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US Army Research Laboratory

Stretchable Conductive Elastomers for Soldier Biosensing Applications: Final Report

by Geoffrey Slipher, Randy Mrozek, W David Hairston, Joseph Conroy, Wosen Wolde, and William Nothwang

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Stretchable Conductive Elastomers for Soldier Biosensing Applications: Final Report

by Geoffrey Slipher and Wosen Wolde
Vehicle Technology Directorate, ARL

W David Hairston
Human Research and Engineering Directorate, ARL

Joseph Conroy and William Nothwang
Sensors and Electron Devices Directorate, ARL

Randy Mrozek
Weapons and Materials Research Directorate, ARL

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13. SUPPLEMENTARY NOTES

14. ABSTRACT

In this report, we communicate the results of a 3-year US Army Research Laboratory Director's Research Initiative project focused on exploring the feasibility of using engineered conducting elastomer materials for interfacing to the human body to collect brain electroencephalographic (EEG) signals. We present a carbon-nanofiber-filled polydimethylsiloxane conductive elastomer material solution (CNF-PDMS) whose electrical impedance can be tuned over more than 4 orders of magnitude and which exhibits a flat electrical impedance shift when subjected to quasi-static compressive strains exceeding 60%. We present experimental results for our conductive elastomer used as a substrate for EEG measurement that indicate signal transmission remains intact when subjected to large compressions in excess of 60%. Specifically, although there is degraded signal-to-noise ratio (SNR) with lower CNF fill ratios, the single-channel hierarchical discriminant component analysis classifier of EEG data acquired using the CNF-PDMS indicates adequate performance across a wide range of compressive strains. Furthermore, we present a purpose-built modular and open-architecture EEG signal-processing board, which we have integrated with a dedicated network analyzer chip for exploring the efficacy of using real-time impedance monitoring for EEG error rejection under field conditions.

15. SUBJECT TERMS

Director's Research Initiative (DRI), stretchable conductors, conductive elastomers, biosensing, electroencephalography, human-machine interface

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1. Introduction

The fundamental scientific question underpinning this Director's Research Initiative (DRI) from the outset was, Is an engineered conductive elastomer system relevant and useful to the US Army Research Laboratory's (ARL) larger biosensing and neuroscience research programs? We have answered this question in the affirmative: conductive elastomers are relevant and potentially useful. To answer this question, we first had to achieve our secondary objective of engineering an electroencephalographic (EEG) signal collection system with integrated conductive elastomer electrodes as the human-computer bioelectronic interface. We have engineered a comfortable, nonintrusive conductive elastomer contact electrode material and demonstrated its efficacy at relaying microvolt level EEG signals using baseline EEG hardware. Furthermore, we have developed an open-architecture, modular EEG signal-processing board with integrated real-time in situ electrical impedance monitoring that will enable the exploration of EEG signal error rejection in future field trials. Through the efforts of this DRI, we have positioned ARL to pursue follow-on research efforts to assemble and validate a potentially field-ready EEG system comprising our conductive elastomer material, EEG system engineering techniques, and modular EEG signal processing board.

2. Approach

Our basic approach was multidisciplinary in nature, including backgrounds in materials science, embedded electronics, neuroscience, solid mechanics, and electrical engineering. At the core of our research effort was a conductive elastomer material design loop, which consisted of a materials synthesis and processing component interacting iteratively with a materials electromechanical characterization component throughout the course of the investigation. In turn, the materials design loop was responsive to requirements and insights coming out of the neuroscience and embedded signals processing efforts. The embedded signals processing effort was, in turn, responsive to the requirements and insights coming out of the neuroscience and material design efforts, respectively. The interdependencies of our multidisciplinary program are illustrated in Fig. 1.

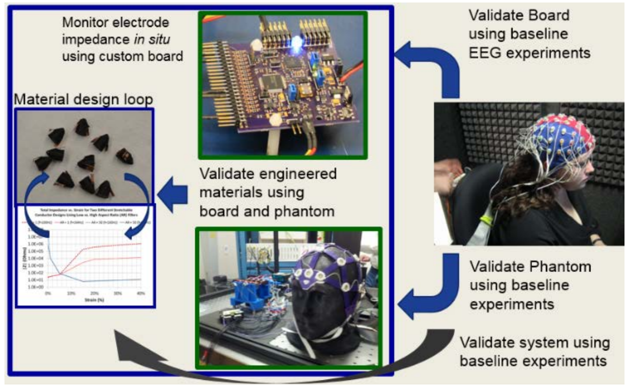

Fig. 1 Interdependencies of our multidisciplinary research approach, comprising a core material design loop interacting iteratively with parallel neuroscience/EEG and embedded signals processing efforts

Our specific work approach adapted substantially from our initial plan over the course of 3 years, and these adaptations are outlined in the following yearly summaries. Highlights of how significantly our approach adapted include 1) a transition from a hybrid tension-shear-compression loading condition to a simple compressive loading condition for the conductive elastomer interface; 2) a change from a styrene-isoprene-styrene (SIS)-based elastomer formulation to a more stable and tunable polydimethylsiloxane based elastomer formulation; 3) a transition from a nickel-based conductive filler to a more biocompatible carbon nanofiber-based conductive filler; and 4) the addition of a dedicated development effort to design, fabricate, and validate our own modular and open-architecture EEG signals processing board with integrated real-time in situ electrical impedance monitoring.

2.1 Year 1 Summary

Year 1 focused primarily on systems analysis and on identifying viable candidate material solutions and materials characterization techniques. Highlights of the acquired in-house capabilities over the course of Year 1 include demonstration of the ability to control sign of the electrical impedance response to deformation through manipulation of constituent materials, development and maturation of electromechanical characterization techniques, and development of 2 new EEG equipment validation techniques. Coming out of Year 1, our identified research priorities for Years 2/3 were as follows:

1) Evaluating our 2-dimensional (2-D) manufactured conductors in the EEG application under tensile loading.

2) Exploring alternate and hybrid materials formulations for improved electrical performance under various loading conditions.

3) Developing the necessary processing techniques to manufacture 3-dimensional (3-D) stretchable conductive components such as the scalp interface.

4) Exploring spatially graded materials; for example, the ultra-soft scalp interface for hair follicle penetration and improved signal-to-noise.

5) Refining our embedded systems for validation of EEG hardware.

6) Identifying online error detection and mitigation methods using custom signals acquisition and processing hardware.

Of the research priorities identified in Year 1, only the fourth—exploration of spatially graded materials—remains unaddressed, and we include it as a recommended potential future research direction at the end of this report.

2.2 Year 2 Summary

During Year 2 we extended the fundamental material science knowledge into the EEG system domain through experimental design, execution, and hardware design and manufacture in order to position the focus for Year 3 on answering the fundamental scientific question underpinning this DRI, Is an engineered conductive elastomer system relevant and useful to ARL's larger biosensing and neuroscience research programs? Work execution over the course of Year 2 achieved 4 main objectives: 1) identifying and evaluating alternate conductive filler materials; 2) EEG experiment design and planning; 3) establishment and evaluation of an EEG phantom simulator with our stretchable conductor materials in-the-loop; and 4) design, fabrication, and initial evaluation of a modular open-architecture EEG signals processing board. In Year 2 we were able to successfully resolve our critical conductive filler material supply issue encountered in Year 1 by exploring a variety of alternative materials and sources, eventually settling on the carbon nanofiber solution as an inexpensive and biocompatible alternative that yields similar control over material electrical impedance performance to the previously used nickel-coated fibers.

2.3 Year 3 Summary

In Year 3, we shifted our approach away from a hybrid tension-compression-based loading scheme to a simple compression-based one for the conductive elastomer. This shift coincided with a decision to focus on the scalp interface electrode as the engineered material component, and a parallel decision to integrate our conductive elastomer material into already-available commercial EEG headgear. As a result of these decisions, we also focused on developing a mold-based production method for the 3-D electrodes. We refined our new material formulation to accurately control electrical impedance over more than 4 orders-of-magnitude. We integrated and validated the real-time in situ electrical impedance measurement capability into our modular open architecture EEG signal processing board. Finally, we designed and executed an EEG classifier experiment using prerecorded human EEG to evaluate performance of our components integrated into a full EEG system while subjected to controlled levels of quasi-static compressive strain up to 60%.

3. Detailed Research Descriptions

We divided our work execution into 3 distinct, parallel, and interdependent research thrusts: 1) materials design and characterization; 2) EEG systems modularization and validation; and 3) embedded signals processing electronics design, fabrication, and validation. More detailed descriptions of each major thrust follow for the full 3-year effort. All 3 research thrusts were multidisciplinary and multidirectorate in their approach and execution.

3.1 Material Synthesis, Processing, and Characterization

Our method of rendering normally electrically insulating elastomers conductive is to fill them with conductive particles with tailored particle-particle interaction, as illustrated in Fig. 2. The series of images depicts the change in particle arrangement and orientation as the material is deformed. As a starting filler material and model system, we initially used nickel-coated carbon fibers (NCCF), which at the time were commercially available from multiple vendors and inexpensive. Magnified images of the resultant composite material and the filler used are also shown in Fig. 2. NCCFs provided an ideal model system with which to explore material processing and synthesis parameters that influence deformed elastomer electronic performance by combining the geometric uniformity of carbon fibers with the electrical performance of metallic nickel. To produce the particle-filled polymer composites, the conductive filler was incorporated into the elastomer matrix using a twin screw extruder and then pressed into sheets using a thermal press. The sheets were then cut with a laser into the final 2-D forms we worked with for Year 1. This process was completely revised for the material used to produce the scalp contact electrodes evaluated in Year 3.

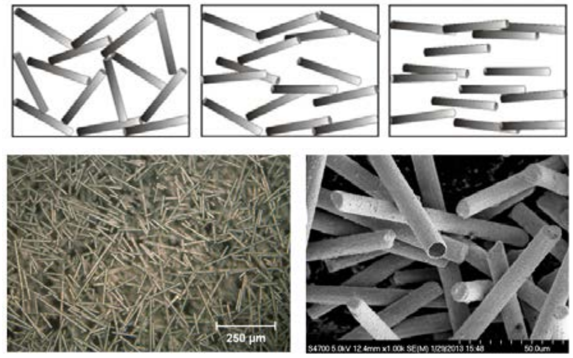

Fig. 2 (top) Conductive filler particle arrangement and orientation as the material is subjected to increasing uniaxial strain from left to right, (bott. l.) optical microscope image of 25 volume-percent NCCF dispersed in SIS, (bott. r.) SEM image of NCCFs as-received

3.1.1 Year 1

During Year 1 we were able to develop extensive knowledge about how to control conductive elastomer response to an applied deformation through manipulation of the filler material aspect ratio (AR) and the filler loading fraction. A high AR enables the material to maintain consistent electrical performance over a broad range of deformation, as illustrated in the chart at the left of Fig. 3. Also demonstrated in Fig. 3 is that AR may be used to not only tune the slope of the electrical response to applied deformation, but also the sign of the response. This is the basis of one of our 2 patent applications arising from the work of Year 1.

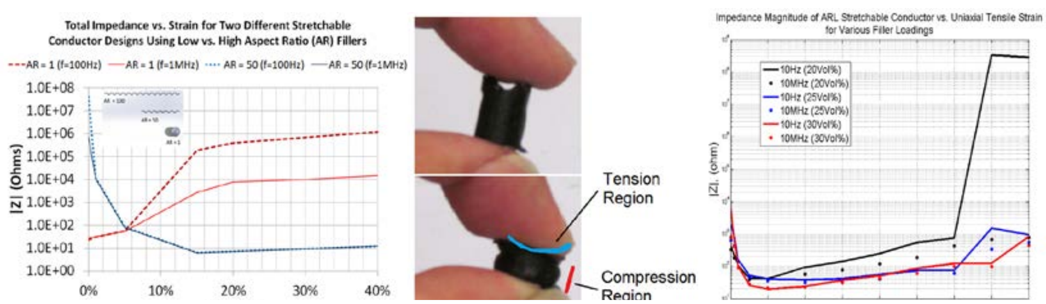

Fig. 3 (l.) Filler aspect ratio influence on sign of the impedance response to an applied deformation, (mid.) uncompressed (top) and compressed (bott.) sensor prototype illustrating the complex multidomain loading that we initially expected the material would be subjected to, (r.) effect of filler loading showing control over slope of response

From a materials engineering and component design perspective, our ability to demonstrate control over the slope and sign of the electrical response to applied deformation is the key advancement enabling us to deal with complex deformations, such as shown in the 2 images at the center of Fig. 3.

We encountered a major setback early on, when our supplier of NCCF filler material suspended production for cost reasons. They informed us that reopening their production line would require a sustained order of 10 metric tons. We subsequently identified and evaluated NCCF materials from multiple alternate vendors. The chart to the right in Fig. 4 shows a comparison of our original vendor (black trace) to 2 of the alternate vendor materials we evaluated (red and blue traces). The difference is dramatic. The response to deformation is of a different sign, and, more importantly, there is no flat region response to deformation.

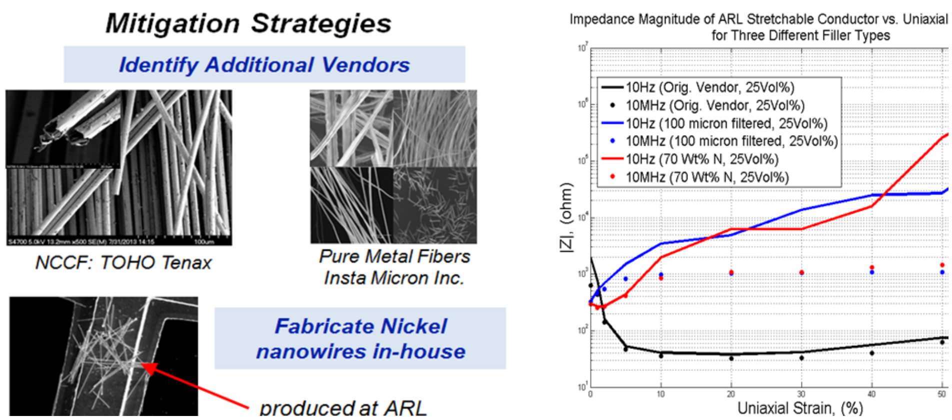

Fig. 4 (l.) Multiprong mitigation strategy; (r.) performance comparison of 2 alternate materials with original vendor material

We developed a multiprong mitigation strategy to deal with the material supply challenge. By the end of Year 1 we had engaged in a process of identifying additional sources for material, exploring the use of materials produced in-house, and in-house postprocessing commercially available materials to improve performance.

3.1.2 Year 2

Based on feedback from the Cognitive and Neuroergonomics Collaborative Technology Alliance Research Management Board (CaN CTA RMB), in Year 2 we shifted our research focus to the development of soft elastomeric materials with controlled conductivity for use as EEG electrodes to contact the scalp. Our development efforts were focused on using materials that have known biocompatibility. The elastomer is composed of a polydimethylsiloxane (PDMS) network. ARL has the in-house ability to tune PDMS mechanical properties via controlled processing over a wide range of modulus (10 kPa to 10 MPa). To the PDMS network we added biocompatible carbon nanofibers (CNF) to provide conductive pathways for the EEG signal. The CNF is inexpensive in comparison with carbon nanotubes, is commercially available in large quantities, and provides the electrical impedance tunability that we required. Representative data for resistance versus volume-percent is presented in Fig. 5 for 2 different heat treatments. In Year 2, we also completed the development of a low cost in-house casting process to produce viable dry, soft, and conductive contact electrode components that can be straightforwardly integrated into existing commercial EEG headgear. In Year 2, we completed design, manufacture, and integration of a purpose-built electromechanical characterization rig (Fig. 6) that would allow us to obtain hyperelastic electromechanical constitutive models of our materials in pure shear, simple compression, and pure tension loading conditions. We believed at the time that multimodal loading characterization would be required since we know that the material electrical performance varies based on the type of loading conditions, and since we anticipated complex hybrid loadings in use. This proved unnecessary when we switched to the simple compression loaded electrodes integrated into commercial EEG headgear for Year 3. Nevertheless, the multimodal electromechanical characterization rig developed during this research effort has proved itself a valuable research tool for ongoing mission projects, most especially the Vehicle Technology Directorate's (VTD) electric field mediated morphing wing research effort.

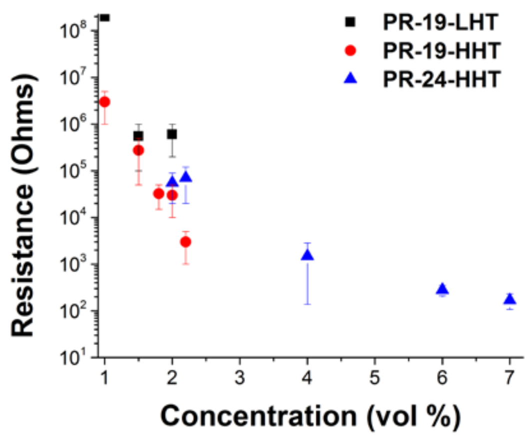

Fig. 5 Resistance values of EEG electrodes as a function of carbon nanofiber grade and concentration

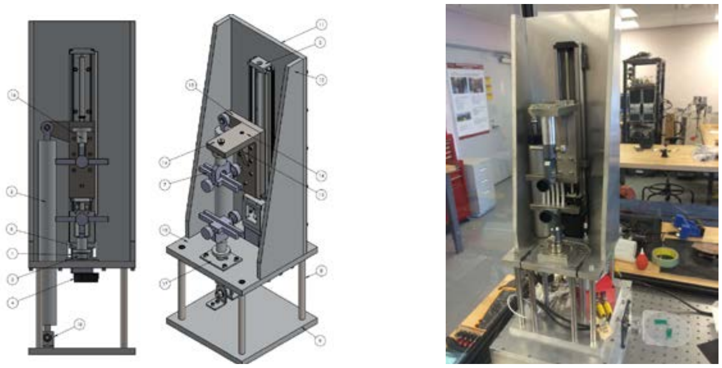

Fig. 6 Purpose built multimodal hyperelastic electromechanical characterization rig, design (left), as-built (right)

3.1.3 Year 3

Our Year 3 efforts continued the development of soft conductive elastomers for use as EEG scalp interface electrodes. The polymer-based electrodes provide practical advantages over the current state-of-the-art "wet" electrodes in both ease of application to the wearer and longer performance times. Our efforts have focused on the use of materials with known biocompatibility. Specifically, the polymer component is composed of PDMS that is an example of a group of materials commonly referred to as "silicones" that have a flexible silicone-oxygen backbone chemistry. Carbon nanofibers are mixed in with the silicone elastomer in sufficient concentrations to promote an electrically percolating network throughout the otherwise insulating material. The electrical impedance performance of the undeformed material can be tailored through the carbon nanofiber concentration over greater than 4 orders of magnitude.

The electrodes were initially composed of an end-linked PDMS with a molecular weight of 117,000 g/mol between cross-links. As the carbon nanofiber concentration increased to 2.4 vol %, the mixture became too viscous to process and represented a practical upper concentration limit. At 2.4 vol % of the PR-19-HHT grade of carbon nanofiber a resistance value on the order of several kiloohms was achieved (Fig. 5). Resistance values between 1 and 10 kΩ have been commonly cited as ideal values for dry EEG electrodes within the EEG community; however, the rationale for that value seems to have more to do with the impedance of the skin and is in contrast with the development of metal electrodes that exhibit much lower resistance values. As a result, it is desirable to extend the resistance range of the developed polymer EEG electrodes to potentially provide insight into defining an optimum electrical performance for applications in EEG sensing.

A low molecular weight nonreactive PDMS oil was added to the electrode formulations at 50 vol % relative to the reactive precursors. The addition of the

PDMS oil decreases the viscosity of the mixture and enables processing at higher carbon nanofiber concentrations. The concentration of nonreactive PDMS oil can be altered to tailor the mechanical stiffness of the resulting electrode, allowing for independent tuning of the electrical and mechanical response. Using 50 vol % of the PDMS oil, concentrations of PR-24-HHT carbon nanofiber—with the highest aspect ratio and conductivity—were able to be processed up to 7 vol %. At 7 vol %, the resistance of the electrode is reduced to approximately 10 Ω (Fig. 6). This enables evaluation of electrode performance over several orders of magnitude. In addition, it is anticipated that 7 vol % is well above the percolation threshold for the carbon nanofiber. This enables the evaluation of how proximity to the electrical percolation threshold impacts the strain-dependent electrical response.

Electrodes were fabricated for integration into commercial EEG headgear. The EEG headgear has a ring where a cylindrical sensor can be placed through the ring and conductive tabs on the sensor contact a conductive surface on the ring to provide low impedance electrical contact. To use the commercial device, the electrodes were fabricated using a clam shell design—the bottom portion is cast into the mold, copper wire or foil is then laid over the cast portion, followed by casting material in the upper portion of the mold to incorporate the copper wire or foil into the center of the electrode (Fig. 7). The copper wires could easily be removed from the sensor with slight pressure so the focus was shifted to the copper foil. Ultimately, it was determined that producing simple cylinders of the material (no embedded metal) produced sufficient contact with the device, provided the diameter was large enough to fit snugly within the device rings.

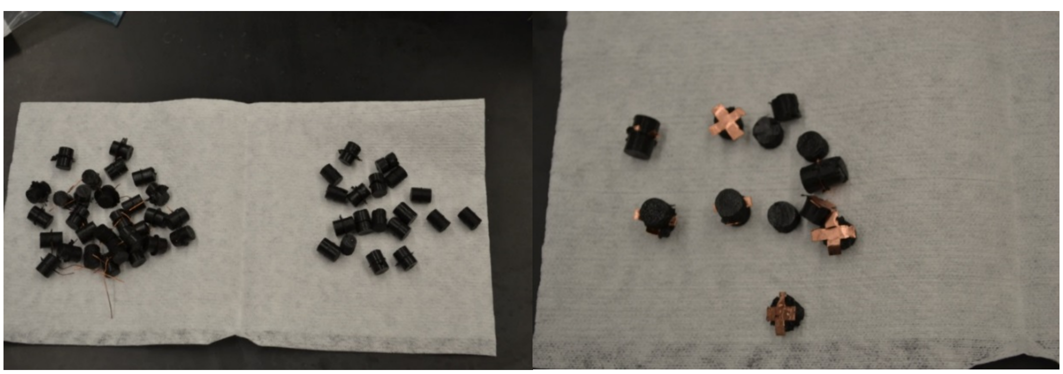

Fig. 7 Photographs of an early batch of the produced electrodes with embedded copper wire (left) and copper foil (right)

Once the synthesis and processing methods for the CNF-PDMS materials were refined, batches of electrodes were produced in 3 volume loadings (3 vol %, 4 vol %, and 7 vol %). These electrodes were characterized electromechanically under simple compressive loading using a E5061B LF-RF network analyzer (NA) supplied by Keysight for the electrical interrogation, and a compressive-tensile stage with a 20-N load cell for the mechanical loading. Representative electromechanical data is presented in Fig. 8. The 4 vol % CNF loading provides the best compromise of softness and electrical impedance response under large compressive deformation. In the next section, we will demonstrate that the 4 vol % also performed well in our EEG classifier evaluations. Of note is the significant observation that, in all cases, increasing compressive strain is associated with a reduction in electrical impedance. The degree of reduction in electrical impedance response to applied compression decreases with increasing CNF volume fraction.

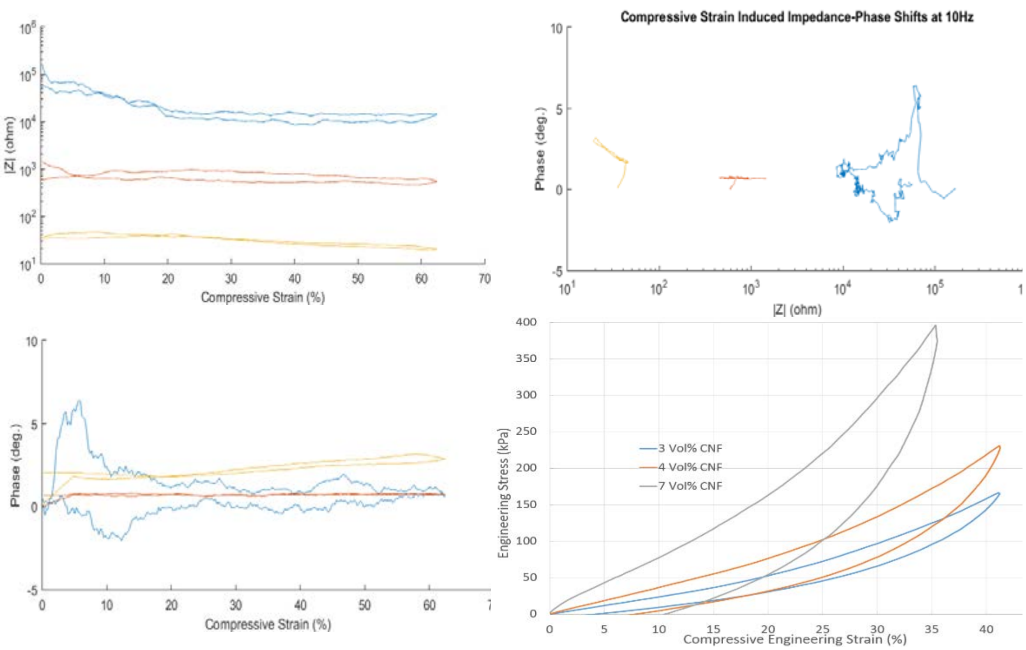

Fig. 8 Electromechanical testing results for 3 different CNF-PDMS electrode formulations: 3 vol %, yellow; 4 vol %, red; 7 vol % orange. NA scan at 10 Hz of impedance magnitude vs. compressive strain over a single loading/unloading cycle (top left), NA scan at 10 Hz of current-voltage phase shift vs. compressive strain over a single loading/unloading cycle (bottom left), NA impedance and phase data at 10 Hz plotted together (top right), and engineering stress-strain data (bottom right).

3.2 EEG System Modularization and Validation

3.2.1 Year 1

The primary focus for Year 1 was to advise and inform the materials development and characterization effort as it spooled up, as well as to establish clear neuroscience/EEG objectives for the remaining 2 years. As we began this project, we were surprised to find that no current standards exist for validating EEG hardware. In Year 1, we identified 2 strategies required for validation: 1) a phantom head and 2) a modular EEG system. We began development of this validation hardware in Year 1 (Fig. 9). Our EEG phantom is capable of generating EEG-level signals to validate full EEG systems on a simulated human head. Our modular EEG system is capable of swapping out individual EEG components, such as our stretchable conductor materials, to isolate individual component influence on overall system performance. These 2 validation systems were further developed in Year 2.

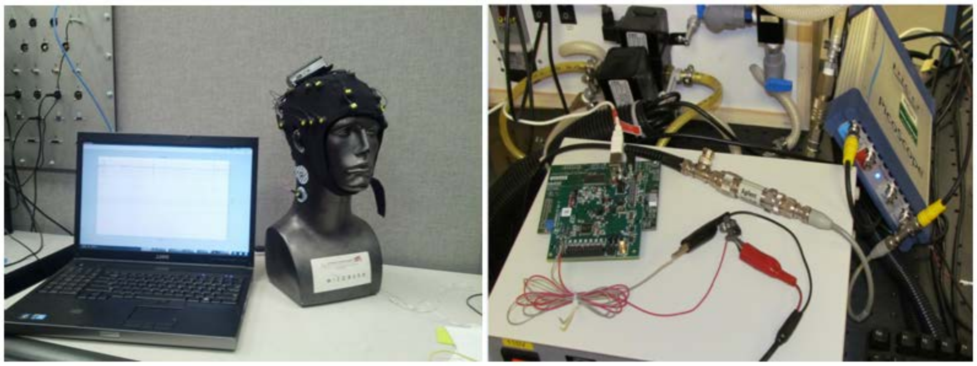

Fig. 9 Our 2 EEG system validation tools: an early version of the EEG phantom (left) and an early version of our modular EEG system (right)

3.2.2 Year 2

An important focus for Year 2 was the development of a protocol and platform for validating efficacy of the materials when used as part of a complete EEG system. Efforts in this domain covered 2 main aspects: 1) development of our own EEG phantom device and other modular hardware components and 2) modular EEG experimental feasibility demonstration using nonoptimized stretchable conductive materials on-hand. Material optimization was achieved during Year 3. The "phantom" design involved the fabrication of a scale and feature accurate replica of an ARL staff-member's head based on MRI data acquired by ARL, 3-D printing of the head in-house, and subsequent conductive coating and instrumentation of the head for use in the modular EEG hardware setup (see Fig. 10). For an initial feasibility test of putative materials used as part of an EEG system, ARL-developed conductive materials subjected to quasi-static tensile strains up to 70% were used as an in-line replacement for the native wiring for 4 channels. The whole modular EEG system was tested using signals transmitted through the phantom head and into the EEG hardware, and acquired using a commercial EEG data acquisition board as a placeholder while our own in-house board was fabricated. The stretchable conductors were evaluated at controlled quasi-static tensile strains between 0% and 70% during EEG signal acquisition. Representative experimental results are presented in Fig. 10 and show the signal correlation of the stretchable channels with the control channels (where a value of one indicates perfect correlation). Importantly, the effects of strain are shown to be sharply stochastic

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(switch-like) in nature, and there is no notable degradation in the signal correlation domain as long as signal transmission is successful, even with significant signal amplitude attenuation. This is important for EEG, where amplitude modulation of the materials during dynamic deformation under field use conditions would otherwise pose challenges to interpretation and usability of the data. The importance of material consistency arising out of well-controlled and understood materials processing is also underscored in the channel-to-channel variation apparent in these initial data, and thus justifies the significant effort expended toward that end as part of this research program.

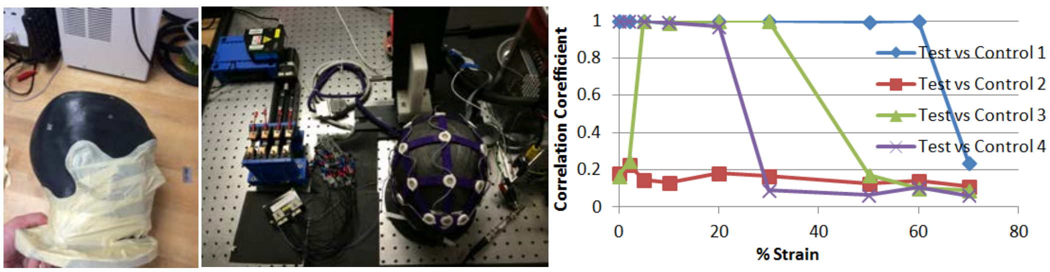

Fig. 10 ARL in-house phantom production (left), EEG modular experiment (center), representative correlation data (right)

3.2.3 Year 3

3.2.3.1 Compression-Related Signal Modulation with Real EEG

A primary focus for the final year was an initial examination of the efficacy of the materials when used as part of a complete system, particularly when formed into 3-D structures to be used as an EEG electrode. Conventional EEG electrodes do not contact the skin and, therefore, rely on electrolytic gels to create the necessary electrical bridge; however, a "dry" design as proposed here requires direct mechanical contact. When in use, maintaining good electrical contact requires some degree of force to stay in place, thus compressing the material, while movement of the subject could also cause subsequent deformation of the material. As a result, it is important to ascertain how deformation of the conductive elastomer influences the effectiveness of signal transduction when used specifically for EEG signals; while we know from the material electrical characterization work described above how mechanical deformation alters electrical impedance through the material, it is not clear how this relates to the actual functional usability within the EEG target application domain.

3.2.3.2 Validation Methods for EEG

Material samples were placed in compressive strain as described above. During each level of compression, a single channel of EEG data from a prior recorded

6-min session was reproduced using a NI USB-6356 and passed through the sample. The reproduced data originate from scalp location P03 of a randomly chosen human subject performing a rapid serial visual presentation (RSVP) task (Cecotti et al. 2015). In this task, images were presented at a rate of 2 Hz, which included a randomized assortment of scene images from a simulated Middle Eastern village. Some images also contained an image of a person holding a weapon; the subject pressed a button when he saw these target images. Reproduced data were recorded using an EEG system based on the ADS1298 chipset by directly connecting the recording system to the sample. In this manner, the general recording setup was roughly similar to what would be used in a typical setting but without variable or noise-inducing components, such as the human subject or conductive gels. In order to assess signal efficacy, we examined 2 levels of signal processing—correlation between the native (input) EEG signal and what is observed using the polymer substrates, and the success rate of a common classifier for EEG state detection.

3.2.3.3 Correlation to Original Signal

The first level of processing we examined was the correlation between the recorded and original signals. This was chosen as a global measure of signal reproduction because it is agnostic to inevitable differences in amplitude due to miscalibration between ADCs or amplifiers, or long-term low-frequency drift, or phase-shifts from minor signal asynchronies. Data files (original and each recorded) were initially band-pass filtered 0.5–45 Hz to remove large trends and line noise artifacts, then divided into 250 independent segments of 1,400 ms each. Pearson correlation coefficient was calculated between them for each segment, then averaged across segments to yield a mean and variance for the record. Note that in this case, variance relates to fluctuations in correlation as the signal naturally changes in amplitude and frequency content over time.

Figure 11 shows changes in mean correlation as a function of compressive strain for each material that was volume-percent tested. Initial inspection of the data yielded lower overall correlation when using the original prerecorded input file as the comparator than when using a control recording condition using conventional metal wiring, suggesting some inherent distortion introduced by the recording system. Correlations were, thus, made using this control record as the comparator in order to bias from this distortion. For 4% and 7%, we observed relatively high and consistent correlation across the range of compression tested, suggesting these mixtures are sufficient for reasonable EEG signal transmission. In contrast, 3% mixture yielded more mixed results. In particular, correlation at no compression (0%) resulted in a poorer correlation, which increased dramatically with compression. It is unclear why results for 3%, 5%, and 10% were so poor, but inspection of these data revealed no apparent presence of signal (just noise). The overall trend with this mixture, however, shows a clear effect of increased signal transduction. Inspection of the time series and spectral content suggest that the de-correlation at lower compressive strain results from a much higher background noise (e.g., 60-Hz line noise and other EMI artifacts) infiltrating the recorded signal.

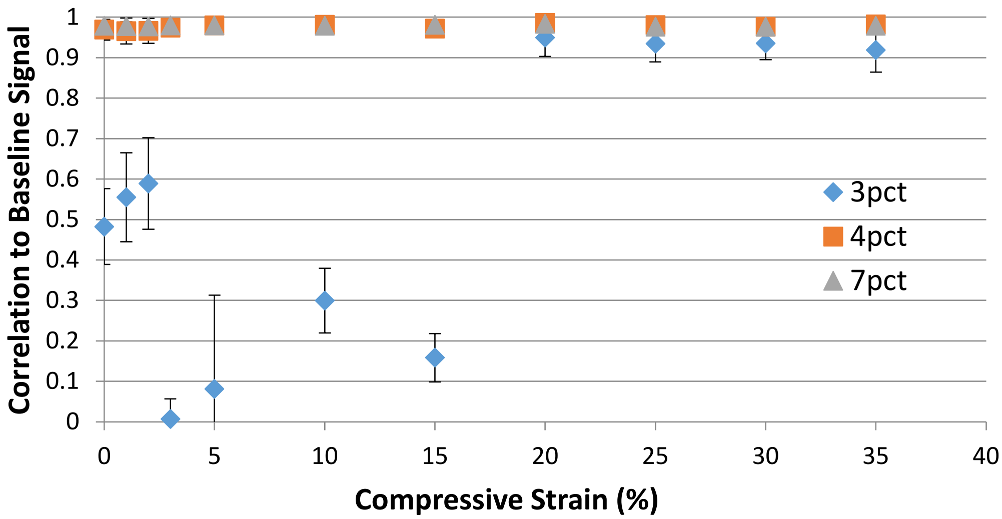

Fig. 11 Baseline correlation vs. quasi-static compressive strain for 3 different volume-percent loadings of CNF in PDMS

3.2.3.4 RSVP Target Response Classification Efficacy

Previous work (Cecotti et al. 2015) has shown that with this type of task, a P300 response to the target images can be reliably detected from the EEG; additionally, responses to the target images can be discriminated using a variety of different classifiers (Marathe et al. 2015). We chose to use a hierarchical discriminant component analysis (HDCA) model for classifying responses to the target and nontarget images (Marathe et al. 2014; Marathe et al. 2015), in particular, because this approach works reasonably well on single-channel EEG data and is fairly robust to minor distortions in signal quality (Hairston and Lawhern 2015). Unlike most EEG classification conditions, which leverage differences between channels as a data feature, in this case it was necessary to run as independent single-channel runs in order to avoid variance from polymer samples—adding confounding variance to the classification performance. Area under the curve (AUC) and correct-rejection hit rate (CR) were calculated based on HDCA results based on recorded re-created EEG data from each level of compression and percent volume loading. These metrics, which stem from ROC analyses, represent 1) the probability that out of the total samples, a randomly chosen positive sample has a higher probability of being positive than a negative sample (AUC), and 2) the rate of which out of all cases where a sample is negatively labeled, it was, in fact, correctly rejected as negative. Together these give a general, robust description of the overall performance of the classifier.

Results for HDCA AUC and CR showed robustness to the above described signal degradation that is typical with lower percent volumes and lower compression. As presented in Fig. 12, both metrics remain fairly high and statistically identical to a baseline control condition where the signal was passed through a conventional metal conductor instead of the polymer substrates. As a comparator, we also used a random noise condition (labeled “noise” in Fig. 11) where the data passed to the HDCA classifier consisted of Gaussian white noise to establish minimum AUC and CR performance based on pure chance. Clearly, all values are above this line. In the samples plotted here, error bars represent standard deviation of results from multiple iterations of the classifier when run on the same data; thus, they represent variance in the classification process and not that of the polymer samples, themselves. Critically, this shows that although some cross-sample variation exists, it is significantly smaller than the internal cross-iteration “noise” of the classification process, itself. Altogether these results suggest that when using HDCA classification for EEG as the target application, there is substantial leeway in the material volume-percent and range of compression under which it is used. However, it also notably calls into question the degree to which this particular metric is valuable for evaluating performance.

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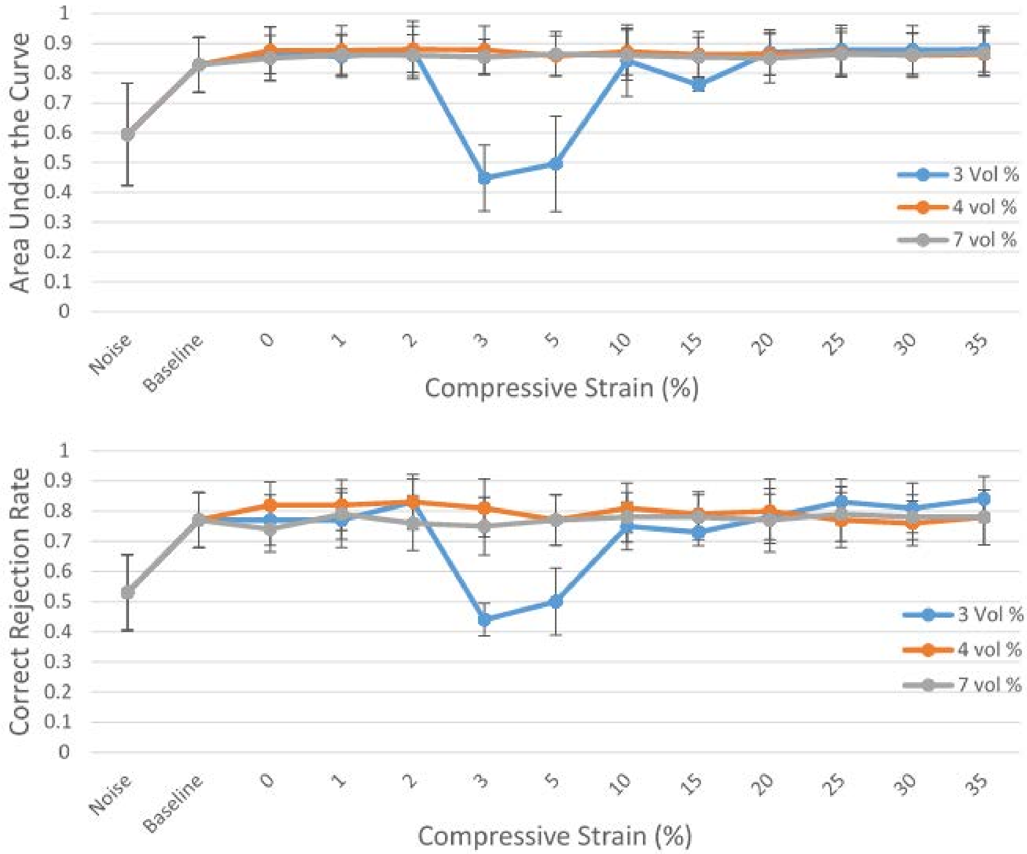

Fig. 12 Results for HDCA AUC (top) and CR (bottom) using 3 different CNF-PDMS formulations and subjected to controlled quasi-static compressive strain

3.3 Embedded EEG Signal Acquisition Electronics Hardware Design, Fabrication, and Validation

3.3.1 Year 1

In Year 1, we identified the need to develop our own in-house EEG signal acquisition electronics as a task to start in Year 2. Traditional EEG data acquisition systems neglect real-time monitoring of electrode contact impedance—at best sampling impedance only periodically during collection. Data acquisition through stretchable conductors with variable impedance from users in dynamic environments necessitates a high update rate and concurrent measurement of electrode impedance with actual EEG data. For these reasons, existing commercial and open-source designs were found to be insufficient for the goals of this DRI. Preliminary requirements for a desired custom design were specified at the end of Year 1.

3.3.2 Year 2

In Year 2, we completed design and fabrication of an ARL custom Texas Instruments ADS1299-based EEG data collection board. The needs met by a custom board, rather than a commercial system, include well-defined and characterized hardware using the latest ADS1299 differential sensing IC and ARL owned source code; a custom board pin-out that supports both traditional EEG and the custom elastomeric leads developed in this project; and support for modular add-on boards to provide for additional channels for EEG acquisition, for on-line impedance estimation, and for usability/integration features such as high speed wireless data transmission and Robot Operating System (ROS) integration. ROS integration allows raw and processed EEG data to be made available to any ROS-enabled real-time applications and provides support for time-synced inertial measurement unit data collection.

3.3.2.1 EEG Board Design Details

The EEG data collection board is designed using the Texas Instrument ADS1299 IC, which is a high-end industry standard for bio-signal data recording and other medical instrumentation devices because of its low-noise PGAs, high-resolution simultaneous sampling of 8 channels, and many different configurations to perform accurate and consistent measurement. It has high-precision, simultaneous multichannel signal acquisition capability. Other important features the ADS1299 IC offers in order to optimize its performance are controls of the gain, the sample rate, and powering on/off individual channels, among others. These configurations are achieved by setting and clearing the specific register values of the ADS1299 IC.

For the ARL board implementation, the ADS1299 IC handles the sample-taking and the conversion of the analog signals to digital samples, and a PIC18LF46K22T microcontroller handles the configuration of the ADS1299 and the data communication between the user or the host computer and the EEG board. Communication between the microcontroller and the host computer can be established in 2 ways. The first is a wired communication method using USB to serial cables, and the second is a wireless communication method employing UDP over Wi-Fi using an ODROID-U3, a small 1.7-GHz Quad-Core processor and 2-GByte RAM Linux Computer.

A real-time data collection and interfacing program was developed in LabVIEW to configure the ADS1299 IC for a specific setting and change any of the register settings at any time online (Fig. 13). The program receives data from the system either through serial or Wi-Fi and displays the measured EEG data from the 8 input channels of the ADS1299. The user has the option to select the communication type, change the register settings of the ADS1299 to modify the measurement setup, and monitor the measured signals through the LabVIEW front panel. This program also allows the user to save raw and processed EEG data to a text file for further analysis.

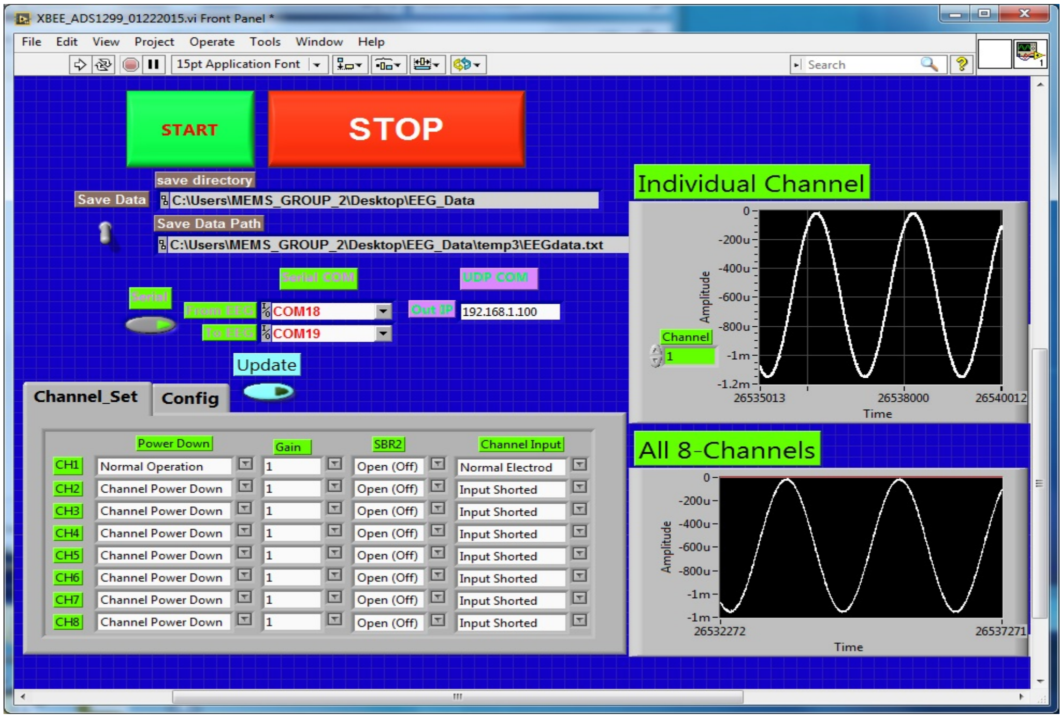

Fig. 13 EEG LabVIEW data collection program

3.3.2.2 EEG Board Validation

To validate the quality of the data collection performance of the EEG Board, we tested 2 of the characteristics the ADS1299—gain and the input-referred noise—using a signal generator to generate simulated small test signals and compared them to the specification on the data sheet. The experimental setup is shown in Fig. 14. Using test signals in range of 20–100 µV in 0.5–50 Hz bandwidth, we were able to measure and record data from all 8 input channels of the EEG board through the LabVIEW program. A sine wave input voltage of 1 mVpp at 0.5 Hz was used to test the gain. With the sample rate set to 1 kHz, 4 measurements were taken with 4 gain settings of 1, 2, 12, and 24. And for the input-referred test, all the positive and negative inputs of the 8 channels were tied together and connected to ground through the SRB2 pin of the ADS1299. Then the output of all 8 channels were measured one at a time. After removing the DC offset and filtering out frequencies above 65 Hz, the noise level measured was 0.17 uVrms. (The data sheet indicates it to be about 0.14 uVrms.)

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Results indicate that the EEG collection accuracy is equivalent to commercial systems, while functionality and modularity exceed commercial systems greatly, facilitating a Year 3 integration of online impedance estimation.

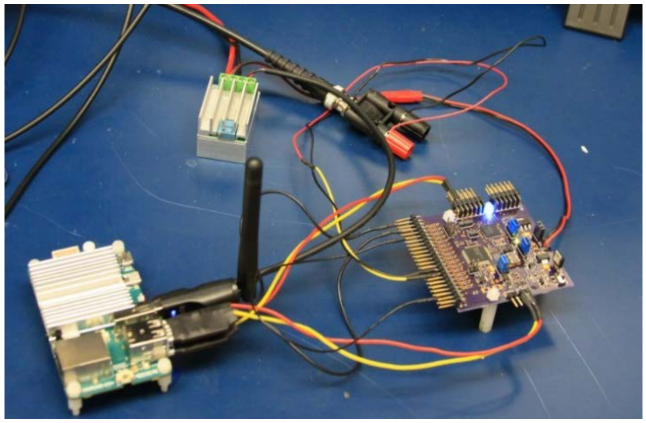

Fig. 14 EEG test setup

3.3.3 Year 3

The primary Year 3 focus for this task area included development and demonstration of a feasible method of online impedance monitoring. Early in this project, it was noted that impedance between the signal source in the scalp and the EEG data acquisition hardware is one of the biggest determiners of signal quality. The goal of this project has been to move towards an EEG collection system that is functional in real-world settings, including accounting for human motion-induced noise from dynamically deforming contact electrodes. Unlike relatively controlled and calm laboratory acquisitions, the impedance can be expected to vary quite quickly and periodically (due to running, jumping, crawling, etc.). As the subject maneuvers through the environment, the elastomeric electrodes will vary widely in contact pressure and shear. Small microvolt-level EEG signals are difficult enough to extract from noise and artifacts in clean environments, hence the requirement was developed for some kind of *metric* that can inform upstream algorithms of the EEG signal quality in real time.

The proposed metric is based on a direct measurement of each channel's impedance concurrently with EEG data collection. Current commercial systems may traditionally permit a one-time initial impedance estimation before collection. More advanced systems may multiplex impedance estimates between acquiring EEG samples. The proposed ARL method of impedance estimation is designed to operate concurrently (and, perhaps, asynchronously) with EEG data acquisition by exploiting a separation in the frequency domain. EEG signals are filtered and acquired at a lower frequency (<1 kHz), and impedance estimation uses injected current at higher frequencies (with zero DC component).

The fundamental concepts for this method of impedance estimation are derived from the long established field of electrical impedance tomography (EIT), where impedance images of solid bodies can be estimated from many electrodes attached to external walls of the target. A novelty in this research is that an active current driving electrode can still simultaneously be a sensor given the frequency domain separation. A major impediment to achieving full EIT analysis for this project is the disambiguation of impedance sources. In traditional EIT analysis, the impedance between the source and the data acquisition is assumed as negligible compared to the internal impedance. For this application, internal impedance is changing constantly, but contact impedance and, possibly, conductive elastomer impedance will be changing, as well. These multiple impedance sources present an observability problem that would require more constraints or types of measurements to solve. For the needs of this application, however, an *absolute* impedance measurement is not needed, but rather a *relative* impedance measurement. We wish to know which electrodes can be best trusted at any moment in time. Another necessary assumption is that the relative impedance metric we observe, indeed, changes monotonically with impedance, itself. Prior research and experience suggests that contact impedance exhibits the widest range of variation, compared to internal impedance and elastomeric conductor impedance, and hence the impedance metric will be dominated by contact impedance variations, which are known from prior research to vary monotonically with scalp pressure.

In our implementation, we used an AD5933 network analyzer IC from Analog Devices to monitor the combined electrode-skin impedance while recording EEG data. The AD5933 injects a small sine wave signal at a programmable frequency range into the path of the EEG measuring loop and measures the returned signal. It then samples the returned signal and performs a discrete Fourier transform onboard, and returns real and imaginary values of the impedance for each frequency. The starting frequency of the injected signal is programed to be at a much higher frequency than the EEG signal frequencies (~DC, 1 kHz) to satisfy the frequency separation requirement hence avoiding interference with the EEG signal, itself. In the representative data presented in Fig. 15, we selected an interrogation frequency between 30 kHz and 40 kHz.

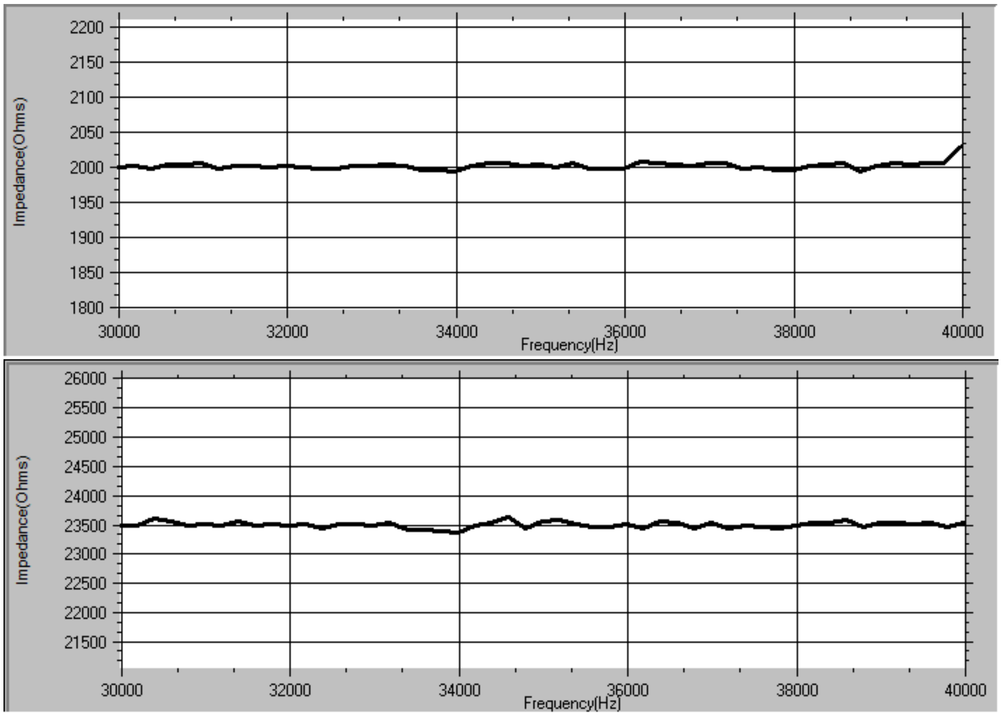

Fig. 15 Network analyzer chip validation results against known impedance values within the impedance range of interest for EEG online impedance monitoring, 2 kOhm (top), and 23.5 kOhm (bottom)

4. Practical and Theoretical Implications for the Research

We have already demonstrated the practical implications of our ability, honed under this DRI, to tune and shape electrical impedance response to an applied deformation of a stretchable electronic component. We have done so through 2 patent applications: one for the materials processing method, and one for an application of the materials to stretchable differential signaling media, as well as with a paper on stretchable solid-state logic switches, accepted for publication in *Applied Physics Letters* (Slipher et al. 2016). The successful stretchable switch research spun out of this DRI project as a failed attempt to make a stretchable conductor material.

In particular, these efforts show promise for the overall usability of flexible, deformable conductors as the basis of fieldable biosensing. Currently, much neuroscience and neurophysiology research is hampered by a reliance on traditional metal-based conductors, which are heavy and rigid, causing limitations due to weight, safety, and comfort; this relegates efforts to fairly restricted, non-realistic scenarios. Results shown here suggest that we should be able to begin honing in on sensors and conductive media that are pliable and comfortable. By maintaining a fairly consistent conductance profile across a range of compressive strain, this means that these materials will be reasonably robust for use as EEG electrodes where the material may need to be under compression in order to maintain a stable connection, and may encounter movement and deflection when in use cases that involve the person moving in a naturalistic manner. The result is that we now have an enabling ability to conduct neuroscience research in more naturalistic "real-life" scenarios that has not previously been possible. We anticipate many more related examples of practical implications of this research to arise in the coming years.

5. Conclusion

We have presented a carbon-nanofiber-filled polydimethylsiloxane conductive elastomer material solution (CNF-PDMS) whose electrical impedance can be tuned over more than 4 orders of magnitude, and which exhibits a flat electrical impedance shift when subjected to quasi-static compressive strains exceeding 60%. We have presented experimental results for our conductive elastomer used as a substrate for EEG measurement that indicate signal transmission remains intact when subjected to large compressions in excess of 60%. Specifically, although there is degraded SNR with lower CNF fill ratios, performance of a single-channel HDCA classifier of EEG data acquired using the CNF-PDMS indicate adequate performance across a wide range of compressive strains. Furthermore, we present a purpose-built modular and open-architecture EEG signal processing board, which we have integrated with a dedicated network analyzer chip for exploring the efficacy of using real-time impedance monitoring for EEG error rejection under field conditions. The net effect of our efforts provide ARL with a unique set of capabilities, the combination of which will enable ARL to initiate a program of EEG evaluation on human subjects under more real-world conditions than are currently possible in laboratory or clinical environments.

6. Proposed New and Continuing Research Directions

We propose the following new or continuing research directions as viable and fruitful avenues of exploration leading out of this DRI.

1) Exploring spatially graded materials—for example, the ultrasoft scalp interface for hair follicle penetration and improved signal to noise. We identified this research objective in Year 1 but did not have an opportunity to address it during the course of our investigations. We feel this research topic would be a significant endeavor, worthy of a 2–3-year basic research focus in and of itself. This research topic is currently being developed as a Foreign Non-Government Cooperative Research and Development

Agreement (CRADA) with the University of Wollongong (AUS) through ARL's technology transfer office.

2) Continue to pursue validation of the modular signal-processing board and real-time impedance monitoring hardware using purposefully designed EEG human experiments, and comparison to comparable baseline EEG experiments. With the inclusion of the impedance monitoring tool for EEG error rejection, a basic research program with this focus could potentially make a significant contribution to the effort to transition EEG techniques out of the laboratory and into real-world environments.

3) Continue to pursue validation of the phantom technique for comparing newly developed EEG hardware and components to baseline standards. This effort is currently under way as a mission program within HRED.

4) Exploring the efficacy and feasibility of autonomously cueing robotic squad members using EEG acquired human state estimation under field conditions using our hardware and approach.

5) Exploring the efficacy and feasibility of engaging the growing maker community with our open architecture EEG signals acquisition and processing electronics design.

6) Continue to explore and develop new applications for ARL's unique knowledge and ability to shape electrical impedance response to applied deformations in stretchable electronic materials. This effort is currently under way as a collaborative mission program between VTD and the Weapons and Materials Research Directorate (WMRD), as well as in an active Open Campus collaboration with Professor Iain Anderson at the University of Auckland.

7. Publications, Patents, and Notable Briefings to-Date

Publications:

1. Slipher GA, Chau NLH, O'Brien B, Mrozek R, Anderson I. A solid-state dielectric elastomer switch for soft logic. Applied Physics Letters, Accepted for publication (MS #L15-08620).

2. Wolde WT, Conroy JK. A customizable and expandable EEG data collection system. Aberdeen Proving Ground (MD): Army Research Laboratory (US); 2016 Mar. Report No.: ARL-TR-7611.

3. Lance B, Hairston WD, Apker G, Whitaker KW, Slipher G, Mrozek R, Kerick, S, Metcalfe J, Manteuffel C, Jaswa M, Canady J, Stachowiak C, McDowell, K. 2012 year-end report on neurotechnologies for in-vehicle applications. Aberdeen Proving Ground (MD): Army Research Laboratory (US); 2013 Jun. Report No.: ARL-SR-267.

4. Silversmith D, Perkons N, Jordan K, Brooks J, Conroy J, Hairston WD, Kerick S, Lance B, McDowell K, Nothwang WD. Recommendations for future experiment. Aberdeen Proving Ground (MD): Army Research Laboratory (US); 2012 Dec. Report No.: ARL-MR-0833.

5. Silversmith D, Perkons N, Jordan K, Brooks J, Conroy J, Hairston WD, Kerick S, Lance B, McDowell K, Nothwang WD. Fusing multiple sensor modalities for complex physiological state monitoring. Aberdeen Proving Ground (MD): Army Research Laboratory (US); Dec 2012. Report No.: ARL-TR-6283.

6. Slipher G. Method and apparatus for precisely applying large planar equi-biaxial strains to a circular membrane. Aberdeen Proving Ground (MD): Army Research Laboratory (US); 2013 Apr. Report No.: ARL-TR-6432.

Patents:

1. Shumaker J, Slipher G, Mrozek R. Differential electronic signal transmission through elastomeric conductors. (ARL Docket 13-20). Full patent application filed and under review as of 2016 Jan 30.

2. Mrozek R, Lenhart J, Slipher G. Stretchable polymer composites with controlled electrical performance during deformation through tailoring strain-dependent filler contact. (ARL Docket 13-37). Full patent application filed and under review as of 2016 Jan 30.

Notable Briefings:

1. Slipher G. Engineering stretchable electronic materials. Invited talk at ASME 2014 Smart Materials, Adaptive Structures and Intelligent Systems; 2014 Sep 08 2014; Newport, RI.

2. Vettel J, Hairston WD, Slipher G, Mrozek R, McDowell K. Translational neuroscience: from the bench to the battlefield, to GEN Dennis Via, CG AMC, 2014 Feb 7.

3. Slipher G. Deformable electronic structures & materials. Briefing to Lisa Sanders, SOCOM Deputy Director, Special Operations Research, Development, and Acquisition Center (SORDAC), 2013 May 1.

4. McDowell K, Lance B, Hairston WD. Translational neuroscience. Presented to Mary Miller, the Deputy Assistant Secretary of the Army, Research and Technology; Aberdeen Proving Ground (MD), May 2013.

5. Slipher G, Mrozek R, Lenhart J. Stretchable elastomeric conductors for instrumentation of precision air drop canopies. Briefing to Richard Benney, NSRDEC, Director, Aerial Delivery Directorate and Directorate professional staff, 2013 Mar 12.

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Marathe AR, Ries AJ, McDowell K. Sliding HDCA: single-trial EEG classification to overcome and quantify temporal variability. Neural Systems and Rehabilitation Engineering, IEEE Transactions on. 2014;22(2):201–211.

Marathe A, Lawhern V, Wu D, Slayback D, Lance B. Improved neural signal classification in a rapid serial visual presentation task using active learning. Neural Systems and Rehabilitation Engineering, IEEE Transactions on. 2014;24(3):333–334.

Slipher GA, Chau NLH, O'Brien B, Mrozek R, Anderson I. A solid-state dielectric elastomer switch for soft logic. Applied Physics Letters, Accepted for publication (MS #L15-08620R).

List of Symbols, Abbreviations, and Acronyms

2-D	2-dimensional
3-D	3-dimensional
AR	aspect ratio
ARL	US Army Research Laboratory
AUC	area under the curve
CaN CTA RMB	Cognitive and Neuroergonomics Collaborative Technology Alliance Research Management Board
CNF	carbon nanofiber
CR	correct-rejection hit rate
CRADA	Cooperative Research and Development Agreement
DRI	Director's Research Initiative
EEG	electroencephalographic
EIT	electrical impedance tomography
HDCA	hierarchical discriminant component analysis
NCCF	nickel-coated carbon fibers
PDMS	polydimethylsiloxane
ROS	robot operating system
RSVP	rapid serial visual presentation
SIS	styrene-isoprene-styrene
VTD	Vehicle Technology Directorate
WMRD	Weapons and Materials Research Directorate

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