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*Original Article*

# Design and Validation of Mechanomyography and Torque Measurement Acquisition System for Skeletal Muscle Function

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***Abstract*** *- Assessment of muscle function is crucial for mitigating the risks of progressive motor weakness, which can ultimately lead to complete muscle impairment. However, existing technologies using commercial dynamometers are expensive, lack open-source availability and are not portable, limiting their accessibility in research settings. This study presents a cost-effective device to record muscle activity and the corresponding elbow joint torque. The device comprises three primary components:1) transducers for Mechanomyography (MMG), 2) torque signals detection, and 3) an application peripheral interface (API) for data acquisition control, visualization, and recording. Both transducers are integrated into an ATMEL ATMEGA 328. The device was validated on 36 able-bodied participants, measuring their MMG and torque across two sessions. Neuromuscular Electrical Stimulation (NMES) was applied to the Biceps Brachii (BB) muscle to induce elbow flexion. Further, submaximal torque and MMG were obtained using a commercial dynamometer and acceleration sensors for comparison. MMG measurements were observed at a maximum mean power frequency beyond 25Hz, while the torque information was found at 10 - 15 % of the Maximum Isometric Contraction (MVIC) induced by NMES. The measurement reliability was assessed using an Interclass Correlation Coefficient (ICC2,1) for elbow joint flexion torque (TQ RMS) and MMG RMS, yielding values between 0.522 and 0. 828. The ICC for the torque measurement device was 0.839, with SEM varying from 3.963 Nm to 11.149 Nm at a CV % of 2.565 to 13.123. These results underscore the potential of the developed device as a reliable, cost-effective alternative, with the added benefit of being replicable using locally available, low-cost electronics.*

*Keywords - Mechanomyography, Joint torque, Muscle function, Tri-axis accelerometer.*

## 1. Introduction

Sensory motor impairment is a leading cause of muscular weakness, which results in acute physical disability and mortality risk [1]. Regular evaluation of muscle function is crucial for assessing muscle status during daily occupational activities and monitoring functional recovery [2]. Existing muscle assessment technology includes the Manual Muscle Test (MMT), which quantifies muscle strength by evaluating the ability of the subject to resist a series of graded levels of resistance. However, the MMT technique suffers from variability among testers, leading to compromised decision-making and hindering the ability to track changing measurements over time [3]. The Handheld Dynamometers (HHD) detect muscle weakness or imbalance from a muscle group while the examiner applies resistance [3]. However, MMT and HHD are prone to errors arising from the tester, affecting the measurements' reliability and reproducibility [4]. In addition, these techniques cannot capture and represent the myographic feature of the muscle essential for tracking visual feedback of muscle activity.

Presently, existing myography assessment techniques such as Electromyography (EMG) [5] are costly, require experienced and trained individuals for their usage and fall short in providing insights into subsequent biomechanical mechanisms arising from muscle contraction, which are essential for joint movement and ability to perform tasks. Additionally, EMG is susceptible to electrical interference, commonly accounted for in medical devices [6]. This increasing demand for more reliable and compatible alternatives has prompted ultrasound imaging technology, which detects the variation in muscle's architecture, muscle thickness, and fascicle length correlating with joint torque [7]. However, ultrasound imaging is confined to laboratory setups due to its hardware and computational demands. Moreover, it cannot effectively detect muscle contraction along the direction of muscle fibers, which is useful for posture and gait analysis in stroke or Parkinson's survival [8].

Research has identified Mechanomyography (MMG) as a complementary approach for assessing skeletal muscle function [9]. MMG offers several advantages, including recording muscle activity from the skin's surface, eliminating the need for hair removal at electrode sites, and non-invasive attachment of sensors to clothing [10, 11]. Applications of MMG have accelerated the characterization of muscle properties, demonstrating its reliability for real-time feedback on subtle changes in muscle tone, which renders it reliable for detecting myopathy [12]. Given the growing demand for neurological interventions, which necessitates thorough screening, MMG's robustness against electrical interference is particularly well-suited for various studies of muscle function [12].

MMG signals can be recorded using various open-source transducers [9], including piezoelectric contact sensors [13], laser displacement sensors, phonomyography [14], Tensiomyography (TMG) [15], and Vibromyography (VMG) techniques. Piezoelectric contact sensors excel in measuring localized muscle vibration, making them ideal for studying specific muscles. In contrast, as non-contact transducers, laser displacement sensors provide comprehensive insights into overall muscle contraction and movement. An accelerometer is another valuable transducer that detects the mechanical vibration generated by muscle contraction and provides information on the frequency characteristics of muscle fibers [12].

This data is crucial for measuring force production and dynamic muscle properties. Accelerometers have been widely used in experimental research on muscle function, movement analysis, sports performance, and rehabilitation. TMG, which uses electrical stimuli to assess mechanical properties, focuses on aspects such as muscle stiffness, rate of contraction, and relaxation profiles [15]. However, the accelerometer complements TMG by capturing spectral characteristics of muscle activity. As such, acceleration MMG contains key neurophysiological insights, giving accelerometers a distinct advantage over other transducers for studying muscle dynamics [12].

Accelerometers are classified based on the type of signal they record and measurement axis configuration. Analog transducers, which detect raw analogue signals, require a signal conditioning process such as amplification, filtering, and analogue to digital conversion before further signal analysis. These processes can influence the signal-to-noise ratio, potentially compromising data quality. In contrast, digital transducers are well-suited for direct integration into digital signal processing systems, enhancing the signal resolution, linearity, and precision. Although accelerometers are versatile sensors used in various applications beyond MMG signal acquisition, they have demonstrated the capability to capture dynamic and isometric muscle activation across different intensity levels [16]. Early studies identified skeletal muscle frequencies ranging from 2 Hz to 120 Hz, corresponding to the dimensional changes in muscle fibers during muscle contraction. These frequencies can be detected using a single-axis [17], dual-axis [18], and tri-axis accelerometer [19]. In muscle function monitoring, the ADXL 345 and ADXL 335 accelerometers have been used to record the contraction of the BB muscle [20], producing MMG signals with consistent features. More recently, an open-source tri-axis accelerometer ADXL 313 was developed, but it has not yet gained attention to assess its performance. Due to its full-range resolution, lightweight design (weight < 2.6 g), flexible reconfigurable range at (10 bits: ± 0.5 g to 13 bits: ± 4 g), and ability to operate at full resolution across any g-range along three axes and given that muscle fibers contraction propagates in three directions at frequencies ranging from 5 Hz to 100 Hz, which falls in the bandwidth of 400 Hz available at ADXL 313, the sensor is particularly suited for MMG signal acquisition [21].

Practically, muscle strength is directly related to muscle activation. Hence, a need arose for devices capable of recording both neural activity and mechanical strength simultaneously. It has been hypothesized that MMG's magnitude correlates with muscle contraction's strength. Previous studies investigating MMG joint torque measurements have employed various recording modalities, including isokinetic dynamometers [22]. However, commercially integrated systems allowing concurrent MMG and torque signal measurement remain scarce. This highlights the gap to develop cost-effective devices that adhere to torque and MMG measurement acquisition outside laboratory environments.

Our literature review identified the FS2050-0000-15000-G force transducer [23] as an optimal solution for detecting varying levels of joint torque information in a standardized neurological evaluation [24]. This transducer operates on a 5V power supply, which makes it compatible with most microcontroller and commercial computing systems. The ATMEL ATMEGA 328 microcontroller, which supports digital and analogue functionalities, is well suited for integrating the ADXL 313 sensor and the FS2050-0000-15000-G force transducer [23].

With recent advancements in MMG and force measurement devices, it is imperative to assess the performance and reliability of these devices for different experimental sessions [25]. Recording MMG and force signals at known levels of dynamic muscle activation can provide valuable insights into the robustness and reliability of these devices [26]. Additionally, assessing the repeatability of these measurements across different days and sessions allows for the analysis of key parameters such as root mean square value of MMG (MMG RMS) and torque (TQ RMS) [26], which is useful for torque estimation. Building on these findings, we developed an in-house MMG and torque data acquisition system for outside laboratory usage. The device underwent rigorous bench-testing for functionality following European Medical Device Regulation and was validated on 36 able-bodied subjects. The BB muscle received Neuromuscular Electrical Stimulation (NMES) for MMG and elbow joint torque measurement to achieve a definite and specific analysis. NMES is crucial for assessing the function of the specific muscle of interest, offering clinicians and muscle trainers the opportunity to evaluate the muscle function under controlled conditions. The BB muscle was selected for this experiment because of its superficial accessibility and ability to accommodate both NMES electrodes and MMG sensors. This anatomical location allows proper MMG sensor placement for reliable data recording with limited interference from adjacent brachialis and brachioradialis synergy. We postulated the following hypotheses:

1) The low-cost MMG and torque data acquisition device provides information reflective of exercise outcomes,
2) The data collected from the device is reliable for use in muscle assessment or intervention,
3) The device can be easily replicated to meet the need of the new applications mitigating the risks associated with the lack of accessible tools for assessing sensorimotor function.

## 2. Materials and Methods

### *2.1. Equipment Selection*

Before developing the device, an extensive literature review assessed the current technology for MMG and torque data acquisition. This involved identifying cost-effective devices available on the market and those employed in early research endeavours. This research focuses on transducers and microcontrollers that are readily accessible and suited for physiological data acquisition procedures. Henceforth, this research centered on three key components:

1) The ADXL 313 accelerometer, chosen for capturing muscle contraction;
2) FS2050-0000-15000-G force transducer, selected for detecting muscle torque at the elbow joint; and
3) the ATMEGA 328 microcontroller which serves as the interface between the sensors and the system’s API for the data acquisition control.

These components were integral to achieving the intended research objectives while maintaining the performance, cost, and accessibility.

#### *2.1.1. Force Sensor*

A force transducer (FS2050 Compression LC1500 GRAM, TE Connectivity, Schaffhausen, Switzerland; full-scale range = 15 Nm; span = 1 – 4V; zero offset = 1V; voltage rating = 5 VDC) (Figure 3) was employed to measure the force applied at a constant lever arm across all subjects.

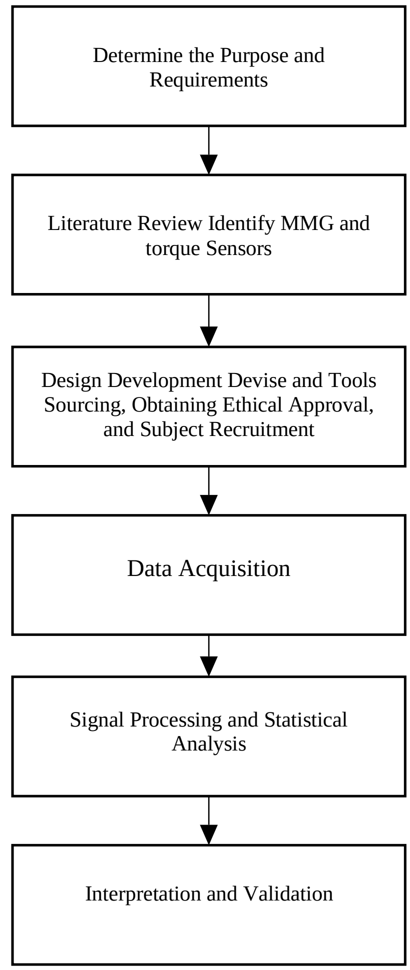


**Fig. 1 Block diagram of the device development**

Following the manufacturer’s instructions [27], the calibration curve was established by applying the sensor’s sensitivity as described in Equation (1) and the output of the sensor (W(kg)) as expressed in Equation (2).

$$Sensitivity = \frac{Span}{Full\ scale\ range} \quad (1)$$

$$\mathrm{W} = \left(\frac{\text{sensor output} - \text{zero offset}}{\text{sensitivity}}\right) \times 10^3 \quad (2)$$

The load cell was securely mounted on a horizontal stand, and the voltage readings were systematically recorded as various weights ranging from $5{\times}10^{-5}$ to 2 kg were successively added to the cell. These voltage measurements were captured using the custom-built Data Acquisition (DAQ) system. Subsequently, the voltage characteristic curve was generated (Figure 3).

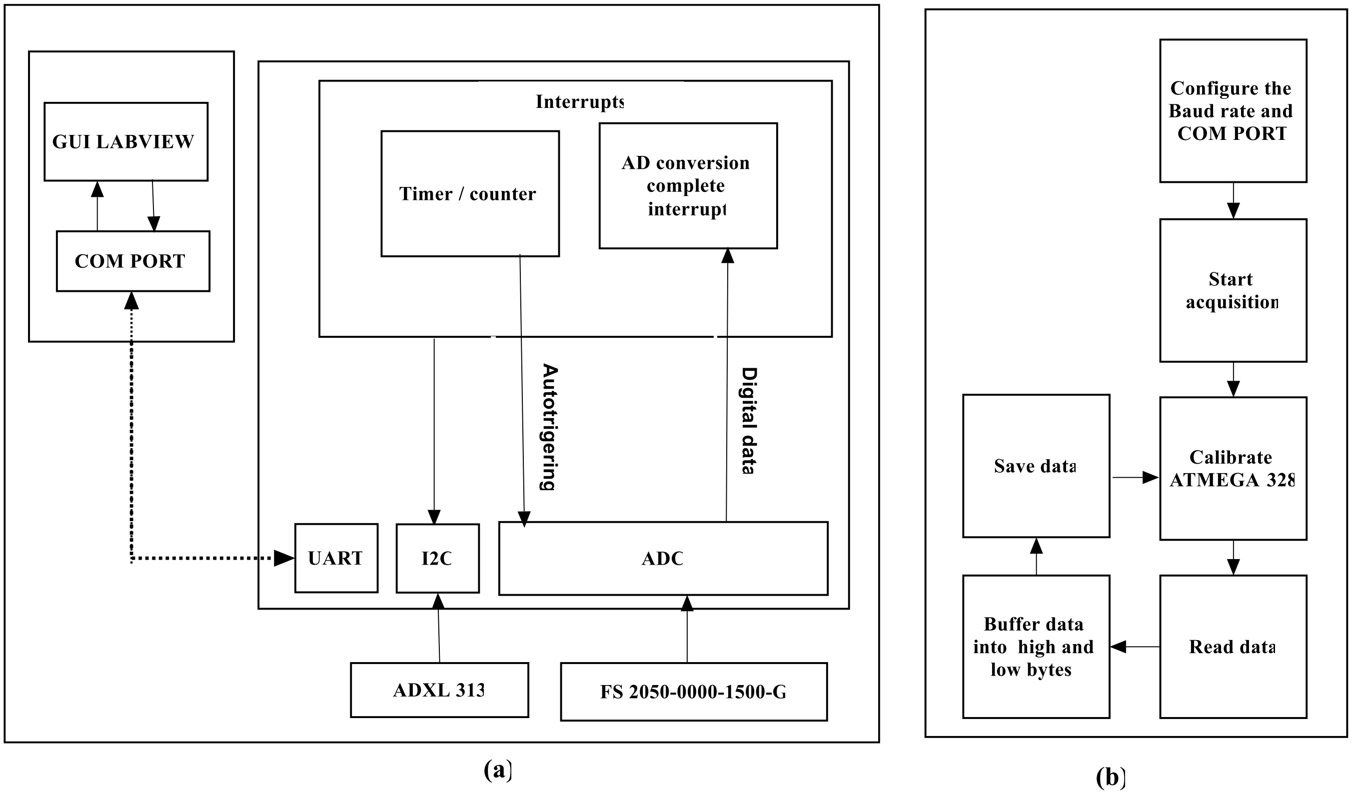


**Fig. 2(a) Flow diagram of the acceleration and force data acquisition unit, and (b) Read and write process.**

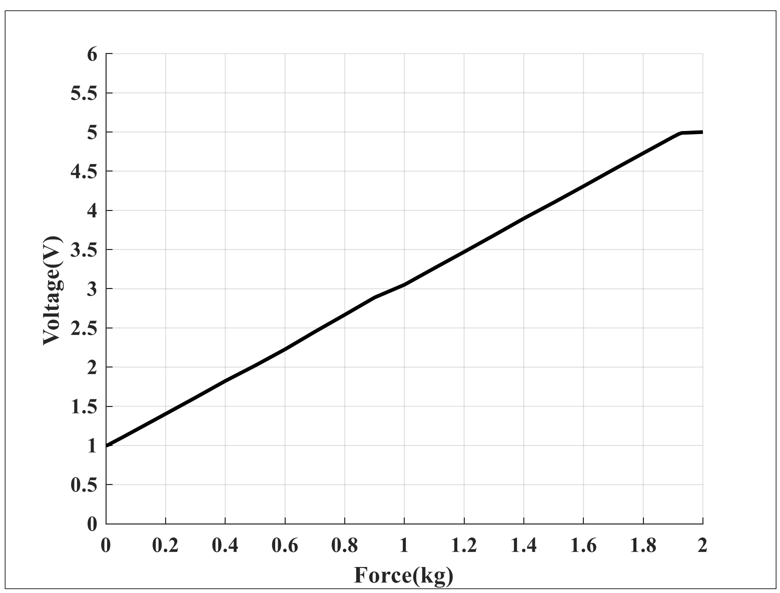


**Fig. 3 Calibration curve of the force sensor**

*2.1.2. Acceleration Sensor*

A digital microelectromechanical triaxial (X, Y, Z) accelerometer ADXL313 (SparkFun, Colorado, USA; weight < 2.6 g; resolution = 10−13 bit; g range = ± 0.5 − ± 4 g; bandwidth = 0 to 3.125 – 1600 Hz; sensitivity = 1024 LSB/g for any g range; dimensions = 5 mm × 5 mm × 1.45 mm LFCSP package; voltage rating = 3.3 V), was used to capture muscle fiber contractions propagating in three axes. The manufacturing design of accelerometers enables the measurement of static and dynamic accelerations. To eliminate the static acceleration and focus on the dynamics of muscle activation, a zero-g bias correction was achieved by adjusting the built-in offset registers. ADXL 313 features a scaling factor of 3.9 mg/LSB, equivalent to ±0.5 g for an 8-bit register. However, this register capacity cannot entirely nullify the 1 g static acceleration. Thereafter, the acceleration output from each of the three axes of the acceleration sensor (Xg, Yg and Zg) was refined by averaging a series of samples per axis. These averaged values, which vary with elbow geometry, are stored as correction factors required to zero out the static acceleration at each axis. The corrected values X, Y, and Z are obtained by adding these stored correction factors (Xg, Yg, and Zg) to the raw accelerometer readings.

*2.1.3. Firmware and Hardware Operation*

The literature on MMG underscores the importance of sampling acceleration data at a rate exceeding the highest frequency in the standard MMG spectrum [28]. A sampling frequency was set using ADC on the ATMEGA 328 microcontroller, along with hardware timers and registers, to meet this requirement. The board was configured to trigger at 1 ms intervals (1kHz) through an interrupt and prescaler mechanism. During operation, acceleration and force data were transmitted to the PC using an Interrupt Service Routine (ISR) running at 1Khz. This synchronous process was initiated by an Interrupt Service Request (IRQ) that triggered both ADC and $I^2c$ communication assisted with a timer and a prescaler set at the compare match register. The entire data acquisition process was coordinated through an API developed in LabView, as illustrated in Figures 2 and 4. This synchronization was implemented using timers and registers, employing a compare match register and prescaler as detailed in Equation (3).

$$\text{Prescaler} = \frac{\text{ADC clock}}{\text{Desired Sample Rate}} + 1 \tag{3}$$

Where the ADC clock is set to 16 MHz, the desired sample rate is 1000 Hz, and the prescaler ranges from 0 to 256 for the 8-bit Timer 1.

***2.2. System Testing and Verification***

The device's functionality was evaluated by assessing the response of the force sensor and the acceleration sensor controlled by the developed firmware. The operation of the force sensor was validated through voltage calibration (Equation 3).

At the same time, the performance of the accelerometer was assessed by analysing the power spectrum density of each axis of the acceleration sensor (Figure 7). In addition, the software and firmware underwent testing for fault recognition, error handling, and accurate execution of the command.

To enhance efficiency, the delay between the execution of data recording and the command functions was optimized by calculating the difference in byte position of data received by the computer from the ATMEGA 328 microcontroller chip. Following successful testing and optimization, the device was deployed for data acquisition.

***2.3. Validation of the Device on Healthy Subjects***

*2.3.1. Experimental Protocol*

Thirty-six healthy male subjects (age 22.24 ± 2.94 years; height 172 ± 0.5 cm; weight 67.01 ± 7.22 kg) attended this study. None of the subjects had a history of neuromuscular disorders, surgical procedures, and skin conditions.

In addition, the subject refrained from physical activity involving muscle training for 72 hours before attending the experiment. After being briefed on the study's objective, all subjects provided a signed informed consent form. The study received ethical approval (NMRR-20-2613-56796 [IIR]) from Malaysia's Medical Research Ethics Committee, which ensured compliance with the principles outlined in the Declaration of Helsinki.

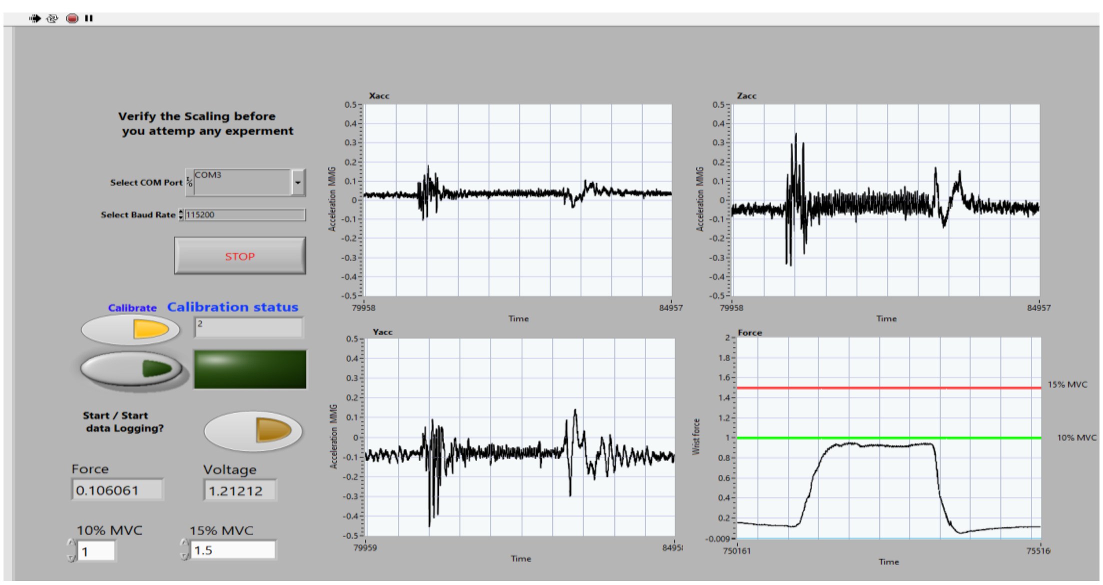


**Fig. 4 Peripheral interface of the data acquisition device**

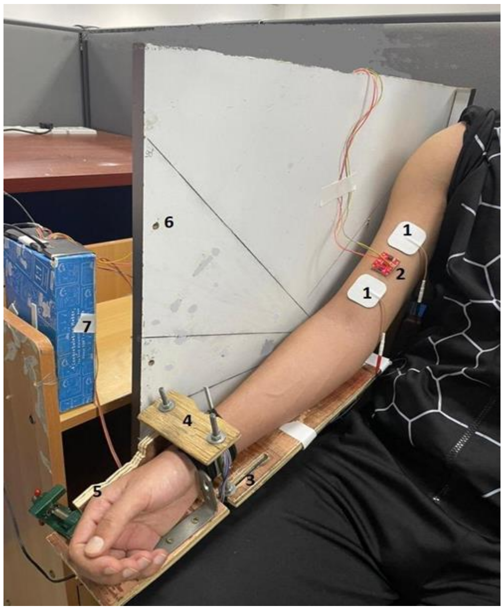


**Fig. 5 Experimental setups of data acquisition for acceleration and torque data acquisition: 1) NMES electrodes, 2) MMG sensor, 3) Adjustable elbow rest and fixture for posture, 4) Support of the force sensor cuff (placed underneath), 5) Fixture of the hand/wrist from flexion, extension abduction, and adduction, 6) Fixture for the adjustable elbow joint angle, and 7) Force and acceleration data acquisition device.**

The experiment involved applying NMES to the BB muscle. Subjects visited the laboratory on three separate days. On the first day, they participated in a warmup protocol, acquainted themselves with generating maximum voluntary contraction and experienced the sensations of NMES. On the second and third days, both acceleration and torque data were concurrently collected while NMES with a frequency of 30 Hz, a pulse width of 110 µs, and current amplitude of 30 mA, ramp time of 1 s, on for 6s and off for 2s, administered to the BB muscle of each subject for 30 s. A minimum of 10 minutes was allowed between 2 consecutive recordings at two different angles, and a five-minute session was provided between 2 trials at a single elbow joint angle. The elbow joint angles and forearm postures were randomized to prevent muscle fatigue. Any recording trial during which the muscle failed to produce at least 10% of the nominally tolerated maximum NMES-evoked muscle contraction force was excluded from the analysis (Figure 6).

Electrical stimulation electrodes were affixed following the guidelines outlined by the International Society of Electromyography and Kinesiology (ISEK), supervised by the medical physician at the experimental site. The muscle belly was located by palpation [29] when the elbow was initially flexed at 90°. Subsequently, the elbow flexion angle was randomly varied among 10°, 30°, 60° and 90°. ADXL-313 acceleration sensor was securely attached to the muscle belly using adhesive tape (3MTM VHBTM 4920, Center St. Paul, MN, USA). The torque was measured using a force transducer FS2050 Compression LC1500 GRAM, TE Connectivity (Schaffhausen, Switzerland). A lockable armrest compensated for the gravitational effect [30] (Figure 5). Data recorded with Arduino script and LabVIEW platforms were sent to the computer hard disk for subsequent offline analysis. All data and subject information were password-protected by the investigators for privacy.

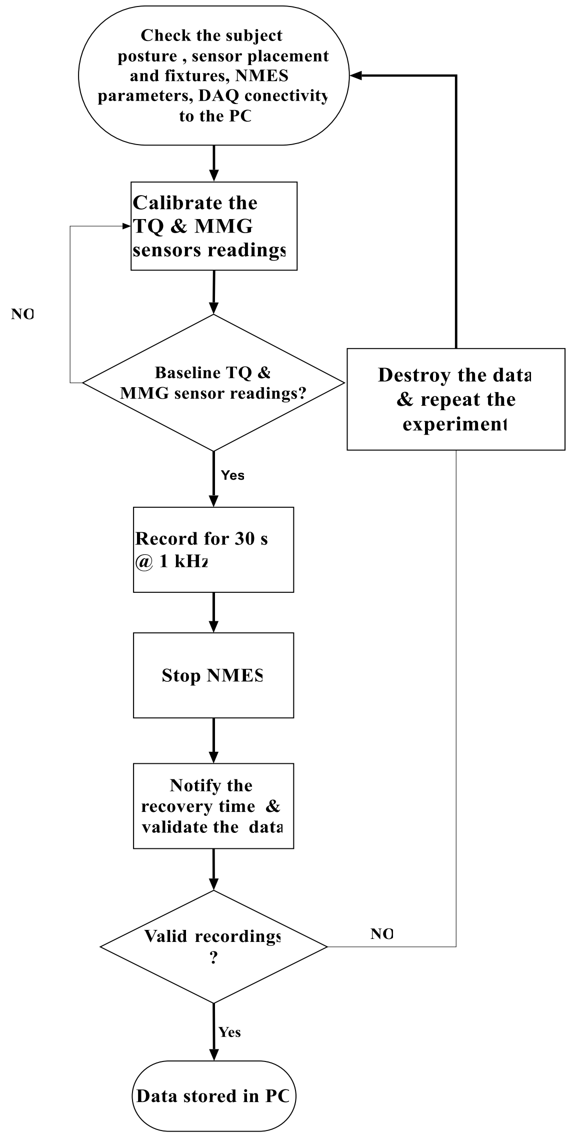


**Fig. 6 Flowchart of the data recording using the device**

### *2.3.2. Signal Processing and Statistical Analysis*

Acceleration data were timestamped and transmitted to the PC at a rate of 1 kHz. To extract the MMG data, acceleration signals were bandpass filtered by a fourth-order Butterworth filter with a frequency range of 5–100 Hz [31]. Subsequently, the power spectrum was conducted to validate extracted MMG from acceleration data. The torque data were cleaned using a fourth-order Butterworth filter at a cutoff frequency of 5 Hz. The analysis segments were obtained by a moving window for 1000 ms and at a threshold of 20% at the start and end of muscle contraction for MMG and torque information. MMG RMS and TQ RMS features were calculated by averaging filtered acceleration and torque data midsection. All features were normalized to the peak contraction values obtained for each subject.

The Shapiro-Wilk test was employed to test the normality of the MMG RMS and TQ RMS. Intra-class Correlation Coefficient (ICC2,1) with a two-way mixed effect model, single measurement was calculated to examine the relative agreement of MMG RMS and TQ RMS over 2 recording sessions [32]. A paired sample t-test was employed to examine the repeatability of the scores obtained from the two sessions. Additionally, the torque measurement device was validated by analyzing its correlation against measurements obtained using a push-pull dynamometer (SF−200, range = 0.5−200 N, resolution = 0.01 N, 100 – 200 V, Aliyiqi, Mainland China, China) under submaximal voluntary contraction equivalent to NMES. Investigated variables were significant for $p < 0.05$. All the statistical tests were conducted using IBM SPSS 25.0 (SPSS Inc., USA).

## 3. Results and Discussion

### 3.1. Results

The tri-axis MMG signals showed MMG to be valid in the range of 5-100Hz. The power spectrum analysis [33] showed a maximum outside tremor range (see Figure 7). Additionally, the MMG sensor exhibited average RMS values of 0.03, 0.07 and 0.08 g in the muscle fibres' longitudinal, lateral, and transverse directions. The reliability of measured outcomes assessed using ICC (2,1) and SEM, MDC, and CV% ranged from 0.522 to 0.823 for TQ RMS and MMG RMS at overall investigated elbow joint and forearm configurations for NMES session. The developed device for torque assessment was further validated by using Pearson's correlation analysis. Measured data from two devices for two trials. The intraclass correlation coefficient for the torque measured by FS2050-1500G was 0.973 and 0.984 for the data recorded by the SF−200 push-pull dynamometer. The ICC (2,1) for both torque measurement devices was 0.839, and the SEM varies from 3.963 Nm to 11.149 Nm at a CV% of 2.565 to 13.123. A paired sample t-test found non-significant differences at MMG RMS and TQ RMS across all trials and testing sessions ($P > 0.05$).

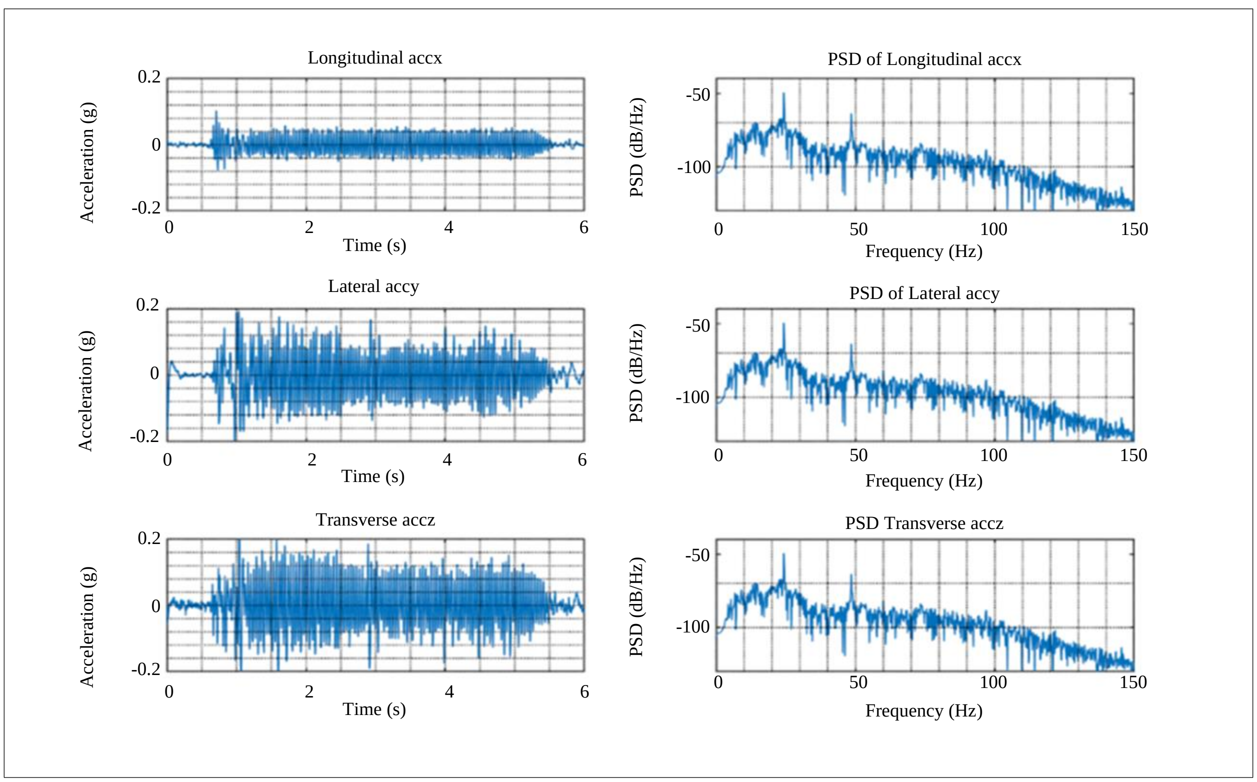


**Fig. 7 NMES -evoked MMG from the right biceps brachii (left) and the PSDs (right)**

### 3.2. Discussion

The BB muscle plays a crucial role in performing activities such as lifting, pushing and pulling. The assessment of elbow flexion strength is a reliable method for evaluating the functionality of the BB muscle. Although isokinetic dynamometers obtained reliable measurements, their dimensions and cost limit their usability outside laboratory settings [34]. In addition, simultaneous recording of myography and torque for muscle function requires high-cost software for firmware development. This research presents the design, testing and validation of a data acquisition device engineered to record muscle activity and torque concurrently. The device was evaluated on able-bodied subjects, measuring the MMG from the BB muscle and elbow joint torque. The magnitude and spectral parameters exhibited reliability compared to established experiments utilizing commercial data acquisition systems.

The inter-test reliability of the torque measurement showed SEM varying between 3.963 and 11.149 Nm. A non-significant difference was obtained from MMG and torque across 2 consecutive experiment trials and daily testing sessions at a 95% confidence interval. These measurements confirm that developed torque measurement devices produce consistent results that are accountable for normal physiological variability of the muscles and are useful to monitor subtle changes in muscle function over time required in ergonomics [35]. The device was developed from an open-source microcontroller and sensors, thus underscoring the potential for the device's replication in assessing other muscle groups, supporting the developers' hypotheses.

The MMG RMS measured across three axes aligns with the findings of Talib et al. [28], which claimed different levels of significance of RMS values observed in the transverse compared to the longitudinal and lateral axes. The peak-to-pick values ranging from 0.1 to 0.4 g are consistent with the results reported in [13], measured explicitly from the BB during isometric and in the isokinetic contraction experiments. Notably, the weight of the MMG sensor was found to reduce the mean frequency [36] and amplitude of MMG [37].

The weight of ADXL 313 <2.6g is reliable with less attenuation. Thus, the study contributed valuable insights into integrating these transducers for bio-signal acquisition. The utilization of the ADXL 313 accelerometer is crucial for characterizing the spectral and temporal features of MMG signals in all propagation directions. These results demonstrate that the obtained muscle strength and MMG measurements are compatible with those derived from the widely accepted VMG sensor, indicating significant potential for monitoring neural behaviours [38].

The applications of MMG technology have broadened to include prosthetic control [39], muscle-strength assessment [40], clinical monitoring [41], and screening respiratory muscle function [42]. With the increasing prevalence of neurological impairments among the geriatric population and the rise of muscle-related conditions linked to occupational and physical activities in young and middle-aged individuals, there is an urgent need to thoroughly investigate the biomechanical aspects of muscles before product design [43]. Many prominent studies on upper limb muscle function rely on hand-held dynamometry, which necessitates both the subject and tester strength, is valid within a limited range of joint positions, and is influenced by body posture [25]. In contrast, our study systematically exposed the elbow to various elbow joint angles and forearm postures, broadening our investigation's scope. While some studies have utilized isometric dynamometers for the wearer's comfort, these devices are primarily restricted to laboratory settings and are not open source [34]. Our design effectively harnesses the potential of microcontrollers, wearable, and mechanical force sensors to create versatile setups for efficiently screening musculoskeletal functions.

The characterization of muscle function using the Fourier analysis [44] and the assessment of MMG amplitude shows a strong correlation with the coefficients of the continuous wavelet transform [45]. These two analysis techniques demonstrate interchangeability, particularly in dynamic muscle studies. Furthermore, employing the MMG spectrum determined by the PSD through Fourier transformation [46] facilitates rapid and efficient computation. Consequently, the acceleration MMG characteristics revealed by the PSD in the present study align with the MMG's established spectrum range, particularly during submaximal contractions. This cost-effective, open-source design, utilizing straightforward tools, holds a strong promise as an affordable solution to mitigate the subjectivity associated with imprecise muscle studies.

Given the limited availability of open-source devices and the scarcity of prior knowledge, this study is the first to validate using a tri-axial ADXL 313 accelerometer and FS2050-000-1500-G transducer for muscle studies. This research confirms that their MMG and torque study application can attain reliable data for both measurements.

Despite potential contributions in myography research such as EMG and VMG, resource constraints hinder research and development in low-income communities. While this investigation has yielded promising insights, further studies are encouraged to evaluate the device's reliability for small muscle groups and their unique functions. Testing has demonstrated the potential of the device to incubate motor training monitoring. Existing modalities relying on VMG are typically confined to laboratory settings and cannot monitor the rate of muscle recruitment in the direction of muscle fibers. Future investigations should improve this device to track muscle re-education in case of acute sensory impairment and post-surgical monitoring.

### *3.3. Study Limitations*

This study has several limitations that necessitate careful interpretation of the findings. First, the device was exclusively tested on healthy, untrained individuals with low muscle contraction, limiting its applicability to unverified populations with muscular strength. Furthermore, the device has not been evaluated on muscle-impaired individuals such as stroke survivors, cerebral palsy and patients recovering from accidents. Future research should focus on validating the device with patients, older adults, and athletes to broaden its generalizability.

Second, the device was tested solely on male subjects, which restricts a comprehensive understanding of its applicability. Although female subjects have demonstrated muscular excitability to NMES, they also exhibit greater muscle response variability than subjects [47].

Third, the device was designed primarily as an isometric measurement system to enhance portability. While isometric methods evaluate muscle strength with a fixed joint angle, isokinetic and isotonic approaches assess muscle strength during joint movement, with the former minimizing the risk of muscle injury. Since movement speed is critical for evaluating daily functional performance, its omission is a limitation of the current device. Future research will expand the device's use to isokinetic measurements and passive motion applications in knee rehabilitation.

Precise muscle strength measurement and trend tracking are crucial in clinical practice. Conventional methods, such as MMT and HHDs, have limited reliability and reproducibility, while isokinetic dynamometers are less accessible in low-income regions [25]. The device developed in this study demonstrated high reliability in measuring elbow flexion strength and muscle function, offering improved accessibility and cost-effectiveness. This makes the device a viable alternative to MMT and IKD for muscle strength assessment.

## 4. Conclusion

The device integrates the force and acceleration sensors on the ATMEGA 328, facilitating the simultaneous recording of MMG and torque information from muscles. Dynamic muscle contractions were evaluated through NMES over two sessions, and experimental trials were designed to assess the repeatability of MMG RMS and TQ RMS. The results consistently demonstrated moderate to high agreement between the measurements obtained from the developed apparatus. These findings underscore the device's suitability for future replication and validation in clinical research settings and the application for machine learning using MMG and torque features across a cohort of different subjects.

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