# Electroadhesive Technologies for Prosthesis Applications

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# CONTENTS

# MOTIVATION

This honors capstone project is part of a larger, encompassing project developing a novel upper limb prosthesis, and one of the major design focuses of this prosthesis project is meeting the target specifications set by human muscle capabilities. Currently in research, commercial, and patent literature, there are no designs that meet the target torque, speed, and range of motion (ROM) targets that would provide sufficient performance to carry out activities of daily living efficiently. This has resulted in high rejection rates of extrinsically powered prostheses (>20%), and those that do use them suffer from a variety of health issues due to over compensatory motion of existing joints to account for these insufficient metrics[1].

The underlying cause of this lack of functionality in prostheses is mostly due to the actuator and transmission architectures employed in existing designs. Specifically, transmissions that have shown promise in larger applications i.e., prosthetic legs, automotive transmissions, etc., do not scale well when implemented in a small assembly such as a prosthetic hand, and lack the proper torque density, specific power, and power density as a result.

This problem has identified a need for a lightweight, compact, and powerful solution for transmission architectures, and one key component our lab has identified in enabling this is electroadhesive actuators. Electroadhesive actuators provide a low power, lightweight, and highly configurable solution for many applications including locking mechanisms and clutches in transmission architectures. By implementing electroadhesives, we hope to increase the functionality of prosthesis transmissions with minimal weight and power costs.

# BACKGROUND

## *Working Principle and Configurations*

The general constitution of electroadhesive devices is the same as that of a conventional capacitor. They typically consist of oppositely charged electrodes encased in an insulating dielectric, however the geometry of these electrodes and dielectric are contingent on the desired functionality of the device. This results in two main configurations of these devices commonly used in literature. The first of which is the double-sided configuration. This configuration most closely resembles that of a conventional parallel plate capacitor, consisting of two electrode plates, each covered with dielectric layer, imposed onto one another as shown in Figure 1.

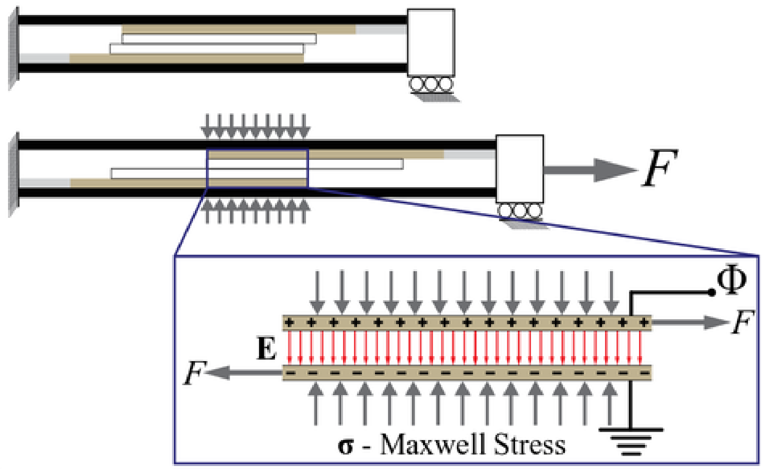


**Figure 1.** *The working principle of the double-sided configuration.* The oppositely charged electrodes induce an electric field, resulting in a pressure that can be used to create locking in shear mode[2].

As seen in Figure 1, this configuration uses two oppositely charged electrodes that induce an electric field between themselves. This electric field results in a pressure better known as Maxwell stress and prevents the electrodes from moving with respect to one another in shear due to friction. A drawback of this configuration is that one of the electrodes must be grounded to function, and in the context of prostheses this is limiting when implemented in applications requiring an extensive ROM or rotary mechanism.

A second configuration that fixes this issue is the co-planar configuration. Unlike the double-sided configuration, the oppositely charged electrodes do not move with respect to one another and are instead interdigitated with each other in the same plane, encased in a single dielectric layer. The working principle is similar to that of the co-planar configuration, however, instead of inducing the electric field across its own electrodes, the fringe electric field causes charge to accumulate in a target substrate (a wall, shaft, or any surface) if it's a conductor or cause a net dipole moment if it is an insulator, resulting in pressure between the device and substrate. This phenomenon is illustrated in Figure 2.

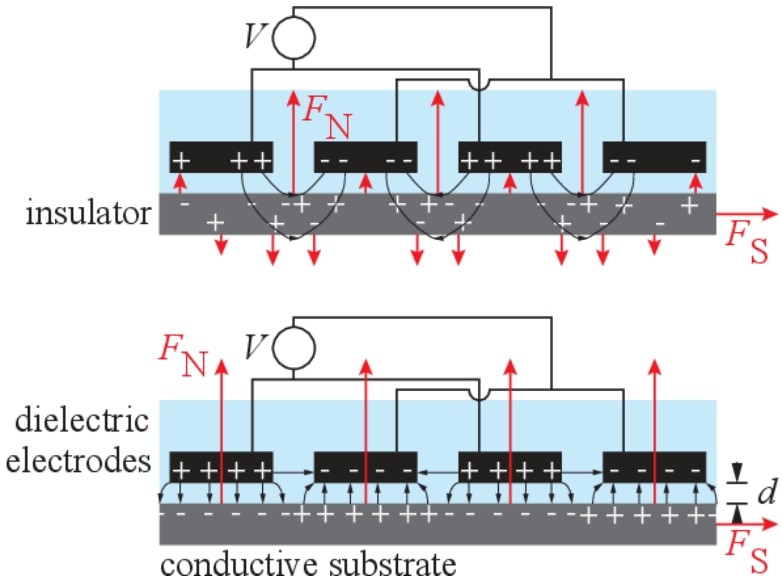


**Figure 2.** *The working principle of the co-planar configuration.* The interdigitated electrodes induce an electric field across a neighboring substrate, resulting in adhesion.

As illustrated in Figure 2, the electric field displaces charges or creates localized dipole moments to create pressure. Since the fringe electric field is the main mechanism that creates adhesion as opposed to a direct potential between electrodes like the double-sided configuration, designs that employ this configuration tend to be weaker. This can be an issue for prosthesis applications that require high shear forces.

## STATE OF THE ART

The electroadhesive designs currently existing in literature are used in a wide variety of applications, including exoskeletons, modular robots, climbing robots, and soft grippers. The materials, electrode geometries, and operating conditions used between these designs varies greatly as there is no clear benchmark design for these devices and requirements between applications varies significantly. For our specific design, the main focus is to maximize the shear force density while also allowing the device to be flexible and be implemented in a rotary mechanism. Based on these requirements, I have narrowed the field of existing electroadhesives to two designs that most closely fulfill our requirements.

### *Hinchet's High Force Density Textile Electroadhesive*

The first design that has had as a major influence on our approach is Dr. Hinchet's textile electroadhesive[3]. This device was developed to be implemented in a virtual reality (VR) haptic feedback glove used to simulate holding objects. The design and implementation of this device is shown in Figure 3.

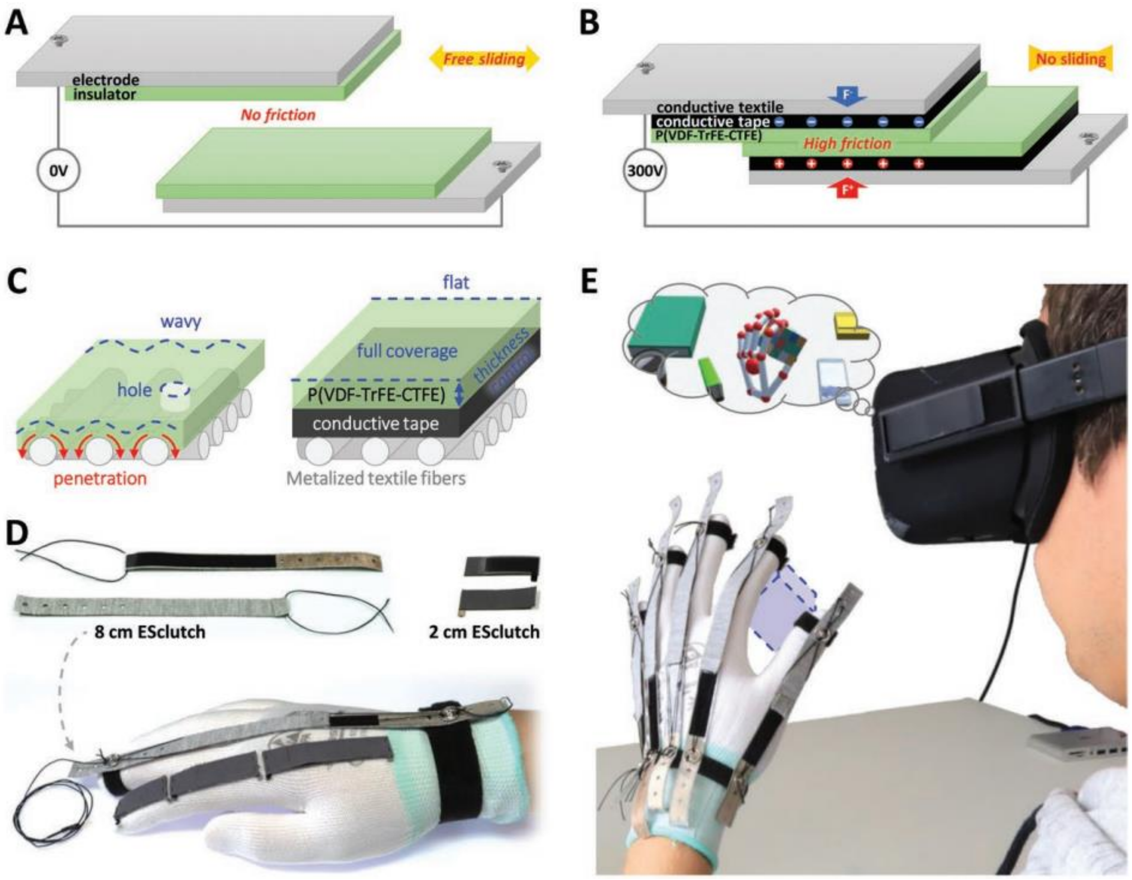


**Figure 3.** *Dr. Hinchet's high force density textile electroadhesive.* This electroadhesive uses a conductive textile electrode with conductive tape covered in P(VDF-TrFE-CTFE) dielectric. When activated, the electroadhesive prevents the hand from closing, simulating the user holding an object.

This design utilizes a metalized fabric as a flexible electrode covered in P(VDF-TrFE-CTFE), a high permittivity ferroelectric polymer that can handle high stresses and wear. This clutch is currently the best performing electroadhesive device, applying shear forces up to 210kPa at a voltage of 300V.

### *Choi's Soft Gripper*

A second top performing design is by Dr. Choi at Sungkyunkwan University[4]. This design is a soft gripper that utilizes a co-planar configuration electrode composed of copper-cladded polyimide film with an anionic waterborne polyurethane dielectric layer. The soft gripper architecture and the device under load are shown in Figure 4.

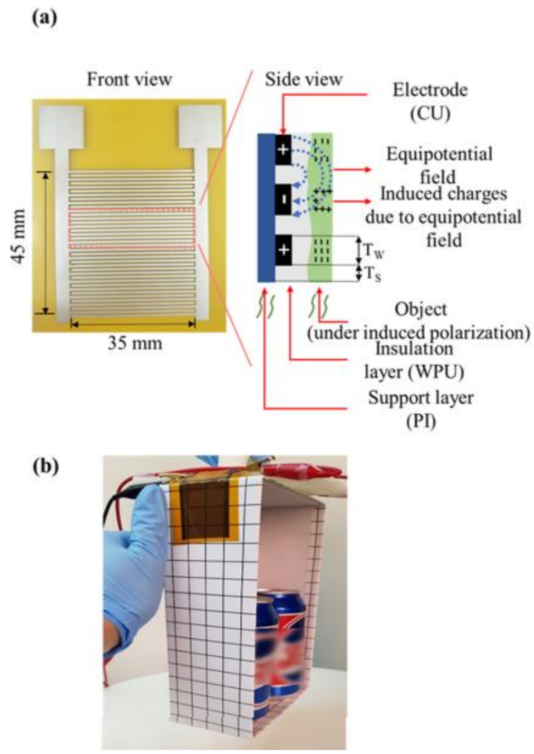


**Figure 4.** *Dr. Choi's Soft gripper.* The soft gripper is shown under load (b) alongside the electrode geometry (a).

In regard to performance of this design, the maximum shear force density is 33.1kPa at an operating voltage of 1kV. This shear force capacity is significantly less than that of Hinchet's design, due to the drawbacks discussed earlier regarding the configuration as well as the material used as the dielectric. The impact of dielectric sourcing will be discussed in greater detail in the approach discussion.

## OBJECTIVE

The main objective with our electroadhesive design is to take the winning characteristics of state-of-the-art designs and incorporate them into the context of our application in the prosthesis. Specifically, we are looking at trends in characteristics among designs that have yielded general high performance. These characteristics include electrode geometry and dimensions, materials used, and operating conditions. We also approach these design characteristics with the consideration of how well these they translate into our application. In general terms, we are to implement this device as a locking mechanism for a rotating component. Therefore, we are not only considering the max force capability, but also the elastic properties, mechanical strength, wear resistance, and other key factors. Furthermore, when sourcing materials and determining electrode geometry, it is important to consider the manufacturability. This is because we want to be able to conduct rapid prototyping and have several prototypes ready to test. Therefore, it is desirable that any solutions we explore are manufacturable in house or require short lead times.

# APPROACH

Our approach to generating a design for our device consists of focusing on two main design drivers: the materials used and the dimensions and configuration of the electrodes. By observing prior art and existing designs it is conclusive that these factors are most effective in optimizing the performance of electroadhesive devices.

## *Electrode Optimization*

When optimizing electrode design, specifically for the co-planar configuration, the relationship between ratios of the dimensions of the device, or dimensional analysis, has been carried out to yield the following relationship[5]:

$$\sigma_{ad} = \varepsilon_0 C\left(\frac{a}{b},\frac{h_1}{b},\frac{t}{b},\frac{\varepsilon_c}{\varepsilon_w}\right)\left(\frac{\varphi}{2b}\right)^2 \quad (1)[5]$$

Where $\sigma_{ad}$ is the pressure exerted by the device, $\varepsilon_0$ is the permittivity of free space, $\varphi$ is the voltage differential between electrodes, and *C* is a dimensionless number determined by the device characteristics as shown in Figure 5.

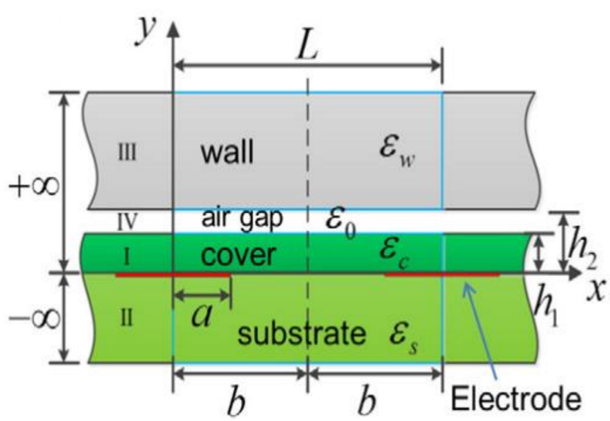


**Figure 5.** *Key electrode dimensions for maximizing shear force.* The width of the electrodes, *a*, their spacing, *b*, electrical permittivity of materials $\varepsilon_w$ and $\varepsilon_c$, and dielectric thickness $h_1$ all impact the strength of the device[4].

The relationships and dimensional analysis between these parameters are further illustrated in Figure 6 and Figure 7.

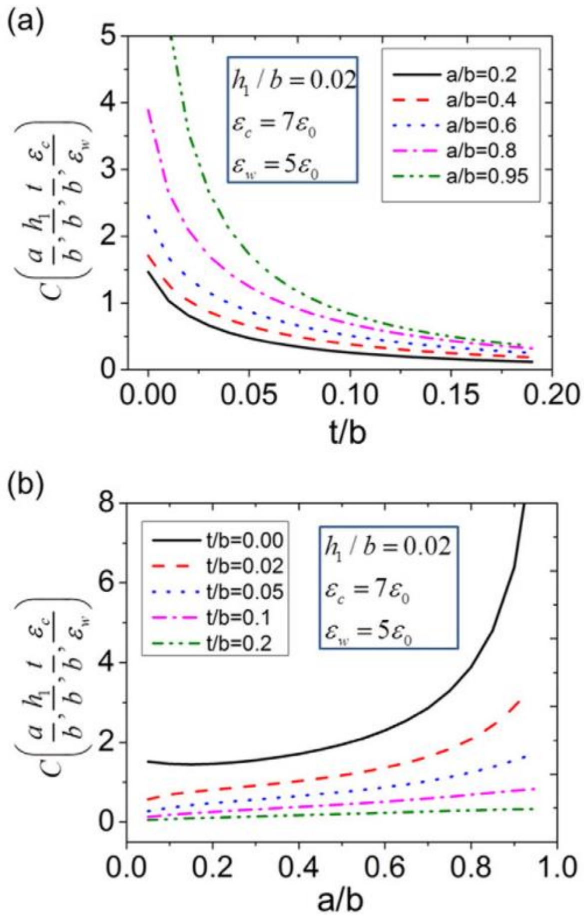


**Figure 6:** Relationship between normalized geometric and material constant $C$ and $\frac{a}{b}$ (b), and $\frac{t}{b}$ (a). It is illustrated above that $C$ increases with increasing $\frac{a}{b}$, and decreases with $\frac{t}{b}$. As $\frac{t}{b}$ increases, the effect of $\frac{a}{b}$ on $C$ diminishes.

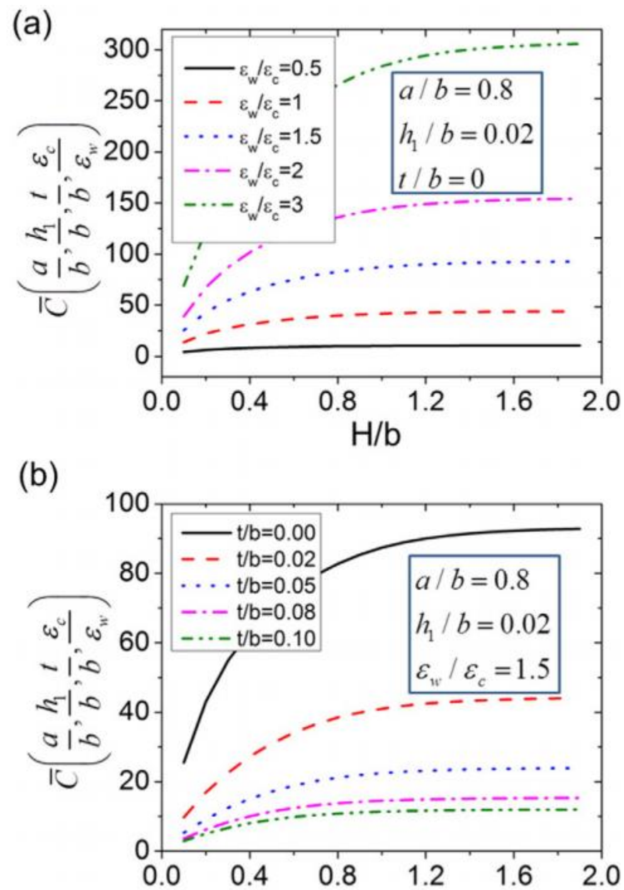


**Figure 7:** The graphs shown above relate $C$ to the normalized substrate thickness, H/b, for varying normalized dielectric constant (a) and normalized air gap (b). For increasing $\frac{\varepsilon_c}{\varepsilon_w}$, $C$ increases, and for both plots (a) and (b), $C$ increases with substrate thickness.

From the dimensional analysis shown above we come to some key conclusions:

- Maximize electrode width to spacing ratio.
- Minimize dielectric thickness to spacing ratio.
- Maximize substrate dielectric constant for a given insulator
- Maximize substrate thickness to electrode spacing ratio (no less than H/b = 2).
- Minimize air gap thickness (very important).
- Maximize possible applied voltage.

Another important metric to consider when optimizing the electrode geometry is maximizing the total boundary edge length along the electrodes[4]. The intuition behind this is that it facilitates the buildup of charge, resulting in stronger electric fields and shear force capabilities. A computational FEA model of an electroadhesive shown in Figure 8 illustrates this affect[6].

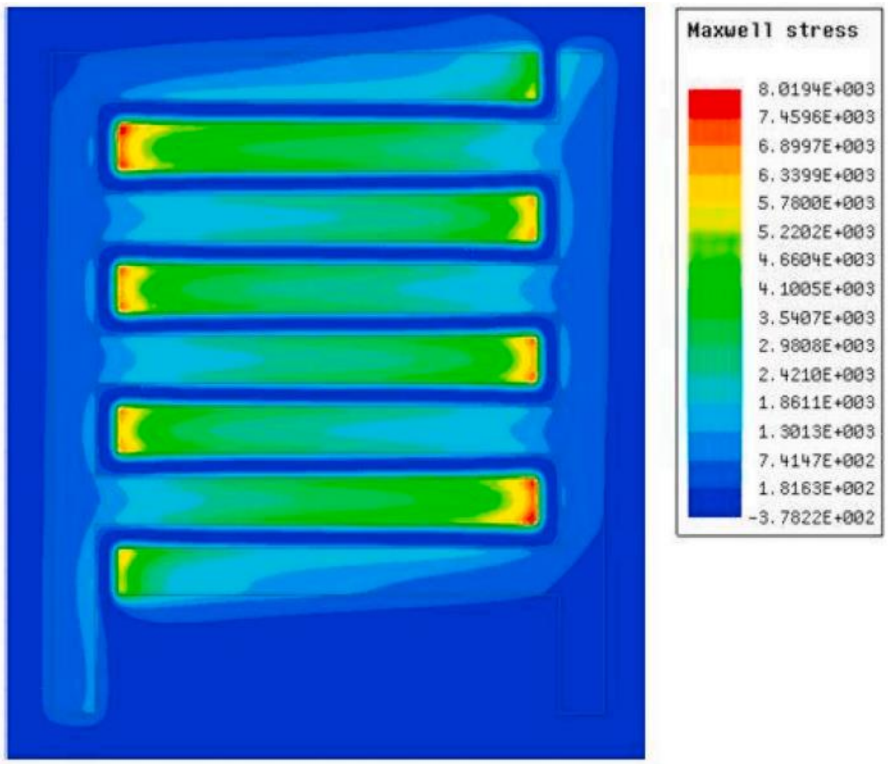


**Figure 8.** *Modeling pressure exerted by electroadhesives.* Pressure exerted by electrodes is maximized at the edges due to increased charge accumulation. Therefore, to take advantage of this effect, the boundary edge length along the electrode teeth should be maximized for a given area of the device.

On paper, these relationships appear rather simple, however, it is important to acknowledge that there are physical limitations and tradeoffs that occur when pushing the performance with these relationships. The first of which is the limitations of the dielectric material used. The dielectric can only handle a certain intensity of electric field i.e., breakdown voltage, so realistically there is a limit to how high this voltage can scale. Furthermore, the method of manufacturing has a significant impact as to how small these electrodes can be, resulting in sub-optimal dimensions.

### *Dielectric Selection*

As discussed, the dielectric used in the device not only limits the operating conditions of the device, but also the performance within these limits as seen in the dimensional analysis. This leads us to prioritize two specific properties of the dielectric selected: the electrical permittivity and the breakdown voltage. It is important to maximize the electrical permittivity to increase the intensity of electric field being passed through the dielectric, as the electric field is the main mechanism for adhesion and co-planar configurations are already at a disadvantage when utilizing the fringe electric field. The breakdown voltage is equally as important as it allows us to scale the voltage to make up for sub-optimal electrode geometries to improve the pressure. A comprehensive plot comparing these properties of commonly used dielectrics in literature[3] is shown in Figure 9.

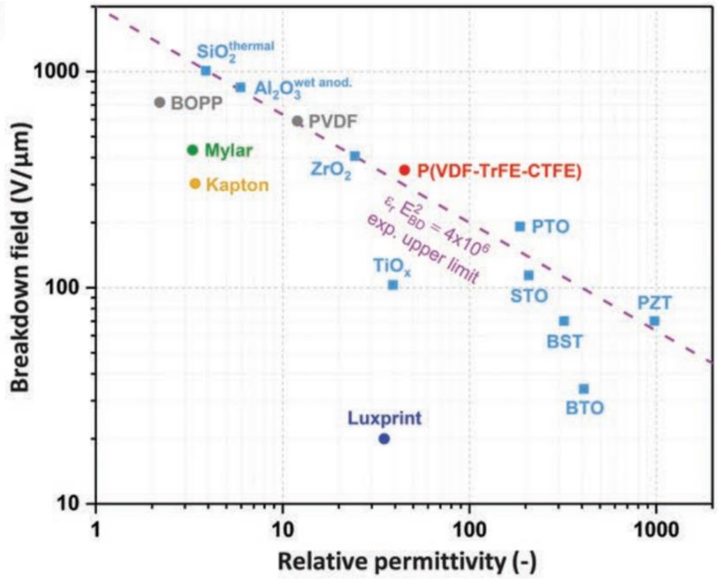


**Figure 9.** *Electrical properties of dielectrics used in literature.* The most attractive dielectrics are those that have both high relative permittivity and breakdown voltage.

As shown in Figure 9, a desirable dielectric material will have both high relative permittivity and high breakdown voltage. One material shown that is particularly attractive is P(VDF-TrFE-CTFE). This polymer has an impressive permittivity of 40+ and breakdown voltage of 300V/micron. This specific polymer is used in Hinchet's high force density textile electroadhesive and is one of the major contributors to its impressive performance. Furthermore, this material is relatively easy fabricate with, as it is thermally cured at a temperature of 102C. This is very feasible as the resources required to thermally cure are readily available.

## INITIAL DESIGN CONCEPT

The intention behind the initial design is to first verify that the design is functional and can be manufactured. Therefore, the initial design concept has not been completely optimized and has been designed to be an indicator of the efficacy of the current design approach and identify potential unexpected issues in the manufacturing process and performance.

### *Materials Selection and Fabrication*

One major difference between many designs in literature was the materials used for the electrodes, dielectric, and base substrate of the device. Because of this, it was imperative to make informed decisions based on the materials research described above. Furthermore, a major challenge in the materials selection process was ensuring that we could fabricate with it in-house. Specifically, our chosen fabrication process consists of using a PCB mill to machine an electrode profile out of 0.002" copper foil super glued to FR-1 blanks. The PCB mill and FR-1 blanks used in our fabrication process are shown in Figure 9.

**Figure 9.** *The Bantam PCB mill and FR-1 PCB blanks.* These are resources readily available in the Duderstadt fabrication studio.

After the electrode profile has been machined into the foil and blank, it is submerged in isopropyl alcohol to remove the remaining super glue and the foil electrodes are removed. These electrodes are again super glued to an ⅛" fabric-reinforced high-temperature silicone rubber base using a 3D-printed jig to ensure proper alignment. This specific base is chosen given that it is highly insulating, mechanically strong, and can withstand high temperatures during dielectric curing.

The next step is applying the dielectric to the electrodes and base. The dielectric used in our device is PiezoTech® P(VDF-TrFE-CTFE) RT-TS dielectric. This material was chosen given its impressive permittivity and breakdown voltage. To apply this dielectric, the P(VDF-TrFE-CTFE) powder is mixed with methyl-ethyl-ketone (MEK) at 14% concentration and applied to the device using a blade coater. Once the dielectric is fully applied, the device is thermally cured in an oven at 102◦C and the fabrication process is complete.

Other components sourced include a DEVMO DC-DC high voltage boost converter. This will allow us to scale the prosthesis supply voltage to the proper operating voltage of the device (300V – 400V). Since this device is very low power (milliwatt scale), we are implementing a simple current regulating circuit using an LM317 as shown in Figure 10.

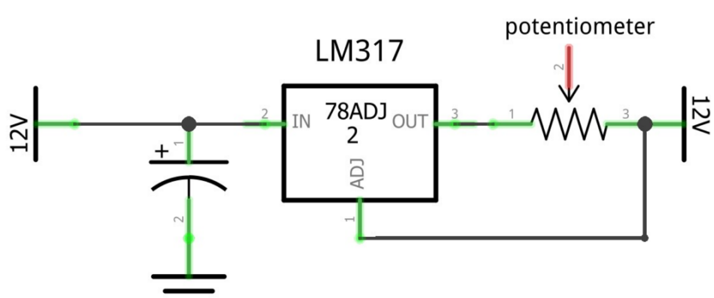


**Figure 10.** *Current regulating circuit.*

This current regulating circuit will also allow the maximum current to be changed via increasing or decreasing the resistance of the potentiometer. This circuit will ensure that the device is not overloaded and that it is safe for the user.

### *Design Parameters*

In terms of physical dimensions of the electroadhesive device, the electroadhesive will employ a co-planar configuration with teeth of width and spacing at 2mm and 0.5mm respectively. This decision was informed by the dimensional analysis above, as a *a/b* ratio of 0.8 was determined to provide high shear force densities while being resistant to voltage breakdown. A dielectric thickness of 25 microns has also been chosen given that a $h_1/b$ ratio of 0.02 was ideal as found by dimensional analyis[5]. An initial fabrication test of the electroadhesive without a dielectric layer is shown in Figure 11.

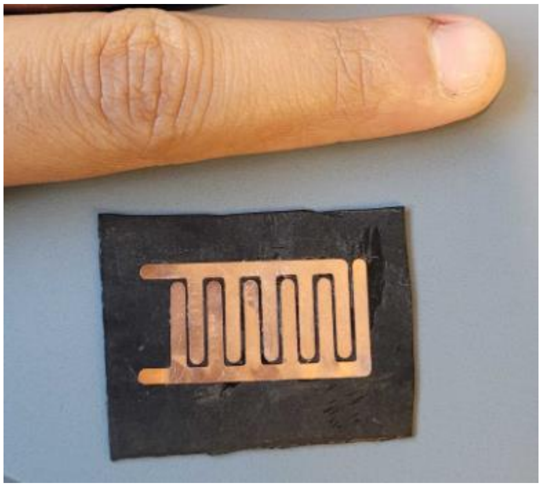

**Figure 11.** *Initial fabrication test of electroadhesive device.* This proof-of-concept trial was used to verify the efficacy of the fabrication process. A median male finger is shown for size comparison.

The overall dimensions of the device have not been finalized as the current focus is to ensure that the device operates as expected to as well as to identify areas for optimization through rapid prototyping of devices with varying dimensions.

## FUTURE WORK

Moving forward in the development of this device, I plan to push the limit of the current manufacturing method to achieve the smallest possible dimensions with the same ratio as the prototype described above. This will maximize the boundary edge length of the electrodes, improving the performance of the device. Once this limit is realized, we plan on fabricating devices of varying dimension ratios and verifying the relationship between these ratios and the resulting performance. From these verification tests, we will choose the best performing design to implement into the prosthesis. Implementation will include changing the overall size and mounting of the device and translating the current electrical subsystem into the prosthesis architecture. To further optimize the device, we plan on incorporating an H-bridge to implement

AC. The use of AC will improve the time of discharging and charging as well as mitigating residual charge buildup in the device.

## ACKNOWLEDGEMENTS

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## REFERENCES

[1] Damerla, R., Braucher, E., et. al., "Design and Testing of a Novel, High-Performance Two DoF Prosthetic Wrist", IEEE Transactions on Medical Robotics and Bionics

[2] Diller et. al, A lightweight, low-power electroadhesive clutch and spring for exoskeleton actuation, 2016 IEEE International Conference on Robotics and Automation

[3] Hinchet, Ronan, High Force Density Textile Electrostatic Clutch, Adv. Mater. Technol. 2020, 5, 1900895

[4] Choi, Kisuk et. al, Lifting-Force Maximization of a Micropatterned Electroadhesive Device Comparable to the Human-Finger Grip, ACS Appl. Electron. Mater. 2020, 2, 1596−1602

[5] Cao, Changyong et. al, Theoretical model and design of electroadhesive pad with interdigitated electrodes, Materials and Design 89 (2016) 485-491

[6] Bigharaz et. al, Increasing force generation in electroadhesive devices through modelling of novel electrode geometries, Journal of Electrostatics 109 (2021) 103540