Torque and mechanomyogram relationships during electrically-evoked isometric quadriiceps contractions in persons with spinal cord injury

Morufu Olusola Ibitoye\textsuperscript{a,b}, Nur Azah Hamzaid\textsuperscript{a,e}, Nazirah Hasnan\textsuperscript{c}, Ahmad Khaired Abdul Wahab\textsuperscript{a}, Md. Anamul Islam\textsuperscript{a}, Victor S.P. Kean\textsuperscript{a}, Glen M. Davis\textsuperscript{a,d}

\textsuperscript{a} Department of Biomedical Engineering, Faculty of Engineering, University of Malaya, 50603 Kuala Lumpur, Malaysia
\textsuperscript{b} Department of Biomedical Engineering, Faculty of Engineering and Technology, University of Ilorin, P. M. B. 1515 Ilorin, Nigeria
\textsuperscript{c} Department of Rehabilitation Medicine, Faculty of Medicine, University of Malaya, Kuala Lumpur 50603, Malaysia
\textsuperscript{d} Clinical Exercise and Rehabilitation Unit, Discipline of Exercise and Sports Sciences, Faculty of Health Sciences, The University of Sydney, Sydney, 2006 NSW, Australia

ARTICLE INFO

Article history:
Received 13 November 2015
Revised 4 May 2016
Accepted 20 May 2016
Available online xxx

Keyword:
Mechanomyogram
Spinal cord injury
NMES-evoked muscle contractions
Muscle torque
Motor unit recruitment

ABSTRACT

The interaction between muscle contractions and joint loading produces torques necessary for movements during activities of daily living. However, during neuromuscular electrical stimulation (NMES)-evoked contractions in persons with spinal cord injury (SCI), a simple and reliable proxy of torque at the muscle level has been minimally investigated. Thus, the purpose of this study was to investigate the relationships between muscle mechanomyographic (MMG) characteristics and NMES-evoked isometric quadriiceps torques in persons with motor complete SCI. Six SCI participants with lesion levels below C4 [(mean (SD) age, 39.2 (7.9) year; stature, 1.71 (0.05) m; and body mass, 69.3 (12.9) kg)] performed randomly ordered NMES-evoked isometric leg muscle contractions at 30°, 60° and 90° knee flexion angles on an isokinetic dynamometer. MMG signals were detected by an accelerometer-based vibromyographic sensor placed over the belly of rectus femoris muscle. The relationship between MMG root mean square (MMG-RMS) and NMES-evoked torque revealed a very high association ($R^2 = 0.91$ at 30°; $R^2 = 0.98$ at 60°; and $R^2 = 0.97$ at 90° knee angles; $P < 0.001$). MMG peak-to-peak (MMG-PTP) and stimulation intensity were less well related ($R^2 = 0.63$ at 30°; $R^2 = 0.67$ at 60°; and $R^2 = 0.45$ at 90° knee angles), although still significantly associated ($P < 0.006$). Test-retest interclass correlation coefficients (ICC) for the dependent variables ranged from 0.82 to 0.97 for NMES-evoked torque, between 0.65 and 0.79 for MMG-RMS, and from 0.67 to 0.73 for MMG-PTP. Their standard error of measurements (SEM) ranged between 10.1% and 31.6% (of mean values) for torque, MMG-RMS and MMG-PTP. The MMG peak frequency (MMG-PF) of 30 Hz approximated the stimulation frequency, indicating NMES-evoked motor unit firing rate. The results demonstrated knee angle differences in the MMG-RMS versus NMES-isometric torque relationship, but a similar torque related pattern for MMG-PF. These findings suggested that MMG was well associated with torque production, reliably tracking the motor unit recruitment pattern during NMES-evoked muscle contractions. The strong positive relationship between MMG signal and NMES-evoked torque production suggested that the MMG might be deployed as a direct proxy for muscle torque or fatigue measurement during leg exercise and functional movements in the SCI population.

© 2016 IPEM. Published by Elsevier Ltd. All rights reserved.

1. Introduction

The study of motor unit (MU) recruitment to evoke force production is of clinical interest, particularly during neuromuscular electrical stimulation (NMES)-evoked contractions of paretic or paralyzed muscles in neurological populations [1]. Incremental MU recruitment during voluntary [2,3] and NMES-evoked contractions [4] has been used to describe muscle force modulation in healthy individuals. However, while NMES-evoked contractions have been utilized for muscle force production in individuals with spinal cord injury (SCI) [5], the mechanical and morphological changes associated with muscle contraction in this population have been poorly documented. To evaluate the effectiveness of NMES interventions, it is important to quantify stimulus-evoked muscle force. In particular, understanding motor recruitment and muscle force characteristics could provide key insights about the contractile properties of the muscle [6] and this has important implications for the use of NMES-evoked contractions in SCI rehabilitation.
NMES in rehabilitation. For example, measuring force or strength changes in persons with SCI can provide evidence of recovery or deterioration of motor output, as well as revealing the efficacy of rehabilitation interventions [7]. Beyond promoting the practical applications of NMES training in maintaining ‘muscle health’ [8], the ability to quantify an acute increase in muscle force production following NMES exercise [9] could widen the application of this assistive technology in the clinical environment.

Traditionally, isokinetic dynamometers have been used to assess muscle force (via joint torque) in a research setting, and they quantify torque throughout the limb range of motion with acceptable reliability [7]. However, these devices lack portability, and are relatively expensive and cumbersome to deploy for assessments in the clinical or home environment. The estimation of the muscle torque from other muscle characteristics, particularly bio-potentials, becomes an attractive option.

An indirect estimation of torque production has been assessed from electromyography (EMG) [10–12], but sensitivity of the signal to the external electromagnetic interference and skin impedance changes due to perspiration [13] presents significant limitations [14]. Additionally, the reliability of EMG estimation of muscle torque generation during NMES-evoked contraction remains debatable [15], largely due to the size of stimulation artifact current in relation to the EMG signal [10]. Thus, quantification of stimulus-evoked force production by EMG alone during neurostimulation is deficient [16].

A mechanical “counterpart” of the electrical activity of active motor units as measured by EMG (i.e., muscle mechanomyogram; MMG) has been proposed for muscle torque assessments [17,18]. During skeletal muscle contractions, the generated MMG signal is a function of the following mechanisms: (1) a slow bulk movement of the muscle at the initiation of the contraction, (2) smaller subsequent lateral oscillations occurring at the resonant frequencies of the muscle, and (3) dimensional changes of active muscle fiber” [17]. Therefore, MMG reflects the mechanical activity of physiological phenomena underlying muscle contractions. MMG quantifies neuromuscular performance, and has been used to gain insights into muscle capability during voluntary [17] and stimulated contractions [4]. In healthy humans, Petitjean et al. [4] reported a positive linear relationship between MMG amplitude and MU recruitment (i.e., muscle torque) during incremental NMES-evoked contractions of the first dorsal interosseous muscle (FDI). The authors suggested that the influence of the muscle fiber type may have been responsible for the pattern observed. Consequently, the MMG-torque relationship is both muscle fiber-type composition [19] and structure [20] dependent. In addition, MMG frequency content provides information regarding the firing rates/frequency of the active motor units during voluntary and NMES-evoked contractions [21]. Therefore, simultaneous investigation of the time and frequency contents of MMG signal has been used to interpret motor control strategy that is responsible for muscle force modulation during voluntary [17] and NMES-evoked [14] muscle contractions. Thus, the torque output during NMES-evoked muscle contractions depends on the degree of MU recruitment, their firing rates [4] and the contractile properties of the activated MUs [22]. Nevertheless, clear interpretation of the specific influence of MU recruitment and their firing rates on MMG characteristics during NMES-evoked contractions in individuals with SCI has been minimally investigated.

Thus, the aims of this study were: (1) to quantify the degree of association between MMG signal and isometric torque of the rectus femoris (RF) muscle during incremental NMES-evoked muscle contractions at 30°, 60°, and 90° knee angles; and, (2) to investigate the reliability of MMG signal recorded over RF muscles in persons with SCI. The quadriceps muscle was selected because of its well-established relevance for the study of knee joint torque dynamics [23], could be readily stimulated in paraplegia [24], and could be easily compared with existing data on voluntary contractions [19,25]. RF was selected to represent quadriceps group because it is the major contributor to the NMES-evoked quadriceps muscle torque during knee extension [25,26]. To our knowledge, no previous studies have reported the relationship between MMG parameters and the quadriceps torque production during incremental NMES-evoked isometric contractions in persons with SCI.

We hypothesized that MMG would be a reliable proxy of incremental torque production, since a positive linear relationship has been previously demonstrated in FDI, which is of comparable muscle fiber morphology to the quadriceps [4] with a predominance of type II fibers in this muscle group after SCI [24]. Furthermore, a significant correlation between the MMG signal and muscle torque production would a priori support the validity of the signal as a proxy of muscle performance, particularly when a direct measurement of torque might be impractical [15], such as in activities of daily living.

2. Method

2.1. Participants

Nine chronic motor complete (American Spinal Injury Association Impairment Scale A and B) SCI [27] participants with neurological lesions below C4 were recruited at the Department of Rehabilitation Medicine, University of Malaya Medical Centre, Kuala Lumpur, Malaysia. Their written informed consent was obtained after a full disclosure of the rational and procedures of the experiment in compliance with the declaration of Helsinki. They were duly informed about the possible sources and discomforts of the dynamometer assessment and electrical stimulation, and were advised of their rights of withdrawal from the study at any time. Individuals with severe spasticity, joint contracture or lower motor neuron lesion that might adversely affect the production of modest quadriceps torque were excluded from participation. Also excluded were any participants who, as a result of incremental NMES current amplitude, produced no relative increase in their stimulus-evoked torque values. Of the nine participants recruited at the outset, only seven successfully completed the full test battery. However, a further participant was excluded due to lack of increase in relative torque in response to increasing NMES current intensity. Therefore, the data of the remaining six participants (Table 1) has been included for analysis. All participants retained quadriceps spinal reflexes, and they could sit up on a dynamometer’s chair with backrest. As part of clinical conditioning exercises [28], at the time of the investigation, participants were already involved in NMES cycle training (2 to 3 times per week for at least 7 weeks), but were asked to refrain from the training for at least 48 hours before testing.

2.2. Experimental protocol

Participants were secured to a calibrated isokinetic dynamometer (System 4; Biodex Medical Systems, Shirley, NY, USA) by an inextensible restraining straps over the thigh, pelvis and the trunk to minimize extraneous movements [29] and to ensure only isometric contractions of the quadriceps could be performed (Fig. 1) [28]. Based on safety considerations of not putting bone health at risk [30], and to analyze the muscle torque in a range that will mimic functionally relevant mode such as in standing up, we utilized maximal torque calculations based on the empirical data of Kagaya and colleagues [31]. Those investigators suggested that the knee extensor’s moment should not exceed that required for NMES standing in persons with SCI. Thus, careful attempts were made to keep the NMES-evoked muscle torque production within...
Table 1
Participants’ physical characteristics.

<table>
<thead>
<tr>
<th>Participants</th>
<th>Gender</th>
<th>Age (y)</th>
<th>Body mass (kg)</th>
<th>Height (m)</th>
<th>NLL</th>
<th>AIS</th>
<th>TSI (Yrs.)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>M</td>
<td>49</td>
<td>79.6</td>
<td>1.74</td>
<td>T1</td>
<td>A</td>
<td>11</td>
</tr>
<tr>
<td>2</td>
<td>F</td>
<td>47</td>
<td>82.0</td>
<td>1.62</td>
<td>T4</td>
<td>B</td>
<td>24</td>
</tr>
<tr>
<td>3</td>
<td>M</td>
<td>28</td>
<td>62.4</td>
<td>1.71</td>
<td>C7</td>
<td>B</td>
<td>14</td>
</tr>
<tr>
<td>4</td>
<td>M</td>
<td>44</td>
<td>71.6</td>
<td>1.79</td>
<td>C6/C7</td>
<td>B</td>
<td>2.5</td>
</tr>
<tr>
<td>5</td>
<td>M</td>
<td>34</td>
<td>75.9</td>
<td>1.70</td>
<td>C6</td>
<td>A</td>
<td>17</td>
</tr>
<tr>
<td>6</td>
<td>M</td>
<td>33</td>
<td>44.0</td>
<td>1.71</td>
<td>C5/C6</td>
<td>A</td>
<td>13</td>
</tr>
</tbody>
</table>

Mean ± SD: 39.2 ± 7.9, 69.3 ± 12.9, 1.71 ± 0.05, 13.6 ± 6.5


Fig. 1. Experimental set-up showing the MMG and NMES electrode placement over the quadriceps muscle in a representative participant. ‘A’, and ‘B’ are the cathode and anode electrodes, respectively, of the neuromuscular electrical stimulator while ‘C’ represents the accelerometer sensor that was used to collect the MMG signal.

A range that would not risk bone integrity. Although not all participants could be NMES-provoked to produce the maximum torque, the maximum torque production in each participant was limited to 75 Nm as suggested by Gerrits et al. [24]. This value approximated an average value of knee extensor moment (torque) sufficient for NMES-assisted standing [31], and was attained around 100–120 mA current amplitude for most participants.

2.3. Familiarization

A familiarization session was conducted (at least a day prior to testing) to acquaint participants with the NMES-evoked isometric assessment procedures on the dynamometer. Thereafter, participants attended the laboratory on two different test days, separated by 48 h, for each of the knee angles assessed, to quantify test-retest reliability.

2.4. Stimulation protocol

Through palpation and visual inspection, the isolated activation of knee extensors was ensured to establish that muscle activation was primarily from quadriceps, using 9 cm × 15 cm self-adhesive stimulating electrodes (Hasomed GmbH, D 39114 Magdeburg, Germany) [32]. The cathode NMES electrode was placed 8 cm distal to the inguinal area, over the RF belly near the expected location of the motor points [33], and the position of electrode was then slightly adjusted to a location whereby the maximal response, based on a palpable muscle response and force production, to stimulation could be identified. The anode electrode was affixed 5 cm proximal to the patella as recommended by Levin and co-workers [16]. For both the distal and proximal anatomical landmarks, motor points represent the location where the motor branch of a nerve accesses the muscle belly from where the maximal muscle force could be obtained for a given electrical stimulus [33,34]. Indelible marker was used to identify the electrode position for accurate placement between trials and testing days.

NMES was used to evoke isometric contractions of the knee extensors with the hip flexed at about 90°, and the knee flexed to 90°, 60° or 30° according to our protocol. NMES electrodes were connected to a neuromuscular stimulator delivering square-wave pulses of current amplitude between 50 and 120 mA. During each stimulus-evoked contraction, a train of electrical stimulation (i.e., repeated bursts of pulses at 30 Hz, and 400 μs pulse duration, with increasing stimulation amplitude (mA); RehaStim™, Hasomed GmbH, Magdeburg, Germany) was imposed for 4 s to potentiate the quadriceps activation.

2.5. Measurements

2.5.1. NMES-evoked isometric torque measurement

Following a submaximal warmup trials wherein the muscle belly was palpated to ensure accurate MMG sensor fixation, about 4 s of randomly ordered NMES-evoked submaximal-to-maximal torque levels were imposed on the participants’ quadriceps muscle at 30°, 60°, or 90° knee angles (0° = full knee extension). The incremental torque was evoked by stimulation intensity from 50 mA to 120 mA in 10 mA increments for each participant. Previous studies have shown that this protocol elicits optimized outcomes as it does not evoke premature muscle fatigue [28]. Torque production (Nm) was quantified in real-time by the dynamometer and data were ‘gravity-corrected’ by subtracting passive torque produced by the leg mass affixed to the lever arm. This was effected automatically following the positioning of the knee angle at 45° from the full knee extension angle (0°) while each participant was instructed to remain relaxed. The recorded limb weight was automatically used by the dynamometer to ‘negate the gravity effect’ on the collected torque data (Biodex (V.4X) operation manual). To eliminate any order effect, the administration of contraction intensities (mA) and knee angles were randomized. A 10-min recovery was allowed between trials to reduce the risk of cumulative muscle fatigue [35].

2.5.2. MMG measurements

Simultaneously with the torque measurement, MMG signals from the RF were obtained using an accelerometer-based vibromyographic sensor (Sonometrics BPS-II VMG transducer, compatible with Biopac MP150 platform, sensitivity 30 V/g) attached by
means of double-sided adhesive tapes (3M Center St. Paul, MN, USA) [36] directly on the muscle belly (i.e., at the midpoint between the inguinal crease and the superior border of the patella [37]), in order to obtain the maximum muscle surface oscillation (Fig. 1). Before attaching the MMG sensor, the skin was shaved (as needed) and cleaned with alcohol swabs. As it was sometimes difficult to identify the precise location of the quadriceps’ muscle belly (due to muscle atrophy and adipose tissue thickening), the determination of the MMG location was assisted by electrical stimulation of the muscle. During this procedure, a stimulation current amplitude of 50 mA (pulse width = 400μs, frequency = 30 Hz) was administered to identify the probable muscle belly by visual inspection and palpation. This location was standardized for subsequent trials by indelible marker. The pattern of MMG signals and the torque production during NMES-evoked muscle contractions is as shown in Fig. 2.

2.5.3. Signal processing

Signals were acquired and analyzed using AcqKnowledge data acquisition and analysis software (MP150, BIOPAC Systems, Santa Barbara, CA, Inc, USA) and a customized programme in LabVIEW (Version 12.0, National Instruments, Austin, TX, USA). The raw MMG signals were acquired at a sampling rate of 2 kHz and digitally band-pass filtered (20–200 Hz), to supress the influence of artifacts associated with tremor and body movement [38,39], for offline analysis. The peak torque (PT) obtained from a dynamometer was calculated for each NMES-evoked contraction level/stimulation intensity. The PT of the participants, MMG root mean square (MMG-RMS), peak-to-peak (MMG-PTP) amplitude and MMG frequency characteristic – peak frequency (MMG-PF) were extracted from the NMES-evoked isometric contraction measurement from 1-s epochs of MMG signal around the peak torque (location of probable maximum muscle recruitment) at each contraction intensity. The middle 1-s epoch was selected to avoid the on-transient of force rise at the beginning and off-transient during the end of muscle contractions [40]. The PT value and the MMG parameters at maximum stimulation intensity (120 mA) were used to normalize their relative submaximal values, at each knee flexion angle, to allow comparison between participants.

2.5.4. Data analysis

The test-retest reliability of measurements between days were quantified by intraclass correlation coefficient (ICC), using a two-way mixed effects model, and standard error of measurements (SEM) [41] calculated as a percentage of relative mean values (i.e., to examine the relative and absolute consistency of the parameters). Thereafter, paired samples t-tests were performed on the dependent variables to determine whether there was a significant mean difference between the test and retest scores. A data normality test was conducted using Shapiro–Wilk statistic, and the data were normally distributed except for a few (i.e., peak torque at 60° knee angle) that were skewed due to our small sample size (Table 2). Based on the recommendation of Munro [42], the interpretation of the ICCs were as follows: >0.90, very high reliability; 0.70–0.89, high reliability; 0.50–0.69, moderate reliability. Polynomial regression analyses were conducted to investigate whether NMES-evoked torque versus stimulation intensity, MMG-RMS versus NMES-evoked torque, and MMG-PTP versus stimulation intensity (at 8 levels of torque production) were significantly correlated. Using the procedures described by Beck and colleagues [43], polynomial regression analyses were also used to examine the model of best fit for these relationships. The highest coefficient of determination ($R^2$) was used to determine the goodness of fit of a particular regression model, within the polynomial class, to the data. Prior to regression analyses, MMG data were expressed as a percentage of their values at maximum stimulation intensity level. A statistical software package (IBM SPSS for Windows Version 20, NY, USA) and Microsoft Office Excel 2013 (Microsoft, Redmond, WA, USA)
were used for data analyses. Exact statistically significant values \((P)\) were reported in the manuscript text and the data were shown as mean± standard deviation.

3. Results

After completing the full test battery and with one participant, out of seven, meeting exclusion criteria, six participants, whose physical characteristics appeared in Table 1, were included in the analyses presented herein.

3.1. Reliability

In Table 2, the ICC, SEM% and their respective probabilities for all the investigated parameters were presented. Based on the normative categories previously described, ICC ranged from “moderate to very-high reliability” (i.e., 0.65 to 0.97). For SEM%, the values ranged from 10.1 to 31.6% of their relative mean values. Paired sample t-tests indicated that there were no significant differences between the mean values of any parameters \((P > 0.05)\).

3.2. Torque production

Fig. 2 is an example of the graph of NMES-evoked torque and raw MMG signal from RF. Fig. 3 depicts MMG recordings at 50 mA (low torque production) and 100 mA (high torque production) and the corresponding spectra responses at 60° knee angle. Fig. 4 shows the relationship between NMES-evoked torques as a function of stimulation intensity (mA) for the three knee angles.

3.3. MMG and contraction intensity

Fig. 5 depicts the relationships between the normalized MMG-RMS and NMES-evoked torque expressed in %PT (20, 40, 60, 80 and 100 %PT) for 30°, 60° and 90° knee flexion angles at eight levels of contraction intensities (50, 60, 70, 80, 90, 100, 110, 120 mA).

In Fig. 6, the relationship between MMG-PTP and stimulation intensity at the three knee angles were shown. There were apparent positive linear relationships between the two parameters at 30° (\(R^2 = 0.63, P < 0.001\)), 60° (\(R^2 = 0.67, P = 0.006\)); and 90° (\(R^2 = 0.45, P = 0.001\)) knee angles.

![Fig. 3. MMG recordings of RF at 50 mA (A) and 100 mA (B) neurostimulation current amplitude and the corresponding spectra at 60° knee flexion angle. The MMG-PF approximated the stimulation frequency of 30 Hz at both 50 mA and 100 mA contraction intensity levels, however, harmonics of the peak frequency characterizes the stimulation intensity of 100 mA.](image-url)
4. Discussion

To our knowledge, this is the first study to investigate the degree of association between mechanomyographic characteristics and isometric NMES-evoked muscle torque in persons with motor 'complete' SCI. A moderate to very high test-retest reliability, together with strong, positive correlations between the contraction intensity and MMG output, indicated that underlying muscle mechanical changes could be reliably tracked by the MMG signal. This finding was in agreement with a previous investigation [22] in healthy volunteers, and supported the validity of the MMG signal to quantify muscle contractile properties and performance of denervated quadriceps muscle re-activated by NMES-evoked contractions.

4.1. MMG sensor reliability

In our study, two measures of MMG amplitude (MMG-RMS and MMG-PTP) demonstrated a comparable level of relative (ICC) and absolute (SEM) reliability (Table 2). Although MMG-RMS has been more widely used to 'track' muscle effort, MMG-PTP could similarly follow underlying mechanical changes reliably during NMES-evoked contractions [44]. This explains why recent investigations [45,46] have adopted the MMG-PTP parameter to track NMES-evoked fatiguing contractions in healthy individuals. Additionally, based on the assumption [47,48] that ICC greater than 0.8 could be considered good enough for clinical application, our findings provide an insight into the potential clinical efficacy of the MMG signal for real-time muscle performance grading during NMES-evoked exercises.

4.2. NMES-evoked torque production

The NMES-evoked torque production was in the order of 60° > 30° > 90° knee flexion angles and in conformity with the previous investigation [49] and was shown to be reliable between
quadriceps muscles. Previous studies [4,20,36,42,45] have demonstrated strong positive linear or non-linear correlations between MMG signal parameters and stimulus-evoked isometric torque in healthy persons. To our knowledge, only a single investigation [51] has utilized MMG amplitude to quantify NMES-evoked contraction levels (as related to cycling ride time) in quadriceps muscle in persons with SCI. However, the present study has uniquely investigated the relationship between MMG amplitude and NMES-evoked isometric torque at submaximal-to-maximal contraction intensities in order to infer characteristics about torque production in a SCI population. Such information may be of practical application during NMES-evoked standing up, prolonged stance or stepping where direct measurement of torque is necessary for fatigue estimation, but is impractical to measure without a dynamometer.

Although the torque values obtained in the present investigation were much lower when compared with the previously reported data on healthy volunteers during voluntary contractions [52], a parallel increase in MMG amplitude with incremental NMES-evoked torque production was equally evident. This suggests that the sensitivity of MMG amplitude to stimulus-evoked muscle contractions is independent of the level of torque production. Furthermore, the pattern of MMG amplitude response was in good agreement with the previously reported patterns in healthy volunteers [4,20,22]. Specifically, MMG-RMS increased linearly up to 80 %PT (for the investigated knee angles) before the appearance of plateau due to the muscle stiffness and the associated force fusion—an manifestation of muscle mechanical changes during contraction [14,22]. This result suggests a possibility that MMG amplitude follows the contraction intensity independently of knee angle up to ~80 %PT. This might be sufficient for the implementation of a muscle performance feedback in situations where torque information is required to modulate NMES contractions to optimize functional outcomes in persons with SCI as previously suggested by Gobbo et al. [45].

Furthermore, MMG-PTP has equally been used to monitor the muscle mechanical changes (i.e., “the viscoelastic characteristics of the series elastic component”) [53] during NMES-evoked contractions [4,44]. Petitjean et al. [4], has previously established a positive linear relationship between MMG-PTP and stimulation intensity in FDI and demonstrated that MMG-PTP could reflect the summation of the contracting MU. The authors suggested an orderly recruitment of the MU as the reason behind their observation. Our results confirm this finding and showed a comparable positive linear correlation between the MMG-PTP and stimulation intensity at the three knee angles. In all, MMG-RMS and MMG-PTP can be used as conjugate proxies of muscle force.

4.3.2. MMG frequency

Unlike in voluntary contractions [54], the increase in contraction intensity did not appear to influence the MMG peak frequency during NMES-evoked muscle contractions. At higher stimulation intensity/torque level, multimodal frequency characterized the MMG spectrum (Fig. 3) and their values appeared to be the harmonics of the fundamental/peak frequency. The apparent correspondence of the MMG peak frequency with the stimulation frequency (Fig. 3) suggested that the MMG frequency may represent the MU firing frequency. This observation arose from the knowledge that the NMES-evoked muscle contraction is synchronous and because the participants of the present study had motor complete SCI, their muscle contraction was entirely involuntarily modulated. This implied that the recorded MMG signal was mainly generated by the synchronously recruited muscle fibers [14]. This explanation is in good agreement with previous studies [20,55] which also demonstrated that the MMG peak frequency matched the stimulation frequency of contracting muscles. While a clearer interpretation of this pattern is beyond the scope of our study, Stokes and
Cooper [55] and Yostakte et al. [20] have previously suggested that this phenomenon might be a function of the type of muscle, properties of the MMG transducer, and valid mainly during unfused contractions [20]. In all, the MMG-PF may only approximate the NMES-evoked firing frequency of muscles (at submaximal torque production levels) of predominant fast twitch muscle fiber type, such as in denervated RF.

4.3.3. Influence of knee flexion angles on mechanomyographic response

Although all the investigated knee angles revealed strong relationships between the MMG amplitude and NMES-evoked isometric torque, the pattern of relationship was knee angle specific (Fig. 5). This is in agreement with a previous investigation by Ebersole and co-workers [52], whereby the relationship between the production of the quadriceps voluntary isometric torque and its MMG-RMS was knee angle dependent. They suggested that such differences might be due to the variations in muscle stiffness or MU activation strategies as a reflection of length-tension relationship.

4.3.4. Correlations among MMG and NMES-evoked torque

Strong, positive correlations were observed between: MMG-RMS and NMES-evoked torque; MMG-PTP and stimulation intensity, at all the three knee angles. The correlations might be attributed to the sensitivity of MMG amplitude to the incremental force modulation. This has a direct implication on the possibility of MMG to track the force modulation in denervated muscles with predominant fast twitch fiber type [24,56] and supports the earlier suggestion [20] of examining muscle’s cellular composition with MMG signal. This explanation is consistent with early voluntary isometric contraction studies of the quadriceps [19,25] which identified that the MMG-RMS of muscles with predominant type II fiber could increase up to 100% of irrespective of the type of MMG sensor used [25]. Additionally, during NMES-evoked contractions of the quadriceps a positive MMG-RMS linear relationship of up to 80% has also been reported in healthy volunteers [20]. Our results with MMG-RMS attenuation at around 80% reaffirmed those earlier findings and showed that comparable correlations could be obtained in denervated quadriceps muscle during NMES-evoked contractions.

4.4. Potential clinical applications

MMG might be used as a non-invasive measure of muscle effort to quantify the effectiveness of NMES training for neurological populations. Additionally, the signal might be used as a real-time proto-dynamometer to quantify muscle performance during activity of daily living – especially as a feedback signal to up-titrating NMES current amplitude or pulse duration to optimize NMES activities. Our findings support the suitability of MMG as a practical NMES control signal, as it only requires the calculation of MMG amplitude parameters for practical implementation [45]. However, MMG responses to NMES-evoked torque production in other functionally relevant muscles, specifically of different fiber distribution, need to be investigated. Such information will give further insight into the physiological relevance of the signal.

The following limitations were acknowledged in our study design: (1) our findings were dependent to the protocol that we used, (i.e. modulating the current amplitude while keeping the pulse width and frequency constant), (2) the MMG cycle training baseline of at least seven weeks was adopted based on the previous recommendation [28], but a longer duration of training may have yielded different results, (3) our investigation was based on a sample size (n) of six and some data distributions were non-normally skewed. Although the sample size was modest in comparison with other studies that have employed individuals with motor complete spinal cord lesions, the size might have affected our findings. A larger test population is warranted in the future study to identify how broadly the present findings could be generalized.

5. Conclusion

We have evaluated the pattern of relationships between the MMG signals and NMES-evoked isometric contractions to study the motor unit recruitment strategy in motor complete denervated quadriceps, and have demonstrated acceptable reliability of MMG measurements. Useful insights inferred from these findings are: (1) MMG signals were sensitive to the incremental NMES-evoked muscle torque measured by a commercial dynamometer (i.e. a “gold standard”), and as a physically small sensor, the MMG could be a reliable proxy for these dynamometer measurements, (2) MMG signals correlations with NMES-evoked muscle torque could be used to assess the denervated quadriceps mechanical changes during submaximal-to-maximal NMES-evoked muscle contractions. The application of MMG amplitude as a proxy of stimulus-evoked isometric muscle force and relevance as a biofeedback source in NMES-evoked activities are evident. Whether these results could be generalized to other muscles and mode of contractions, specifically, during critical activities – such as NMES-supported standing, is a topic of further research. Testing of such hypothesis remains a promising perspective, particularly since automated MMG is clinically more relevant, effective and safe when compared with traditional “user-controlled” approach [57].

Funding

The study was financially supported by the Ministry of Higher Education, Malaysia and the University of Malaya through HIR Grant No. UM.C/625/1/HIR/MOHE/ENG/39.

Ethical approval

The study was approved by the University of Malaya Medical Ethics Committee (Approval NO:1003.14(1)).

Conflict of interest

None

Acknowledgment

We are indebted to Dr. Khameel. B. Mustapha of The University of Nottingham, Malaysia Campus for his technical guidance, Asyaa, S.R; Rozman, N.N; Othman, N.S; Nadzri, M.H and Ibrahim, I. are gratefully acknowledged for facilitating the successful execution of muscle conditioning exercise. The authors especially thank the participants who gave their time to this investigation.

References

During ischemic events, muscle tone is maintained through passive mechanisms, including stretch reflexes, myotatic reflexes, and tonic muscle activity. The role of these mechanisms in maintaining muscle tone during ischemic events is not well understood. This study aimed to investigate the mechanisms that maintain muscle tone during ischemic events in human muscle.

Materials and Methods

A total of 10 healthy individuals were recruited for this study. Muscle tone was assessed using electromyography (EMG) and pressure algometry. EMG was recorded from the anterior tibial muscle using surface electrodes placed over the muscle belly. Pressure algometry was performed using a mercury-filled pressure algometer.

Results

The results showed that muscle tone was maintained during ischemic events through the activation of the stretch reflex. The stretch reflex was triggered by a decrease in muscle length, which activated the muscle spindles and produced an increase in muscle tone. The results also showed that muscle tone was maintained through the activation of the myotatic reflex, which is a stretch reflex that occurs when the muscle is stretched to its passive length.

Conclusion

The results of this study suggest that muscle tone is maintained during ischemic events through the activation of stretch and myotatic reflexes. Further studies are needed to understand the role of these mechanisms in maintaining muscle tone during ischemic events in human muscle.