Printed soft-electronics for remote body monitoring

Matti Mantysalo*[a], Tiina Vuorinen[a], Vala Jeyhani[b], Antti Vehkaoja[b]
[a]Tampere University of Technology, Faculty of Computing and Electrical Engineering,
Korkeakoulunkatu 3, FI 33720, Tampere, Finland;
[b] Tampere University of Technology, Faculty of Biomedical Sciences
*matti.mantysalo@tut.fi

ABSTRACT

Wearable electronics has emerged into the consumer markets over the past few years. Wrist worn and textile integrated devices are the most common apparatuses for unobtrusive monitoring in sports and wellness sectors. Disposable patches and bandages, however, represent the new era of wearable electronics. Soft and stretchable electronics is the enabling technology of this paradigm shift. It can conform to temporary transfer tattoo and deform with the skin without detachment or fracture. In this paper, we focus on screen-printed soft-electronics for remote body monitoring. We will present a fabrication process of a skin conformable electrode bandage designed for long-term outpatient electrocardiography (ECG) monitoring. The soft bandage is designed to be attached to the patient chest and miniaturized data collection device is connected to the bandage via Micro-USB connector. The fabricated bandage is tested in short exercise as well as continued long-term (72 hours) monitoring during normal daily activities. The attained quality of the measured ECG signals is fully satisfactory for rhythm-based cardiac analysis also during moderate-intensity exercise. After pre-processing, the signals could be used also for more profound morphological analysis of ECG wave shapes.

Keywords: Printed electronics, soft electronics, stretchable, skin-conformable, health monitoring

1. INTRODUCTION

Wearable electronics applications have started to emerge in the consumer markets over the past decade. Wearable electronics together with Internet-of-Things (IoT) technologies provide scalable and easily implementable way to measure one's (person's or animal's) vital signs. This has increased the interest of consumer and healthcare industries towards mobile monitoring. Several patient groups would tremendously benefit from continuous wireless monitoring of vital signs. For example, cardiac patients may experience unpleasant sensations from the heart outside the hospital environment, and it might be crucial to be able to verify the degree of criticalness immediately when the symptoms arise. To improve the skin/sensor interface and wearability of the devices (comfort and ease of application) in these tracking situations, the development is transitioning from rigid and planar electronic systems towards more adaptable, skin-mounted electronics[1, 2]. Usage of ultra-thin and ultra-bendable circuit board and conductor materials makes the devices light and able to adjust and attach to irregular surfaces such as textiles or human skin[3, 4]. However, the use of stretchable substrates as circuit board material poses new challenges in the manufacturing process. For example, the operational temperatures, chemical inertness, and mechanical properties of soft substrate materials are generally weaker than those of commonly used non-stretchable but flexible substrates.

The material for stretchable interconnects is required to possess a high electrical conductivity and high mechanical deformability. There are many approaches to enhance the stretchability of materials and one of the most common way in printed electronics is to use composite materials. In composites, conductive (e.g. metal and carbon) flakes or nanomaterials (e.g. metal nanoparticles, metal nanowires, carbon nanotubes, or graphene) are combined with elastomeric materials[5]. In this way, the conductive particles form the current path network, while the elastomer provides the high elongation. This method provides lower conductance than bulk conductor material, but much higher than polymer conductors. However, while the higher conductive material concentration improves the conductivity of the composite material, it will also increase its brittleness and rigidness. Hence, depending on the application, the ratio between the materials dictates the electro-mechanical properties of the outcome. As an example, Suikkola[6] et al. demonstrated screen-printed interconnects with sheet resistances of 36 mΩ/□ and they were able to stretched the interconnects up to 70% (single pull) elongation, and more than 1000 cycles with 20% elongation. This is suitable for many textile applications[7].

In this paper, we report a fabrication process of the screen-printed skin-mounted bandage for outpatient monitoring. The interconnects and electrodes of the bandage are manufactured by screen-printing stretchable silver/silver chloride ink on a 50 µm thick thermoplastic polyurethane (TPU) substrate. The substrate was then heat laminated with a transparent adhesive film to provide an adhesive layer towards the skin. A Micro-USB connector is used to connect the bandage to a commercially available mobile data acquisition device. The resulted skin-mounted system is designed for measuring the ECG using a small area electrode configuration called EAS, that is a subset of an EASI lead system introduced by Dower et al[8]. The EAS electrode setup was selected to provide unobtrusive ambulatory monitoring with high accuracy and completeness of the measurement setup[9]. Using this small area electrode configuration makes it possible to integrate a small, but rigid ECG measurement module with the stretchable bandage type of a substrate, and because of this, it can be used also in ambulatory situations. We also show preliminary test results that demonstrate the applicability of our stretchable ECG bandage in both, measurement during moderate intensity exercise and in long-term monitoring.

2. METHODOLOGY

2.1 Materials

CI-4040 (ECM) ink was used in the silver electrodes and circuitry fabrication. The ink is a commercially available stretchable silver/silver chloride screen printable ink designed for use in medical electrodes. The ink contains 40 – 50 wt% silver powder and 5 – 15 wt% silver chloride powder diluted in a diethylene glycol ethyl ether acetate solvent. This material was chosen due to its good conductivity, stretchablity, right Ag:AgCl ratio for sensing electrodes, and low annealing temperature of 120 °C.

The substrate material used in the study is commercially available Epurex Platilon U 4201 (Covestro). This is a soft thermoplastic polyurethane (TPU) film with a thickness of 50 µm. Main reasons for selecting this substrate are the relatively high softening temperature (155 – 185 °C), which is higher than the annealing temperature of the selected ink, softness, its thermoplastic properties, and the compatibility with the printing process. For example, polydimethylsiloxane (PDMS), commonly used in stretchable electronics, has low surface energy which makes the printing process more challenging.

2.2 Screen printer

TIC SCF-300 screen printer is used to manufacture the stretchable interconnects and electrodes. The printer is equipped with a screen that has a polymer mesh stretched over an aluminum frame. The parameters of the screen are presented in Table 1. A square-edge shaped squeegee with a Shore hardness of 75 was used in the manufacturing process.

Table 1. Screen parameters

Mesh material	Polyester
Frame material	Aluminum
Mesh count (threads/cm)	79
Mesh opening (µm)	69
Thread diameter (µm)	55
Stretching angle (◦)	22.5

2.3 Fabrication of ECG bandage

The fabrication process was started by cutting TPU film into sheets. After that, the sheets were manually pre-stretched approximately 5 % and mounted on a rigid temporary aluminum plate carrier. The mounting process is needed to ensure that the TPU sheet will stay steadily flat throughout the printing and temperature annealing processes. Prior to the printing process, the surface of the sheets were cleaned with isopropyl alcohol (IPA) to make sure that there is no dust or stains that could impair the printing quality. The ink, stored below 10 °C, was let to warm up by keeping it in a room temperature for one hour and gently stirred with a spatula for 1-2 minutes.

After substrate and ink preparations, the electrode and interconnection patterns were screen-printed on the TPU. Prepared patterns were annealed in a convection oven at 130 °C for 30 minutes. The ink manufacturer recommends a curing temperature of 120 °C with a curing time of 10 minutes. However, our annealing setup included the additional aluminum carriers which were seen to influence on required annealing temperature and time.

After thermal annealing of the conductors, the TPU was released from the aluminum carrier, and Opsite Flexifix transparent adhesive film was heat laminated on top of the print. Opsite film has an acrylic adhesive layer underneath a polyurethane layer and this adhesive layer is used to attach the bandage to the skin. Before the lamination process, an electrode sized patterns and a connector area were cut out from Opsite film to enable the electrodes to be in contact with the skin and the adapter, respectively. In addition. Hydrogel coins were mounted on top of the electrodes to ensure a stable skin-electrode interface during the measurement.

Finally, an additional circuit board adapter was mounted on the electrode bandage. The adapter contains zero insertion force (ZIF) and Micro-USB connectors connected together with 100 kΩ series resistors. The circuit board adapter is attached to the bandage with the ZIF and the measurement device will be connected to the Micro-USB plug. The resistors will ensure the electrical safety of the measurement subject in case of an electronics fault in the measurement device. Figure 1 presents the fabrication process of the printed electrodes.

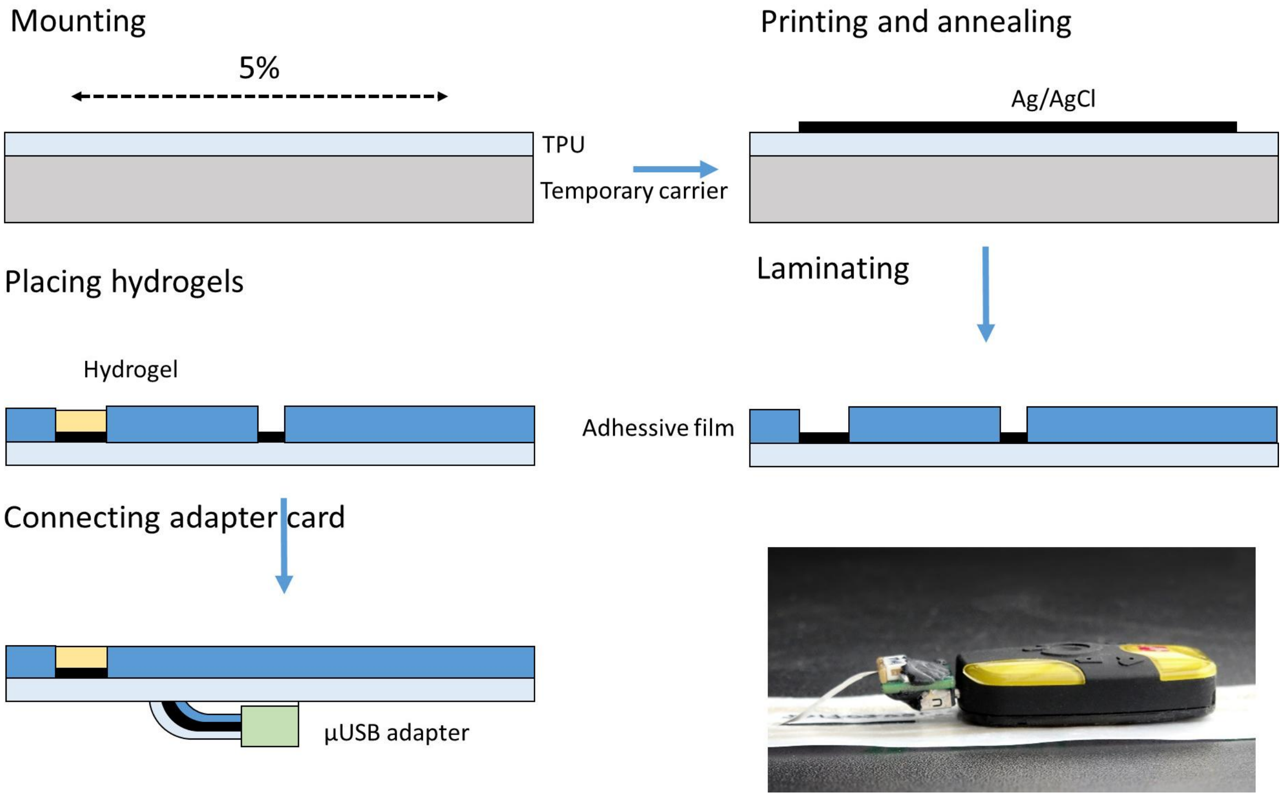

Figure 1. The fabrication process of the bandage. 1) TPU is stretched approximately 5% and mounted on the rigid aluminum carrier, 2) Ag/AgCl ink is printed and annealed, 3) lamination of the adhesive film, 4) placing of the hydrogels, and 5) mounting of the Micro-USB adapter.

2.4 Test procedure

In the EAS configuration, the three electrodes are located as follows: electrode E at the lower part of the sternum, electrode A at the standard ECG V5 electrode location, and electrode S at manubrium. Figure 2 presents the monitoring system and the printed electrodes secured in place for the measurements.

A male volunteer, with no diagnosed cardiac problems, performed the tests with the printed electrodes. A normal skin preparation procedure for ECG measurements was followed prior to the measurements: body hair was removed, skin was cleaned with a disinfectant and dead skin cells were scraped off with a rough sponge around the electrode areas. The measurement protocol started with an exercise phase on a treadmill, which was followed by a long-term monitoring phase up to three days (72 hours).

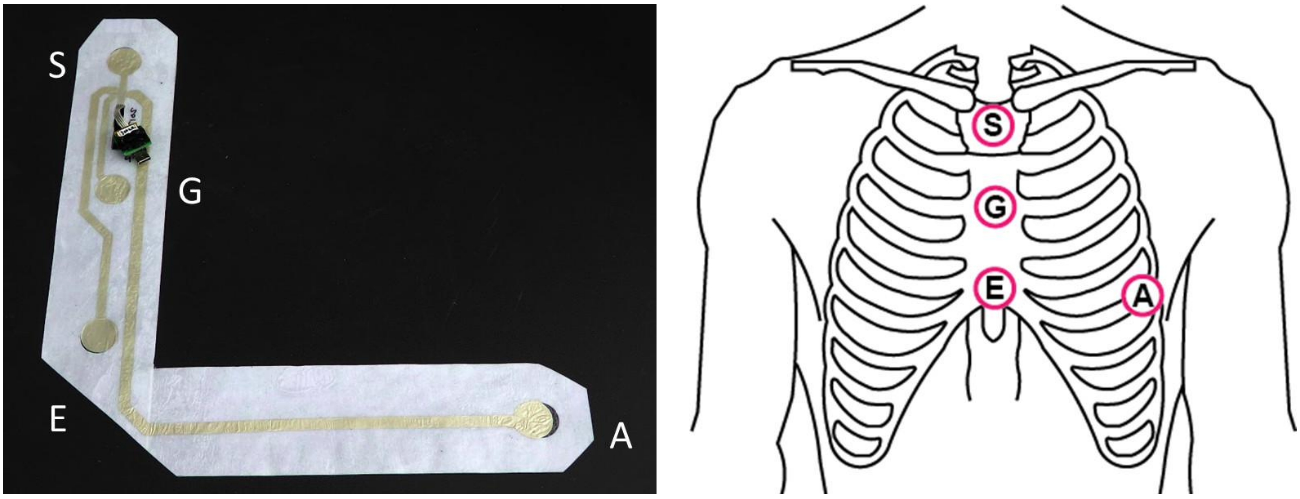

Figure 2. The monitoring system and printed electrode placement.

3. RESULTS AND DISCUSSION

Figure 3 shows ten second extracts of the EAS signal leads from the four measurement conditions: laying, slow walking (3 km/h), brisk walking (6 km/h), and running (10 km/h). The raw ECG signals of the three leads are presented with different colors. The A-E, E-S, and A-S signal leads are the green line (top), the blue line (middle), and the red (bottom), respectively. The signal plots show slight increase in noise level as a function of increased intensity of the activity. However, it should be noted that the signal-to-noise ratio is excellent in all cases.

Figure 4 presents ten second extracts of the ECG signal leads after 24 hours, 48 hours, 72 hours, and continuous data stream from 60 to 72 hours. As seen from the figure, the signal quality is extremely good even after two days measurement. However, the signals showed occasional moments of saturation with slightly increasing frequency towards the end of the measurement. These saturation spikes are caused by the disconnection between the skin-electrode interface. It was noted that activities of higher intensity increased the frequency and duration of disconnections. After 66 hours of measurement, the amount of discontinuity increased significantly as seen in Figure 4. However, the quality of the measured ECG signals is fully satisfactory to compute heart rate and many other important cardiac parameters when the skin-electrode connection returns as shown in Figure 4. During these moments, there is no significant change in the noise level of the ECG signals when compared with the beginning of monitoring period. After proper preprocessing, the measured signals also offer potential for morphology analysis of ECG wave shapes. The increasing amount of electrode disconnections is therefore currently the biggest challenge for the long-term use of our printed ECG bandage. Our assumption is that the amount of electrode disconnections could be affected by the selection of the electrode-skin adhesion material. Evaluation of different skin-compatible, non-irritating adhesion materials as well as testing with larger number of both healthy subjects and patients suffering from cardiac problems are part of our future work.

ACKNOWLEDGEMENTS

This work was funded by the Finnish Funding Agency for Innovation (Tekes) as a part of project VitalSens (decision ID 40103/14) and Academy of Finland (Grant nos. 288945 and 294119). T.V. would like to thank KAUTE Foundation and Eemil Aaltonen Foundation for support.

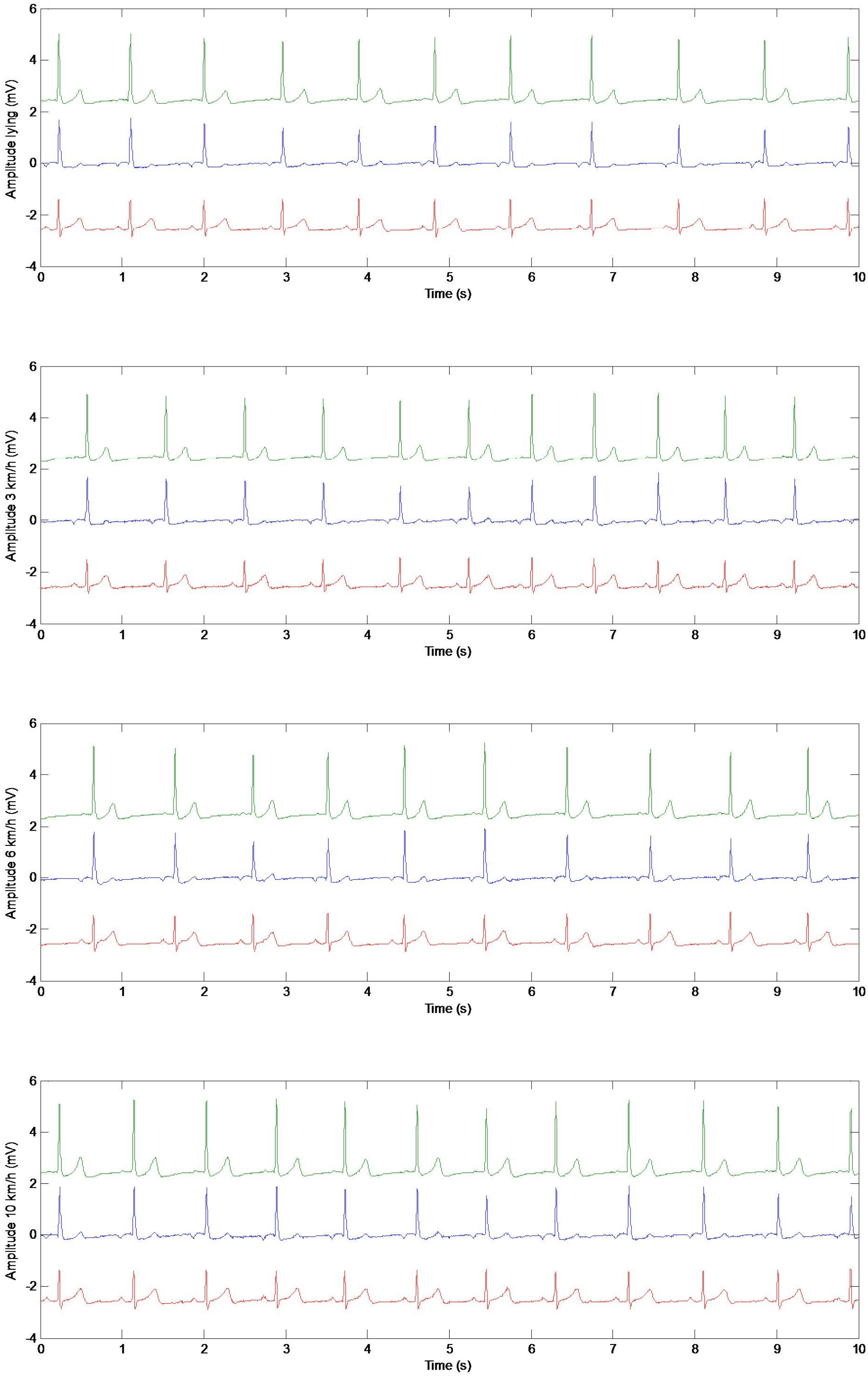

Figure 3. A-E (top green), E-S (middle blue), and A-S (bottom red) signal leads during laying, 3 km/h walking, 6 km/h walking, and 10 km/h running.

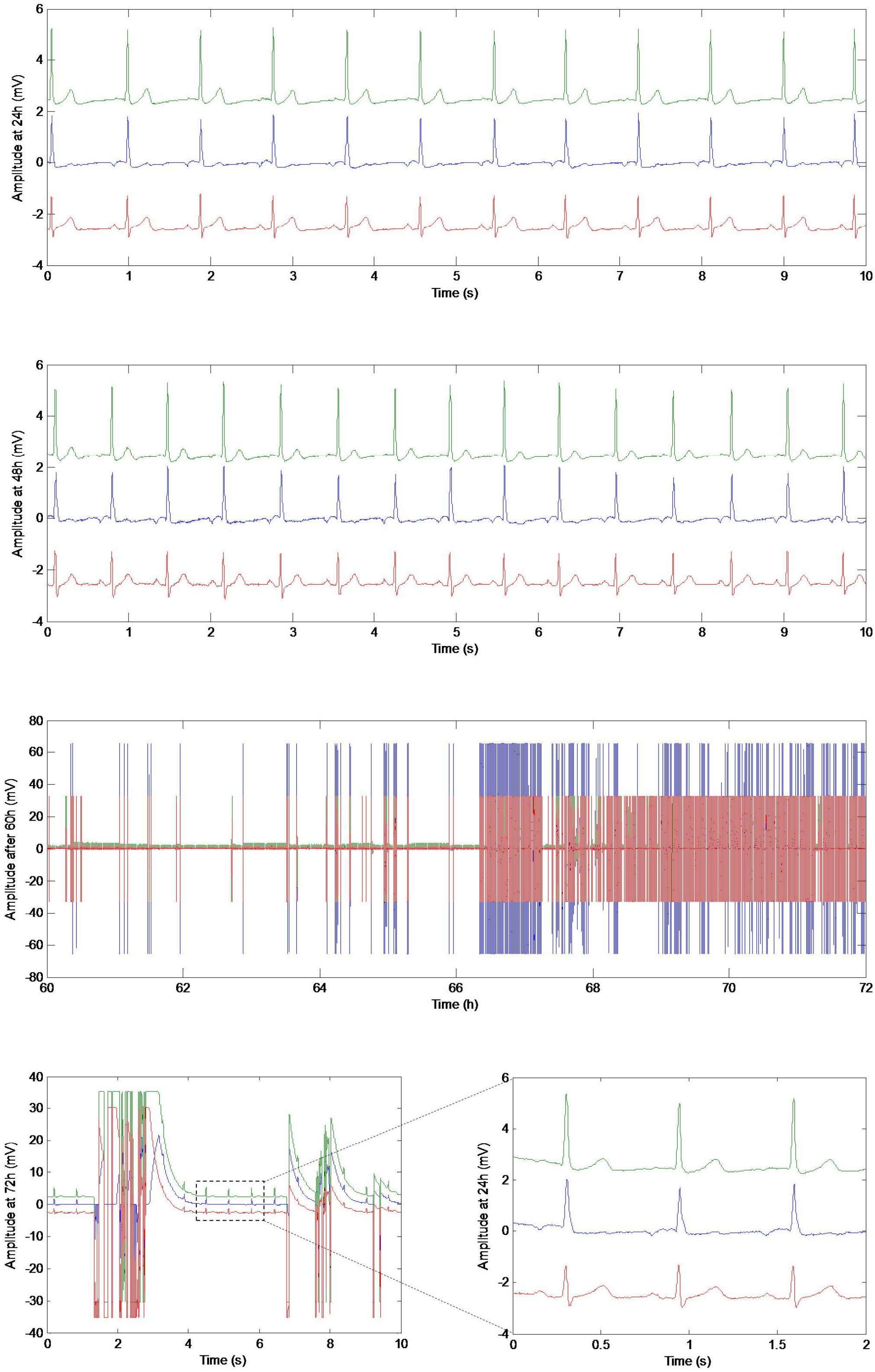

Figure 4. A-E (top green), E-S (middle blue), and A-S (bottom red) signal leads after 24 h, 48 h, data stream between 60 – 72 h, and signal leads after 72 h.

REFERENCES

[1] Sillanpaa, H. et al. "Integration of inkjet and RF SoC technologies to fabricate wireless physiological monitoring system." In Proc. of Electronic System and Technology Conference (ESTC) , Helsinki, Finland, Sept. 16-18, (2014).

[2] G. Yang et al. "Bio-Patch Design and Implementation Based on a Low-Power System-on-Chip and Paper-Based Inkjet Printing Technology." IEEE Transactions on Information Technology in Biomedicine Vol. 16 (2012), pp. 1043-1050.

[3] Shixuan Yang, et al. "“Cut-and-Paste” Manufacture of Multiparametric Epidermal Sensor Systems”, Adv. Mater. 2015

[4] Huyghe, B., Rogier, H., Vanfleteren, J. & Axisa, F. “Design and manufacturing of stretchable high-frequency interconnects”, IEEE Transactions on Advanced Packaging 31, 802–808 (2008).

[5] Yao, S. & Zhu, Y. Nanomaterial-Enabled Stretchable Conductors: Strategies, Materials and Devices. Advanced Materials (2015).

[6] Suikkola, J., et al., “Screen Printing Fabrication and Characterization of Stretchable Electronics”, Scientific Reports 6, Article number: 25784 (2016), doi:10.1038/srep25784

[7] Zheng, Y. L., et al, “Unobtrusive sensing and wearable devices for health informatics”, IEEE Trans. Biomed. Eng., vol. 61, no. 5, pp. 1538–1554, 2014.

[8] Fung, E., et al. "Electrocardiographic Patch Devices and Contemporary Wireless Cardiac Monitoring", Frontiers in Physiology. Vol. 6, No. 149 (2005).

[9] Vuorinen, T., et al.” Printed, Skin-Mounted Hybrid System for ECG Measurements”, In Proc. of Electronic System and Technology Conference (ESTC), Grenoble, France, Sept. 13-16