



## MECHANOMYOGRAPHIC AMPLITUDE TRACKS MUSCLE MECHANICAL RESPONSES DURING NMES-EVOKED TORQUE PRODUCTION

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### ARTICLE INFORMATION

Submitted 18 Nov., 2023  
Revised 29 January, 2024  
Accepted 5 February, 2024

#### Keywords:

Electrical stimulation  
Muscle force  
Mechanomyography  
Accelerometer  
Motor unit recruitment

### ABSTRACT

This study investigates the pattern of relationship between mechanomyographic (MMG) signal amplitude and neuromuscular electrical stimulation (NMES)-evoked muscle contractions in healthy volunteers. Recent evidence has suggested that NMES-evoked muscle contractions enhance motor ability for physical performance in healthy individuals, maintain muscle health, and offset secondary complications of inactivity in persons with neurological conditions. However, the traditional NMES operation is inefficient, thus making the NMES outcomes less than optimal. Responsible for this is the lack of mechanical responses of muscle from the electrically-evoked limb during NMES-evoked contractions. This study investigated the use of a muscle contraction signal (mechanomyography, MMG) generated during NMES-evoked contractions in healthy knee extensors in tracking the muscle mechanical responses during NMES-evoked force production. Six healthy males underwent NMES-evoked submaximal-to-maximal isometric contractions, in random order, of their knee extensors at 30deg and 90deg knee angles on an isokinetic dynamometer. The MMG signal was acquired through an accelerometer sensor affixed to the rectus femoris (RF) muscle belly. Paired samples t-tests was used to compare the relationship between the two knee angles (90deg and 30deg) tested. Results show that the mechanomyographic amplitude (MMG-RMS) increased with increase in torque production and stimulation intensity up to 50 mA and 60 mA at knee flexion angles of 90deg and 30deg, respectively, before the appearance of plateau-like contractions—an indication of fusion of contracting muscle fibre at maximal contraction level. Although there were knee angle differences in the pattern of relationship between MMG and stimulation intensity, the effect was only statistically significant ( $p \leq 0.05$ ) for the torque versus stimulation intensity plot. These results suggest that MMG-RMS tracks motor unit recruitment strategy that might be responsible for muscle force/joint torque modulation during electrically stimulated contractions. The MMG signals may, therefore, have a clear application as an indicator of muscle force in areas where muscle force is needed and impractical to measure

### 1.0 Introduction

Although healthy volunteers, athletes (Micke et al., 2023) and non-athletes (Yoo et al., 2023), have demonstrated improved muscle strength and joint torque production following neuromuscular electrical stimulation (NMES)-evoked contractions (Filipovic et al., 2012), the open-loop mode of operation of commercially available NMES systems precludes the optimal application of the technology for routine clinical practices (Gil-Castillo et al., 2020; Keller et al., 2002). Alternatively, mostly through simulation studies, the closed-loop controlled of NMES

systems promise to automate the modulation of stimulation parameters for optimal outcomes (Gil-Castillo et al., 2020). However, for effective and efficient implementation, a closed-loop system requires reliable feedback from the controlled limb (Jensen et al., 2002). Practically, the information on the muscle force production which is required, in real-time, to automate the operation of such NMES systems (Peckham and Knutson, 2005) is difficult and may be impossible to measure directly for this application (Li et al., 2018).

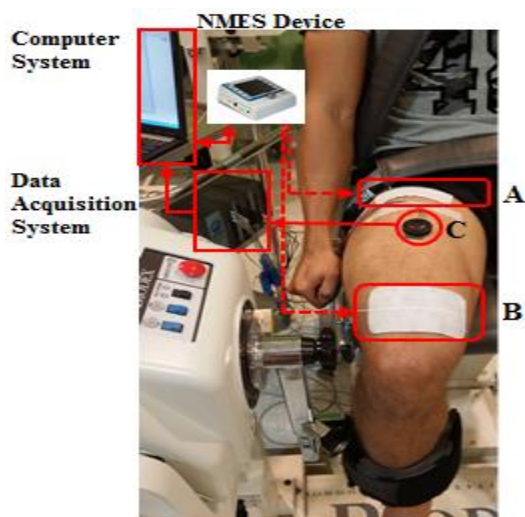
The inability of the conventional muscle function assessment modalities to non-invasively measure the mechanical responses (force, fatigue, or joint position) sequel to the motor control of muscle contractions (motor units (MU) recruitment and firing rates) suggests the need for a complementary alternative (Ibitoye et al., 2014). Muscle mechanomyography (MMG), due to a low frequency of muscle dimensional changes (Cè et al., 2022; Orizio, 1993), has been established as a reliable measure of muscle surface oscillation during NMES-evoked muscle contractions (Brandenberger et al., 2023; Orizio et al., 1997). Impetus from the earlier work of Orizio et al. (1997) and that of Gobbo et al. (2006), have also suggested a good relationship between muscle contractile properties and MMG parameters during NMES-evoked contractions. Apart from the ease of attachment of MMG sensors onto the skin surface, MMG measurements are immune from stimulation artefact and represent the summation of the activities of active motor units—useful to estimate the neural activation level and the associated muscle force modulation (Beck et al., 2004). However, the probable nonlinear relationship between the NMES-evoked contractions and the generated muscle force/joint torque (Rabischong and Chavet, 1997) as well as with the MMG amplitude features (due to the muscle fibre fusion at a high contraction intensity level) (Orizio et al., 2003) warrants that the MMG signal be continuously reassessed during clinically relevant muscle tasks. This may provide insight into the possibility of MMG signal deployment to interpret physiological phenomena underlying NMES-evoked muscle force production.

Earlier, Windhorst et al. (1986) suggested that there is a reduction in the “mechanical effect of MU activation” during stimulated contractions in cat medial gastrocnemius muscle. In humans, Orizio et al. (2003), tracked the suggested pattern with MMG signal during sustained voluntary contractions of biceps brachii muscle. These authors recommended the application of MMG for improved identification of muscle activity during NMES-stimulated contractions. Previously, in our laboratory, we have identified a “strong relationship between MMG signals and torque during NMES-induced contractions” in spinal cord injury survivors (Ibitoye et al., 2016). Using a machine learning method, we have also estimated knee torque from MMG signals (and other relevant input parameters) in comparison to the torques generated from the isokinetic dynamometer (Ibitoye et al., 2016). To our knowledge, no analysis of the MMG amplitude (MMG-Root Mean Square (MMG-RMS)) only at different levels of NMES-evoked isometric contractions and at different knee angles has been examined to evaluate muscle mechanical responses using traditional statistical methods. The objective of this study was to examine whether the MMG signal amplitude could track the mechanical responses of the knee extensors during NMES-evoked contractions for force/torque production. Additionally, the study also sought to investigate if there are statistically significant effects of knee angle on the MMG signal amplitude relationship with the incremental NMES-evoked muscle contractions.

## **2. Methodology**

### **2.1 Study Participant**

The University of Malaya Medical Ethics Committee (Approval No: 1003.14 (1)) approved the experimental procedure performed in this study. The schematic representation of the experimental setup is as presented in Figure 1. Six healthy male participants [mean (SD) age, 23.2 (1.3) years; height, 173.0 (0.1) cm; body mass, 72.2 (5.2) kg] volunteered for this study. All study participants consented to participate in the experiment.



**Figure 1.** Schematic of the experimental setup on an isokinetic dynamometer with “A”, and “B” representing the cathode and anode electrode, respectively of the NMS. “C” stands for the accelerometer sensor (to acquire MMG signals).

## 2.2 Experimental Procedure

Following initial familiarization sessions to habituate the participants to the NMES-evoked muscle contractions of moderate-vigorous intensity, they were safely secured, during the experiment, to an isokinetic dynamometer (Biodex Medical System (System 4), Shirley, NY, USA) using a rigid “restraining straps” over the pelvis, trunk and the thigh to reduce body movement during testing in accordance with the standard practice (Brown and Weir, 2001) (Figure 1). This procedure was necessary to prevent participants’ voluntary contributions to the muscle contractions and joint movements as much as practically possible.

For electrical stimulus contractions, electrical pulses (“square-wave pulses” at 30 Hz frequency and pulse duration of 400  $\mu$ s, of increasing current amplitude from 20 mA to 80 mA) was imposed for 4 secs to allow sufficient contraction duration for data collection (Orizio et al., 1992). NMES pulses were administered via a computer-controlled neurostimulator (RehaStim<sup>TM</sup>, Hasomed GmbH, Germany) through 9×15 cm<sup>2</sup> electrically-conductive and self-adhesive electrodes (Hasomed GmbH, Germany) on the quadriceps muscle of each participant’s dominant leg (Adams et al., 1993) (Figure 1). This method was used in this study because a similar stimulation pattern has been previously recommended (Babault et al., 2001) stimulated contractions without premature muscle fatigue.

## 2.3 Measurements

### 2.3.1 Incremental Peak Torque

As described earlier, following the administration of a series of randomly ordered NMES-evoked submaximal isometric muscle actions delivered using stimulation current from 20mA to 80mA in 10mA increments (seven different levels of stimulation intensities) for 4 secs (Ibitoye et al., 2016), joint torques were produced. The NMES-evoked torque productions were obtained from the isokinetic dynamometer on which the participants were safely secured. To prevent voluntary effort by the participants during testing, there was no visual feedback displayed by the dynamometer to the participants. This was in addition to the instructions given to the participants not to contribute or prevent NMES-evoked muscle contractions through any voluntary effort. This was because the experiment was only meant to assess the torque production by NMES. In each trial, the value of the highest NMES-evoked torque was used as the neurostimulation-evoked peak torque (PT). From the PTs’ values, percentages of submaximal torque levels were subsequently calculated for analysis. In order to minimize

potential muscle fatigue, a recovery duration of 10-min was allowed after each trial before the next.

### **2.3.1 Mechanomyographic signal**

While the generated torques were being measured by the isokinetic dynamometer (Figure 1), MMG signals were simultaneously being collected from the rectus femoris (RF) muscle of the quadriceps with an accelerometer sensor (Sonostics BPS-II VMG transducer used in Biopac MPI50 environment) affixed using an adhesive tape (Barry, 1991). The attachment of the accelerometer sensor was directly onto the belly of RF muscle. This was meant to obtain the maximum muscle contraction through surface oscillation of the muscle (Barry, 1991). RF was selected for investigation because it is functionally relevant and a major contributor to the common actions of daily activity, such as standing—where it is considered to be near isometric contraction during knee extension (Antonelli et al., 2009).

### **2.3.3 Signal Processing**

The acquired MMG signals (sampling rate: 2 kHz) were band-pass filtered digitally (20-200 Hz). The signals were amplified and stored through AcqKnowledge data acquisition and analysis software (MPI50, Biopac Systems, Santa Barbara, CA, Inc.) for further analysis using programs written in the LabVIEW programming software environment (version 12.0, National Instruments, Austin, TX, USA) (Ibitoye et al., 2016). From the MMG and torque signals, the 2sec segment was selected (the middle position where maximum muscle recruitment might have occurred). This segment selection was based on the standard procedure (Orizio et al., 1992).

Thereafter, the MMG-RMS was extracted for further analysis. To account for any potential inter-individual variability (due to muscle composition-related factors during submaximal electrical stimulus-evoked contractions in each participant), the MMG-RMS and PT for the highest stimulation intensity were used to normalize their respective values at submaximal levels. This procedure is common in related studies (Beck et al., 2005; Keller et al., 2018).

### **2.3.4 Statistical Analysis**

The comparison between the two knee angles tested (90deg and 30deg) on MMG-RMS or PT versus stimulation intensity data, at each stimulation intensity level, was examined using paired samples *t*-tests. Microsoft Office Excel 2013 (Microsoft, Redmond, WA, USA) and SPSS (IBM SPSS for Windows Version 20, NY, USA) were used for the statistical analysis. Statistical significant (i.e.,  $p \leq 0.05$ ) was assumed for comparison between knee angles at each stimulation intensity level.

## **3. Results and Discussion**

Figure 2 presents the plots of normalized NMES-evoked peak torque versus stimulation intensity (A) and normalized MMG-RMS versus stimulation intensity (B) at 90deg and 30deg knee angles. The statistical analysis results of the data presented was also indicated on Figure 2 where the paired samples *t*-tests was used to statistically compared the relationship between the two knee angles tested. The peak torque was significantly different between the tested knee angles at particular lower and higher stimulation intensities ( $p < 0.05$ , (Figure 2 A)) but the MMG-RMS between the tested angles was found with no significant difference (Figure 2 B). Specifically, in Figure 2, the significant increase in the PT at 90deg over 30deg knee angle at (A) for 70 mA and 80 mA reflected as a decline in the MMG amplitude in (B). On (B), at 60 mA to 80 mA, the higher the torque production the lower the MMG amplitude. This is suggestive of muscle mechanical response during high stimulation intensity where muscle stiffness could lead to a decline in MMG amplitude (Beck et al., 2005).

### **3.1 Torque Response, MMG Response versus Stimulation Intensity**

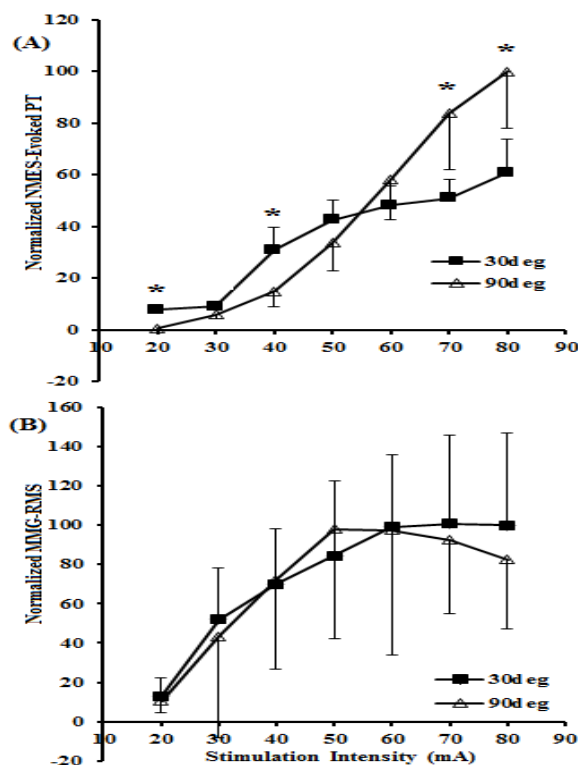
The results presented in Figure 2 based on the two knee angles investigated, revealed that at low stimulation intensities (from 20-50 mA), isometric torque production increased with

reduction in the knee flexion angle, but increased with increasing knee flexion angle from 50-80 mA (Figure 2 (A)). This pattern was also mirrored by the MMG versus stimulation intensity plot (Figure 2 (B)) up to 40 mA, except for the appearance of a plateau-like response of MMG-RMS at 60 mA. Specifically, at 90deg knee angle, the MMG amplitude increased up to 50 mA stimulation intensity, and declined thereafter up to the stimulation intensity of 80 mA while at 30deg knee angle, the MMG amplitude increased up to 60 mA stimulation intensity before plateauing up to the stimulation intensity of 80 mA. It can be proposed that, although not statistically significant at stimulation contraction levels, the different patterns of the MMG amplitude versus stimulation intensity were due to knee flexion angle differences in muscle stiffness and intramuscular fluid pressure as earlier suggested (Ebersole et al., 1999; Orizio et al., 2003).

The MMG (Figure 2) decline at high stimulation intensity has been suggested to be due to the “fusion of motor unit twitches” (Beck et al., 2004; Orizio et al., 2003) and it is thought to be as a result of the high level of muscle/ intramuscular stiffness. This is attributable to the profound indication of muscle mechanical response to high stimulation intensity/twitches (Orizio et al., 2003). Essentially, the results obtained in this study has established a relationship between MMG amplitude (RMS) and muscle mechanical responses (PT) during low and high stimulation intensity for NMES-elicited muscle force production in healthy volunteers.

### 3.2 Knee Angles and MMG Response

Although the knee angle did not have statistically significant effects on the pattern of response of MMG amplitude and stimulation intensity levels, from 50 mA to 80 mA the decline in MMG amplitude to high contraction intensity tended to be more pronounced and appeared earlier in 90-degree knee angle (Figure 2 (B)). This may suggest that the higher the torque production the higher the tendency for muscle stiffness which affects the muscle dimensional changes as reflected by a decline in MMG amplitude.



**Figure 2.** Plots of Normalized NMES-evoked peak torque versus stimulation intensity (A) and Normalized MMG-RMS versus stimulation intensity (B) at 90deg and 30deg knee angles. \* $p \leq 0.05$  for comparison between knee angles at each stimulation intensity level. Data are presented as mean $\pm$ SE; N=6.

Therefore, there seem to be knee angle effects on the contracting MU mechanical characteristics which could explain the MU activation pattern during NMES-evoked muscle contractions at different joint angles. Moreover, the extent of muscle stiffness observed from 50 mA may be due to the possibility of firing a greater quantity of fast twitch motor units in the rectus femoris. This explanation is in accordance with that of Barry (1991) during the stimulated contractions of the abductor pollicis brevis muscle.

#### **4. Conclusion**

MMG amplitude (RMS) relationship with the incremental NMES-evoked torque production of knee extensors has been investigated to study whether MMG-RMS might be used to track the muscle mechanical responses (PT). It has been identified that based on its sensitivity to the incremental stimulation amplitude and torque production, the MMG signals may be deployed to interpret physiological phenomena underlying NMES-evoked torque production. The use of MMG-RMS to track incremental knee torque production will be an interesting tool for clinicians and other allied health professionals who could not afford to use the expensive isokinetic dynamometer—conventionally required to measure muscle force via joint torque. Also, these results may have implications in using MMG-RMS as a force/torque sensor for application in advanced prostheses in individuals with prosthetic support but without neurological lesions. In the future, a larger number of test participants will be recruited for a better interpretation of the MMG response to NMES-evoked torque production in both the healthy and denervated muscles.

#### **Acknowledgment**

This research study was supported by “HIR Grant Number: UM.C/625/1/HIR/MOHE/ENG/39”. We are grateful to the experimental participants for their time and cooperation.

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