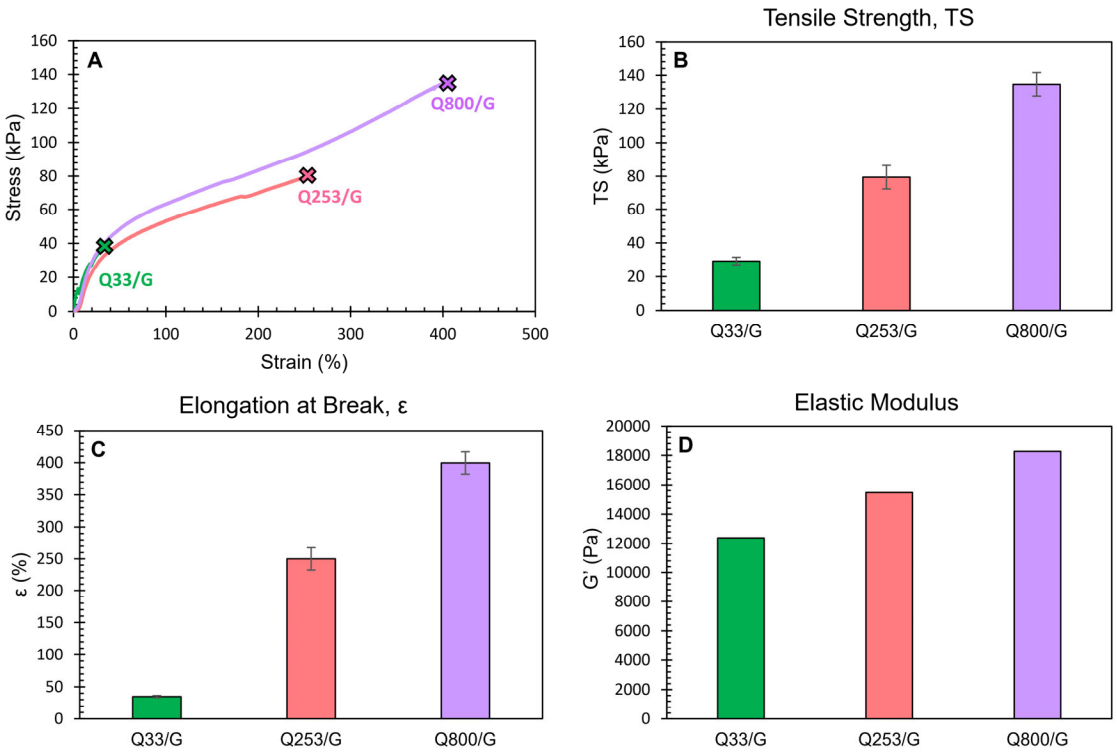

Figure 3.6. Tensile and rheological testing comparing the optimization of the QDM network in our QDM/Gln IPN gels. (A) Overall stress v. strain curves, (B) Tensile Strength (*TS*), (C) elongation at break (ε), and (D) G' (elastic modulus).

The contrast between the mechanical response of our original QDM gel and the strengthened Q800/G gel is visually shown by Figure 3.7. The original QDM gel of length 2 cm breaks when stretched by just a tiny amount because its ε is just 14%. In comparison, the Q800/G gel is shown stretched between the jaws of the Instron to a length exceeding twice its original length of 2 cm and it is yet to break (note that its ε is 400%).

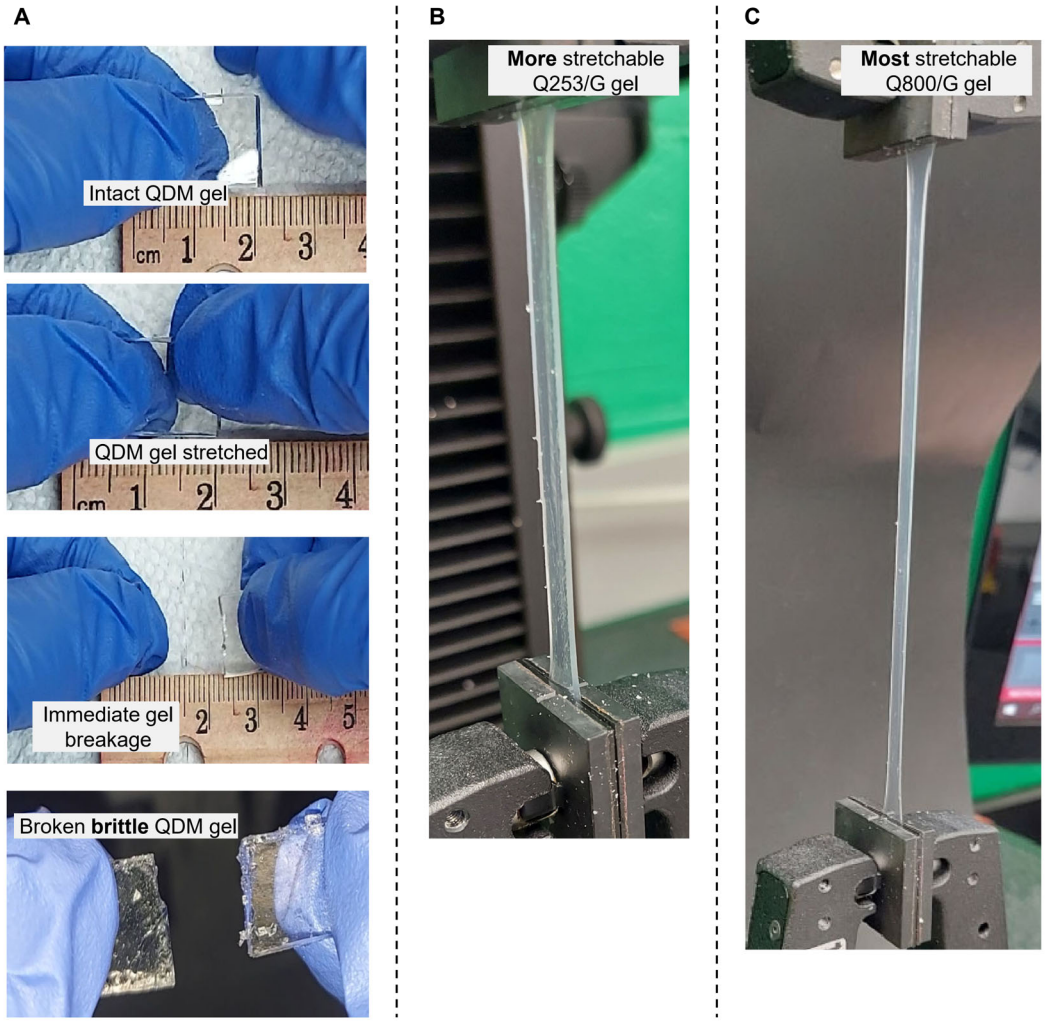

Figure 3.7. Comparison from control QDM gel to strong and stretchable IPN QDM/Gln gels. (A) Our original QDM gels were brittle and could only be stretched slightly before breaking. (B) By incorporating a gelatin network our QDM/Gln IPN gels became stronger and more stretchable. (C) By further optimizing the IPN the gels became even more stretchable.

Next, electroadhesive strength was studied for the above gels. We tested two scenarios: gel-gel and also gel-tissue. For the gel-gel case, the chosen cationic gel was sandwiched with an anionic sodium acrylate (SA) gel and placed under 10 V DC for 20 s to achieve EA. For the gel-tissue case, we procured chicken intestines. A segment from the intestinal wall was cut to the same size as the gels, and this was contacted with the above cationic gel. This sandwich was again placed

under 10 V DC for 20 s to induce EA between gel and tissue. Following EA, the pull-off adhesive strength was measured for both sets of samples and we report these in Figure 3.8 as σ_{GG} for the gel-gel case and σ_{GT} for the gel-tissue case.

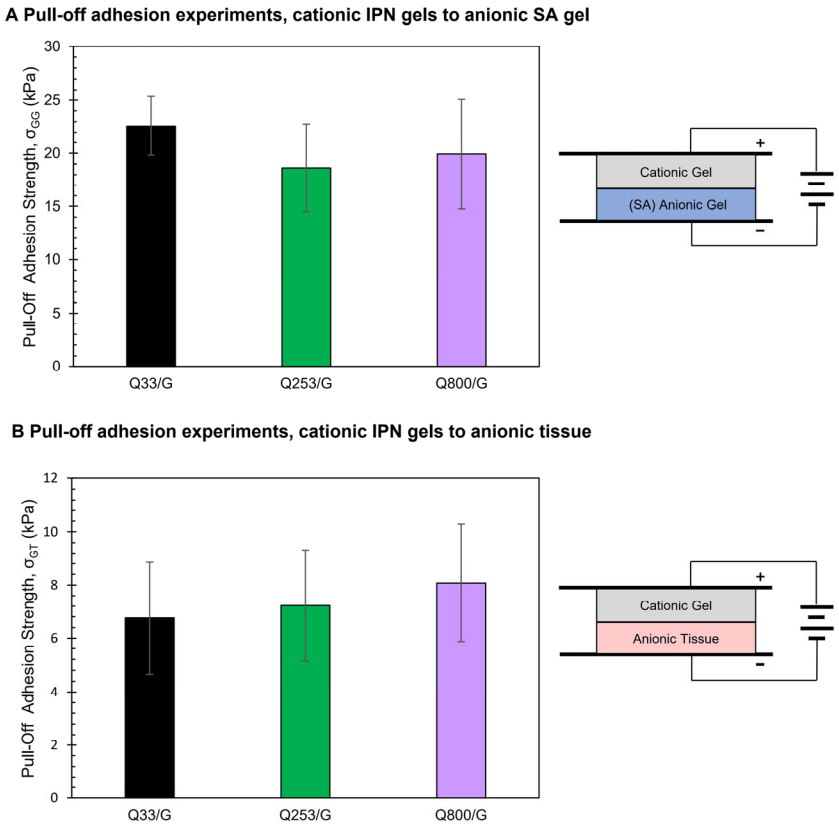

Figure 3.8. Comparing the electroadhesive strength of our QDM/Gln IPN gels.
(A) Pull-off adhesion testing was done utilizing the anionic gel sodium acrylate (SA), (B) as well as utilizing anionic tissue.

From Figure 3.8, we find that the adhesive strength of our new "loose" Q800/G gel is comparable to that of the Q33/G as well as of the original QDM. This is found to be the case both for gel-gel and gel-tissue adhesion. For the gel-gel case, all three cationic gels exhibit σ_{GG} around 20 kPa. When electroadhered to tissue, all three cationic gels exhibit σ_{GT} between 7 and 8 kPa. This is an important result, and it means that the strengthening of the gels (by adding gelatin and loosening the QDM network) can be done without sacrificing the electroadhesive property of the

gels. Note that the adhesive strength is definitely lower in the gel-tissue case compared to the gel-gel case, and this was also observed in our earlier study, where the tissue studied was bovine aorta.

3.3.4 Burst-Pressure Testing

Our final assessment to determine surgical feasibility of our new gels is to conduct burst-pressure testing. As discussed in Figure 1.3 burst pressure adhesive testing can be utilized to determine the strength of a gel to tissue adhesive bond, as well as determine a gel's possible applications in surgery. Burst testing involves the introduction of a fluid flow into a stopped tube using a peristaltic pump, the pressure along this tube can then be read using a pressure sensor, as shown schematically in Figure 1.3. In the case of our experiments, the "tube" utilized is chicken intestines. Our experiments involve three pressure readings per cut size (2.0 mm, 3.0 mm, and 4.0 mm) as denoted in Figure 3.9. The first reading corresponds to a "before" reading, which describes a baseline fluid flow pressure within the intestine before a cut is introduced and where there is no leakage in the tube. The second reading corresponds to the pressure after a cut has been introduced into the wall of the intestine. As seen in Figure 3.9 the pressure reading post-cut drops to 0 mmHg for all cut sizes used in this experiment. The final reading corresponds to after the cut has been patched using our cationic gels. This final reading also corresponds to P_{burst}, in which water is allowed to flow into the tube unimpeded resulting in a buildup of pressure until finally the patch fails in some way and leakage occurs. A higher P_{burst} thus corresponds to a more successful repair. Within Figure 3.9 is also a red dashed line along the pressure 80 mmHg. This pressure represents the ideal peristaltic pressure we would like to achieve for our gel patches, corresponding to a fully descended intestinal bowel. A "successful repair", as denoted in Figure 3.9, is one that is able to

provide a repair strong enough to allow water to flow at or exceeding 80 mmHg, all else is a "non-successful repair".

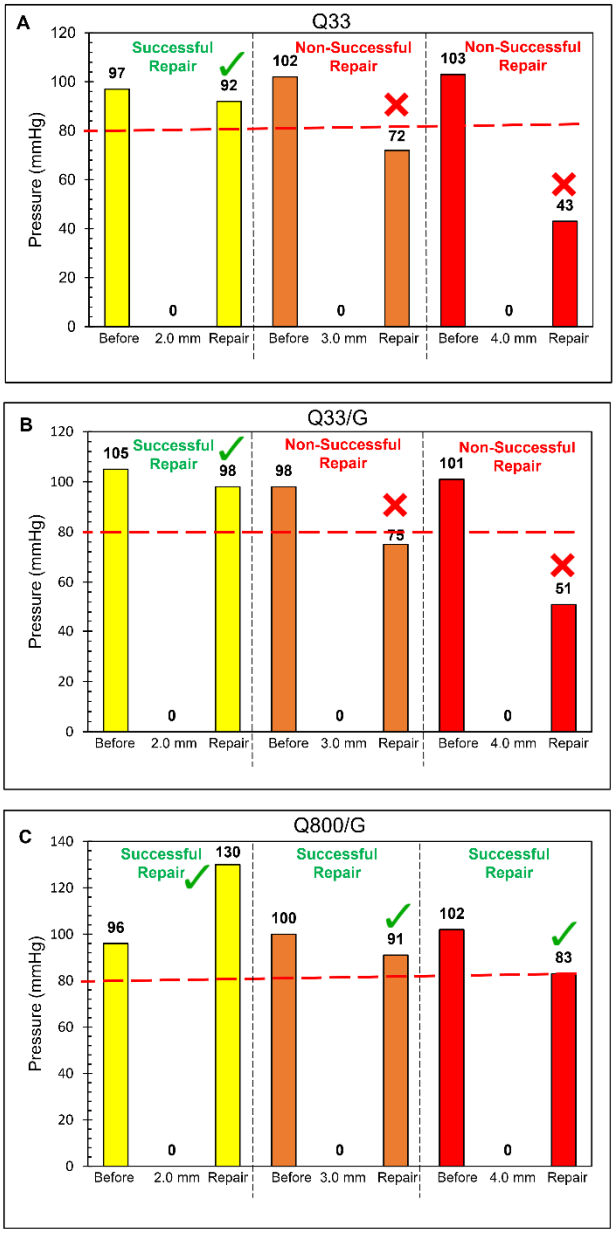

Figure 3.9. Conducting burst-pressure experiments on three gels for cut sizes of 2.0 mm, 3.0 mm, and 4.0 mm. P_{burst} was determined for the brittle QDM control Q33 (A), non-optimized QDM/Gln IPN Q33/G (B), and optimized QDM/Gln IPN Q800/G.

As shown in Figure 3.9 we have conducted burst pressure experiments on 3 gels with 3 cut sizes. These gels include our original brittle QDM gel, denoted as Q33 as described in section

3.3.3. Our next gel is our original QDM gel but with a 2% concentration of gelatin incorporated, denoted as Q33/G. Our final gel is the optimized QDM/gelatin gel, also described in section 3.3.3, denoted as Q800/G. As shown in Figure 3.9A our original Q33 gels are brittle and are unable to seal to a P_{burst} of 80 mmHg or higher for cut sizes larger than 2 mm. With cut sizes of 3 and 4 mm, this gel only achieves P_{burst} values of 72 and 43 mmHg respectively. This corresponds to gel failure mode 1, as discussed in Figure 1.3. In which brittle gels are unable to withstand higher pressures of fluid flow, resulting in the gel breaking and a leakage occurring. This mode of failure can also be seen in 3.9B, where our Q33/G QDM/Gln gel is slightly more successful, with higher P_{burst} values of 75 and 51 mmHg for cut sizes of 3 and 4 mm respectively. However, this gel is still unable to achieve fluid flow pressures equal to or exceeding those that would be found in the body. The failure of Q33/G demonstrates that the addition of gelation to form a IPN is not sufficient to create a gel strong enough to seal wounds in tissues.

Our final gel is Q800/G which utilized an optimized IPN network, as described in section 3.3.3. As shown in Figure 3.9C this gel is able to not only repair cuts of 3.0 mm in size, but also those of 4.0 mm in size, with P_{burst} values of 91 and 83 mmHg respectively. This demonstrates that it is not enough to have an electroadhesively strong gel patch it is also necessary for the patch to be mechanically strong enough to withstand the fluid flows that would be inherent within an in-vivo environment. Thus, by fine-tuning the IPN network of the Q33/G QDM/Gln gel to synthesize a Q800/G gel we have been able to create a gel that is able to perform optimally in surgeries.

3.4 Conclusions

The work discussed in this chapter shows how to systematically synthesize a strong and stretchable electroadhesive gel for surgical applications. We began this discussion by highlighting

how incorporating a second polymer additive (of gelatin, agarose, or xanthan gum) affected the mechanical properties of our original brittle cationic (QDM) gels. By utilizing tensile and rheological testing we were able to quantify the mechanical increase with both the tensile strength (*TS*) and elongation-at-break (ε) of these gels. It was shown that by incorporating any second polymer network into our QDM gels a stronger gel was achieved. However, after quantifying the electroadhesive strength of these gels, using pull-off testing, it was found that both xanthan gum and agarose decreased the gel's ability to adhere to negatively charged materials. It was thereby determined that gelatin was the optimal second polymer for a strong electroadhesive gel. We then discussed how the percentage of gelatin affected the mechanical and adhesive abilities of the QDM gel. It was found that by increasing the percentage of gelatin past 2% that the mechanical strength of the gel increased but the electroadhesive strength decreased significantly. We were thereby able to conclude that to make a mechanically strong gel, without sacrificing electroadhesive strength, 2% gelatin was the optimal polymer additive. Following these experiments, with knowledge of the constraints evident in a higher gelatin concentration, we were then able to look at other ways to optimize the mechanical strength of our QDM/Gln gel. We then demonstrated that by making a "looser" QDM network we were able to achieve a gel that has a ~20 times higher *TS* and ε when compared to original QDM gels. These gels also were able to achieve comparable adhesion with anionic gel as well as chicken intestinal tissue, when compared to our original brittle QDM gels. Finally, we were able to demonstrate how these new optimized QDM/Gln electroadhesive gels were able to achieve successful P_{burst} values at cut sizes of up to 4.0 mm. This is in contrast to both our orginial brittle QDM gels, as well as the non-optimized QDM/Gln gels.

4: CONCLUSIONS & RECOMMENDATIONS

4.1. Conclusions

In this thesis, we have demonstrated that a mechanically robust electroadhesive gel can be synthesized to withstand high levels of burst pressure without mechanically breaking or exhibiting an adhesive failure. We have synthesized electroadhesive gels that are optimized for surgical applications as well as demonstrated how gel mechanical strength can be improved without compromising electroadhesive ability.

In order to increase the strength of our original brittle QDM gels we first attempted to incorporate several types of polymers into the gels as a second network. By performing tensile and rheological experiments we were able to quantify the increased mechanical strength of these new IPN electroadhesive gels. However, by performing pull-off adhesive testing it was found that several of these polymers decreased electroadhesive strength to a high degree. One polymer that did not demonstrate this decrease in electroadhesive ability was gelatin.

As gelatin was found to be the optimal second polymer for our electroadhesive gels, we then sought to find the ideal weight percent of gelatin within our gels. To do this we preformed tensile and rheological testing which demonstrated an increase in gel mechanical strength as the weight percent of gelatin was increased. However, it was found through pull-off adhesive testing that by increasing the gelatin weight percent past 2% electroadhesive strength decreased to non-practicable levels.

As shown through the studies comparing different types of polymers incorporated into our gels, and again through the comparison of gel weight percentages on adhesive and mechanical

strength, 2% gelatin appeared to be the optimal second network for our electroadhesive gels. To further mechanically strengthen this gel, we were able to "loosen" the QDM network within the gel in order to optimize the QDM/Gln IPN. This resulted in a strong and stretchable electroadhesive gel that exhibited ~ 20× higher elongation at break (ε) and tensile strength (TS) compared to QDM alone. This gel was also able to adhere to both anionic gel and animal tissue with adhesive strengths comparable to our original QDM gels.

To test the surgical practicality of this gel we conducted a series of burst pressure experiments for cut sizes of 2.0 mm, 3.0 mm, and 4.0 mm. It was demonstrated that this stronger cationic gel was able to repair cut sizes of up to 4.0 mm with a repair P_{burst} up to or exceeding 80 mmHg (the normal intestinal fluid pressure in the body). This is in contrast to burst-testing failures of both the original QDM gel as well as the non-optimized QDM/Gln IPN. Thus, we have demonstrated the synthesis of a strong and stretchable electroadhesive gel that is able to seal wounds at pressures comparable to those in the body, making it a functional adhesive patch for surgical applications.

4.2. Recommendations

While our approach has optimized a mechanically strong, electroadhesive gel, further work can be done to create even stronger gels that can seal larger wounds in animal tissue. In addition, further optimization of different types of polymers into the second network of our electroadhesive gels should be studied. While high concentrations of gelatin was found to decrease the electroadhesive capabilities of the gel, it is possible that other polymers such as agarose could be studied at lower concentrations to test gel electroadhesive ability. Overall, the reason as to why

higher concentrations of certain polymers (such as agarose and gelatin) decreased electroadhesive strength is largely unknown and this question could lead a mechanistic study of electroadhesion in tougher gels and other materials. Could simply increasing the crosslinking agent in a normal QDM gel (to make a tougher gel and a "tighter" network) decrease electroadhesion in a similar vein? Could decreasing agarose from 2% to lower concentrations allow for a more electroadhesive, but strong gel? Is how "soft" or "hard" a gel is a factor in how electroadhesive it is? There are many directions for future studies on why certain gels do not electroadhere and other gels do. In addition, the longevity of our gels in animal tissue also should be studied due to gelatin's relatively low melting temperatures. If one of our QDM/Gln gels were left at human body temperature for an extended period of time would the electroadhesive properties change? By becoming a "softer" gel as the gel is warmed will the electroadhesive abilities decrease? These are important avenues to explore due to our gel patch's potential as a surgical adhesive.

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