Development of a portable labor-contraction monitor based on mechanomyography

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Abstract. *Mechanomyography is a technique that measures the mechanical signal observed from the muscle surface during the contraction. It is expected to be a promising tool to evaluate uterine dynamics even though the uterus contraction is known to be involuntary. This study aimed to describe the development of a portable device based on mechanomyography to monitor uterine contraction. To assess the system was performed isometric contractions of the rectus femoris muscle with varying loads. The acquired data showed that the system could identify each contraction's onset, offset, and amplitude. The sensors and the system architecture proved to be a promising tool for the desired application.*

1. Introduction

Monitoring the uterine contraction is a standard approach during pregnancy and labor, once it can anticipate the outcomes of the birth delivery. Inadequate uterine contractions are the leading cause of dystocia, characterized by difficult labor that is not progressing [Bernitz et al. 2014]. Dystocia accounts for ~60% of cesarean delivery indications, associated with an increased risk of maternal mortality [Edwards et al. 2019].

The external tocography (TOCO) and the manual procedure for recognizing uterine dynamics are the most common methods to assess uterine contractile activity. Both approaches lack information about those contractions' intensity and have low accuracy and reliability [Furley 2012]. TOCO can suffer from a high level of misreading and presents a technical limitation in obese parturients [Euliano et al., 2013].

A widely studied method for evaluating muscle contraction is mechanomyography (MMG). This method is based on detecting the muscle surface oscillation due to the mechanical activity of the motor units [Islam et al. 2018].

Beck et al. (2005) defined the MMG signal based on three main mechanisms: the gross lateral movements, lateral oscillations of the muscle at its resonant frequency, and dimensional changes of the active fibers. Meanwhile, the myometrium, mainly composed of smooth cells, is expected to initiate a mechanical contraction like that of other muscle tissues. Still, it can present circularly and longitudinally orientated muscle layers highly interwoven with associated characteristics, such as a slow, intense, and prolonged contraction [Dunford et al. 2019], presenting a particular state in which the uterine muscle fibers remain shortened after contracting, called retraction.

Although the contractile proteins in the myometrium do not have the same organization as can be seen in skeletal muscles, the exact mechanism of contraction occurs. Namely, the filaments sliding relative to one another [Wray 1993].

No application of MMG for uterine contraction monitoring has been reported in the literature. This is probably because most studies are related to voluntary contractions based on the analysis of skeletal muscles instead of on smooth muscle cells. Despite that, some works use accelerometry to assess fetal movement during pregnancy [Lai et al. 2018].

Therefore, this work aims to describe the development of a light-weighted device that monitors uterine contraction during labor based on surface MMG by using two different approaches: accelerometry (MMG_{ACC}) and sound (MMG_{MIC}), and also attest the device capability of relating the MMG signal amplitude with different contractions efforts levels.

2. Materials and Methods

The MMG recording can use different transducers, following two approaches: The first is based on the muscle fiber displacement, e.g., accelerometer, piezoelectric contact sensor, and laser displacement sensor; the second is based on the sound produced during muscle contraction, e.g., microphones [Beck 2010].

Even for typical applications of MMG, such as the study of muscle condition and/or state, there is no consensus in the literature on the transducer that should be used. Orizio and Massimiliano (2006) highlight the choice of the sensor transducer should consider the advantages and drawbacks of each option.

Most MMG measurements use accelerometer sensors for different reasons, such as low cost, light-weight, small dimension, high reliability, and also the ease to compare data from other studies once it is converted to accelerometer unit (m/s^2) [Orizio and Massimiliano 2006]; [Watakabe et al. 2003]. On the other hand, the MMG signals are highly susceptible to artifact movements, and microphone usage is preferred when recording dynamic activities that are more influenced by those artifacts [Posatskiy and Chau 2012]. Therefore, the proposed system uses the two different transducers to further comparison and studies of its application.

The system is subdivided into two main blocks: (1) Control Module Unit (CMU), composed of a communication unit which has a Bluetooth module for wireless communication, and a Microcontroller Unit (MCU) (Nordic Semiconductors, nRF52832) and contains the system power supply, a 3.7V lithium-polymer battery; (2) Sensor Unit, composed by two tri-axial accelerometers (STMicroelectronics, LSM6DSOX) and two microphones (TKD Invensense, ICS-40300), see Figure 1.

The CMU uses the I_2C (Inter-integrated Circuit) protocol to communicate with accelerometers, and the microphones uses an analog-digital converter (ADC) of 12-bit resolution, which also monitors the battery charge level. The signal acquisition sampling rate was 1 kHz, and afterward the MMG signal was digitally filtered (zero-phase 4th-order Butterworth filter) with a passband of 5–100 Hz [Ibitoye et al. 2014].

Data are transferred to the laptop via Bluetooth connection, and a throughput of 16Kbytes/s was achieved.

Based on the acquired data sent to the laptop for offline processing, the following parameters are expected to be extracted: (a) Onset, the instant of the begging of muscle activation, (b) Offset, which comprises the end of the muscle activation, (c) amplitude, calculated through the root mean square (RMS) amplitude. Later on, those parameters will be the basis for extracting the parameters used to analyze the quality of uterine contraction: frequency, duration of contraction, and intensity.

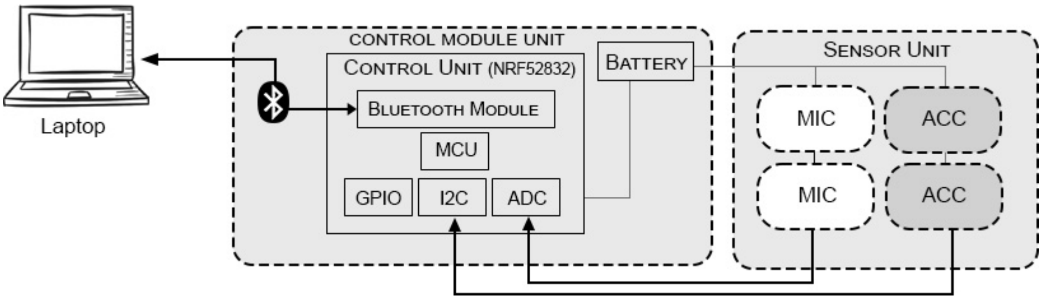

Figure 1: Hardware block diagram, Control Module Unit (CMU), and Sensor Unit.

2.1 Microphone and Accelerometer Design Aspects

Many authors suggest the need to use the microphone on the top-end of an acoustic chamber to enhance the detection of low-frequency signals induced by muscle vibration [Beck 2010]; [Fara et al. 2013]. Thus, the acquired MMG signal is strongly influenced by the geometry of the air chamber.

The MMG_{MIC} uses a MEMS microphone with an extended low-frequency response (6 Hz to 20 kHz) combined with a conical chamber design proposed by [Posatskiy 2011], with a 7-millimeter diameter and a 5-millimeter height, which aimed to improve signal quality. Figures 2-A2 and 2-A3 illustrate the sound chamber.

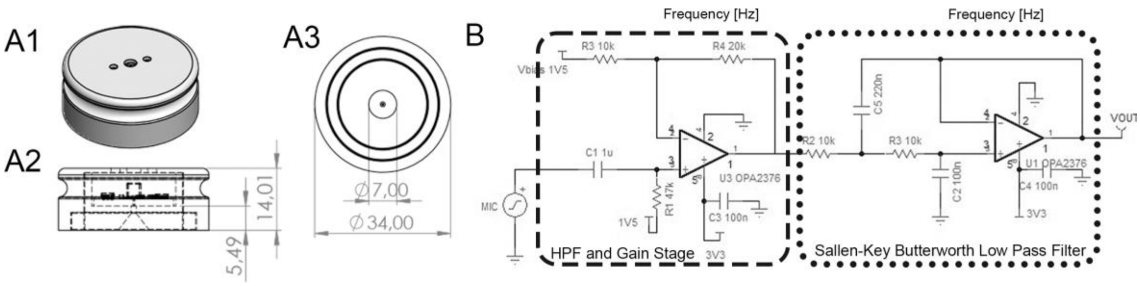

Figure 2: (A1) Microphone sound chamber (A2) lateral view (A3) top view. All length dimensions are shown in mm. (B) Microphone circuit, composed by HPF and gain stage and a 2nd order Sallen Key Butterworth Low Pass Filter.

The MMG_{MIC} circuit comprises a microphone with a pre-amplifier, a high pass filter responsible for attenuating any DC component from the microphone. A gain stage with 9 dB was added to match the microphone's dynamic range; see Figure 2B highlighted by the dashed line. The filter second stage is composed of a Sallen-key Butterworth low pass filter given a cut-off frequency of approximately 200Hz, highlighted by the dotted line; it was chosen a higher cut-off frequency than the known MMG frequency components 100Hz since the signal is not well-known. The circuit was simulated using TINA (Texas Instruments). The microphone signal was simulated using a voltage generator with a 0.8V DC level and a sine wave of 0.5V amplitude.

One of the essential specifications of the MMG Accelerometer is how the weight of the used sensor can affect the MMG signal, suppressing muscle activity. Watakabe et al. (2003) show that with the increase of accelerometer weight, the MMG signal is gradually distorted. Therefore, the accelerometer developed weighs 4.5g with enclosure.

3. Acquisition Protocol

To assess the mechanomyography function of the system, it was performed measurements during isometric contractions of the rectus femoris muscle with the varying load during knee extension movements. The volunteer did not report any current or ongoing neuromuscular diseases or musculoskeletal problems.

The volunteer was in a sat position and performed five consecutive knee full extensions, during 5 seconds of contraction and 15 seconds of rest. This protocol was repeated three times for different loads (no-load, 2Kg, and 4Kg) with at least 5 min of intervals between each series.

After the acquisition, the signal was rectified and segmented through the manual selection of the start and end of each contraction. The correspondent root mean squared value (RMS) was computed to obtain the indices of muscle force for each contraction series of each load. Pearson's correlation coefficient assessed correlation between MMG_{RMS} and loads.

4. Results

The developed circuit board was printed in a two-layer board, as shown in Figure 3 – A1, A2, A3, and the enclosures were printed in a 3D printer. Both MMG_{MIC} were held in place onto the subject skin by an elastic belt, and the MMG_{MODULE} and MMG_{ACC} were attached with double-sided adhesive tape as shown in Figure 3-B.

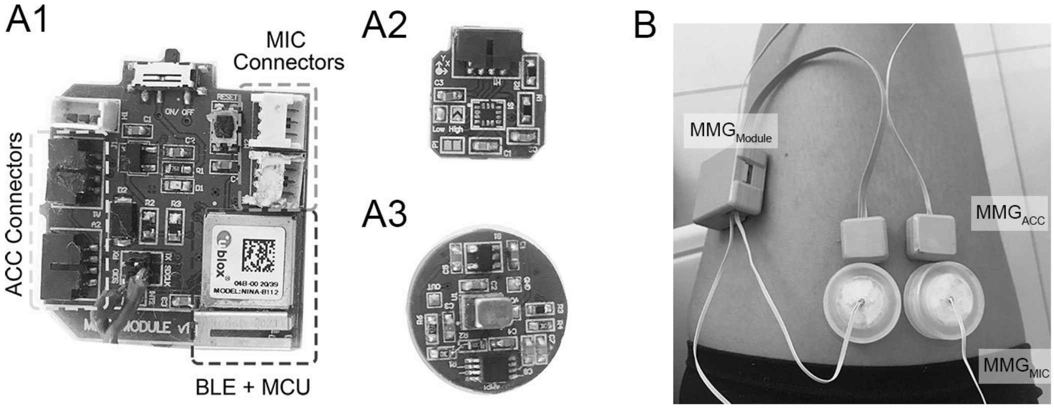

Figure 3: (A1) MMG_{Module} with main parts highlighted (A2) MMG_{ACC} (A3) MMG_{MIC} (B) Arrangement of MMG system placed in the rectus femoris muscle.

Although the MMG signal at the different detection points might not be the same, it was observed that only the amplitude was affected, but the duration of contraction and correlation between the RMS value and different loads can be considered similar, see Figure 4-A. As expected, the RMS was higher with the microphone than with the ACC [Krueger et al. 2014]; in order to compare the data was necessary to normalize the values.

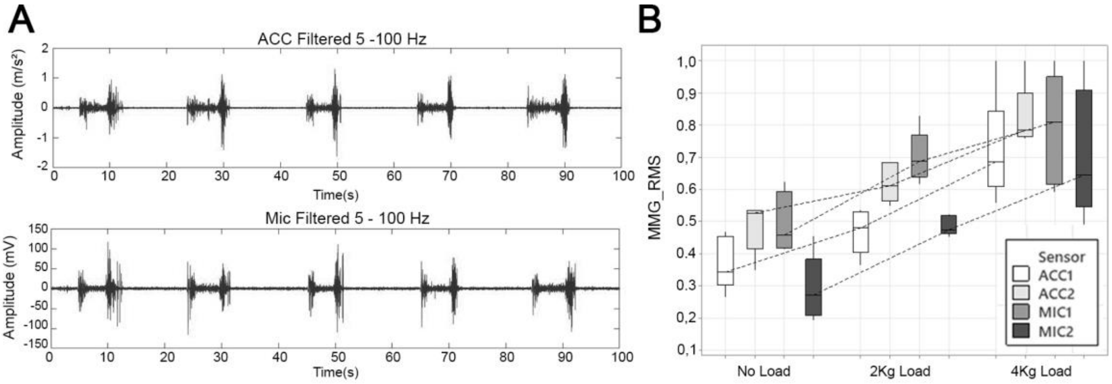

Figure 4: (A) Filtered Signal, MMG_{ACC}, and MMG_{MIC} (B) Normalized MMG_{RMS} for different loads for each sensor.

Figure 4-B shows the relation between MMG_{RMS} normalized of all sensors for the tested loads. Pearson's correlation between normalized MMG_{RMS} and loads was significative ($p<0.001$) for the ACC_1, ACC_2 and MIC_2 and ($p<0.005$) for the MIC_1, with correlation coefficient of 0.794 (0.475; 0.929) for ACC_1, 0.878 (0.666; 0.959) for ACC_2, 0.727 (0.343; 0.903) for MIC_1, and 0.825 (0.543; 0.940) for MIC_2.

As expected, the MMG_{RMS} showed a continual growth towards the load increment, which agrees with previous works [Ibitoye et al. 2014], confirming the relationship between the MMG signal produced and the muscle effort necessary to perform the contraction. Due to the small sample, it's not possible to infer about the standard deviation (SD) difference between loads. However, it's expected to get a higher SD when performing higher loads, due to the signal amplitude variation. We hypothesized that the MIC_2 had lower RMS median values in all loads due to the sensor positioning, taking into consideration a small area in which all 4 sensors were placed.

5. Conclusion

This paper proposed a new device topology based on MMG to monitor the uterine contraction during labor, providing greater information reliability and more comfort to the parturient due to its low weight and dimensions.

The MMG device produced reliable signals in terms of describing the muscle activity onset and offset and it was confirmed the relationship between the MMG signal produced and contraction forces required for different loads.

The results showed that the developed system was able to detect muscle contraction, both with the inertial sensor and with the microphone. For the next steps, the system will be applied to volunteers during labor and the data collected will be compared with the external tocography.

6. References

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