Multi-Material Non-Planar Additive Manufacturing of Conformal Electronics on Curvilinear Surfaces

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ABSTRACT

Non-planar additive manufacturing (AM) technologies, such as microextrusion 3D printing processes, offer the ability to fabricate conformal electronics with impressive structure and function on curvilinear substrates. Although various available methods offer conformal 3D printing capability on objects with limited geometric complexity, a number of challenges remain to improve feature resolution, throughput, materials compatibility, resultant function and properties of printed components, and application to substrates of varying topography. Hence, the overall objective of this dissertation was to create new non-planar AM processes that are compatible with personalized and anatomical computer-aided design workflows for the fabrication of conformal electronics and form-fitting wearables.

After reviewing the current state of knowledge and state of the art, significant challenges in non-planar AM have been identified as: 1) limited non-planar AM path planning capability that synergizes with personalized or anatomical object surface modification, 2) limited approaches for printed and non-printed component integration on non-planar substrates. To address these challenges, a template-based reverse engineering workflow is proposed for conformal 3D printing electronics and form-fitting wearable devices on anatomical structures. This work was organized into three complementary tasks that enhance non-planar AM capabilities:
1) To achieve anatomical tissue-sensor integration, 3D scanning-based point cloud data acquisition and customized 3D printable conductive ink are proposed for capturing the topographical information of patient-specific malformations and integrating conformal sensing electronics across anatomical tissue-device interface.

2) To fabricate conformal antennas on flexible thin-film polymer substrates, a versatile method for microextrusion 3D printing of conformal antennas on thin film-based structures of random topography is proposed to control the ink deposition process across the curvilinear surfaces of freeform Kapton-based origami.

3) To simplify the fabrication process of form-fitting wearable devices with fiber-based form factors and self-powered capability, an innovative 3D printing process is proposed to achieve coaxial multi-material extrusion of metal-elastomer triboelectric fibers.

By developing new advanced non-planar printing processes and conformal toolpath programming strategies, the utility of non-planar AM could be further expanded for fabricating various personalized implantable and wearable multi-functional systems, including novel 3D electronics. In summary, this work advances capability in additive manufacturing processes by providing new advances in multi-material extrusion processes and personalized device design and manufacturing workflows.
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GENERAL AUDIENCE ABSTRACT

The ability to assemble electronic devices on three-dimensional objects with complex geometry is essential for developing next-generation wearable devices. Additive manufacturing processes, commonly referred to as 3D printing, now offer the ability to fabricate conformal electronics on surfaces and objects with non-planar geometry. This dissertation aims to expand non-planar 3D printing capabilities for applications to objects with anatomical or personalized structures, such as patient-specific malformation and origami.

The proposed methods in this dissertation are focused on addressing challenges, such as the acquisition of object 3D topographical data, material selection, and tool path programming for objects that exhibit anatomical geometry. The utility of the proposed methods is demonstrated with practical applications to 3D-printed conformal electronics and wearable devices for monitoring human behavior and organ healthcare.

This dissertation contributes to improving manufacturing capability and outcomes of 3D-printed form-fitting wearable and implantable devices. Future work may emphasize developing biocompatible functional ink and toolpath programming algorithms with real-time adaptation capability.
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1 Introduction

1.1 Background and motivation

The ability to manufacture three-dimensional (3D) electronics on curvilinear surfaces, known as 3D conformal electronics, is essential for developing next-generation smart skins, epidermal electronics, wearable sensors, implantable biomedical devices, and wireless transmission for the Internet of Things (IoT) systems[1-7]. Processing advances combined with advances in soft materials, functional inks, and flexible structures now enable conformal electronics fabrication by transfer printing[4, 8-10], self-assembly techniques[11-16], non-planar lithography[17-20], and laser direct structuring[2, 21-23]. However, such techniques obtain limitations with regard to substrate type and form-factors[3]. For example, while a very useful technique in various applications, transfer printing is difficult to apply to multifaceted objects that high curvature or sharp changes in surface slope. Transfer printing techniques can also involve challenges associated with robust interfacial adhesion to the substrate, long-term stability of the transferred electronics, and repeatable positioning on curved surfaces[3]. While attractive techniques, self-assembly techniques are confined to limited applications because of the requirements on the substrate’s material properties, pre-processing steps, and processing conditions. Additionally, other factors, such as the cost, material waste, resolution, material selection, manufacturing process complexity, and scalability, impact the selection of the aforementioned conformal electronics manufacturing methods. It remains a goal and challenge to create versatile computer-aided manufacturing processes for conformal electronics fabrication that synergize with personalized design applications and anatomical substrates[2].
Recently, additive manufacturing (AM) techniques have been investigated for functional material deposition on curvilinear surfaces[30-32], including aerosols[24, 25], liquid metals[9, 26, 27], metallic inks[28, 29]. For instance, inkjet printing has enabled the deposition of conductive aerosol droplets on cylindrical and orthogonal surfaces as interconnects with a resolution of 10 microns[25]. Microextrusion printing has also be leveraged to fabricate conformal antennas on both convex and concave surfaces of a hemisphere[28]. In addition, hybrid robotic embedding and 3D printing processes have achieved one-step fabrication of wearable devices, including conductive interconnects, insulating matrix, and functional non-fluidic electronic components[33]. Compared with other manufacturing processes, AM processes offer various advantages, including low cost, compatibility with a broad palette of functional materials, integration with computer-aided design processes, and programmability. Nevertheless, challenges facing non-planar AM still exist regarding ink formulation[34], controlled ink deposition[3], and matching of geometry and mechanical properties between the printed functional materials and substrates[35]. Advances in multi-material AM processes are continuing to enable the fabrication of novel high-performance conformal electronics on complex curvilinear substrates for wearable applications[31, 33, 35].

1.2 Research objectives

As discussed in Section 1.1, AM processes exhibit promising characteristics for fabricating high-performance electronics systems on anatomical surfaces and objects. Despite this potential, 3D-printed conformal electronics are still in the preliminary stages of development. Significant challenges of non-planar (conformal) AM include: 1) limited non-planar path planning capability that synergizes with personalized or anatomical object
surface modification; and 2) limited approaches for printed and non-printed component integration on non-planar substrates. The overall objective of this dissertation was to create novel non-planar AM processes that are compatible with personalized computer-aided design workflows for the fabrication of form-fitting wearable electronics. More specifically, this dissertation addresses the following three objectives:

1. **Reverse engineering of topographical information for personalized anatomical targets**: What is the best approach for acquiring bio-CAD (anatomical geometry) data for personalized anatomical targets, such as human organs and limbs (e.g., those exhibiting patient-specific malformation)?

2. **Direct write on flexible non-planar thin-film substrates**: What AM process design can achieve conformal direct write on thin-film substrates that exhibit randomly non-planar topographical features, such as grooves and wrinkles?

3. **Conformal non-fluidic electronics integration with wearable and anatomical systems**: How can reverse engineering be leveraged to guide path planning for conformal electronics integration with existing personalized wearable devices?

By achieving these three objectives, the utility of non-planar AM could be further expanded for fabricating “smart” (e.g., wireless and sensor-embedded) personalized wearable systems on diverse substrates that exhibit anatomical structure using a one-pot manufacturing approach.

### 1.3 Dissertation organization

This dissertation is organized as follows. A brief overview of this work with a detailed description of each research task is described in Chapter 2.1. A comprehensive literature
review of the current state-of-the-art in non-planar AM is provided in Chapter 2. In Chapter 3, the proposed framework of developing a personalized form-fitting tissue-device interface with integrated conformal electronics is described. Chapter 4 presents a novel manufacturing approach for fabricating conformal antennas on free-form Kapton films. A novel multi-material 3D printing process is presented in Chapter 5 for the fabrication and integration of wearable 3D constructs using co-axial flexible triboelectric fibers. Finally, Chapter 6 summarizes the contribution of this dissertation and identifies potential future research directions.
2 Research overview and literature review

2.1 Research overview

As introduced in Chapter 1, this dissertation contributes to developing non-planar AM processes that enable the design and fabrication of conformal electronics and form-fitting wearable devices based on personalized curvilinear objects. As shown in Figure 2-1, four essential components are identified for non-planar AM processes when fabricating conformal electronics, including reverse engineering of topographical data from curvilinear target objects, formulating novel functional inks, and tool path for non-planar electronic circuits design, and multi-material integration. Innovations are desired in each aspect to enable the fabrication of conformal electronics and form-fitting wearables on freeform surfaces. In this dissertation, a 3D scanning-enabled reverse engineering process is proposed to acquire the surface curvature of anatomical targets, such as patient-specific malformations and origami structures. A method for the fabrication of conformal antennas on flexible freeform substrates is provided for free-form wireless transmission. Moreover, a novel multi-material ‘ink’ system for 3D printing of flexible triboelectric sensors is provided and utilized in speech recognition and organ deformation monitoring applications.
Figure 2-1: The overview of essential components for non-planar AM processes when fabricating conformal electronics.

Three tasks have been designed to address the aforementioned challenges in conformal electronics and form-fitting wearables area described in Section 1.1 (Figure 2-2):

1) **Task 1**: To achieve anatomical tissue-sensor integration, a process based on point cloud data acquisition and conductive material 3D printing is proposed for capturing the topographical information of patient-specific malformations and integrating conformal pressure sensors across anatomical tissue-device interface.

2) **Task 2**: To fabricate conformal antennas on freeform thin-film structures, a versatile method for microextrusion 3D printing of conformal antennas is proposed and applied to randomly wrinkled and folded films, including Kapton origami substrates.
3) Task 3: To establish novel workflows for self-powering and triboelectric sensing, coaxial multi-material extrusion of elastomers and metallic wires is proposed to enable the integration capability of form-fitting wearable devices.

![Diagram of research tasks]

**Figure 2-2:** The overview of the proposed research tasks for non-planar AM processes.

With these three research tasks, the innovation of this dissertation work can be summarized as the following four aspects,

1. First, this work presents a novel framework and hybrid additive manufacturing processes to fabricate anatomical human-machine interface (AHMI) with integrated conformal sensing components by the proposed scanning-enabled computer-aided design process.

2. Second, it reports a versatile method for microextrusion 3D printing of conformal antennas on freeform thin-film structures.

3. Third, a novel 3D printing process is developed for simultaneously coaxial multi-material extrusion and in-process assembly of triboelectric fibers.
Finally, it provides a critical review of 3D-printed conformal electronics and wearable devices for monitoring human behavior, organ healthcare, and wireless transmission, as well as associated fabrication processes.

2.2 Literature review

In this section, a review of the current state-of-the-art in non-planar AM is introduced. Specifically, recent progress in non-planar AM using functional inks (e.g., conductive inks) is described in Section 2.2.1. Subsequently, Section 2.2.2 discusses existing conformal toolpath programming methodologies. Trends, future directions, and challenges in the field of conformal electronics AM are described in 2.2.3.

2.2.1 Recent progress in non-planar AM for flexible electronics

Non-planar AM, which is also called conformal printing, is a technology to deposit functional materials conformally on curvilinear surfaces in a non-planar fashion. Comparing with the conventional 3D printing process, the advantages of non-planar AM include but not limited to improving the surfaces smoothness with fewer efforts in post-processing[32, 36], reducing the total mass and assembly complexity of the wearable devices[37], and enhancing the mechanical properties of printed products[38]. As shown in Figure 2-3, the existing non-planar AM techniques can be mainly categorized into three groups based on the nature of the material delivery process, including inkjet-based printing, microextrusion-based printing, and hybrid printing. Although laser-based printing has also been involved in fabricating conformal electronics on arbitrary substrates, a platting process is generally required to build-up the conductive circuits according to the 3D patterns left after the laser ablation. In this case, laser-based printing is not included under the scope of this section.
Figure 2-3: Types of non-planar additive manufacturing approaches, including inkjet-based (a), microextrusion-based (b), and hybrid printing (c).

1. **Inkjet-based printing**: The history of using inkjet-based printing techniques to fabricate three-dimensional conductive traces on the curvilinear surface could be backtracking since 2005[39]. By dispensing the droplet of silver nitrate-based aqueous solution ink combined with the printing substrate's controllable movement, 3D conductive traces are fabricated on the outer surfaces of a small glass cylinder and the inner surface of a bugle-shaped glass cup[39]. Conventionally, inkjet printing techniques employed pulsed pressure, either controlled by thermal variation or piezoelectric deformation, to form discrete ink droplets
that could be precisely dispensed on non-planar substrates. However, the limited quality of the final product has been observed because of the low spatial resolution and spreading of the low viscosity inks on curved surfaces[40, 41]. The Aerosol Jet (AJ) process is one of the most popular inkjet-based 3D printing processes that enabled fabricating interconnects and conductive traces with feature sizes ranging from 10 µm to several centimeters on non-planar surfaces, such as stepped, curved, and orthogonal surfaces[24, 42]. The AJ printing technique firstly creates aerosol ink droplets by atomizing the functional inks with pneumatic and ultrasonic methods. It then transfers them into a tightly controlled beam by collimating aerosol mist via the aerodynamic focusing technique. By dispensing the focused beam of ink-laden microdroplets onto substrates, conductive traces could be conformally fabricated on arbitrary curvilinear objects. To date, a wide variety of functional inks have been developed for AJ printing, including both conductive (e.g., metallic nanoparticle-, conductive polymer-, and carbon nanotube-based inks) and insulating (e.g., dissolved polymer-, UV epoxy-, and thermoset resin-based inks)[24, 43]. However, most of the printed inks require additional thermal post-treatment to drive away ink solvent, sinter nanoparticles, or induce cross-linking among polymer chains, which highly limited the manufacturing throughput and the selection of substrate materials[24].

Electrohydrodynamic (EHD) jet printing, which uses electric fields to pull the fluids down to the substrate, has also been employed to fabricate submicron (< 1 µm) conformal electronics on non-planar surfaces[44, 45]. Thanks to the different driving force direction, EHD techniques can print 10-1000 times smaller than its nozzle size with much higher viscosity functional inks (1-10000 cP), compared with other inkjet-based printing techniques[44, 46-48]. However, increasing the production rate of EHD printing
techniques is still a main challenge that limited its widespread applications, especially when using smaller diameter jets for high-resolution printing. The interference from adjacent nozzles due to an asymmetric electric field again raises problems when considering the integration of multiple printing heads[49].

(2) Microextrusion-based printing: In contrast, microextrusion 3D printing techniques deposit continuous filaments or paths of functional inks. Ink extrusion is driven by the applied pressure (i.e., pneumatic, piston-based, or mechanical screw-based). Since the last decade, microextrusion-based printing has been adopted to fabricate electrical interconnects[50], conformal antennas[28, 51], 3D structural electronics[52] in a non-planar fashion on or within convex and concave curvilinear substrates. For instance, pneumatic-driven microextrusion-based printing is utilized to generate small conformal antennas on the convex and concave surfaces of a hemispherical glass substrate with a feature size of around 100 µm[28]. Comparing to inkjet-based printing, the programmed toolpath is much essential for the geometry fidelity, conformability, and integrity of the printed traces. Hence, a detailed description of the evolvement of the toolpath programming algorithm, especially for microextrusion-based printing, is presented in Section 2.2.2. Additional, microextrusion-based printing could incorporate with a wider range of material selection with much larger range of viscosities[2, 53], including inorganic conductors (e.g., silver[54], silicon[55], CdSe[56], perovskites[57], and graphene-based 2D materials[58]), organic conductors (e.g., poly(3,4-ethylenedioxythiophene) polystyrene sulfonate (PEDOT:PSS)[54, 59]), and organic semiconductors (e.g., poly(3-hexylthiophene):[6,6]-phenyl C61-butyric acid methyl ester (P3HT:PCBM)[60]). To control the drying properties, thickness uniformity, printability, and rheological properties
of the functional inks, conductive fillers (e.g., platinum-cure silicone[61], thermoplastic polyurethane[33], and polyethylene oxide[51]) and organic solvents (e.g., dichloromethane (DCM) and N, N-dimethylformamide (DMF)[35]) are introduced to increase the intrinsic viscosity while maintaining sufficient conductivity. Recently, ionogels and ionic hydrogels are a new class of conductive functional inks that have been involved in the microextrusion-based printing process to fabricate biocompatible soft electronics for biomedical applications[62-65]. However, balancing the chemical, electrical, mechanical, and rheological properties of the functional ink formulation is still critical research interests in the field.

(3) Hybrid Printing: Hybrid printing, which combines multiple types of 3D printing platforms or with subtractive manufacturing techniques, has been introduced to fully enable the automatic fabrication of high-performance wearable electronics with arbitrary form factors[33, 66-68]. The main benefit of hybrid printing is fully taking advantage of each type and achieving effortless integration of non-printable electronic components with reduced error and minimum labor effort. For instance, the AJ printing technique is combined with the fused deposition modeling (FDM) process to achieve high-resolution 3D structural electronics[25, 69, 70]. Alexander Valentine and colleagues also developed a hybrid 3D printing platform, which combines a microextrusion-based printing platform and pressure-enabled pick-and-place (P&P) technique, to generate wearable sensor arrays and functional devices with a microcontroller, include insulating matrix, conductive interconnections, and embedded passive and active electrical components[33]. Developing novel hybrid printing platforms offers a unique opportunity to expand the implementation and manufacturing throughput of 3D printed functional electronics. However, the
significantly increasing cost for a hybrid 3D printer comparing with a conventional 3D printer could be an inhibitor for widespread applications.

2.2.2 Existing toolpath programming methodologies for non-planar AM

Conventional 3D printing fabricates a part with curvilinear surfaces by stacking layers of two-dimensional construction together, based on a layer-by-layer toolpath program through slicing the computer-aided design (CAD) model in the transverse direction. However, this approach has several limitations, such as mechanical strength[38, 71, 72], material waste[73-75], and insufficient surface finish[76-78]. Hence, conformal toolpath programming methodologies are developed for printing a 3D structure on arbitrary curvilinear surfaces as a promising alternative method to overcome these problems.

Contrary to conventional 3D printing, the z-axis coordinates in the conformal programmed toolpath are continuously altered to construct a 3D curved layer. Based on the design inputs, there are four levels of toolpath programming methodologies integrate with non-planar AM.

As shown in Figure 2-4, the Level 1 conformal toolpath is programmed only based on CAD design geometry. The objective of the Level 1 toolpath is to improve the surface finish by eliminating the stair-step effect and enhancing the mechanical properties of the 3D printed structure[79-82]. The primary approach is to slice the CAD according to the outer surface's curvature rather than a transversal plane. For instance, Alsharhan et al. fabricated thin-walled hemispherical structures with enhanced mechanical properties by programming the toolpath along the tangential line of the hemispherical cap with the nozzle oriented perpendicular to the deposition surfaces[38]. To achieve continuous material dispensing in multiple direction, 6-axis robotic arms or multi-direction rotating printing beds are commonly involved[32, 38, 79]. Moreover, the Level 2 algorithm also takes the predefined
substrate's topographical information as additional input parameters when programming the conformal toolpath[31, 83, 84]. Specifically, the Level 2 algorithm will generate two reference surfaces, one by offsetting the substrate and the other according to the CAD design's geometry. Sequentially, the toolpath will be programmed considering the slicing result from both reference surfaces with composition layers as a transition. For example, Alkadi and colleagues reported a conformal toolpath programming algorithm to fabricate a customized hexagonal-shaped part directly on a hemispherical substrate[31]. However, Level 2 conformal toolpath programming methodologies are limited to fabricating the parts on the predefined 3D substrate since both 3D models of the designed part and the predefined substrate are required.

![Levels of toolpath programming methodologies.](image)

**Figure 2-4:** Levels of toolpath programming methodologies.

To overcome the challenge in acquisition of the target surface geometry, the Level 3 conformal algorithms integrate 3D scanning tools, such as structural-light 3D scanner[85, 86], computer tomography (CT) scanner[87], laser scanner[88, 89], and reverse engineering technologies to acquire the precise topographical information from complex
freeform surfaces, such as organs[86], fingers[61], and patient-specific wound[90]. This enables a more precise determination of the conformal toolpath on an arbitrary surface with a complex trajectory design. For instance, a 3D printed conformal microfluidic device for non-invasive organ assessment is developed based on the reverse-engineered live swine kidney[86]. The accuracy and efficiency of 3D scanning technologies will highly influence the performance of the conformal toolpath. For a comprehensive introduction of 3D scanning technologies, we refer to the following reviews[91-93]. However, it is still challenging to direct printing on live tissues or changeable substrates, which may deform and move during the printing process. Hence, the Level 4 conformal toolpath programming approaches have been developed to enhance the printing quality, achieve real-time compensation and correction of the printed trajectory[35, 51, 94]. Online sensing systems combined with computer-vision and machine-learning algorithms have been employed into 3D printing platforms to online tracking and controlling the material-feeding status and the geometry fidelity of the printed structures. For instance, a vision-based in-situ inspection and control system is developed for piezoelectric-based inkjet printing process to regulate the liquid metal jetting behavior in real-time[95]. Recently, 3D printing on moving freeform surfaces has been demonstrated using microextrusion-based 3D printing, enabled by 3D scanning-based curvature acquisition system and a computer-vision-based close-loop feedback control system[51]. However, considerable computing resource, rapid respond from motion control system, and exceeding numbers of online sensors are required to achieve real-time close-loop feedback control, which limit the broad application of this type of toolpath programming methodologies.
2.2.3 Trends and challenges

In summary, recent progress in non-planar AM offers powerful capabilities with impressive structural versatility and a simple process in fabricating conformal electronics and wearable devices on arbitrary curvilinear surfaces. Although various available methods have been established in fabricating conformal electronics with different strengths in reliability, feature sizes, or material selection, none of them is without its limitations.

Inkjet-based printing is highly attractive for conformal electronics with high resolution and conductivity. However, it is compatible with relatively fewer ink formulations and heat-resistant substrates. By contrast, microextrusion-based printing on curvilinear surfaces shows great material compatibility, simpler post-process, and exceeding geometry versatility after combining with appropriate tool path planning algorithm. Nevertheless, challenges remain to exist in quality, feature resolution, and throughput. Hybrid printing, combining 3D printing techniques and subtractive manufacturing techniques, shows higher applied potential for conformal fabrication electronics with multi-material and high-resolution feature sizes. However, future work is still desired to improve the performance with reduced fabrication costs.

Moreover, new functional inks that integrate biocompatibility, flexibility, and shape programmability are still desired to meet the increasing demands for applications in biomimetic prostheses, wearable healthcare monitoring, and smart-skin sensors. Controlling the drying properties and uniformity of 3D-printed conductive traces on curvilinear surfaces is still challenging for all processes. Other challenges in the field of material types and processing challenges should be further explored to provide sufficient
electrical, mechanical, and rheological properties to improve 3D printed functional wearables' performance.

In addition, there is a continuing need in the exploration of advanced conformal toolpath programming algorithms to improve the geometry fidelity and conformability when directly printed on 3D curvilinear organic substrates. Recent advances in higher levels of toolpath programming approaches assisted by artificial intelligence (AI) are developed to continuously sense and track the state of the printing targets. However, AI integration in non-planar AM is still at an early stage. An interactive interface between AI-enabled toolpath programs and 3D printing robots is still attractive. Moreover, the throughput of the AI-enabled non-planar AM is still limited by the time consumed during surface scanning, reverse engineering, toolpath programming, and real-time compensation. Multi-printhead AM platforms and in-process integration of non-fluidic electronic components are still unconsidered in conformal toolpath programming.

2.3 Research gaps analysis

The research introduced in Section 2.2 is focused on non-planar AM approaches for conformal toolpath programming. As summarized in Section 2.2.3, although the research efforts in this area have provided various pathways to fabricate conformal electronics and wearable devices directly on curvilinear substrates, further investigation is needed to improve the performance of 3D-printed wearable electronics as well as expand the available material palette and functional material compositions. In addition, research efforts should consider complex high-personalized curvilinear targets, such as patient-specific malformation, organ surfaces, and high-flexible thin films. Therefore, the proposed research methodologies in this dissertation seek to contribute to those challenges.
by developing novel non-planar AM processes and conformal toolpath programming strategies.
3 Scanning-enabled low-cost sensor-integration of personalized prosthetics

Interfacing anatomically conformal electronic components, such as sensors, with biology substrates, is central to creating next-generation wearable systems for health care and human augmentation applications. Thus, there is a need to low costly design and manufacture personalized anatomically conformal systems, such as wearable devices and human-machine interfaces (HMIs). Here, we show that a three-dimensional (3D) scanning and 3D printing process enable the design and fabrication of a sensor-integrated anatomical human-machine interface (AHMI) in the form of personalized prosthetic hands that contain anatomically conformal electrode arrays for children affected by amniotic band syndrome, a common birth defect. A methodology for identifying optimal scanning parameters is identified based on local and global metrics of registered point cloud data quality. This method identified an optimal rotational angle step size between adjacent 3D scans. The sensitivity of the optimization process to variations in organic shape (i.e., geometry) is examined by testing other anatomical structures, including models for foot, ear, and kidney. We found that personalization of the prosthetic interface increased the tissue-prosthesis contact area by 408% relative to the non-personalized devices. Conformal 3D printing of carbon nanotube-based polymer inks across the personalized AHMI facilitated the integration of electronic components, specifically, conformal sensor arrays for measuring the pressure distribution across the AHMI (i.e., the tissue-prosthesis interface). We found that the pressure across the AHMI exhibited a non-uniform distribution and became redistributed upon activation of the prosthetic hand’s grasping action. Overall, this work
shows that integrating 3D scanning and 3D printing processes offers the ability to design and fabricate wearable systems containing sensor-integrated AHMIs.

3.1 Introduction

Additive manufacturing, also called 3D printing, has emerged as a valuable fabrication process for creating personalized and anatomical biomedical devices by incorporating medical imaging data and computer-aided design (CAD) tools[96-102]. For example, 3D printed patient-specific anatomical tracheal implants have been developed for pediatric patients born with tracheobronchomalacia[103]. 3D printed anatomical nerve regeneration pathways have also been used to regenerate mixed bifurcating peripheral nerve injuries in rats[99]. In addition to tissue regeneration applications, 3D printing has been used to fabricate patient-specific anatomical models for surgical testing applications[104-107].

Medical imaging data for 3D printing is often collected via magnetic resonance imaging (MRI) and computed tomography (CT) scanning[107-110]. 3D scanning techniques have also been used because of their relatively low cost, portability, flexibility in range and resolution, and user-friendliness[99, 111-114]. Thus, 3D scanners have been used across multiple industries, including healthcare and manufacturing, primarily for design and inspection applications[115-118]. 3D scanning techniques often differ regarding light sources, detectors, and sensing principles, but they are broadly categorized as laser- or patterned-based approaches[119-121]. Structured-light 3D scanning is a patterned-based approach for measuring an object's shape based on the projection and reflection of light patterns[119, 121]. While laser 3D scanning has been used for medical imaging applications, structured-light 3D scanning offer advantages in speed, versatility, and price[121]. Structured-light 3D scanners have also facilitated micrometer- to millimeter-
scale anatomical design of 3D printed anatomical devices[99]. Thus, structured-light 3D scanning has the potential to become a transformative tool for designing anatomically conformal systems, such as prostheses, as it now enables personalization through digital anatomical models (e.g., of a patient’s limb structure) at a lower cost and higher speed than MRI and CT scanning.

Over 1.6 million people are living with limb loss in the United States, and the number is expected to double by the year 2050[122, 123]. Vascular diseases, such as diabetes, are currently the leading cause of limb loss and account for an estimated 54% of cases[122]. In contrast, traumas resulting from events, such as car accidents and improvised explosive devices, account for an estimated 45% of cases[122]. Among 3D printing applications, prosthetic hand fabrication is an emerging area[124-126]. Prosthetic hands can be categorized as electric, myoelectric, and body-powered[124, 127]. For example, one study reviews 58 3D printed upper limb prostheses and find that electric prostheses are superior in gripping tasks because of the capability to make a range of grasp types (e.g., power, precision, hook, spherical, tripod, and lateral grip)[124]. Several researchers also focus on improving prostheses finger movement by integrating servo motors[128], new tendon routing designs[129], and innovative kinematic designs of the thumb[129]. Bionic prosthetic hands derive function from integrating electrical components with the user’s tissue, such as myoelectric control, and they often have bio-inspired geometric and mechanical designs[130]. Bionic hand reconstruction successfully restores hand function in three patients with global brachial plexus injury and lower root avulsions who had no alternative treatment[131]. However, bionic prostheses place a disproportionate economic burden on users, especially the families of children with amniotic band syndrome and
similar congenital disabilities, due to the initial cost and need to make size modifications throughout child development. For example, according to the non-profit organization Amputee Coalition, children generally need a new prosthesis every two years up to the age of 18 due to their bodies' growth[132].

Among congenital disabilities, amniotic band syndrome is widespread, occurring in approximately one of 1,000 births[133]. Amniotic band syndrome often results in limb malformation, commonly to the arm or hand[133]. Body-powered prosthetic hands have been frequently used for children with hand malformations caused by amniotic band syndrome or other congenital abnormalities because of their low cost, simplicity, maintainability relative to bionic prostheses, and a large number of designs available for long transradial amputations[127]. While molding processes offer low-cost approaches for fabricating personalized tissue-prosthesis interfaces that could potentially interfere with non-personalized prostheses, 3D printing has emerged as a disruptive manufacturing process for creating low-cost prostheses for children with amniotic band syndrome[124, 134-137]. For example, online databases have been established to support the 3D printing of low-cost prostheses for children with birth defects, such as amniotic band syndrome (e.g., www.enablingthefuture.org). In parallel, prosthetic management for hand malformations remains an active area of research[138-140], in which the child’s age and fit of the prosthesis are often discussed as factors affecting prosthesis usage and cost. However, while 3D printing can be used for rapid prototyping of low-cost prosthetic hands for children, amniotic band syndrome malformations are highly variable. Thus, personalizing generic digital models of prosthetic components could enable the fabrication of low-cost personalized prostheses for children with amniotic band syndrome.
In addition to fabricating anatomical biomedical devices[110], 3D printing enables continuous material deposition along non-planar tool paths, commonly referred to as conformal 3D printing, typically accomplished by printing support material or directly on an object[111, 130, 141-144]. While conformal 3D printing applications are abundant and still emerging, conformal 3D printing has been used to create novel conformal and bionic devices[130, 145]. For example, microextrusion conformal 3D printing has been used to fabricate organ-conforming microfluidic devices for non-invasive isolation and profiling of biomarkers from whole organs[112] and stretchable tactile sensors[146]. Thus, conformal microextrusion 3D printing approaches could enable integrating electronic features, such as sensors, across anatomical human-machine interfaces (AHMIs), such as those found in personalized wearable systems.

Here, we describe an approach to creating low-cost 3D printed personalized prostheses via an optimized 3D scanning and 3D printing method for applications to children with amniotic band syndrome. In addition to providing a methodology for optimizing 3D scanning parameter selection and designing the personalized prosthesis interface for multiple anatomical structures, conformal 3D printing is utilized to integrate conformal electrode arrays to measure the pressure distribution across the tissue-prosthesis interface during use. 3D scanning and online CAD software are used to create a body-powered prosthetic hand that contained a personalized interface for a 12-year-old child with a distal hand malformation caused by amniotic band syndrome. We find that personalization of the prosthesis geometry increased the tissue-prosthesis contact area. Ultimately, this work provides a new approach to designing and fabricating low-cost 3D printed personalized prostheses with anatomically conformal electronic interfaces. The methods reported here
can potentially fabricate optimized, personalized prostheses and wearable systems for a wide range of fundamental research and industrial applications.

3.2 Material and methodology

3.2.1 Materials

Alja-Safe™ and 300Q fast urethane resin are purchased from Smooth-On. Polymer filament (polylactic acid; PolyLite) is from Lulzbot. Assembly kits for the e-NABLE Raptor Hand are from 3D Universe. Multiwalled carbon nanotubes (CNTs) are from Cheaptubes.com. Polydimethylsiloxane (PDMS; Sylgard 184 Silicone Elastomer Kit) is from Dow Chemical. Copper tape is from 3M.

3.2.2 Reverse engineering of limb malformation via structured-light 3D scanning

Prior to 3D scanning, a cast of the participant’s hand is made using the Alja-SafeTM-300Q resin system following the vendor-provided protocol. Subsequently, the limb's polyurethane replica is scanned using a single camera, single projector structured-light 3D scanning system (SLS-2; HP). Before scanning, the system is calibrated using a 60 mm calibration grid following the vendor-provided protocol. The limb replica is scanned from a side-view with a stationary scanning system. The limb replica is manually rotated after each scan by an angle $\Delta \theta$ using a turntable. The output from each 3D scanning measurement is a point cloud $P$, starting now referred to as a scan.

3.2.3 Calculation of metrics for optimization of 3D scanning parameters

Two metrics are used to assess the quality of registered point cloud data and identify optimal 3D scanning parameters.

(1) Scan overlap ratio (SOR) as a local quality metric:
The limb replica is scanned from 0 - 360° using a constant rotational angle step size (Δθ). Δθ ranged from 5° to θ_{max}, where θ_{max} is the maximum rotational angle at which the two scans could be successfully registered using the vendor-provided auto-alignment algorithm. Thus, this procedure resulted in a set of n scans (i.e., point clouds) \( P_i \) with surface area \( A_i \) for a given value of Δθ, where \( n = 360°/Δθ \).

The overlap area between adjacent scans (e.g., a primary and secondary scan) \( P_1 \) and \( P_i \) (\( A_{1i} = A_1 \cap A_i \)) for a given value of Δθ is calculated using the following procedure, where \( A_1 \) and \( A_i \) are the respective surface areas of the primary and secondary scans. The two scans are first registered using the vendor-provided software’s auto-alignment algorithm. The data in non-intersecting regions are then removed from the second scan using the software’s post-processing toolbox’s trimming function. The primary scan and the truncated second scan \( P_i' \) of surface area \( A_i' \) are then exported to a commercially-available mesh editing software (Meshlab). The first scan (\( A_1 \)) and the trimmed second scan (\( A_i' = A_{1i} \)) are next calculated using a quality measure and computation filter for geometric computing measures within the software. This enables the calculation of SOR as \( A_{1i}/A_1 \).

Iteration of this procedure for different values of Δθ then allows the construction of a plot of SOR vs. Δθ. Given the relationship of SOR vs. Δθ could depend on the initial projector-object orientation (i.e., the scanning perspective), we define the object’s starting orientation as that which produces a primary scan of maximum surface area.

(2) Average registration error (ARE) as a global quality metric:

The effect of Δθ on the dimensional accuracy of registered 3D models is analyzed in terms of the ARE among a globally assembled set of scans \( P_i \) acquired at each Δθ. All scans \( P_i \) for a given value of Δθ are first registered using an iterative closest point (ICP) algorithm
for pairwise local alignment followed by global alignment using a global minimization algorithm that distributed the residual error among all pairs in Meshlab[147-150]. The effect of $\Delta\theta$ on the reconstructed 3D model's dimensional accuracy is then analyzed in terms of the ARE, calculated as the average residual error after the global minimization process (i.e., a global alignment)[147, 149, 150]. The ARE is normalized to facilitate comparison among different replicas based on the maximum value obtained over the ranges of $\Delta\theta$ that lead to a successful global alignment of point cloud data. The aforementioned procedure is repeated using replicas (i.e., molds) of an adult human ear (female), adult human hand (male), adult human foot (male), and adult porcine kidney (female) to examine the dependence of the scanning parameter selection across anatomical structures of varying organic shape. The molds for the ear, hand, and foot are obtained using the aforementioned molding procedure. The mold of the kidney is obtained using previously reported methods. The relationship between SOR and ARE, the respective local and global quality metrics, is then used to identify optimal scanning parameters. Here, the optimal rotational angle step size ($\Delta\theta_{opt}$) produces dimensionally accurate 3D models based on the minimum amount of required point cloud data.

3.2.4 Assessment of tissue-prosthesis contact area

The effect of personalization on the tissue-prosthesis contact area is calculated by assembly modeling with Rhinoceros (Rhino 6), which is an approach for positioning the components using absolute coordinate placement or relative position. The imported CAD models include the personalized palm component, the non-personalized palm component, and the 3D model of the participant’s hand that is constructed via 3D scanning. The prosthesis component's relative position with respect to the limb is set based on the
orientation observed via photography during use by the participant. The tissue-prosthesis contact area \( A_{contact} \) is defined as the intersection area between the digital model of the participant’s limb and each of the two palm components of the prosthetic hand.

3.2.5 Characterization of PDMS-carbon nanotube inks as pressure sensors

Polymer nanocomposite inks are prepared over the concentration range of 1 - 20 wt% in 10:1 base: agent ratio PDMS. The inks are mixed in a centrifugal mixer (ARE-310; Thinky). For testing, a 1 mm thick film of ink \( w \) is 3D printed onto glass slides using the parameters described for conformal electrode printing (we note that the 20 wt% ink is too viscous for extrusion using a digital pressure regulator and thus, is hand-printed). Samples are fabricated using inks of varying CNT filler content. Following 3D printing, the samples are cured. The PDMS-CNT inks’ resistivity is then measured using a four-point probe method (SP4-40085TRJ; Signatone) and a power supply meter (2450 SourceMeter; Keithley) at 1 A. The body resistivity associated with a sample of finite thickness \( w \) is calculated based on an infinite slice assumption using the relation \( V/Iw[\pi/\ln(2)]F(w/s) \), where \( V/I \) is the resistance (here, measured by the source meter), \( w \) is the sample thickness, \( s \) is the four point probe spacing (here, \( s = 1.33 \) mm), and \( F(w/s) \) is a correction factor that approaches unity as \( w \) approaches zero[151]. The pressure sensitivity of the 3D printed polymer electrode arrays is characterized by measuring the resistance of a single pair of electrode terminals under a range of applied forces, given the contact area remained constant throughout the measurement. For testing, a couple of polymer electrodes are 3D printed onto a glass substrate using the same parameters that are used for conformal 3D printing. Electrical contact between the electrode junctions is created by first placing copper tape across the electrode terminals. Subsequently, the resistance of the two-
electrode circuit is measured using a multimeter (Fluke, 289 True RMS Multimeter) as the applied pressure across the electrode junction is varied by placing calibrated weights from a calibration set (Neewer 205) on top of the electrodes. The applied force is calculated as the product of the mass of the weight and the acceleration due to gravity. Details on electrode area are provided in the following section.

3.2.6 Conformal 3D printing of anatomical electrode arrays

Conformal electrode arrays are 3D printed using a PDMS-CNT composite ink in 10:1 base: agent ratio PDMS. A CNT filler content of 15 wt% is used for printing. A 2D tool path for the conformal electrode arrays is first designed in a commercially-available CAD/computer-aided manufacturing (CAM) software (Rhino 6; Rhinoceros). The conformal electrode array contains one pressure sensor per metacarpal in hand. Each pressure sensor consists of a pair of 4 mm diameter pad electrodes (center-to-center distance = 4 mm) and two associated conductive leads of different lengths (26 and 15 mm). Thus, the anatomically conformal electrode array contains a total of five pressure sensors and ten electrodes. Subsequently, the 3D tool path associated with the conformal electrode array is obtained using the associated 2D tool path of the array and the personalized palm component's digital model using Rhino 6. Before printing the conformal electrode arrays, the inner surface of the personalized palm component is coated using a thin layer of PDMS (10:1 base: agent ratio). PDMS is applied using a paintbrush to smooth the surface for subsequent conformal printing and promote the PDMS-CNT ink adhesion. The ink is loaded into 3cc syringe printing barrels with a 16-gauge tapered tip to fabricate the conformal electrode arrays. The electrode arrays are then printed using a custom low-cost microextrusion 3D printing system created by mounting the microextrusion printing barrel
to the fused deposition modeling (FDM) printer already present in the low-cost plastic 3D printing systems that are used for prosthesis 3D printing. The 3D printer from Lulzbot is used in all conformal 3D printing studies. Deposition of the CNT-PDMS ink during printing is accomplished using a digital pressure regulator (DC100; Fisnar). The conformal electrodes are printed at a speed of 3.3 mm/s using a pressure of 2 psi. The personalized palm component is then heated overnight at 90 °C to cure the ink. Following curing, the diameter of the 3D printed conformal filaments is measured based on photographs of the 3D printed conformal electrode arrays using the feature measurement tool in a commercially-available image processing software (ImageJ; NIH).

3.2.7 Measurement of the pressure distribution across the AHMI

Before placing their hand in the personalized prosthesis, the dorsal side of the participant’s hand is wrapped in a flexible thin film of plastic covered with copper tape to eliminate potential effects of skin moisture on the sensor signal. Subsequently, the straps on the prosthetic hand are adjusted to fit the participant. The participant is then asked to place the prosthetic hand on a rigid flat table with the palmar side facing upward, and the wrist relaxed in a straight position, referred to as the ‘relaxed’ position. The response of each sensor (1-5) is then measured by recording the resistance across each electrode pair. The signal reported for each electrode is the average of \( n = 4 \) measurements recorded over a one-minute interval. The participant is then asked to flex their wrist, which creates a grasping and flexing action in the prosthetic hand, while the response from each pressure sensor in the array is recorded using the aforementioned procedure. This is referred to as the ‘flexed’ position. During this procedure, the fingers of the prosthetic hand are scanned.
from a side perspective to verify that personalization did not impede the body-powered grasping action.

### 3.3 Results and discussion

Amniotic band syndrome results from limb entanglement with amniotic fibers *in utero*. This condition typically results in a malformation of limbs, such as hands and feet. As shown in Figure 3-1, the premise of this work is that 3D scanning and 3D printing can facilitate the design and fabrication of low-cost personalized prosthetic hands with anatomically conformal electronic interfaces for children with amniotic band syndrome.

![Figure 3-1: The overall research framework for the developed personalized wearable systems via 3D scanning and 3D printing technology](image)

#### 3.3.1 Identification of optimal scanning parameters via local and global metrics

Having demonstrated the principle of using 3D scanning to reverse engineer 3D digital models of limb geometry, we next examine a procedure for identifying the optimal scanning parameters. While there could be multiple approaches for optimizing scanning parameters, here, we focus on identifying scanning parameters, specifically $\Delta \theta$, that produce dimensionally accurate 3D models based on the minimum amount of required point cloud data. This is an important consideration as the value of $\Delta \theta$ affects the amount...
of point cloud data generated. We remind the reader that the number of scans requiring global alignment is based directly on the value of $\Delta \theta$ as $n = 360^\circ / \Delta \theta$. As shown in Figure 3-2a and b, model assembly from point cloud data is based on the principle of collecting successive scans that provide sufficient similarities in structure for the scans to be registered (i.e., aligned). However, while many values of $\Delta \theta$ can produce overlapping primary and secondary scans, we examine whether it is possible to identify an optimal value of $\Delta \theta$ based on the objective of minimizing the amount of data needed to assemble a dimensionally accurate 3D model. Specifically, it is of interest to determine the maximum rotational angle step size ($\Delta \theta_{opt}$) between successive scans as this would minimize the total number of collected scans ($n$), and thus, the required computing power and the post-processing time. We note that this is an important consideration as previous studies using 3D scanning in 3D printing applications typically report the rotational angle step size used for reverse engineering[56, 112, 152] but do not discuss optimization of scanning parameter selection. Thus, we next examine an approach for identifying $\Delta \theta_{opt}$ for reverse engineering 3D models associated with hand malformations caused by amniotic band syndrome.
**Figure 3-2:** Methodology for selecting optimal scanning parameters based on scan quality assessment metrics and sensitivity to multiple anatomical structures.  

**a)** Description of the scan overlap ratio (SOR) as a local quality metric.  

**b)** Description of average registration error (ARE) as a global quality metric.  

**c)** The relationship between scan quality metrics and the scanning parameter of interest ($\Delta \theta$).  

**d)** Sensitivity analysis of local and global alignment thresholds to variations in anatomical structure (abbreviation: amniotic band syndrome (A.B.S.)).

As shown in Figure 3-2c, the scan overlap ratio (SOR) decreases from 1.0 to 0.45 over the range of $\Delta \theta = 0$ to $105^\circ$. We find that a step size greater than $105^\circ$ did not facilitate scan registration, which provides a threshold value for $\Delta \theta$. Thus, this is referred to as a ‘local’ threshold because it did not involve data acquired from the object’s entire form, but rather
two scans of the object separated by a rotational step $\Delta \theta$ that represent its local geometry. The data exhibits two linear regions characterized by slopes of different magnitude. As shown in Figure 3-2c, the crossover point between the two linear regions occurs at $\Delta \theta = 85^\circ$. Given changes in $\Delta \theta$ caused relatively larger changes in SOR above the crossover point (i.e., the slope of SOR vs. $\Delta \theta$ exhibits a larger absolute value above the crossover point), this suggests that the location of the crossover point could serve as $\Delta \theta_{opt}$ in applications when an assembly of a complete 3D model is not required.

For applications requiring assembly of a complete 3D model, we examine the dependence of the average registration error (ARE) on $\Delta \theta$. While SOR provides insight into scan quality based on an object’s partial geometry, ARE provides a measure of the alignment quality among a set of globally assembled scans, which influences the model’s dimensional accuracy. As shown in Figure 3-2b, the number of scans ($n$) in a complete set that is registered to generate a 3D model is dependent on $\Delta \theta$. As shown in Figure 3-2c, ARE increased relatively linearly over the range of $\Delta \theta = 15 – 55^\circ$. The study also shows that step sizes greater than $\Delta \theta = 55^\circ$ did not facilitate the convergence of the registration algorithm, which provides a second threshold for $\Delta \theta$, referred to as the global threshold in Figure 3-2c. Thus, given the global threshold occurred within the region of local stability, this suggests that the location could serve as $\Delta \theta_{opt}$, specifically the maximum step size in rotational angle.

Given that the previous methodology could be applied to a number of anatomical structures in future applications, we next examine the trends of the local and global thresholds across multiple anatomical structures. We select different structures to represent a range of geometric shape factors, length scales, and feature sizes, including an adult foot, hand, and
ear, as well as an adult porcine kidney. As shown in Figure 3-2d, the local and global thresholds varied considerably across the set of objects. The anatomical structures with the smallest features, such as the ear, exhibit the lowest relative values of local and global thresholds at 70 and 20°, respectively. We also find that the global thresholds among structures with highly dissimilar geometry (e.g., the ear and the foot) differed substantially. For example, the global thresholds for the ear and foot are 20 and 90°, respectively. This result illustrates a methodology for identifying optimal scanning parameters since it suggests that the optimal scanning parameters are dependent on the object’s size and geometry.

3.3.2 3D printing of anatomically conformal electrode arrays and sensor characterization

As shown in Figure 3-3a, we find that personalization increases the tissue-prosthesis contact area ($A_{contact}$) by 408% relative to the non-personalized design. Specifically, the tissue prosthesis contact areas of the non-personalized and personalized designs are 768 and 3,132 mm$^2$, respectively. This result suggests that personalization could reduce the pressure on a user’s limb relative to the pressure found in a non-personalized prosthesis.
Figure 3-3: Integration of anatomically conformal electrode arrays into the personalized tissue interface via conformal 3D printing. 

a) Comparison of contact area ($A_{\text{contact}}$) between personalized and non-personalized prosthetics. 
b) Schematic showing the path design of conformal electrode arrays. 
c) Experimental data showing the effect of CNT filler content on the resistivity of the PDMS-CNT ink. 
d) Photograph of the 3D printed conformal electrode array corresponding to the tool path shown in panel (b).

In addition to potentially reducing the pressure exerted on the user’s limb, increasing the tissue-prostheses contact area has implications regarding a potential improvement to prosthesis comfort and function. For example, increasing the tissue-prostheses contact area creates new opportunities for integrating components required in bionic systems, such as sensors. The ability to understand the pressure distribution across the AHMI via integrated sensors would provide useful information for understanding the biomechanics associated
with personalized wearable systems and improving their comfort and function. To illustrate the potential for integrating sensors into the prosthetic hand in a low-cost fabrication format for potential pressure mapping applications, we next design a conformal electrode array to interface with the user’s hand's dorsal side. As shown schematically in Figure 3-3b, the five pairs of conformal electrodes extended longitudinally along the hand’s metacarpals.

We next examine the effect of CNT filler content on the resistivity of cured PDMS-CNT inks based on their previous use in strain and pressure sensing applications[153-155]. The data in Figure 3-3c show the effect of CNT filler content on the PDMS-CNT ink's resistivity. We find that the resistivity of the PDMS-CNT ink decreases with increasing CNT content. As shown in Figure 3-3c, the resistivity exhibits a sharp change at a CNT content of approximately 10%. Beyond this range, the resistivity remains relatively constant at values below 1kO-m. Such values compare reasonably with those obtained from CNT filler contents of 7–8% used in previous PDMS-CNT-based pressure sensing studies[153]. In that study, they find the devices exhibit a pressure sensitivity of 500 Pa, which is smaller than the pressure associated with a small touch to the human skin (10 kPa)[153, 156, 157]. The applied forces used for testing ranges from 0–1.4 N in steps of 350 mN, with equivalent pressures ranging from 0 to 9 kPa[153], which compare reasonably with the other applications in biomonitoring and electronic skin that require high sensitivity in the low-pressure regime <10 kPa[158]. Thus, we examine the 3D printed sensors’ response over a similar range of applied forces.

While a CNT filler content of 10 wt% could have provided a useful sensor for mapping pressure distributions across the personalized interfaces of prosthetic hands based on previous research, the resulting PDMS-CMT ink did not exhibit the rheological properties
needed to facilitate conformal 3D printing of high-quality electrodes. Given conformal 3D printing involves the deposition of inks on non-flat surfaces (i.e., non-planar 3D printing), the material can potentially flow down the surface through a falling film effect (i.e., flow under a gravitational load) resulting in a poor quality of the conformally 3D printed material. For example, spreading effects due to material flow on curved substrates have been previously discussed in research on 3D printed conformal antennas[159]. Inks that exhibit Hershel-Bulkley-type rheological properties are widely accepted as ideal candidates for 3D printing, especially for conformal printing applications. For example, they enable the 3D printing of free-standing macroscopic structures on flat and non-flat surfaces whose final form after curing exhibits a high dimensional accuracy with the originating path code (i.e., digital model). In our previous work, we find that Room-Temperature-Vulcanizing (RTV) silicone exhibits yield stress sufficient to enable conformal 3D printing of form-fitting anatomical microfluidic devices for non-invasive isolation and profiling of biomarkers from organs[112]. Thus, it is of interest to determine the concentration at which the PDMS-CNT system exhibits a sharp change in viscosity or yield stress, as such would identify an optimal CNT filler content for conformal 3D printing. We find that a CNT filler content of 15 wt% resulted in inks that exhibit yield stresses capable of preventing ink flow after deposition. Thus, given a filler content of 15 wt% is also above the conductivity threshold identified in Figure 3-3c, this concentration is selected for 3D printing of the conformal electrode arrays as it provided sufficient yield stress for 3D printing and conductivity for pressure sensing based on previous work. One previous study finds that PDMS-CNT composites exhibit a sharp increase in viscosity at filler contents ranging from 3–4 wt%[160]. We note that the wide range could be due to
differences in: 1) the type and source of CNTs; 2) the concentration of the PDMS (i.e., base-hardener ratio); and 3) the processing method utilized for preparation (e.g., mixing techniques)[160]. Figure 3-3d shows a photograph of the conformally 3D printed PDMS-CNT electrode array containing five pressure sensors in the personalized 3D printed prosthetic hand.

3.3.3 Spatially-resolved pressure sensing at the personalized prosthesis-tissue interface using conformally printed sensor arrays

Figure 3-4a shows that the 3D printed electrodes function as pressure sensors over the range of 0 to 980 mN based on previous work[153]. The sensor’s response above 980 mN is not investigated in this study. As shown in Figure 3-4a, the resistance across the electrode terminals changes from $447 \pm 61$ to $7.8 \pm 0.1$ MΩ over the range of applied forces. A power-law model fits the data with a high confidence level ($R^2 = 0.92$). Having validated the anatomically conformal electrode’s ability to function as 3D printed pressure sensors in a controlled setting, we next examine the response of the electrode array integrated within the 3D printed personalized prosthetic hand to better understand the pressure distribution that occurred across the user’s limb while wearing the prosthesis and during use.
**Figure 3-4:** Sensor characterization and mapping of pressure distributions across the AHMI using the anatomically conformal sensor array. **a)** Validation of the 3D printed sensor’s dynamic range. **b)** Response of the anatomically conformal 3D printed sensor array to body-powered actuation of the prosthetic hand’s grasping action.

As shown in Figure 3-4b, the resistances, and thus, forces and pressures measured across each electrode are significantly different depending on whether the hand is in a ‘relaxed’ or ‘flexed’ position. For example, the resistance measured across each electrode pair while the participant’s wrist is relaxed (i.e., straight) ranges from $326 \pm 82$ to $15.3 \pm 8.0$ MΩ. Considering the relationship between sensor resistance and corresponding applied force shown in Figure 3-4, the minimum and maximum forces, and thus, pressures, are measured on electrodes 2 and 5, respectively. This indicates that while the geometric fit between the prosthetic hand and the user’s limb exhibits anatomical matching, the pressure distribution is non-uniform across the interface (here across the hand’s dorsal side). As shown in Figure 3-4b, a non-uniform distribution became redistributed when the participant engages the prosthetic hand into the flexed position. For example, the resistances measured exhibits a similar range of $461 \pm 21$ to $165 \pm 81$ MΩ, but the minimum and maximum pressures are found in different locations relative to the relaxed position (electrodes 3 and 2, respectively). This result suggests that while the pressure distribution remained non-uniform across the participant’s limb while the prosthetic hand is in both the relaxed and flexed positions, the forces are redistributed during the transition between the two positions. We remind the reader that these conclusions assume that any potential effects of curvature on pressure sensitivity among different sensors are negligible. To further substantiate this claim, consistency among the sensor performance is also verified by
measuring a baseline in the sensor’s response to a control load (here a light tactile load), which averaged $179 \pm 60 \, \text{M}\Omega$ across all sensors.

### 3.4 Conclusions

Here, we show that the integration of 3D scanning with 3D printing enables the personalization of low-cost prosthetic hands with anatomically conformal electronic interfaces for children with amniotic band syndrome. Specifically, a 3D scanning-CAD process is used to create a personalized palm component associated with a widely used non-personalized prosthesis that enables a form-fitting interface with the participant’s anatomy. We also report a method for the identification of optimal scanning parameters based on the use of local and global scan quality metrics. Personalization increases the tissue-prosthesis contact area, which enabled the integration of electronic components for pressure mapping through a low-cost conformal 3D printing format. We observe that the pressure distribution across the personalized tissue interface is non-uniform and redistributed during wrist flexion. Overall, this work shows that integrating 3D scanning and 3D printing processes can design and fabricate low-cost personalized and anatomical wearable systems rapidly. It also suggests that 3D scanning and 3D printing create a useful computer-aided design and manufacturing framework for improving our understanding of the effect of personalization on wearable system comfort and function.
4 Conformal 3D printing of electronics on Kapton-based origami and Kapton films of random topography

Flexible antennas, as a critical component to enable wireless transmission and communication, have also been gaining widespread interest for ubiquitous healthcare, the Internet of Things (IoT), and aerospace applications. The most common approach involves the deposition of electronic materials on planar substrates, which are subsequently formed to target freeform surfaces. However, such ‘fabricate-and-form’ techniques limit the design space and material selection. Hence, integration of topographical information of freeform targets to the fabrication processes for conformal antenna manufacturing may considerably expand conformal design opportunities. Here, we present a conformal 3D printing process for fabricating non-planar antennas on Kapton films of near-arbitrary topography. We show that structured-light scanning processes facilitate conformal microextrusion 3D printing across Kapton films that exhibit a range of topographical features, including wrinkles and folds. The utility of the process was demonstrated by conformal printing of non-planar spiral, Hilbert-curve loop, and triangular monopole antennas on randomly wrinkled and folded Kapton films and origami. These prototypic 3D-printed conformal antennas on flexible Kapton substrates suggest that reverse engineering techniques can support the design and additive manufacturing of novel 3D-printed electronics on freeform and geometric thin-film structures.

4.1 Introduction

Flexible electronics are the basis of various emerging devices and technologies, including wearable electronics[7, 161-164], electronic skins[165-169], bionic biomedical devices[145, 170, 171], soft robots[172-175], and the Internet of Things (IoT)[176-178].
Flexible antennas, as critical components for wireless transmission and communication, are receiving considerable interest based on extensive healthcare[179-185], IoT[186-188], and aerospace applications[189-191]. Various printing techniques have been utilized for the fabrication of flexible antennas, including screen-printing[192, 193], inkjet printing[194-196], and liquid metal-filled microfluidics[197, 198]. These techniques have enabled antenna fabrication on planar flexible substrates, such as polyethyleneimine (PEI)[199, 200], polyethylene terephthalate (PET)[201-203], and polydimethylsiloxane (PDMS)[204, 205]. The most common approach for non-planar antenna fabrication involves the growth, deposition, or integration of electronic materials on planar substrates, which are subsequently shaped to a non-planar or 3D geometries through a forming process (e.g., to target freeform surfaces). However, while such ‘fabricate-and-form’ techniques for flexible electronics design and manufacturing offer considerable promise, the electronic components often undergo deformation associated with the forming process, which can limit the design space and material selection. For example, conformal antenna performance can be adversely affected by substrate deformation and result in a degradation of radiation characteristics[206-208]. Hence, while flexible conformal antennas based on fabricate-and-form techniques have various advantages, the integration of topographical information from a given freeform target to the conformal antenna fabrication processes may considerably expand antenna design opportunities (e.g., regarding material selection and substrate integration).

Three-dimensional (3D) printing processes have expanded conformal electronics manufacturing capability on curvilinear substrates and freeform objects[33, 209-211]. For instance, Aerosol Jet printing enabled the fabrication of 10 μm wide conformal
interconnects on cylindrical and orthogonal surfaces[25]. Microextrusion 3D printing has also been leveraged to fabricate functional electronics on non-planar surfaces, such as curved glass substrates[28, 54, 56, 212], personalized prosthetics[213], human skin[51, 61, 62], and organs[164]. However, it remains challenging to fabricate continuous trajectories across substrates with near-arbitrary topography, such as films with randomly distributed wrinkles and folds. Although a range of functional material systems is available for flexible electronics 3D printing that offers excellent electrical properties (e.g., conductivity), such as metallic nanoparticles[211, 214], conductive polymers[59], and graphene[215, 216], such systems often require post-processing steps and exhibit poor mechanical properties (e.g., rheological or brittle behavior that offers poor processability and durability, respectively).

Here, we provide a versatile method for microextrusion 3D printing of electronics on polymer thin films that exhibit random non-planar topography. The utility of the process for conformal electronics 3D printing was demonstrated by the fabrication of conformal spiral coil, Hilbert-curve loop, and triangular monopole antennas on Kapton films and origami that contain various random topographical features (e.g., wrinkles and folds). 3D-printed conformal antennas on folded and randomly wrinkled Kapton films and origami were characterized through S11 frequency response. These prototypic 3D-printed conformal antennas on polyimide-based thin-film substrates suggest that reverse engineering techniques can support the design and additive manufacturing of novel printed 3D electronics on thin-film freeform substrates.
4.2 Materials and methods

4.2.1 Materials

Pluronic F-127 was purchased from Sigma-Aldrich. Conductive silver epoxy (AA-DUCT 907) was from Atom Adhesives. Kapton tape (polyimide film) was from Uline.

4.2.2 Fabrication of Kapton-based origami and Kapton films of random topography

Flat Kapton films that provided the basis of all structures examined in this study were prepared by depositing parallel strips of Kapton tape on a paper substrate. Kapton-based folded, wrinkled, and origami structures were prepared as follows. Folded Kapton films were inspired based on reported folding patterns[217]. Randomly wrinkled Kapton films were prepared by forming an initially flat Kapton film into a ball. Kapton origami cranes were prepared based on reported folding patterns and methods (https://www.origami.me/crane/).

4.2.3 Conformal tool path planning

Kapton-based structures were firstly imaged using a single camera-projector structured-light scanning system (HP 3D Structured Light Scanner Pro S3; HP) following our previously reported protocols[213]. The two-dimensional (2D) conductive paths associated with spiral coil, Hilbert-curve loop[218], and triangular monopole[206] antennas were designed using a commercially-available 3D computer graphics and computer-aided design (CAD) software (Rhino 6; Rhinoceros). The non-planar tool paths associated with the conformal antennas were subsequently generated from the point cloud data and 2D tool path using CAD software.
4.2.4 Non-planar path planning validation

Pluronic F127 hydrogel (30 wt.%) was prepared using de-ionized water. Red dye was added to enhance the contrast between the printed hydrogel and the Kapton film. Prior to printing, a chilled 30 wt% Pluronic F127 solution was transferred to a dispensing barrel (10 cc; Nordson EFD) and subsequently allowed to warm to room temperature (thus, forming a hydrogel). Conformal spiral coil antennas were printed with a 27-gauge taped tip (Nordson) using a custom microextrusion 3D printing system[164]. The custom 3D printing system included a three-axis robot (MPS75SL; Aerotech), a motion controller (A3200; Aerotech), and a digital pressure regulator (Ultimus V; Nordson). Printing was performed at a speed of 5 mm/s and extrusion pressure of 12 psi.

4.2.5 Conformal 3D printing of non-planar antennas

Following preparation, conductive silver epoxy was loaded into a dispensing barrel (3 cc; Nordson EFD). Conformal antennas were then printed with a 20-gauge taped nozzle (Nordson) using the aforementioned microextrusion 3D printing system at a speed of 3 mm/s and extrusion pressure of 7 psi. The 3D-printed antennas were subsequently cured at room temperature for 24 hours.

4.2.6 Characterization of 3D-printed conformal antennas on curvilinear Kapton thin-film substrates

SubMiniature version A (SMA) connectors were bonded to the 3D-printed antennas using conductive epoxy. The $S_{11}$ frequency response of the 3D-printed conformal antennas across the 0 to 3 GHz frequency range was measured using a spectrum analyzer (FSH6, Rohde & Schwarz) with a VSWR bridge and power divider (FSH-Z2, 10 MHz to 3GHz, 50 Ω; Rohde
& Schwarz). Prior to measurements, the system was calibrated with a standard calibration kit. The SMA connection and testing apparatus are shown in Figure 4-1.

Figure 4-1: SMA connections and measurement apparatus. Photographs of the SMA connection with a conformal printed Hilbert-curve loop antenna (a), triangular monopole antenna (b), and modified loop antenna on Kapton-based origami (c), with zoomed views (d, e, f, respectively). g) photographs of measurement apparatus.
4.3 Results and discussion

4.3.1 Motivation for conformal 3D printing on Kapton

Among substrates, Kapton polyimide films are high-value targets for electronics integration based on their unique combination of material properties as an engineering material. For example, Kapton exhibits a low dielectric loss factor over a wide frequency range, high tensile strength, and high-temperature tolerance[34]. Given Kapton is processed as a film, it is widely utilized in thin film-based form factors that can exhibit near-arbitrary topography. For example, Kapton films have already been widely employed in curvilinear targets, such as aircraft[219], biomedical devices[220, 221], biosensors[222].

The majority of conformal 3D printing applications have utilized simple geometric objects or required CAD models of the objects on which printing will be done. Such constraints limit the potential for materials integration with a wide range of structures and objects, such as one-of-a-kind systems, that are common in nature and various engineering applications. As shown in Figure 4-2, the conformal 3D printing process facilitates microextrusion 3D printing on thin-film structures with near-arbitrary topography based on point cloud representations of the film acquired by 3D scanning. This ‘scan-plan-print’ process eliminates the need for substrates and antenna deformation following fabrication, which is the basis of commonly used fabricate-and-form approaches involving planar substrates.
4.3.2 Validation of conformal 3D printing on wrinkled and folded Kapton films

Based on its ability to form free-standing 3D structures[143, 223-225], we first validated the non-planar tool paths to be used for conformal antenna printing by conformal 3D printing of Pluronic F127 hydrogel on randomly wrinkled Kapton film. A photograph and associated point cloud representation of the randomly wrinkled Kapton film are shown in Figure 4-3a and b, respectively. The right panel of Figure 4-3b shows the non-planar tool path associated with a projected spiral coil antenna. As shown in Figure 4-3c and d, the process enabled conformal microextrusion 3D printing of Pluronic F127 hydrogel on the Kapton film that exhibited wrinkled surfaces with random non-planar topography.

Having established the capability for conformal 3D printing of spiral paths across randomly wrinkled Kapton films, we next examined the ability to deposit material on folded Kapton films that exhibited discontinuities in the slope (see Figure 4-3e). As shown in Figure 4-3f to h, the 3D scanning-driven non-planar path planning approach also enabled conformal 3D printing of spiral paths across folded Kapton films that exhibited topographical features associated with slope discontinuities (i.e., sharp grooves and peaks) (Figure 4-3f). Photographs of the printed path are shown in Figure 4-3g and h. These data demonstrated
that the proposed methodology could achieve conformal 3D printing of a test material across Kapton films with variable 3D topography.

Figure 4-3: Path planning validation for antenna 3D printing on the Kapton film of random topographies (wrinkled and folded). a) A Photograph of the randomly crumpled Kapton film. b) Highlight the reverse-engineered film and the conformal programmed toolpath on crumpled Kapton film. c) A photograph of the conformal 3D printed Pluronic trajectories on the crumpled Kapton film with zoomed views (d). e) A Photograph of the folded
Kapton film with a star feature. f) Highlight the reverse-engineered film and the conformal programmed toolpath on the folded Kapton film. g) A photograph of the conformal 3D printed Pluronic trajectories on the folded Kapton film with zoomed views (h).

4.3.3 Conformal antenna 3D printing on non-planar wrinkled and folded Kapton films

Having validated the ability to 3D print a test material in paths associated with a commonly utilized antenna (spiral) on non-planar Kapton films that exhibit a range of topographical features, we next examined the microextrusion 3D printing of various conformal antennas on Kapton films that exhibit a range of non-planar topographical features. Conductive silver epoxy was selected for conformal antenna printing on flexible Kapton thin-film substrates because of its improved mechanical properties compared with silver nanoparticle-based inks, which are relatively brittle. As shown in Figure 4-4, a Hilbert-curve loop antenna (Figure 4-4a to c) and a triangular monopole antenna (Figure 4-4d to f) were design and printed on curved and folded Kapton films, respectively. The antenna designs were selected based on their previous use in multiband flexible antenna applications on planar substrates[206, 218]. The $S_{11}$ frequency response for each antenna is presented in Figure 4-4c and f. The Hilbert-curve loop antenna exhibited resonance at 1.39 and 2.45 GHz with $S_{11}$ less than -10 dB (see Figure 4-4c), while the monopole antenna exhibited resonance at 0.84 and 1.28 GHz with $S_{11}$ less than -10 dB (see Figure 4-4f). The measured resonant frequencies agree reasonably with previously reported values[206, 218], suggesting that the process enabled the fabrication of antennas on the flexible non-planar Kapton thin-film substrates.
Figure 4-4: Conformal 3D printed antennas and frequency response. a) A digital model of a conformal Hilbert-curve loop antenna on a folded substrate. b) A photograph of the printed Hilbert-curve loop antenna on the folded Kapton film with correlated $S_{11}$ frequency response (c). d) A digital model of a conformal triangular monopole antenna on a curved substrate. e) A photograph of the printed triangular monopolar antenna on the curved Kapton film with correlated $S_{11}$ frequency response (f).

4.3.4 Conformal antenna 3D printing on Kapton origami

In order to demonstrate the versatility of the approach for conformal antenna integration with Kapton-based structures that exhibit complex geometry, we next examined the integration of conformal antennas with Kapton origami. Electronics integration with origami structures has led to novel technologies, including a non-invasive remote-controlled miniature robot for patching stomach wounds[226], a foldable three-degrees-of-freedom force-feedback robot for human-robot interaction[227], an origami-inspired hexagonal bellow-shaped structure exhibiting peristaltic locomotion[228]. Figure 4-5a shows a photograph of a Kapton-based origami crane that served as the non-planar antenna.
substrate. The associated point cloud data, 2D tool path associated with the antenna, and resultant conformal toolpath are shown in Figure 4-5b and c. Visual inspection of the 3D-printed path (see Figure 4-5d and e) suggests that the approach provides quality conformal 3D printing of functional materials on Kapton structures with complex 3D geometry. As shown by the $S_{11}$ frequency response in Figure 4-5f, the conformal antenna on the Kapton crane origami wing exhibited a resonant frequency near 1.59 and 2.71 GHz with $S_{11}$ less than -10 dB.

**Figure 4-5:** Conformal antenna on Kapton-based origami crane. a) A photograph of the Kapton-based origami crane substrate. b) Reverse engineered digital model of the origami crane with modified loop antenna design. c) Highlight the conformal programmed toolpath based on the reverse-engineered model. d) A photograph of the conformal 3D printed antenna on the wing of an origami crane with a zoomed view (e). f) $S_{11}$ frequency response of the conformal printed antenna.
4.4 Conclusion

A novel conformal 3D printing process for antenna integration with Kapton films of near-arbitrary topography is presented. The utility of this methodology has demonstrated the fabrication of various conformal antennas on Kapton films that exhibited a range of topographical features, including wrinkles and folds. Characterization of 3D-printed conformal antenna $S_{11}$ frequency responses was provided on Kapton films of varying topography, including Kapton-based origami. Ultimately, this work shows that reverse engineering-driven additive manufacturing methods offers the ability to rapidly design and fabricate low-cost conformal electronics on freeform films.
5 3D printing of functional triboelectric fiber with coaxial core-cladding structure

In addition to take anatomical topographical information into consideration, in-process integration of the non-printed electronic components, such as sensors, metallic wires, and chips, is critical to fully automated conformal fabrication electronics and wearable devices. Here, we report an innovated microextrusion-based process for integrating metallic wires within elastomeric constructs at room temperature to rapidly fabricate triboelectric nanogenerator (TENG) devices, with various form factors including stretchable membranes, meshes, and hollow 3D structures on planar, rotating, and non-planar anatomical substrates. The triboelectric performance of single 3D-printed elastomeric metal-core silicone-copper (Cu) (cladding-core) fibers and 3D-printed membranes is quantified by cyclic loading tests, which showed maximum power densities of 31.39 and 23.94 mW m-2, respectively. The utility of the flexible silicone-Cu TENG fibers and 3D printing process is demonstrated through applications to wearable mechanosensors for organ and human activity monitoring, specifically monitoring perfused organs and speech recognition in the absence of sound production by the speaker (i.e., ‘silent speech’), respectively. In combination with machine-learning signal processing algorithms, 3D-printed wearable triboelectric mechanosensors, in the form of stretchable form-fitting meshes and membranes, enabled real-time monitoring of perfusion-induced kidney edema and speech recognition in the absence of sound production by human subjects (99% word classification accuracy). Overall, this work expands the conductive and functional materials palette for 3D printing and encourages the use of 3D-printed triboelectric devices for self-powered sensing applications in biomanufacturing, medicine, and defense.
5.1 Introduction

The use of wearable electronics has grown substantially, owing to their promising applications, ranging from healthcare monitoring to communications[229-231]. However, conventional power sources, such as rechargeable electrochemical batteries, impose limitations on device weight, size, and usage time, delaying the development and deployment of practical and sustainable wearable electronics[229]. One promising path to overcome these limitations is self-powered electronic systems based on integrated energy-harvesting components. Triboelectric nanogenerators (TENGs) have gained considerable attention for their ability to convert mechanical to electrical energy, based on triboelectrification and electrostatic induction effects[229, 232-235]. Among the multiple forms of TENGs, fiber-based TENGs (FTENG) are attractive for various applications, given fibers are fundamental elements of complicated structures[236-245]. For example, fibers can be integrated and assembled into high-dimensional structures by processes such as weaving and knitting. However, existing FTENG fabrication approaches are often complicated because of multi-step coating and spinning processes[246, 247]. Moreover, in many applications, form-fitting structures are desired for irregular shapes, challenging to fabricate using traditional fiber assembly processes. For example, in organ monitoring applications, an anatomically conforming mesh is preferable to planar textiles. However, it remains a challenge to fabricate stable high-dimensional devices with soft 1D fibers. Thus, new processes for the production and assembly of FTENG could create advanced triboelectric-based devices, such as by rapid prototyping of form-fitting wearable systems. Multi-material 3D printing processes have been leveraged extensively for the fabrication of structural electronics[248-252], bionics[145, 253-255], and wearable devices[61, 213,
256, 257]. For example, stereolithography processes have enabled the fabrication of gaming pieces composed of polymer-embedded electronics, including LEDs, microprocessors, accelerometers, and silver interconnects[249]. Micro-extrusion 3D printing processes have allowed the fabrication of bionic tissues and active 3D electronics, such as tissues that contain integrated stretchable antennas[145] and lenses containing integrated light-emitting diodes (LEDs)[56], respectively. However, the use of conductive and functional inks poses challenges to the design and fabrication of 3D-printed TENG fibers and triboelectric devices because of high resistance, high cost, high-temperature post-processing steps (e.g., sintering), and poor mechanical properties, including limited flexibility and durability under cyclical loading[258]. Thus, it is desirable to expand the conductive and functional materials palette for 3D printing processes. To overcome the aforementioned limitations associated with the use of conductive and functional inks in electronics 3D printing, studies have been conducted to explore the feasibility of robotically interweaving high-quality drawn wires with 3D-printed constructs. For example, an integrated micro-extrusion 3D printing and pick-and-place process enabled the integration of platinum wires into silicone scaffolds, which is applied to the fabrication of custom-sized nerve cuffs[68, 259]. A fused deposition modeling process with active wire integration capabilities is developed for encapsulating conductive metal wires in an extrudable matrix of styrene block copolymers[258]. While important from the perspective of integrating high-quality wires (i.e., conducting materials) with 3D-printed constructs, the use of thermoplastics[258, 260] makes the process relatively unattractive from the perspective of fabricating triboelectric devices (e.g., because of their poor performance as triboelectrically-negative materials and limited elasticity). In contrast, silicone rubber, a
widely 3D-printed material in micro-extrusion processes[68, 86, 152, 259, 261], has been widely recognized as a promising candidate for use in triboelectric systems, due to its high electronegativity, temperature-independent properties, resilience (i.e., long lifetime), and elasticity, which can generate a relatively large (and useful) charge upon contact with human skin through triboelectric effects[262-265]. However, it is currently difficult to achieve wire encapsulation within elastomeric materials, such as silicone.

Here, we present a fabrication process for the production and 3D printing of elastomeric metal-core silicone-copper (Cu) TENG fibers using a coaxial micro-extrusion process. Fabrication of 2D and 3D constructs - via 3D printing on stationary and moving substrates, including membranes, meshes, and hollow 3D structures - is demonstrated with capacitor charging and powering of LEDs. The utility of the flexible TENG fiber and 3D printing process is herein examined through applications to wearable mechanosensors to organ and human activity monitoring, specifically, monitoring of perfused organs and speech recognition in the absence of sound production by the speaker, which we refer to here as ‘silent speech.’ 3D-printed mechanosensors, in the form of anatomical organ-conforming meshes, are leveraged for real-time monitoring of perfusion-induced kidney edema, a serious problem encountered in organ preservation and transplantation. We demonstrate that 3D-printed self-powered wearable mechanosensors in the form of stretchable membranes enable speech recognition in the absence of sound production and image-based facial expression monitoring by the speaker. These prototypic applications suggest that 3D-printed elastomeric metal-core silicone-Cu TENG fibers could provide the basis for developing high-performance triboelectric devices across a range of applications, including
healthcare and human behavior monitoring, such as real-time monitoring of pain and “silent speech.”

5.2 Materials and methods

5.2.1 Materials
Silicone (SI 595 CL) is from Loctite. Cu wire (bare uninsulated, 36 and 40 AWG) and Al wire (bare uninsulated, 28 AWG) is from WesBell Electronics, Inc. Glass pins (extra fine) are from Dritz. Polylactic acid 3D printing filament (PolyLite) is from Polymaker. Phosphate-buffered saline (PBS) is from Sigma Aldrich. Ultrapure deionized water (DIW) is from a commercially available DIW system (Direct-Q 3UV; Millipore).

5.2.2 Customized manifolds design
Customized manifolds for 3D printing of elastomeric metal-core TENG fibers are designed with a computer-aided design (CAD) software (Onshape). Each manifold is designed as a hollow two-part structure. The manifold's top and bottom portions served as a source of wire (core material) and a die for passive wire feeding based on drag extruded silicone that surrounded the wire, respectively. The bottom portion of the manifold also provided structural integration with the surrounding dispensing barrel in which the elastomer is contained (see Figure 5-1a). The 3D design and engineering drawing of the customized manifold are shown in Figure 5-1b and c. The manifolds are fabricated using a commercially-available desktop 3D printer (LulzBot mini 2; LulzBot) using vendor-provided slicing software (Cura; LulzBot) and protocols (see Figure 5-1d and e).
Figure 5-1: Design of Customized Manifolds and Printing Substrates. a) Schematic of the mechanism of coaxial fiber microextrusion with customized manifolds. b) CAD model of the two-part customized manifolds. c) Design drawing of the assembled customized manifolds. Photographs of the fabrication process (d) and the finished customized manifolds (e). CAD design of customized substrates with anchor pins for fabricating cuboid- (f), star-shaped 3D structures (g), and fiber-based sensing pads (h).

5.2.3 Coaxial multi-material micro-extrusion 3D printing processes metal-core fibers

Elastomeric metal-core TENG fibers and devices are fabricated using a custom micro-extrusion 3D printing system, which consisted of a three-axis robot (MPS75SL; Aerotech), a digital pressure regulator (Ultimus V; Nordson), a motion controller (A3200; Aerotech),
and coaxial extrusion nozzle (i.e., dispensing barrel-manifold assembly). Cu wire (36 AWG), which served as the metallic core material, is first loaded on the top portion of the manifold. The preloaded manifold is then transferred into a dispensing barrel (10 cc; Nordson EFD), with an 18-gauge tapered tip. Silicone, which served as the elastomeric cladding material, is subsequently loaded in the coaxial extrusion barrel. Before printing, the metal core is anchored on the substrate by locally curing the cladding. TENG fibers are printed by a continuous extrusion of silicone using a pressure of 15 psi at a constant vertical feed rate of 2 mm/s. Following printing, the fibers are cured at room temperature. Fibers of varying composition are fabricated by changing the core material (28 AWG Al wire and 40 AWG Cu wire) and the printing nozzle diameter (16- and 20-gauge tapered tips) and appropriately modulating the extrusion pressure in the range of 10 - 20 psi.

Hollow 3D structures, specifically cylinders and cones, are fabricated by fiber 3D printing on continuously rotating glass mandrels (radius = 3 mm; frequency = 45 RPM) using a 16-gauge tapered tip and extrusion pressure of 20 psi. Following anchoring of the core material to the stationary mandrel, the structures are fabricated by fiber 3D printing with linear horizontal motion along the axial dimension of the mandrel in the presence of continuous mandrel rotation at 0.7 - 2 mm/s. Wristbands are fabricated by fiber 3D printing on a rotating polylactic acid mandrel (radius = 70 mm; frequency = 45 RPM) using a 16-gauge tapered tip, extrusion pressure of 55 psi, and printing speed of 1.4 mm/s. The printed structures are cured overnight prior to release from the mandrels.

Cuboid- and star-shaped 3D structures are fabricated on planar, stationary 3D-printed substrates that exhibited four and five anchoring pins, respectively (pin patterns are provided in Figure 5-1f and g, respectively). Substrates are designed using commercially-
available CAD software and desktop 3D printer (Onshape and LulzBot mini 2, respectively) using vendor-provided slicing software (Cura; LulzBot) and protocols. The structures are 3D printed using an 18-gauge tapered tip, extrusion pressure of 15 psi, and printing speed of 2 mm/s using a custom tool path. Changes in fiber trajectory during 3D printing are facilitated by fiber interweaving among the substrate anchoring pins by manual toolpath programming. The printed structures are cured overnight before releasing from the substrates.

TENG fiber-based membrane sensors for silent speech studies are printed on a planar substrate (see Figure 5-1h) that contained a 50 × 120 × 3 mm³ cavity and 30 edge anchors using an 18-gauge tapered tip, extrusion pressure of 15 psi, and printing speed of 2 mm/s. The toolpath consisted of a zig-zag pattern with a 45-degree inclination relative to the substrate edge. An additional layer of silicone is printed on top of the patch to smoothen the surface. The printed structures are cured overnight before releasing from the substrates.

TENG fiber-based mesh sensors for organ monitoring studies are conformally printed on 3D-printed models of porcine kidneys[86]. Conformal tool paths are manually programmed based on uniform mesh geometry that spanned the bottom half of the kidney. Glass pins are mounted to the kidney model to provide fiber anchor points within the non-planar tool path. Printing is performed using an 18-gauge tapered tip, extrusion pressure of 15 psi, and printing speed of 2 mm/s. The printed structures are cured overnight prior to release from the substrates.

5.2.4 Characterization of 3D-printed TENG fibers and devices

The cross-section and the diameter of 3D-printed fibers are characterized using a microscope (Axio Zoom. V16; ZEISS). The active material is driven by a linear motor
(LinMot E1200) in the cyclic loading tests, as shown in Figure 5-2. Fibers and wristbands are evaluated while clamped on a force plate (Vernier FP-BTA). The applied load is controlled using a sensor console (LabQuest 2) and the software (Logger Pro). Commercial acrylic plates (McMaster-Carr) are applied as moving materials. The moving frequency and contact force are maintained at 8 Hz and 50 N by a linear motor. A programmable electrometer (Keithley 6514) is used to measure the short-circuit current, open-circuit voltage, transferred charge, and current in the quantitative electrical output and the self-powered sensors studies. The data are exported in real-time by a data acquisition card (National Instrument USB-6211), LabVIEW, and Matlab.

![Schematic of Dynamic Loading Configuration](image)

**Figure 5-2:** Schematic of Dynamic Loading Configuration for Characterizing the Triboelectric Performance of 3D-printed Silicone-Cu TENG Fibers.

A custom apparatus is used to investigate the effect of humidity on the TENG performance. A humidity sensor is placed adjacent to the fiber, which is both enclosed in an environment chamber that contained a humidifier. The sensor and the humidifier are connected to the humidity controller to achieve varying setpoints in the chamber humidity.
5.2.5 Characterization of perfusion-induced organ edema via real-time monitoring of 3D-printed TENG fiber-based devices

Adult porcine kidneys are obtained from a local abattoir in strict accordance with good animal practice as defined by the relevant national and local animal welfare bodies and approved by Virginia Tech as previously reported[86]. Briefly, kidneys are dissected from the detached viscera and stored in an insulated container during transportation to the experiment site. The transportation time is 2 hours. Prior to machine perfusion, residual fat tissue around the organ is removed. The renal artery of the kidney is subsequently anastomosed to plastic tubing (7 mm diameter). The system tubing is then connected in series with a variable-speed peristaltic pump (Cole-Parmer) and feed reservoir that contained a PBS solution. Organ mechanosensing is done using 3D-printed TENG fiber-based mesh placed under the kidney, separated by a thin layer of insulating material (Parafilm). Data acquisition began five minutes prior to activating the perfusion process (i.e., initiating the flow of PBS solution) to establish an initial baseline in the sensor response. Subsequently, the kidneys (n = 3) underwent normothermic perfusion for 1 hour in a single-pass flow mode using room temperature PBS solution as the perfusate at a flow rate of 7.8 mL min\(^{-1}\). Following the 1-hour perfusion interval, the perfusate flow is stopped, and the sensor response is continuously monitored for the next hour.

5.2.6 Characterization of perfusion-induced organ edema via real-time monitoring of 3D organ surface displacement

Perfused kidneys are continuously imaged from a top-down perspective throughout the perfusion process using a single camera-projector structured-light scanning system (HP 3D Structured Light Scanner Pro S3; HP). The system is calibrated in advance following vendor-provided protocols using a 60 mm calibration grid. Scans are collected every two
minutes throughout the perfusion process, which resulted in a set of point clouds that quantify the out-of-plane displacement of the kidney during perfusion. The transient displacement is calculated as the distance between the first scan and subsequent scans. Quantification of the separation distance between the two point-clouds is performed using a commercially-available 3D CAD modeling software (Rhino 6; Rhinoceros). Specifically, a point object ($P$) is manually created above the point clouds ($S_i$), which marked the location of the midpoint of the organ in the x- and y-axes based on top-down projection. Following projection of the same point on each scan using the Project command (i.e., the projection of point $P$ on scan $S_i$ resulted in the point $P_i$), the absolute organ surface displacement ($d$) of scan $S_i$ is then defined as the distance between the projected point $P_i$ and $P_0$, where $P_0$ is the projection of point $P$ on the initial scan $S_0$. The initial organ surface level ($h$) is defined as the distance between $P_0$ and the height of the substrate on which the organ is resting as identified from the scanning data. Thus, the relative surface displacement is calculated as the ratio of the absolute height change to the original height (i.e., $d/h \times 100\%$).

5.2.7 Real-time silent speech

Studies associated with silent speech recognition are done using 3D-printed TENG fiber-based membranes affixed to the user’s face using a surgical mask. For silent speech studies, the participant said the number “three,” the letter “D,” or the word “print” silently (i.e., performing the physical act of speaking but without sound production) during which the short-circuit current ($I_{sc}$) of the membrane is continuously monitored. The participant remained still and silent with a neutral facial expression at other times throughout the experiment. An anti-aliasing filter is designed to filter signals above the Nyquist frequency
Custom Matlab scripts provided communication with the data acquisition card to obtain real-time electrical signals. $I_{sc}$ signals associated with the user’s facial muscle movements are detected using a threshold method. The thresholds are determined by the frequency-domain noise level. Measurements are acquired at a sampling rate of 1000 Hz. Acquired signals are digitally filtered with a band-pass filter (0.1 to 20 Hz) and a notch filter at 60 Hz and its harmonics. Note, the notch filter is added to provide additional filtering from 60 Hz noise associated with power outlets.

To train the machine learning algorithm, we first reduced the dimensionality of the filtered signal via Principal Component Analysis (PCA) and Individual Component Analysis (ICA). Depending on the subject’s jaw shape and muscle movements, we observed varying effectiveness of the dimensionality reduction. If the accuracy of prediction increased by 1% or greater, the dimensionality reduction step is utilized in the training algorithm. To obtain the classifier coefficients, we implemented various supervised learning methods and compared the accuracy rates. The methods included linear and quadratic discriminant analysis, linear, quadratic, and Gaussian support vector machine, and K-nearest Neighbor models. In the real-time feedback system, the machine learning model that exhibited the highest degree of accuracy in the training algorithm is used.

5.3 Results

5.3.1 3D printing processes

As shown in Figure 5-3a, a coaxial multi-material micro-extrusion process enabled 3D printing of elastomeric metal-core triboelectric fibers. Cu wire and silicone served as the metallic core and elastomeric cladding, respectively. Elastomeric metal-core silicone-Cu
fibers provide attractive materials for creating self-powered wearable triboelectric devices based on the current generated by electron transfer from proximity to or mechanical contact with biological tissue, such as skin. Coaxial micro-extrusion of elastomeric silicone-Cu fibers involved a custom wire-feed manifold that enabled passive wire drawing through terminal anchoring of the fiber on the printing substrate. The micro-extrusion nozzle served as the extrusion die and provided wire alignment and encapsulation within the elastomeric cladding. Design and fabrication details associated with the custom manifold are provided in Figure 5-1. As shown in Figure 5-3b, continuous extrusion of silicone in combination with the extruder's continuous vertical motion (i.e., vertical 3D printing) resulted in the production of metal-core elastomeric silicone-Cu fibers. A photograph of the 3D-printed silicone-Cu fiber and micrograph of the fiber cross-section is shown in Figure 5-3c and d, respectively. As shown in Figure 5-3c, the 3D-printed silicone-Cu fibers are highly flexible due to the low bending modulus of the Cu wire and the silicone's high elasticity cladding. The fiber diameter (840 ± 8 μm) reasonably approximates the inner nozzle diameter (838 μm), indicating a minimal die-swell effect during extrusion. In addition to fiber 3D printing on planar stationary substrates, elastomeric metal-core fibers are also 3D printed on continuously rotating substrates. As shown in Figure 5-3e, fiber printing on continuously rotating mandrels enabled the fabrication of 3D hollow structures, including cylinders and cones (see Figure 5-3f and g, respectively). The process is scalable through modification of the mandrel diameter, thus enabling the fabrication of wearable systems, including wristbands (see Figure 5-3h). As shown in Figure 5-3i, the process also enabled the fabrication of 3D constructs on planar stationary substrates that contained distributed anchors, including cuboid- and star-shaped structures (see Figure 5-3j and k, respectively).
The process also offered control over the fiber and core diameters. For example, elastomeric metal-core silicone-Cu fibers are fabricated across a range of outer fiber diameters from 510 to 1,560 μm that contained Cu core diameters ranging from 79 to 320 μm, respectively. Scanning electron micrographs of fiber cross-sections printed with varied core-shell sizes are shown in Figure 5-4. As shown in Figure 5-4, the process resulted in an asymmetric cladding of the core wore, which is attributed to asymmetry in the wire feed mechanism. Thus, core positioning is constrained by the die design.
Figure 5-3: Description of Elastomeric Metal-core Triboelectric Fiber 3D Printing. a) Concept of 3D printing elastomeric metal-core silicone-Cu fibers. b) Schematic illustrating micro-extrusion 3D printing of silicone-Cu fibers through a terminal anchoring process. c) Photographs of flexible 3D-printed elastomeric metal-core silicone-Cu fibers. d) Micrograph of the fiber cross-section (silicone cladding, Cu core). e) Schematic illustrating the fabrication of 3D hollow structures via 3D printing on continuously rotating substrates. Photographs of a 3D-printed hollow cylinder (f) and cone (g) triboelectric constructs. h) Demonstration of device scalability through the fabrication of triboelectric wristbands. i) Schematic illustrating 3D printing on planar substrates containing distributed anchors. Photographs of the 3D-printed triboelectric cuboid- (j) and star-shaped (k) structures.
Figure 5-4: Cross-sectional scanning electron micrographs of fibers printed with varied core-shell sizes.

5.3.2 Triboelectric performance

3D-printed elastomeric metal-core silicone-Cu fibers and devices can be characterized in various measurement formats. As wearable triboelectric mechanosensors described in the following sections, the Cu wire is directly connected to the test samples (e.g., the skin or the organ), and the triboelectric fiber worked in the contact-separation mode. The working mechanism is shown in Figure 5-5a. The skin served as the first triboelectric material and the ground, and the silicone cladding of the TENG fiber served as the second triboelectric material. The Cu core of the TENG fiber served as the electrode. Triboelectric fibers facilitate energy harvesting based on the coupling of triboelectrification and electrical induction effects. Silicone is among the most negatively charged materials in the triboelectric series[266], and thus, provides an excellent candidate for wearable triboelectric devices driven via skin contact. As shown in Figure 5-5a, contact between the skin and the silicone-Cu fiber results in electrons in the skin being attracted to the silicone layer of the TENG, because the latter lies in a more negative location in the triboelectric
series. When the skin moves away, the accumulated negative charge on silicone induces a positive charge in the Cu wire for compensation, which creates a current flow from the skin to the fiber. Similarly, when the skin re-contacts the silicone, the current returns to the skin. No current is present when equilibrium has been reached. Thus, triboelectric fibers generate alternating current associated with repetitive triboelectric charge transfer cycles, during which the electrical potential of the Cu wire increases with decreasing separation distance between the two triboelectric materials (e.g., skin and silicone). The amplitude of the induced current depends on the amount of the transferred charge and the frequency of the contact event. Static finite element simulations of the working mechanism are shown in Figure 5-5b in terms of the electrical potential distribution established by the two interacting triboelectric materials. As shown in Figure 5-5b, when the distance between the materials decreased, the electrical potential of the Cu wire increased, and vice versa.
Figure 5-5: Characterization of 3D-printed Elastomeric Metal-core Silicone-Cu TENG Fiber and Device Triboelectric Performance.  
a) Schematic illustrating the working mechanism of the 3D-printed silicone-Cu TENG fibers in the contact-separation mode.  
b) Numerical simulation of the electrical potential distribution created upon dynamic contact-separation of silicone and skin.  
c-e) Short-circuit current, transferred charge, and open-
circuit voltage of the 3D-printed silicone-Cu TENG fibers under cyclic loading (contact area = 18 mm$^2$; load = 50 N; frequency = 8 Hz).  

f) Current generated for different external loads and corresponding power densities.  
g) Short-circuit current generated in the durability tests (i.e., cyclic loading studies - contact area = 18 mm$^2$; load = 50 N; frequency = 8 Hz).  
h) Circuit for capacitor charging and LED powering using a silicone-Cu TENG fiber generator.  
i) Charging curves for two commercial capacitors using a single silicone-Cu TENG fiber.  
j) Photograph showing powering of 20 LEDs using a single silicone-Cu TENG fiber.

The triboelectric responses of single elastomeric metal-core silicone-Cu fibers (diameter = 700 µm; length = 2.5 cm) and 3D-printed wristbands (testing contact area = 1 x 1 cm$^2$) are quantified with cyclic loading tests. The triboelectric responses of the single silicone-Cu fibers and 3D-printed wristbands in terms of the short-circuit current ($I_{SC}$), transferred charge ($Q$), and open-circuit voltage ($V_{OC}$) are provided in Figure 5-5c to e and Figure 5-6a to c, respectively. The single fibers exhibited $I_{SC}$, $V_{OC}$, and $Q$ maxima of 0.38 μA, 5.75 V, and 2.65 nC, respectively, while the wristbands exhibited maxima of 0.46 μA, 8.01 V, and 3.97 nC, respectively. Studies are also conducted using various loads that allowed current flow to examine the corresponding power density ($PD$). The corresponding $PD$ is calculated as $PD = \frac{I^2R}{A}$, where $I$ is the current, $R$ is the resistance of the external load, and $A$ is the contact area. As can be seen in Figure 5-5f and 5-6d, the single TENG fibers and 3D-printed wristbands exhibit a maximum $PD$ of 31.39 and 23.94 mW m$^{-2}$, respectively. The decreased maximum $PD$ of the wristband relative to the single fiber is associated with the wristband’s relatively decreased Cu-to-silicone volume ratio, which caused a relatively lower induced charge for the same loading conditions.
Figure 5-6: Triboelectric Response of a 3D-printed Wristband. a-c) Short-circuit current ($I_{sc}$), transferred charge ($Q_{SC}$), and open-circuit voltage ($V_{oc}$) of the 3D-printed TENG. d) Currents with different external loads and corresponding power densities ($PD$) (scale bar = 2 cm). The contact area is ~100 mm$^2$.

To further verify the durability of the 3D-printed TENG fibers, a 5000-cycle loading test is performed using a force amplitude of 50 N. We selected 50 N as the applied force amplitude in the cyclic loading study based on its established use as the upper limit of the dynamic range associated with force sensors for human motion monitoring applications[267-269]. As shown in Figure 5-5g, no visible decay in $I_{SC}$ is observed after 5000 loading cycles, which indicated that the silicone-Cu TENG fibers and 3D-printed constructs could serve as reliable transducers for sensing and energy harvesting applications. We note that it is critical to examine device durability under the loading conditions expected in applications (e.g., magnitude and frequency of the dynamic load)[270, 271].

The circuit for a 3D-printed silicone-Cu TENG fiber-based energy harvester and the measured charging curves associated with the charging of two commercial capacitors using a single silicone-Cu TENG fiber are shown in Figure 5-5h and i. The voltage saturated at
55 V after 4.5 min and 10.1 min for the 0.1 and 0.22 μF capacitors, respectively. We also showed that the generated energy could be consumed instantaneously. As shown in Figure 5-5j, a single silicone-Cu TENG fiber is sufficient for powering 20 LEDs. Given biomedical applications may establish humid testing environments, we also investigated the effect of humidity on the fiber output. The experimental apparatus and temporal fiber responses are shown in Figure 5-7a and b, respectively. The data show that changes in relative humidity caused a minimal effect on the maximum short-circuit current, which decreased from 0.41 to 0.35 μA upon a relative humidity increase from 20% to 70%. Importantly, the fiber remained functional in the presence of humid environments.

![Figure 5-7](imageURL)

**Figure 5-7:** Effect of environment humidity on TENG fiber output. a) Schematic of the apparatus for measuring the effect of environmental humidity on TENG output. b) Short-circuit current generated by a single fiber under different humidities.
5.3.3 Organ monitoring

Having shown that the extruded elastomeric metal-core fibers are highly flexible, can be 3D printed in various structures, and serve as reliable TENGs, we next examined if the response of 3D-printed TENG fiber-based triboelectric devices could enable sensing of mechanical motions associated with perfusion-induced organ edema, a significant problem encountered in *ex vivo* machine perfusion-based organ preservation processes[272, 273]. In addition to a suite of conformal bioanalytical devices for non-invasive isolation of biomarkers from perfused organs and biosensors for compositional analysis of perfusate and microfluidic biopsy samples[86], next-generation *ex vivo* machine perfusion systems by necessity should incorporate low-power sensors for real-time monitoring of organ edema (*i.e.*, swelling). As shown in Figure 5-8a to c, the process enabled the fabrication of an organ-conforming mesh-based triboelectric mechanosensor by conformal 3D printing of TENG fibers on anatomical models of porcine kidneys. During the perfusion test, the contact area between the mesh and the kidney dynamically changes for three reasons. First, natural variance in the size and shape of kidneys from different animals establishes unavoidable discontinuities (*i.e.*, gaps) between the mesh and the kidneys that are tested (*i.e.*, while the mesh geometry is indeed fabricated based on a porcine kidney template, the kidney used for templating is not the same as the kidneys used for testing). Second, the fiber surfaces contain inherent variance in surface topography. Third, the organ and the silicone cladding can deform due to perfusion-induced increases in contact force. Such swelling-associated changes in the organ-mesh contact area drove electron transfer between perfused kidney and the form-fitting triboelectric sensor, thus changing the electrical potential distribution. As shown in Figure 5-8d (point cloud data), the perfused kidneys swelled ~ 20% in height over the course of a 2-hour normothermic machine-
perfusion interval. As shown in Figure 5-8e, the real-time $V_{oc}$ response of the sensor agreed reasonably with the surface displacement measured by 3D scanning (i.e., point cloud data) throughout the preservation interval, which contained a 5-minute baseline equilibration period (with no perfusion) followed by a 1-hour machine perfusion period. Following the 1-hour perfusion period, the perfusate flow is stopped and the kidney is continuously monitored throughout a further 1-hour post-perfusion period. Both the sensor $V_{oc}$ and the kidney surface displacement increased monotonically throughout the preservation period. Stopping the perfusate flow caused a continuous decrease in both signals over the course of the next 60 min. $V_{oc}$ increased to a maximum of $\sim 70$ V, which occurred at the end of the perfusion period, and ultimately stabilized at $\sim 20$ V at $t \sim 120$ min. Kidney displacement increased by a maximum of 32.7%, which similarly occurred at the end of the perfusion period and decreased to a value of 19.7% at the end of the post-perfusion period and in the absence of perfusate flow. While the total surface displacement and the voltage response reached a maximum value at the time at which the perfusate flow is stopped, the voltage response of the TENG mesh stabilized $\sim 10$ minutes prior to the maximum of total surface displacement. This result could be attributed to various factors, including reaching the upper end of the TENG device dynamic range, change in the mechanisms by which total surface displacement affects change in the device-organ contact area, or sensitivity of the TENG device to other perfusion-induced physiological changes in the perfused kidney (such as perfusion-induced injury). Given the organ swelling response is analyzed in terms of a total surface displacement, we attribute the discrepancy in rates of decrease in the sensor response and total surface displacement to changes in the top region of the kidney that did not couple with change in device-organ contact area. In
summary, the data in Figure 5-8e suggest that 3D-printed form-fitting constructs composed of silicone-Cu TENG fibers provide attractive self-powered, wearable mechanosensors for organ preservation and biomanufacturing applications, specifically real-time sensing of perfusion-induced edema.
Figure 5-8: Form-fitting Organ-conforming Stretchable TENG Fiber-based Mesh for Monitoring of Perfused Organs. a) Schematic illustrating conformal 3D printing of elastomeric metal-core TENG fibers on objects with an organic shape, specifically, a 3D-printed porcine kidney model, for fabrication of form-fitting wearable triboelectric devices. b) Photograph of the custom machine perfusion apparatus. c) Photograph of the 3D-printed kidney-conforming TENG fiber-based mesh sensor. d) Representative point cloud data acquired via 3D scanning of perfused porcine kidneys at $t = 0$ (green), 64 (purple), and 124 min (yellow). e) Real-time responses of organ displacement associated with perfusion-induced edema acquired using 3D scanning shown with the corresponding $V_{OC}$ response of the 3D-printed TENG fiber-based mesh sensor.

5.3.4 Silent speech

Having shown that silicone-Cu TENG fibers enabled real-time monitoring of small deformations in organs, we next examined monitoring of human activities that involve small amounts of motion. ‘Silent communication,’ also referred to as ‘silent speech’ or ‘silent talk[274],’ is defined as sound-free communication among humans (i.e., verbal communication among humans in the absence of sound production by the speaker). We next investigated if a wearable 3D-printed triboelectric device could reliably detect and classify user speech in the absence of sound production by the speaker without the use of image-based facial expression monitoring. As shown in Figure 5-9a, integration of a 3D-printed TENG fiber-based membrane (1.8 mm thick) in a surgical mask provided effective mechanical coupling between the device and the speaker’s face. Movement of the user’s face caused membrane deformation and thus change in device-skin contact area. Interwoven 3D-printed TENG fibers enabled the transduction of facial movements associated with silent speech to electrical response through the triboelectric effect (e.g., $I_{SC}$) (see Figure 5-9b and c). The sensor response generated by user’s facial movements are subsequently used for real-time speech recognition via filtering, feature extraction, and
classification based on the silent word spoken (the computational framework associated with speech classification is provided in Figure 5-10).

Figure 5-9: 3D-printed Stretchable Wearable TENG fiber-based Membrane for ‘Silent Speech’ (i.e., speech recognition in the absence of sound production by the speaker). a) Photograph of a human subject wearing the triboelectric membrane-integrated facemask. b) Photograph of the stretchable TENG fiber-based membrane’s integrated transduction
elements (TENG fibers) with a zoomed view (c). d-e) Highlights of the TENG fiber-based membrane orthogonal stretchability with zoomed views showing fiber orientation in the absence and presence of strain. f) Raw sensor data generated by the worn triboelectric device during a silent speech. g-i) Filtered and averaged (dotted line) triboelectric responses corresponding to silently speaking the number “three,” the letter “D,” and the word “print.” j) Accuracy of the online classification system for different training sample sizes.

**Figure 5-10**: Framework of Signal Processing for Real-time Silent Communication (scale bar = 2 cm).

As shown in Figure 5-9d and e, the 3D-printed TENG fiber-based membranes exhibited an elastic response up to engineering strains of ~20% in both vertical and horizontal directions. In contrast to single fibers, which are highly flexible but exhibited limited stretchability constrained by the elasticity of the fiber’s metallic core, 3D-printed constructs could be printed with toolpaths that resulted in highly stretchable devices (e.g., serpentine patterned). While the influence of silicone-copper adhesion is not observed in the tensile testing studies, additional material processing techniques could be used for improving silicone-copper adhesion strength for applications that may generate asymmetric strain in the core and cladding materials. For example, plasma treatment of copper and other metals can be used
to improve the adhesion strength of polymer coatings, including epoxy resins and organopolysiloxanes[275, 276].

Figure 5-9f shows the raw sensor data generated through the cumulative triboelectric effect in the sensor, which contained 60, 120, and 180 Hz noise. The filtered data corresponding to the user speaking the number “three,” the letter “D,” and the word “print” with no sound production are presented in Figure 5-9 g to i. Each sound produced a distinguishable waveform, suggesting that the platform may provide opportunities for silent speech-based communication by combination with time-series data classification methods. The observation of distinguishable signals associated with each of the three sounds (three, D, print) is consistent among multiple human subjects (n = 3).

We trained various supervised machine learning models to classify each word that is silently spoken, including Linear Discriminant Analysis, Linear Support Vector Machine (SVM), Quadratic SVM, Gaussian SVM, and K-nearest Neighbors models. Various models, including Linear SVM, Quadratic SVM, Gaussian SVM, and K-nearest Neighbors, enabled recognition of the silently spoken word (i.e., word classification in the absence of sound production by the speaker) with greater than 95% accuracy. The Linear Discriminant Analysis model exhibited the lowest word classification accuracy of 74.8%. The Quadratic SVM, Linear SVM, and K-nearest Neighbor models exhibited relatively higher word classification accuracies of 98.4, 98.4, and 98.1%, respectively. The Gaussian SVM model yielded the highest word classification accuracy of 99.2%. Figure 5-9j illustrates the effect of the training sample size on the word classification accuracy for the Gaussian SVM model. Word classification accuracies exceed 95% accuracy for training sample sizes greater than 85 samples.
5.4 Discussion

The coaxial micro-extrusion wire encapsulation process provides a unique capability for producing fibers with elastomeric claddings. Silicone and Cu are selected as the cladding and core, respectively, based on their relative positions in the triboelectric series. For example, silicone rubber is among the most negatively changed materials in the triboelectric series, while skin is among the most positively charged materials[266, 277].

The 3D-printed conformal mesh also exhibits advantages relative to the use of shape-adaptive membrane TENGs for organ monitoring applications. First, while shape-adaptive membranes can conform to objects with a simple organic shape, such as the arm, they must be crumpled or folded to conform with objects that exhibit complex organic shapes, such as internal organs[278-281]. In contrast, the 3D-printed conformal mesh can fit accurately with objects that exhibit complex organic shapes as they are fabricated based on a template of the object. Second, conductive liquids or coated metal nanowires are usually employed as electrodes to maintain both elasticity and conductivity in shape-adaptive membranes[278, 279, 281, 282], which increases the device cost and internal resistance. Alternatively, the elasticity resides in the space between adjacent fibers for the case of the 3D-printed TENG mesh, which enables the metal core to serve as the electrode and results in a relatively decreased device cost and internal resistance. Thus, elastomeric metal-core silicone-Cu TENG fibers provide attractive material properties for wearable triboelectric systems.

In addition to the large negative charge of silicone, making it an excellent candidate for triboelectric devices, silicone is an elastomer, which offers desirable mechanical properties in resultant fibers and devices. For example, as shown in Figure 5-3c and 5-9d, silicone
claddings lead to highly flexible single TENG fibers and stretchable 3D-printed triboelectric devices. As shown in Figure 5-9d and e, the 3D-printed triboelectric membranes with serpentine toolpaths exhibited an elastic response up to an engineering strain ($\Delta L/L_0$) of $\sim$20%, where $\Delta L$ is the length change and $L_0$ is the initial length. Silicone claddings also offer improved biomechanical matching characteristics relative to other fibers. For example, the Young’s modulus of silicone is 440 kPa[152], which is in the range of the Young’s modulus of skin ($E = 420 – 850$ kPa)[283] and lower than other thermoplastic claddings, such as polysulfone ($E \sim 2,600$ kPa).

The rheological properties of elastomers make them excellent candidates for micro-extrusion 3D printing. For example, elastomers can exhibit Herschel-Bulkley rheological properties, defined as a power-law fluid with yield stress. Uncured silicone elastomers can exhibit yield stresses that are sufficient to enable 3D printing of free-standing constructs. In addition to desirable rheological properties, silicone also exhibits high self-adhesion and substrate adhesion, which facilitates layer-by-layer assembly of 3D structures and conformal 3D printing, respectively. As shown in Figs. 5-3 and 5-8, silicone-Cu TENG fibers can be assembled into various structures and form factors via 3D printing, including 3D constructs and form-fitting systems. When considering the material and mechanical properties of the silicone-Cu TENG fiber system, this work represents an advance in 3D printing of fiber-based functional materials and devices, which commonly exhibit single-fiber or woven device formats composed of fibers with relatively more rigid thermoplastic claddings.

The continued, unmet demand for high-quality transplantable organs, such as kidneys, remains a driving force for the creation of novel organ preservation processes and sensors
for real-time monitoring of organs[284], from real-time organ bioanalysis to real-time characterization of organ biophysical and mechanical properties. Perfusion-induced organ edema (swelling) remains an important problem for organ preservation. While edema during the reperfusion phase is expected, excessive edema is detrimental to organ health[285]. Kidneys, as highly vascularized organs, will swell during perfusion caused by the reintroduction of fluid and tissue edema[285]. In Fig. 5-8, we showed that 3D-printed TENG fiber-based meshes enabled real-time monitoring of machine perfused kidney swelling response (up to increases of 32.7%). The observed swelling response associated with perfusion-induced edema is consistent with previous reports[285]. The advantages of this 3D-printed triboelectric sensor in an organ preservation setting are self-powering capability, form-fitting design, and a real-time monitoring capability (sampling rate = 1 kHz). The detection of organ swelling during machine perfusion could allow interventions that may lead to better organ preservation.

Systems for silent communication, such as silent speech interfaces, have various applications, including assisted communication among soldiers and individuals affected by speech-related disabilities[274, 286, 287]. A silent speech interface is traditionally defined as a device that allows speech communication without using the sound made when individuals vocalize their speech sounds, regardless of whether a sound is produced. The most used silent speech interfaces are based on simultaneous monitoring of sound production and facial expression. Thus, they require high-dimensional image data and image processing methods, which are typically computationally intensive, as well as the use of imaging systems for facial monitoring, which increases power demands, creates the need for conventional power supplies, and limits system portability and human integration.
Alternatively, 3D-printed silicone-Cu TENG fiber-based devices provide a self-powered wearable system for silent speech that offers various advantages, such as those associated with wearability, durability, power consumption, and compatibility with data-driven signal processing methods, such as machine learning. Overall, 3D-printed silicone-Cu TENG fiber-based devices provide attractive systems for silent speech without the need for sound production or image-based facial expression monitoring.

5.5 Conclusion

In conclusion, we report a process for 3D printing elastomeric metal-core TENG fibers based on the silicone-Cu system. The process enabled the in-process assembly of metallic wires and rapid prototyping of self-powered wearable triboelectric systems. The utility and sensitivity of 3D-printed silicone-Cu TENG fibers and resultant triboelectric devices are demonstrated through applications to mechanosensing in organ and human activity monitoring. 3D-printed wearable triboelectric devices and supervised learning algorithms enabled high-accuracy real-time ‘silent speech’ (i.e., speech in the absence of sound production by the user and image-based facial expression monitoring). Ultimately, the ability to 3D print elastomeric metal-core TENG fibers on stationary or moving planar and non-planar substrates has broad implications in the fabrication of wearable triboelectric devices.
6 Conclusions and outlook

This dissertation aims to explore the potential of utilizing non-planar AM to fabricate conformal electronics and wearable devices based on anatomical substrates. Several innovative 3D printing processes are developed to overcome the challenges in substrate versatility, material versatility, and sensor integration. A summary of the contribution and discussion of the future work are presented in this chapter.

6.1 Research contributions and conclusions

The major contribution associated with this dissertation is the development and application of a ‘scan-plan-print’ workflow for conformal electronics microextrusion 3D printing on anatomical objects, including organs, prosthetics, and origami. Specifically, contributions of this research could be summarized from the following perspectives (Figure 6-1):

1) **Enhancing reverse engineering:** A 3D scanning-assisted computer-aided design (CAD) process is developed to successfully acquire the topographical information of patient-specific malformation and fabricate personalized prosthetics with conformal pressure sensing electrodes at low cost (Chapter 3). To explore the influence of scanning parameters on the quality of reverse engineered model, a methodology for identifying optimal scanning parameters is determined based on local and global metrics of registered point cloud data quality (Chapter 3). The sensitivity of the optimization process to variations in organic shape (i.e., geometry) is examined by testing various anatomical structures.

2) **Expanding available material palette:** A PDMS-CNT-based conductive ink is applied for pressure sensing below 500 Pa at a tissue-prosthesis interface (Chapter
3). This material is printed conformally on the anatomical human-machine interfaces to track the change in pressure distribution under different wearing postures. In addition to newly developed functional inks, the available material palette could also be expanded by adapting with innovative printing platforms, such as conductive epoxy (Chapter 4) and coaxial elastomer-metal fibers (Chapter 5).

3) **Conformal toolpath programming**: To further explore the substrate versatility of non-planar AM, a versatile method for microextrusion 3D printing of conformal antennas on arbitrary surfaces is proposed to fabricate conformal antennas on freeform Kapton films (Chapter 4).

4) **Innovative printing platforms**: A hybrid FDM-microextrusion printing platform is proposed to fabricate low-cost plastic personalized prosthetics with integrated pressure sensors (Chapter 3). Moreover, an innovative coaxial microextrusion printing platform is presented to produce elastomeric metal-core silicone-copper (Cu) triboelectric nanogenerator-based fibers (Chapter 5). This platform is compatible with printing devices with various form factors, including membranes, meshes, and hollow 3D structures, via 3D printing on stationary and moving substrates.
6.2 Directions of future work

The utility of the newly established non-planar AM processes has been demonstrated through practical applications, including “smart” prosthetics, monitoring human behavior, organ healthcare, and origami-based wireless transmission. Future work may emphasize continuing contributions to improve the performance of 3D printing functional devices with reduced costs. A conformal toolpath with real-time adaptation capability is a vital future direction to move towards directly fabricating wearable electronics and biomedical devices on human skin. Monitoring and controlling the quality of the product for the biomanufacturing process using in-process built conformal electronics is also a valuable topic to be further explored. Compared with conventional planar printed antennas, the effect of three-dimensional structure on power transmission performance still lacks systematic investigation. Moreover, 3D printed triboelectric-based wearable devices' practicality should also be further accommodated for speech-less pain monitoring.
References


[89] K. Binder, "In situ bioprinting of the skin," Wake Forest University, 2011.


[203] Z. Kang, Y. Zhang, and M. Zhou, "AgNPs@ CNTs/Ag hybrid films on thiolated PET substrate for flexible electronics," *Chemical Engineering Journal*, vol. 368, pp. 223-234, 2019.


