

Advancements in Wearable MagnetoCardioGraphy (MCG) Sensors

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Abstract—MagnetoCardioGraphy (MCG) is a technique used to detect the heart's naturally emanated magnetic fields. Given the weak level of these signals, state-of-the-art MCG devices are costly, bulky, and/or require shielding. More recently, our research has demonstrated wearable MCG sensors that overcome these limitations. In this paper, we review the operating principle of these wearable MCG sensors and discuss the advancements in technology over the past years. The ultimate goal is to inspire further design optimization and clinical applications.

Index Terms—Coils, heart beats, magnetic field, magnetocardiography, sensor.

I. INTRODUCTION

MagnetoCardioGraphy (MCG) uses sensors to detect the weak magnetic field of the heart in a non-invasive manner [1]. A major advantage compared to ElectroCardioGraphy (ECG) is that it is contact-less, implying that irregularities arising from inconsistent or variable electrode-skin connections do not manifest [2]. Concurrently, given that biological tissues are non-magnetic, MCG signals propagate unaltered towards the outside of the body, unlike ECG. Indeed, since the first MCG recording in 1963 [3], MCG has become attractive due to its importance in diagnosing critical cardiovascular diseases, including 3D mapping of the heart, detection of myocardial ischemia, and more [4].

The first MCG sensor consisted of two large magnetic sensor coils [3]. However, its size and limited spatial resolution prohibited the use of this technology in the clinical environment. Notably, the extremely weak magnetic field of the heart, which is in the order of 10^{-15} T, poses a major challenge on MCG sensor design [5]. For comparison, the Earth's magnetic field is around 10^{-6} T. To sense such low signals, MCG recordings should be performed in low-noise setups, such as inside a magnetically shielded room. Although this approach is effective, it increases the cost and reduces portability. For example, superconducting quantum interference devices (SQUIDS) utilize a shielded environment and employ the principles of Josephson Junctions and electron quantum tunneling, as well as the concept that magnetic flux passing through a superconducting loop is quantized [6]. Though SQUIDS have shown to be extremely sensitive (detecting magnetic fields down to fT [7]), they are very high in cost (in the range of tens of thousands of dollars), bulky, and sophisticated to fabricate.

In an effort to reduce the cost and footprint of MCG, [8] introduced the first portable MCG device. The device relied on Faraday's law and utilized an array of coils to capture the heart's magnetic field. The end prototype consisted of an array of 12 coils having a diameter of 7 cm each. The sensor

operated in unshielded environments and was capable of detecting the cardiac magnetic field with sufficient sensitivity. However, though the footprint of the device is suitable for the pre-hospital and hospital environment, it cannot be considered as wearable. The coils also included cores that increased the weight of the sensor, further limiting its wearability.

More recently, we reported the first wearable MCG sensors. Over the past few years, our sensors have been refined in terms of hardware design and algorithmic post-processing. In this paper, we report an overview of these advancements with a goal to inspire further optimization.

II. OPERATING PRINCIPLE OF WEARABLE MCG SENSORS

The approach relies on an array of miniaturized, air-core coils that passively couple to the heart's magnetic field, followed by advanced digital signal processing (DSP) to denoise the signal. More specifically, the operating principle is based on Faraday's law, which states that the voltage induced on a coil from a time-varying magnetic field is:

$$V = AN \frac{dB(t)}{dt} \quad (1)$$

where N is the number of coil windings, A is the effective cross sectional area of the coil, and $B(t)$ is the time-varying magnetic field that is naturally emanated by the heart.

The constituent coils of the sensor are designed based on the model of a tightly wound air core coil. With a goal to optimize sensitivity, it was recently demonstrated that the ratio of the coil's inner diameter (D_i) to the ratio of the coil's outer diameter (D) has to be approximately equal to 0.6 ($\frac{D_i}{D} \approx 0.6$), and the ratio of the length of the coil (l) to the outer diameter of the coil (D) should be approximately equal to 0.7 ($\frac{l}{D} \approx 0.7$) [9].

III. ADVANCEMENTS IN SENSOR DESIGN

Referring to Table I, the first wearable MCG sensor in [9], was 4.67 times smaller in diameter compared to [8] and functioned using 4 coils instead of 12. The coils used were tightly wound air core induction coils, making the design lighter. The chosen inner and outer diameters of the coils were 9.3 mm and 15 mm, respectively, with a length of 11 mm. In an effort to increase sensitivity in the axial direction, [10] later modified the ratios $\frac{D_i}{D}$ to 0.56 and $\frac{l}{D}$ to 0.72, resulting in $D = 16.6$ mm and $l = 12$ mm while keeping $D_i = 9.3$ mm. The new dimensions led to a decrease in the noise density by 92% from [9], when operating the coil at a frequency of 100 Hz. Each of the coils in [9] and [10] were connected to an amplifier board with a gain of 1000 that was placed away

TABLE I
COIL COMPARISON FOR MCG SENSORS

Coil	Number	D(cm)	Di(cm)	Length(cm)	Magnetic Core
[8]	12	7	2.975	4.83	Yes
[9]	4	1.5	0.93	1.1	No
[10]	1	1.66	0.93	1.2	No
[11]	7	1.66	0.93	1.2	No

from the sensors to reduce electronic noise. The output of the amplifier was inputted into an Analog-to-Digital Converter (ADC), so that the recorded signals can be processed in MATLAB.

IV. ADVANCEMENTS IN SIGNAL PROCESSING

Aside from optimizing the coil design, improvements in the used DSP techniques have also been reported. Initially, [9] band-pass filtered the raw MCG signals collected from the coils in the range [8-35] Hz. After that, the signal was cut into windows while using the R-peaks of an accompanying ECG sensor as a fiducial point. These windows were averaged for each coil separately. Once done, the signals from all coils were averaged together to produce a single MCG signal, i.e., averaged in time and across all coils. The purpose was to remove uncorrelated noise and properly visualize the QRS complex. The procedure aimed at utilizing the least number of coils after which the signal could not be retrieved.

In a later work, [10], we introduced additional steps to the DSP. A notch filter was added to notch out the noise components at multiples of 60 Hz. Also, Ensemble Empirical Mode Decomposition (EEMD) was applied to denoise the final averaged signal. These techniques permitted the detection of lower-level signals as compared to [9] using only one (1) coil. However, the recording had to be performed over 24 mins so that the time averaging step would produce a clean signal.

In our most recent work [11], we managed to eliminate the averaging of windows over time. The only averaging done was across the coils. By utilizing 7 coils and a band-pass filter, the resultant averaged signal over all the coils allowed the real-time detection of MCG signals.

V. CLINICAL APPLICATIONS

To date, our wearable MCG sensors have been used for applications in cognitive workload classification [12]. Cognitive workload is the amount of mental effort an individual exerts when performing a task. Data was collected on 11 human subjects performing low and high cognitive workload tasks (i.e., two levels). Using MCG measurements, different Heart Rate Variability (HRV) metrics were calculated. The obtained results agreed with the literature such that low cognitive

workload is associated with higher HRV and high cognitive workload is associated with lower HRV. As MCG sensor reliability and sensitivity improves, several additional clinical applications are envisioned.

VI. CONCLUSION

The ability of wearable MCG sensors to seamlessly monitor the heart's magnetic field brings forward unprecedented opportunities in healthcare. This paper summarized our work to date in this field, including hardware implementation, signal processing, and clinical applications. Future work will focus on the signal processing side to further denoise the recorded signals and make them robust against external noise, such as motion artifacts.

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