Contactless EMG sensors embroidered onto textile

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Abstract— **To obtain maximum unobtrusiveness with sensors for monitoring health parameters on the human body, two technical solutions are combined. First we propose contactless sensors for capacitive electromyography measurements. Secondly, the sensors are integrated into textile, so complete fusion with a wearable garment is enabled. We are presenting the first successful measurements with such sensors.**

Keywords— **surface electromyography, capacitive transducer, embroidery, textile electronics, interconnect**

I. Introduction

An excellent embodiment of a health monitoring device is a wearable system. Clothes are natural possessions and are part of the processes and routines in our daily life. The technological drive is to integrate sensors and electronics into textiles in such a way that the usage and advantages of cloths are maintained. Therefore a high level of textile integration has to be combined with aspects of reliability, comfort and washing resistance.

Sensors suitable for integration in clothes should be non-invasive and must be capable of monitoring health and wellness parameters. The common approach is to take relatively simple measuring techniques and to use signal processing and multi-parameter analysis to derive the physiological parameters of interest. In the ConText project [1-3], the universal measurement method of surface electromyography (sEMG) is used from which information about fatigue is derived by signal processing. Interpretation of the sEMG signal may be assisted by electrocardiography (ECG) signals and the output of movement/position sensors.

In ConText we develop a vest that contactlessly measures an sEMG signal with textile integrated sensors. So the vest can be worn over other clothes and will still extract electrophysiological signals like fatigue. The vest can be used by untrained individuals and therefore enables 24 hour monitoring at home and at work.

In ConText a number of different integration technologies are investigated: weaving of conductive yarn, printing of conductive material onto fabric, lamination of conductive and non-conductive fabric layers and embroidery of conductive yarn. This paper presents the efforts to make textile circuits and interconnections using embroidery.

II. Contactless surface EMG

All living cells are surrounded by membranes. These membranes are selectively permeable for various ions and may actively transport them through the membrane resulting into a membrane potential. Nerve cells and muscle fibres are depolarized when activated by a certain threshold voltage. The result is the propagation of a depolarization wave along the nerve and muscle fibre [4]. Such an electrical wave over the muscle fibre is the direct cause of muscular contraction and is subsequently followed by relaxation. The quick combination of contraction and relaxation of a muscle fibre is referred to as "twitch". Since all muscle fibres in a muscle do not twitch simultaneously, the overall observed potential over a muscle is the random summation of multiple single fibre action potentials. This random signal is conducted to the surface of the skin by means of volume conduction.

Surface electromyography (sEMG) electrodes are placed on the skin in order to record the muscle potentials. A common configuration is a set-up of two electrodes along a muscle contacting the skin using conductive gel. A problem with this set-up is that the interface potential between the skin and the solid electrode is undefined. In addition, it is uncomfortable to tape such electrodes onto the skin. Therefore, contactless electrodes are proposed in literature to monitor signals of the heart [5-7], which is also a muscle. These electrodes detect an electric displacement current by coupling capacitively to the body instead of detecting a Nernstian current; therefore, they require no electrical contact with the skin. The possibility to avoid direct skin contact reduces the skin irritations problems.

III. Contactless EMG measurements using PCB electrodes

Fig. 1 shows the bipolar set-up with two contactless electrodes with which the first EMG experiments are successfully performed [8]. The first measurements are done using electrodes which are not yet integrated into textile, but consist of 12 mm circular shapes on a printed circuit board.

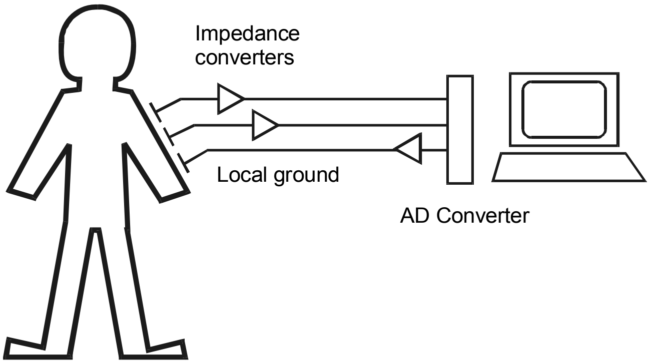

Fig. 1 Set-up for contactlessly sensing EMG signals

Because of the capacitive coupling, the impedance of the sensor is extremely high. The result is that environmental noise is easily picked up. This problem is solved by placing an impedance converter directly on top of the electrode. The electrode is actively shielded by feeding back the output signal of the local amplifier to a metal cap over the electrode. The individual sensor output signals are fed into an analogue to digital converter after anti-aliasing filtering.

Fig. 2 shows the measured EMG signal on the biceps while lifting a weight of 2.5 kg with a 90 degrees bended arm. The contactless sensors have an electrode spacing of 37 mm and a gain of 11. It is compared to a commercial active sEMG electrode (B&L Engineering type BL-AE-N) having a spacing of 20.6 mm and a gain of 346. In Fig. 2, the signals are normalized by the gains to give the skin surface voltage. Note that it is not possible to perform the two recordings simultaneously.

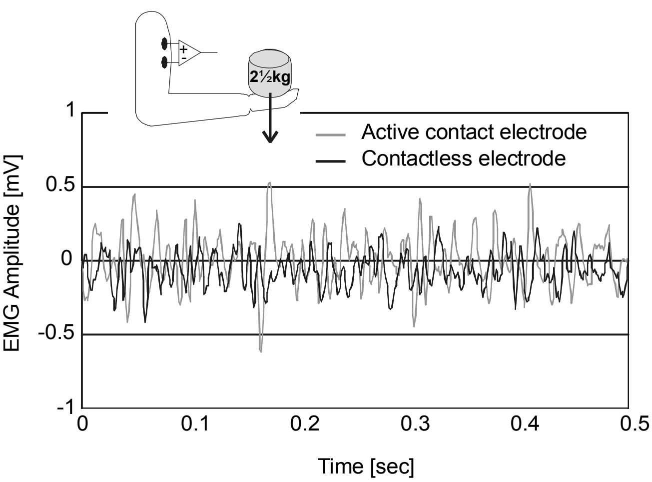

Fig. 2 Recorded EMG signals with both an active- and a contactless electrode

In fig. 3, the spectra are shown for the two measurement methods using the same data set as in fig. 2. As a reference, the spectra during rest are plotted as well. We can see that the contactless electrodes and the commercial active sEMG electrodes provide similar signal levels and shapes. Only the bottom noise during rest is a little bit higher. The bandwidth of the contactless electrode set-up is adequate and comparable to the reference measurement.

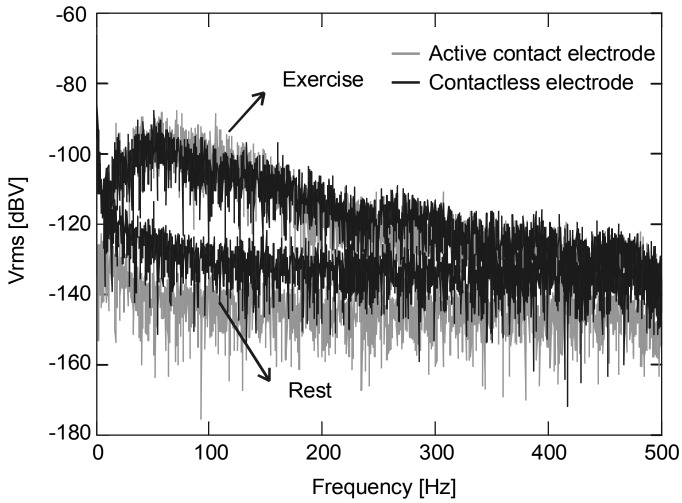

Fig. 3 Spectra of the recorded EMG signals with both an active- and contactless electrode

IV. The electronic module

Textile structures are orders of magnitude larger than electronics structures. Therefore a dimension adapter is required. As proposed in [9] we used a so called interposer for that purpose. This is a flexible polymer substrate which carries the electronics and has conductive pads to connect the conductive textile structures.

Although textile structures are big it is advisable to miniaturize the electronics to archive an overall textile character and to improve the reliability. E.g. thinned silicon chips are flexible and therefore in a mechanically stressed environment more reliable. However miniaturization is very expensive and it is often difficult to find bare dies. Therefore we have only produced a limitedly miniaturized sensor electronics module for the first tests with an amplifier in an SO8 package as shown on the picture.

A 25µm polyimide flex foil was used as substrate and was structured with a 25µm thin layer of copper-nickel-gold metallisation and coated with a 15µm layer of solder-resist mask. Components were only placed on the top side. The substrate was folded to form a ground shielding around the electronics to protect them form RF noise. All the areas that were not used have been covered with the ground plane to make this shielding most effective.

Unfortunately the whole package is rather thick as some of the components were not available in smaller size. The largest part is the amplifier with a thickness of 1.7mm.

To make an embroidered interconnection to external textile structures as described in [10] the pads on the substrate must be conductive on the top side. As the module has to be folded before embroidery, the folded part has openings above the metallised pads so that the embroidery needle punches through the pads only.

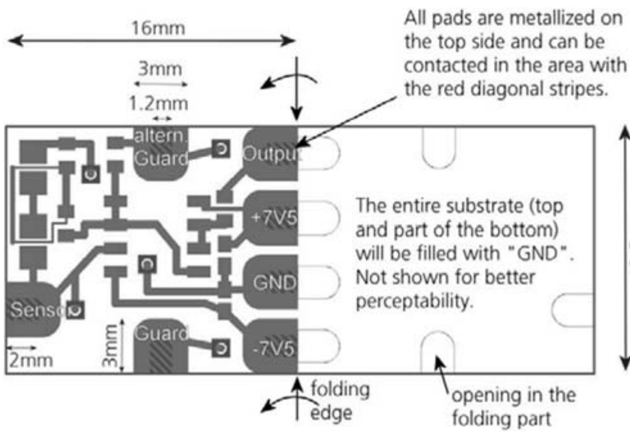

Fig. 4 Design of the flexible substrate

This module is still about as big as the whole sensor when it is manufactured from one piece of flexible substrate (incl. electronics and the sensor disk) and rather thick, which is certainly not desired for the final product. At this first stage of the project an expensive flip chips miniaturization was not worth the risk as the design still needed to prove that it works in textile generally. Generally a bipolar EMG-lead consists always of two disks. It is thinkable that one module of approximately the same size as this one serves as amplifier for both. This will already make the package appear smaller.

Fig. 5 Flexible substrate

V. Embroidering the Senor Electrode and the Interconnection

In [11] embroidery with conductive yarn is described as a means to produce conductive textile structures on fabric. Additionally in [12] embroidery was found to be an excellent technology to interconnect such embroidered circuits with electronic modules.

The fabrication of embroidered contactless EMG sensors additionally requires a multilayer design. In this project it was found that isolating layers of embroidery can be constructed. As shown in Figure 7 three layers of non-conductive embroidery are required to make sure that the layer below (the disk) and layer above (the guard cap) are isolated from another. To achieve a successful isolation it is essential that each new layer is embroidered perpendicular to the layer below (be it an isolating one or a non-isolating one). Furthermore for the construction of this special sensor it is fundamental that the top layer – the guard cap – is embroidered with a non-conduction bobbin-thread and with a conducting top-thread. Otherwise the guard cap would entirely shield the sensor and it wouldn't sense anything.

Note that the layer one (the disk) is embroidered with a step stitch that means that the needle connects top and bottom thread every 2mm. This is not possible in the layers above because these threads must not go through the conductive areas of layer one. Therefore the other layers are embroidered with a satin stitch which means that the needle goes through the fabric only at the endpoints of a desired structure (e.g. on the circumference of the guard cap in fig. 8).

The first conductive layer shown in figure 6 connects the embroidered circuit with the flexible substrate. This requires that at least the top thread is conductive. However, better results have been achieved with conductive top and bobbin thread.

The sensor connects to the computer via snap fasteners which are crimped through the embroidery as shown in figure 9.

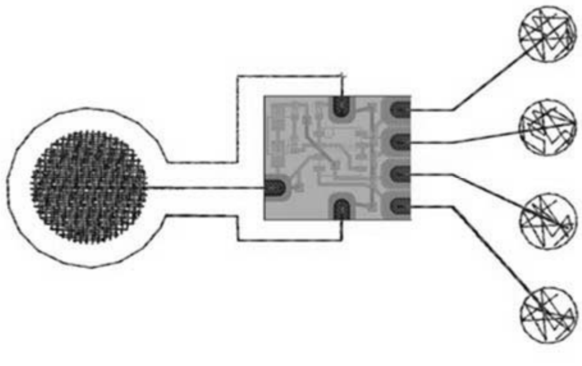

Fig. 6 Folded substrate with first embroidery layer; conductive yarn for the guard connection and the capacitive disk

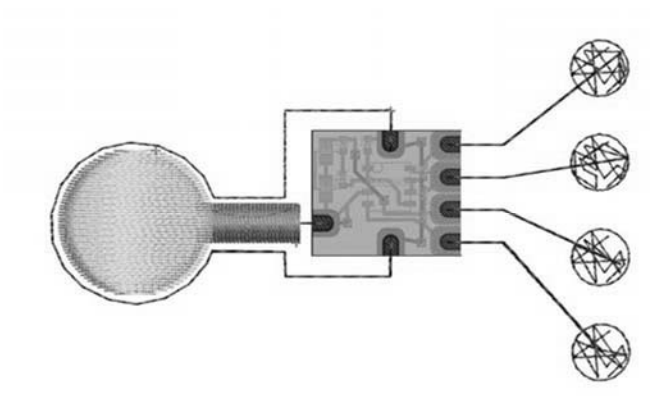

Fig. 7 The second, third and forth layer are non-conducting to isolate the sensor disk from the guard cap

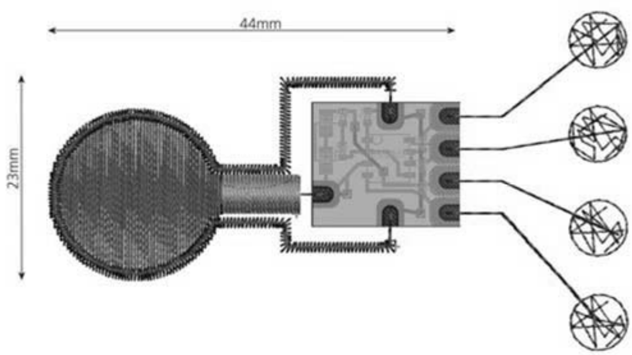

Fig. 8 Fifth layer with conductive yarn for the guard cap

Embroidery is a very demanding process for the yarn. The needle thread is bent by 180° around the needle. The thread is pulled through the eye of the needle and other small parts at a high speed. This can result in fussiness of the yarn or even a breaking. Yarn must be designed for embroidery.

Currently only a very limited number of conductive yarns are available for embroidery. One product series comes from Statex. Good results have been achieved with Shieldex 117/f17 2-ply. However the conductivity of around 500Ohm/m is rather low. A better conductivity can be expected from the ELITEX series by TITV which is based on Shieldex and further galvanized. Good results have been achieved with ELITEX PA/Ag 110/f34 2-ply.

For the interconnection as described in [11] it is necessary that the top-thread is conductive on the surface (everywhere). This means the technology cannot make use of spun yarn that consists of non-conductive fibres spun together with a copper wire as the position of the wire is arbitrary. These yarns cannot be embroidered anyway.

Another pre-condition for this technology is the use of flexible electronic substrates because the needle punches a hole through the substrate. This is not possible with thick FR4 substrates.

An encapsulation of these interconnects is required and will be investigated in the coming months.

Currently we believe that the actual contact mechanism is purely a mechanical one. The thread lies on the metallisation of the pad and is trapped in the gap of the wrenched substrate. Furthermore the contact could be improved with encapsulation, pressing the thread down on the substrate.

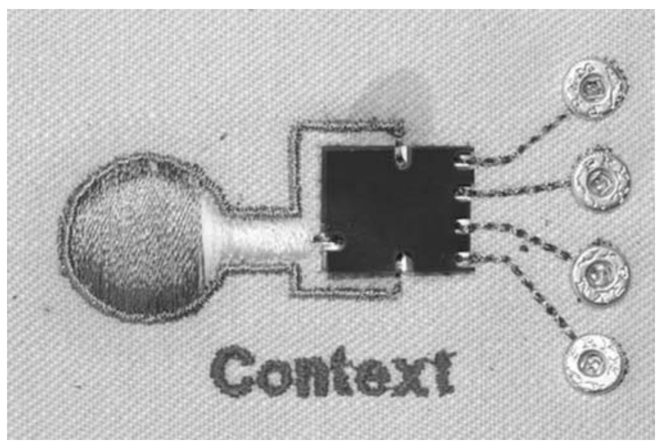

Fig. 9 Embroidered Sensor with interconnections to the electronic module and snap fasteners as interface to the computer

VI. MESUREMENTS WITH THE EMBROIDERED SENSORS

To evaluate the embroidered capacitive transducers, without having the problems of motion artefacts and noise of the human body, an artificial muscle model was used. Fig. 10 shows the hardware model. The muscle itself is emulated by a strip of moderately resistive paper. By two aluminium beams, an electrical current can be forced through this muscle. On top of the muscle, a leather chamois is used which mimics the human skin. On the chamois we can put several types of textile on which in its turn the sensor is placed. So, the model does imitate the contactless behaviour and the distributed shape of a buried muscle, but does not include the human tissue volume conductor properties.

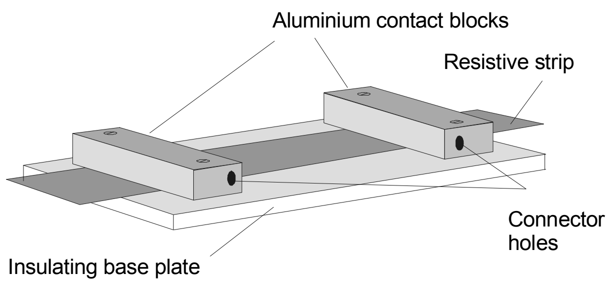

Fig. 10 Artificial muscle model

A waveform generator was connected to the artificial muscle. A square wave of 1 V_{pp} with a frequency of 20 Hz was generated. Fig. 11 shows the recorded signal by using a single embroidered sensor on the artificial muscle. Note that the envelope of the recorded square wave shows a 50 Hz noise signal. This is the result of the single-sensor approach. By using a single sensor with respect to a grounded reference, we will see 50 Hz noise as picked up capacitively from the environment. This will be cancelled when using the set-up of Fig 1.

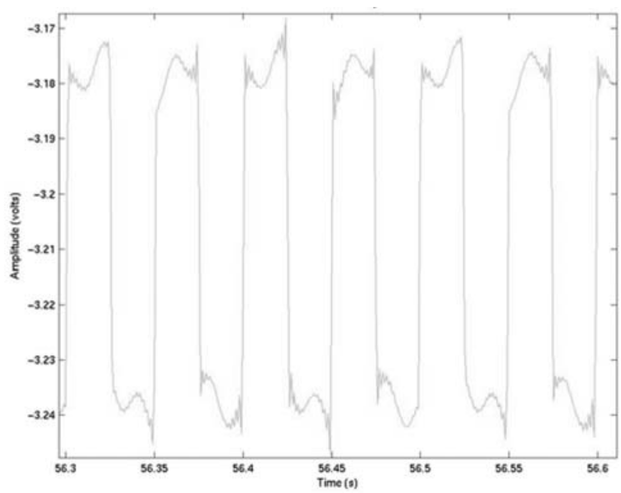

Fig. 11 Capacitively recorded square wave

In Fig. 12, the human EMG is measured on the biceps using two electrodes of the type of Fig. 9. The set-up is similar as used in section III. At t = 0 sec and t = 100 sec, a contraction of the biceps was applied. We can see that muscular activity is clearly detected by the textile embroidered sensors. Some motion artefacts are visible as spikes on the signal.

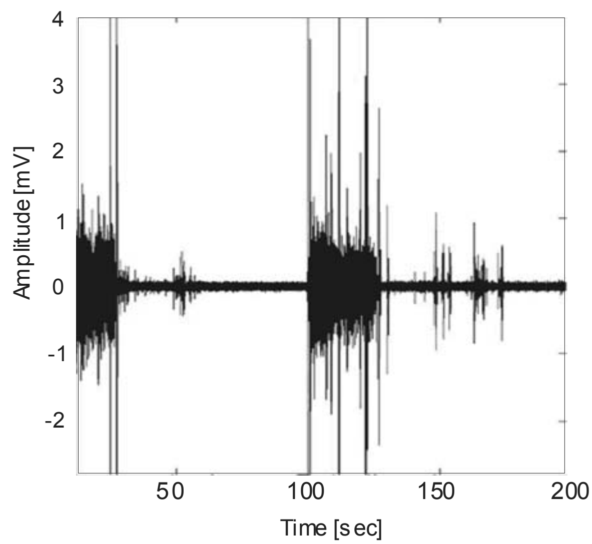

Fig. 12 Capacitively recorded bipolar EMG on a human biceps

VII. Conclusions

It could be demonstrated that contactless EMG electrodes deliver similar signals as contact electrodes and allow at least the differentiation of the states "rest" and "exercise". Furthermore a textile integration based on embroidery was presented and approved by electrical measurements in a test environment.

In the next steps we will try to reduce motion artefacts and improve the robustness of the EMG signal. Once this works the encapsulation and the reliability of the sensor will be tested.

Acknowledgment

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