Mechanomyography and muscle function assessment: A review of current state and prospects

Morufu Olusola Ibitoye a,b, Nur Azah Hamzaid a, Jorge M. Zuniga c, Ahmad Khairi Abdul Wahab a

a Department of Biomedical Engineering, Faculty of Engineering, University of Malaya, 50603 Kuala Lumpur, Malaysia
b Department of Biomedical Engineering, Faculty of Engineering and Technology, University of Ilorin, P. M. B. 1515 Ilorin, Nigeria
c Department of Exercise Science, Creighton University, 2500 California Plaza, Kiewit Fitness center 228, Omaha, NE 68178, United States

A R T I C L E   I N F O
Article history:
Received 31 October 2013
Accepted 8 April 2014

Keywords:
Muscle function assessment
Mechanomyogram
Electromyogram
Voluntary and evoked muscle contraction

A B S T R A C T
Previous studies have explored to saturation the efficacy of the conventional signal (such as electromyogram) for muscle function assessment and found its clinical impact limited. Increasing demand for reliable muscle function assessment modalities continues to prompt further investigation into other complementary alternatives. Application of mechanomyographic signal to quantify muscle performance has been proposed due to its inherent mechanical nature and ability to assess muscle function non-invasively while preserving muscular neurophysiologic information. Mechanomyogram is gaining accelerated applications in evaluating the properties of muscle under voluntary and evoked muscle contraction with prospects in clinical practices. As a complementary modality and the mechanical counterpart to electromyogram; mechanomyogram has gained significant acceptance in analysis of isometric and dynamic muscle actions. Substantial studies have also documented the effectiveness of mechanomyographic signal to assess muscle performance but none involved comprehensive appraisal of the state of the art applications with highlights on the future prospect and potential integration into the clinical practices. Motivated by the dearth of such critical review, we assessed the literature to investigate its principle applications, current applications, challenges and future directions. Based on our findings, the importance of rigorous scientific and clinical validation of the signal is highlighted. It is also evident that as a robust complement to electromyogram, mechanomyographic signal may possess unprecedented potentials and further investigation will be enlightening.

1. Introduction

1.1. Historical background

The origin of muscle sound in relation to the muscle physiologic condition could be traced to 1665 by Grimaldi (1665), who referred to the sound as “motion of animal spirits”, a formulation similar to other esoteric explanations of natural phenomena from that time (Rhatigan et al., 1986). In 1810, Wollaston was able to describe quantitatively with analogy the frequency of muscle sound within the range of 14–35 Hz (Wollaston, 1810). The characteristics of time and frequency domain analysis of the muscle sound during evoked and voluntary contraction was initiated by Oster and Jaffe (1980) and Oster (1984). They described the dominant frequency of muscle sound within 25 plus or minus 2.5 Hz. With the aid of stethoscope and microphone, the investigators verified that sound (a form of mechanomyogram (MMG)) is an intrinsic property of muscle contraction.

Rhatigan et al. (1986) and Orizio et al. (1990) in separate attempts to justify the mechanomyogram (MMG) relevance in muscle research, showed that the predominant power in MMG frequency spectrum of humans’ biceps brachii under isometric contraction was within 10–22 Hz. Other investigators subsequently demonstrated that the lowest limit of frequency reported in human is approximately 2 Hz while the highest is 120 Hz (Beck et al., 2008). Consequently, MMG was validated as the complementary mechanical signal to the established electromyogram (EMG) to investigate neuromuscular activities (Stokes and Blythe, 2001) especially when noninvasive estimation of the muscle physiology underlying contraction and the fatigue phenomenon is sought (Esposito et al., 1998).

However, due to the emergence of sensitive, lightweight, inexpensive sensors, and advanced signal analysis techniques, acquisition of low frequency vibration of muscular activities in the form of MMG has been made feasible. The signal has been verified to be detectable at the skin surface during dimensional changes of active muscle fiber that generate pressure waves due to voluntary or evoked contraction and demonstrated to possess rich information of the underlying neuromuscular parameters leading to contraction and thus reliable in muscle function assessments (Orizio et al., 2003a). It has been suggested that
the signal may reflect the three main physiological phenomena including the gross lateral movement of the contracting muscle at the initiation of contraction, smaller, subsequent vibrations at the resonance frequency of the muscle, and the dimensional changes in the active muscle fiber (Barry and Cole, 1990; Orizio et al., 2003a). In general, impetus from the classical work on muscle sound revealed that its properties i.e. vibration, acceleration and dimensional changes can be used to assess muscle contraction and performances.

Currently, electromyogram as the conventional signal modality to monitor skeletal muscle activities is unable to generate sufficient information regarding mechanical index of muscle contraction (Sasidhar et al., 2013) thus possessing limited information about the neural control of muscle function (Farina et al., 2004). The signal is also not sufficiently suitable to quantify muscle function during electrically evoked dynamic muscle action (Braz et al., 2009). The exploration of complementary paradigm that is sensitive to the muscle mechanical activities and devoid of inherent electrical noise, such as MMG, is therefore warranted. Inference from various scientific research verified that MMG signal can be used for the following: (i) muscle fiber typing (Herda et al., 2010); (ii) assessment of muscle force (Sarlabous et al., 2013); (iii) muscle fatigue (Hendrix et al., 2010); (iv) indication of resonance frequency of muscle (Barry and Cole, 1990); and (v) assessment of contractile properties (Gorelick and Brown, 2007). It has been observed that the mechanical nature of the muscle fiber activities that leads to contraction (Gerdlle et al., 1999) could be better discerned and represented with signal response that is inherently mechanical.

With cognisance attention specifically to emerging area of applications of MMG due to its resistance to electrical interferences, and flexibility in its sensing technology, literature survey highlighted the robustness of MMG signal that is often underestimated. Aside from few isolated studies with variations, such as that of Herda and Cooper who demonstrated the inability of mechanomyographic amplitude-force relationship to distinguish differences in voluntary activation capacity between individuals (Herda and Cooper, 2013), it is apparent that the analysis of variables of MMG signal in time and frequency domains has been demonstrated to assess various aspects of neuromuscular functions (Malek and Coburn, 2012) and could be used to investigate motor unit activities (recruitments and firing rates) i.e. the force generation mechanism (Cè et al., 2013; Yoshitake et al., 2001) thus can be used to assess different conditions of skeletal muscle activities (Islam et al., 2013).

1.2. Basic principle of detection

The great surge of research activity noticed lately on the MMG application was due to the advancement in sensor technology and signal analysis techniques. Typically, MMG signal, characterized by frequency distribution below 100 Hz (Orizio, 1992), originates from the skeletal muscle contraction that leads to shortening of the muscle fiber length and increase in the fiber diameter (Farina et al., 2008). The vibration of muscle fibers and their dimensional changes during activation create pressure waves that could be detected on the skin surface; as an acceleration obtained by sensors such as accelerometer or as a skin displacement acquired by piezoelectric contact sensor (Watakabe et al., 1998), laser distance sensor (Orizio et al., 1999a), or condenser microphone (Orizio, 2004). Generally, the typical motor response of the muscle during contraction stage that is often accompanied by the transverse diameter changes to the muscle belly (Orizio et al., 2003b) could be readily obtained as MMG signals on the skin surface.

A recent interest in understanding the mechanism behind the MMG signal generation has led to an investigation of the relationship between the acceleration of motor unit and the surface spatial distribution. It has been inferred from the muscle anatomy and the theory of wave propagation that the muscle fiber contraction (fiber shortening and increase in the fiber diameter) (Farina et al., 2008) generates a time dependent spatial distribution of acceleration on the surface of the skin (Cescon et al., 2007). The spatial distribution of acceleration due to the activity of a single motor unit is termed motor unit acceleration map and it is dependent on the muscle morphology and architecture (Bichler, 2000). The limitation of using single channel MMG features for the assessment of the motor unit control strategies was highlighted by Madeleine and co-workers because of the substantial effect of the sensor’s position on the relationship between force and MMG signal and they suggested that a two-dimensional array of accelerometers may contribute to a better understanding of the origin of MMG signal (Madeleine et al., 2006). However, the alteration of the acceleration of the skin by the weight of multiple sensors may have limited the observation to large muscles where such weight may only have an insignificant effect (Farina et al., 2008).

By implication, the weight of the sensor and the signal detection site are of vital importance for the MMG signal integrity. The sensor placement has been generally validated to have the strongest response near the muscle belly and increasingly weaker toward the tendon (Frangioni et al., 1987). Equally skin fold thickness may affect the MMG signal response because of the low pass filtering effect of the tissue between the target muscle and the detection sensor (Jaskólska et al., 2004). Therefore the response of the signal depends significantly on the consistency of sensor location between trials.

The technical characteristics of the signal have also been shown to be dependent on the types and nature of the transducer adopted for acquisition (Beck et al., 2006). Investigators have continued to examine the signal characteristics of the differences in the variables indicated by MMG from different sensors. Fig. 1 shows the two major sinusoidal mechanical variables of the signal with different transducers.

![Graphical representation of the key mechanical variables of MMG signals during voluntary contractions of the biceps brachii muscles where panel a represents the amplitude response of the condensed microphone transducer and panel b is the amplitude response of accelerometer transducer.](image)

Reprinted with permission from Watakabe et al. (2001).
The MMG spectra derived from condenser microphone are generally favored over accelerometer alternative because of the less motion artifact in the former during dynamic contractions (Posatskiy and Chau, 2012; Watakabe et al., 2001). The major drawback of a condenser microphone is the influence of the air chamber on its temporal and spectral responses (Watakabe et al., 2001). The flat frequency range of accelerometer at the MMG signal range (Watakabe et al., 1998), ease of fixation to the skin surface, lightweight (less than 5 g) (Watakabe et al., 2003), and measurement in physical unit (meters per second square) are construed to favor its current wide patronage in most MMG studies (Orizio et al., 2003a). A piezoelectric contact sensor has been used especially for dynamic contraction but its relatively large mass (above 40 g) and inconsistent mechanical coupling to the skin surface continued to preclude its wide application (Orizio, 2000). Investigators have suggested that the amplitude response of the piezoelectric contact sensor is sensitive to contact pressure which implied different amplitude responses over trials (Smith and Stokes, 1993). Silva and Chau (2003) suggested the application of a coupled transducer pair for MMG acquisition to facilitate motion artifact reduction and enhance the distinction between the noise and MMG signal parameters. Muscle actions may therefore be interpreted differently depending on the nature of the transducer adopted.

Once an appropriate transducer is adopted for the acquisition of the signal, Cescon et al. (2004a) demonstrated that the spectral features of MMG signal reflect recruited muscle fibers’ mechanical properties and depend on motor unit activation during voluntary contraction. Equally, Yoshitake and colleagues verified the significance of MMG signal features to information about contractile properties of active motor unit in humans (Yoshitake et al., 2002). Therefore, MMG signal represents the general mechanical activities of summated active muscle fiber (Orizio et al., 1996).

The update on the MMG sensing technology includes recent application of vibromyographic (VMG) transducer for a quantitative measure of muscle performance during muscle contraction (Biopac systems Inc. Goleta, CA, USA) (Stone, 2013). Recently, there were renewed interests in the possibility of simultaneous measurement of electrical behavior (EMG) and mechanical behavior (MMG) of the active skeletal muscle with a common sensor for the purpose of compactness (Chang et al., 2012; Kimoto and Yamada, 2012). In general, standardized method and availability of reliable sensors will play a constructive role in the reliability of MMG signal parameters. Therefore, attention to the basic principles of signal acquisition will continue to be necessary in order to enhance the validity and reliability of the signal in generating neuromuscular information. Fig. 2 depicts the sensor arrangement for acquisition of MMG by a lightweight sensor with relevance in isometric and dynamic muscle activities (Zagar and Križaj, 2005).

Based on the MMG acquisition reliability, the predominant frequency (below 20 Hz) of its power spectrum is known to correlate positively with the sensor firmness to the skin surface (Bolton et al., 1989). Ensuring constant contact pressure between and within trials is vital for reliability assessment (Posatskiy and Chau, 2012). Equally, the positioning of the sensor on the belly of the muscle of interest (Cescon et al., 2004a) with good contact pressure influences the force versus MMG amplitude relationship positively (Smith and Stokes, 1993). According to the empirical result of Kitazaki and Griffin (1995), the weight of the sensor should not be felt on the muscle to obtain the actual muscle vibration due to the contraction. The possible effect of skinfold thickness and mass of adipose tissue between muscle of interest and sensor location on the skin surface during cycle ergometer was investigated by Zuniga and colleagues and the effect was shown to be insignificant (Zuniga et al., 2011). Given the insensitivity of MMG signal to electrical noise, it does not critically require signal conditioning circuit at the detection stage (Perry et al., 2001). Thus, subject’s comfort is enhanced and experimental set-up time is reduced. Fig. 3 depicts the typical arrangement of MMG sensors.

1.3. Basic MMG signal processing

In order to obtain the signal with an acceptable integrity, investigators have recommended the analysis of mechanomyographic signals in the light of the established electromyographic signal processing techniques. Thus, the extraction of MMG signal variables could be performed in time or frequency domain and in time-frequency representations. Diverse methods of time domain or frequency domain marked the beginning of the investigation into the signal analysis of MMG. However, different types of joint time–frequency representation analysis have been suggested lately to extract the mechanomyographic signal features due to the non-stationarity of the signal (Beck et al., 2008). Typical, in the time domain, the amplitude variables of the signal are voltage values detected from specific points on the signal. Such as the peak to peak (PTP), root mean square (RMS) and average rectified values (ARV). The amplitude characteristics of the signal are essentially an indication of the changes in motor unit recruitment during contraction (Orizio et al., 2003a).

The standard practice includes the computation of power spectra of MMG signals using a fast Fourier transform (FFT) or discrete Fourier transform (DFT) algorithm to obtain the frequency domain equivalence of the signal. Essentially, the median power frequency (MPF) and mean power frequency (MPF) are the most common variables often computed from the frequency domain (Madeleine et al., 2006). Generally, the MMG frequency variables reflect changes in the global firing rate of unfused activated motor units during contraction (Ryan et al., 2008). The mean power frequency is favored to represent power spectrum of MMG because it is a vital metric to examine muscle mechanical changes underlying the muscle activation (Diemont et al., 1988). During a particular muscle action, changes in the muscle length, the number of active motor units, and the thickness of the tissue between the muscle and MMG sensor can affect MMG amplitude and frequency contents (Frangioni et al., 1987; Jaksölska et al., 2004). However, as both the extracted amplitude and frequency contents of MMG signals can be used to estimate the motor control strategies during muscle actions (Beck et al., 2008), the selection of appropriate signal processing techniques on the basis of their estimation accuracy and precision to analyze MMG signals is vital.

1.4. MMG parameters and its relationship with the muscle contraction

MMG records and quantifies lateral oscillation of contracting muscle fiber (Orizio et al., 1989). This modality has been shown to generate sufficient information about motor unit recruitment (Akataki et al., 2001). During muscle contraction, the trend of MMG parameters has been verified to reflect the alteration in motor unit activation strategy (Itoh et al., 2004). During isometric contractions, investigators have continuously demonstrated an increase in MMG amplitude over time when the submaximal force production is low (between 10% and 40% maximum voluntary contraction (MVC)). While at high level of muscle force production, i.e. 50–80% MVC, either no change or a decrease in MMG amplitude has been reported (Perry-Rana et al., 2002; Rodríguez et al., 1993) and a progressive decrease in MMG amplitude at higher level of muscle effort (80–100% MVC) has been described (Orizio et al., 1989). Oster and Jaffe (1980) demonstrated a linear relationship between the contraction force of the forearm flexor muscle and the root mean squared (RMS) amplitude of mechanomyogram. Therefore, the amplitude of MMG signal has been shown to correlate with motor unit activation during voluntary contraction (Kimura et al., 2004) and electrically evoked contraction (Beck et al., 2005). There are also documented studies regarding the mean power frequency (MPF) of MMG relationship with motor unit firing rate strategy underlying muscle force generation (Akataki et al., 2001) during contraction, and fatigue tracking (Ebersole et al., 2006) during fatiguing contraction. MMG is particularly sensitive to fatigue induced changes in mechanical properties of the muscle (Orizio et al., 1999b). Collectively, in isometric contraction, MMG signal has shown reliability in quantifying the
differences in absolute muscle force generation with potential application in the assessment of muscular performance in healthy and diseased muscles (Matheson et al., 1997; Sarver and Seliktar, 2000) (Fig. 4). In dynamic contraction, MMG amplitude has equally been used to indicate torque generating capacity of the skeletal muscle. The non-uniformity of the MMG amplitude of the individual muscle (vastus lateralis, vastus medialis and radius femoris) in the quadriceps muscle groups could be used to discriminate the contribution of the individual muscle to the force production in the dynamic contraction of the muscle (Perry-Rana et al., 2002) and the linear relationship between MMG amplitude and the muscle effort (Shinohara et al., 1997).

Since motor units are the basic functional block of neuromuscular system (Cescon et al., 2004b), analysis of their firing pattern enables understanding of strategies adopted by the central nervous system to control different muscle activities and conditions such as the force generation and fatigue (Bigland-Ritchie et al., 1983; Zennaro et al., 2003). Investigations of MMG signal have revealed a significant relationship between its parameters and motor unit activation strategy (Beck et al., 2004; Orizio et al., 2003a). Consequently, the signal has been thoroughly verified to generate mechanical information on the neuromuscular contraction, and specifically to examine muscle function of clinical and research relevancies (Orizio, 1992). The signal has also been validated to possess rich information about muscular characteristics of individuals with muscular weakness (Orizio et al., 1996). The analysis of its amplitude and frequency domain has consistently been assessed to discriminate muscle fiber types (Marchetti et al., 1992). Although the research efforts on MMG signal have been predominantly on isometric muscle actions, convincing evidence is accumulating to assert the reliability of the signal as a viable dynamic muscle function assessment tool as consistently demonstrated by the close correlation between the MMG amplitude and muscle power output (Beck et al., 2005; Cramer et al., 2002a). In addition, a positive relationship has been established between MMG amplitude and velocity of contraction with an inverse relationship with the peak torque i.e. velocity related dissociation between MMG amplitude and the peak torque (Evetovich et al., 1997) (Fig. 5). However, Islam et al. (2013) in their review observed that the selection of MMG parameter for muscle assessment is a function of protocol adopted, area of application and the muscle of interest.

2. Applications

2.1. State-of-the-art in applications of MMG signal in muscle research

Based on the reviewed empirical data, MMG has been widely demonstrated as a control signal (Antonelli et al., 2009), and used to monitor training induced changes in muscle power output and other areas of muscle function assessments (Cramer et al., 2002b). The signal has gradually evolved as a useful tool in clinical diagnosis in various rehabilitation interventions (Beck et al., 2005). Under a general framework, MMG applications in assessing muscle strength and efforts have been applied as a tool to diagnose neuromuscular diseases (Marusiak et al., 2009), to facilitate the establishment of appropriate rehabilitation programs (Gorelick, 2006), and as diagnostic tool to prevent musculoskeletal disorder (Kassolik et al., 2009). Investigators have also verified that MMG signal could detect modifications to mechanical processes of muscle activation due to the muscle fatigue (Ebersole and Malek, 2008; Rana, 2006), force, and could be used to estimate muscle fiber composition (Dahmane et al., 2001). Table 1 presents the studies of MMG signal applications in the examination of neuromuscular disorder. The studies that recruited human subjects with clear protocols and data presentations were included. Collectively, the validation of MMG signal to assess muscle function (Beck et al., 2010; Cramer et al., 2003) has been verified by various attempts as presented in Table 2 included studies with appreciable sample sizes of healthy volunteers, clear data presentation and excluded studies on non-human subjects.

Nevertheless, with all these areas of application (diagnostic tool, control signal, assessment tool, indicator of fatigue threshold among others), MMG signal is still far from full exploration when compared to other non-invasive means of muscle function assessment modalities. Although the significant impact of the signal is promising and its application has been fairly demonstrated, the practical implementation and multi-center validation on the target population remain inadequate.

Fig 2. Typical arrangement of MMG measurement with accelerometer integrated displacement sensor. Conversion of analog MMG signal was shown with graphical display in displacement versus time. $D_m$ represents maximum displacement, $T_{D_m}$ means time at which $D_m$ occurred and $T_r$ is half relaxation time. Reprinted with permission from Žagar and Križaj (2003).
2.2. MMG as the mechanical counterpart of EMG

The application of EMG recording to indicate the level of muscle activation or assess muscle fatigue has been widely investigated (Farina et al., 2004; Jiang et al., 2012; Shi et al., 2007; Vøllestad, 1997). However, the clinical impact of the electrical information from EMG measurements has been consistently reported to be limited without the manifestation of the mechanical muscle effort. The combination of the two signal modalities (EMG and MMG) has been suggested to provide the unique information about the electrical and mechanical aspect of the muscle performance (Cramer et al., 2002a). The investigation of MMG variables to analyze muscle output is therefore considered complementary to EMG assessments. Among the most reported research attempts to simultaneously adopt measurement of EMG and MMG in order to verify the differences between electrical and mechanical actions is that of Stokes and Dalton (1991). On the other hand, electromechanical and phonomechanical dynamic contraction delay factors were examined by Petitjean and colleagues using the same approach (Petitjean et al., 1992). Much later, simultaneous measurement of EMG and MMG was validated to provide unique information on motor control strategies that was essentially used to modulate the production of torque in dynamic muscle activities (Beck et al., 2004). We also found sufficient evidence in the literature to claim that both electrical and mechanical responses of fatigued and atrophied muscle are reduced when compared to fresh and healthy muscle through simultaneous assessment of EMG and MMG parameters (Barry et al., 1990). Consequently, combined application of EMG and MMG (electromechanical coupling) has a diagnostic importance. Significant impetus on the confirmation of reliability of MMG as a competent complement to EMG in muscle research has been thoroughly elucidated by Stokes and Dalton (1991).

Investigators have generally demonstrated some physiological measurements to establish the validity of MMG signal as the mechanical counterpart of the electromyogram (EMG). MMG reflects the first mechanical event after muscle activation (Orizio et al., 1989), before elongation of the series of elastic component of the skeletal muscle (Cè et al., 2014). The time delay between the onset of the EMG and MMG signals has been shown to provide information on the propagation of the action potential along the sarcolemma of the muscle fiber to the cross-bridge formation. During the skeletal muscle contraction, the time delay from the onset of muscle electrical activation (EMG) and the onset of mechanical manifestation in form of measurable tension has been labeled electromechanical delay (Cavanagh and Komi, 1979), which is between 30 and 100 ms (Cavanagh and Komi, 1979). Thus, during the performance of explosive and fast movements it is possible to observe termination of the EMG activity before force can be detected (Cavanagh and Komi, 1979). Several mechanical events contribute to electromechanical delay during muscle contraction, such as: (a) propagation of the action potential over the sarcolemma; (b) calcium release by the sarcoplasmic reticulum, with calcium binding to troponin to allow the cross-bridge formation; and (c) the myosin

Fig. 3. Arrangement of MMG sensors. (a) MMG acquisition with only one sensor as opposed to EMG with three electrodes including reference electrode (Hendrix et al., 2010), (b) placement of two miniature accelerometers on vastus lateralis (VL) and rectus femoris (RF) muscles (Ryan et al., 2008), (c) simple setup of a vibromyogram’s transducer; no skin preparation and no adhesive tapes are required; http://www.biopac.com/ResearchApplications.asp?Level=1&Aid=68. Simplicity of setup is observed in MMG acquisition in upper and lower limbs, and during isometric and dynamic muscle contractions.

Fig. 4. An example of the test retest reliability of the MMG signal obtained with piezoelectric accelerometer. The MMG (RMS)–root mean square values of amplitude for the (HF)-high force and (LF)-low force were acquired on the two separate test days during voluntary isometric contraction of the five quadriceps force values. Trial 1 and trial 2 represent the test and the retest data.

Reprinted with permission from Matheson et al. (1997).
heads pull with subsequent force transmission to the tendon insertion point through the elongation of the series elastic components of the muscle tendon unit. Therefore, factors affecting one or more of these processes may potentially influence the duration of electromechanical delay.

Changes in electromechanical delay occur with increasing contraction intensity, after mechanical stress of the muscle–tendon unit, after fatiguing tasks and with variations in muscle temperatures (Rampichini et al., 2013). It has been suggested (Cè et al., 2014) that muscle fatigue may increase the electromechanical delay by affecting the speed of motor unit action potentials, and thus, delaying its propagation across the sarcolemma affecting the calcium efflux from sarcoplasmic reticulum, which in turn affects the cross-bridge cycling rate. In addition, fatigue has been shown to increase muscle–tendon unit compliance due to repetitive cycles of stretches and relaxations. The increased compliance of the aponeurosis and other elastic components suggests a longer time for force transmission to the bone, giving further contribution to overall electromechanical delay increase due to fatigue (Cè et al., 2014; Rampichini et al., 2013; Rodriguez-Falces and Place, 2013).

Following the electromechanical delay that occurs during the contraction phase of the skeletal muscle, there is another important mechanical phenomenon that occurs during the relaxation phase. This electromechanical event is characterized by latency between the cessation of muscle electrical activity and the beginning of force decay (Viitasalo and Komi, 1981). This time lag between the cessation of muscle electrical activity and the beginning of force decay has been labeled relaxation electromechanical delay (Ferris-Hood et al., 1996) and has been suggested to be readily estimated by MMG signal (Stokes, 1993). The relaxation electromechanical delay includes several electrochemical events, such as: (a) cessation of motor unit action potential propagation across the sarcolemma; (b) calcium reuptake by the sarcoplasmic reticulum and termination of actomyosin interaction; and (c) non-force-generating cross-bridges. The electrochemical events related to the relaxation of electromechanical delay directly affect certain mechanical properties of the muscle, specifically the return of the sarcomeres toward their resting length and the release of the series elastic components previously stretched during contraction (Cè et al., 2014; Ferris-Hood et al., 1996). The understanding of these processes can provide useful information for clinicians and rehabilitation experts. Several neuromuscular disorders (Table 3), such as myotonic dystrophy, affect not only the muscle contraction but also the relaxation phase. The impaired activation or relaxation of contractile activity is due to intrinsic muscle dysfunction, such as diseases associated with myotonia or periodic paralysis (Lewis and Haller, 1989). Furthermore, assessment of the delays of electromechanical events that occur during the contraction and relaxation phases may represent a useful tool to detect possible physiological adaptations induced by physical exercise and may be useful to follow the effects of therapeutic and rehabilitative treatments focused on restoring normal muscle function (Cè et al., 2014; Lewis and Haller, 1989).

Combining all the reported attempts, sufficient evidences have been documented to justify the enhanced efficacy of muscle function assessment when MMG is simultaneously applied as the mechanical counterpart of the EMG signal (Al-Mulla et al., 2011; Cramer et al., 2005). In Table 3, studies were included only if the researchers recruited human subjects to examine the potential of the MMG as a control signal in muscle machine interfaces.

### 3. General muscle function assessment

The broad classification of the muscle function assessment using MMG signal is isometric and dynamic contraction. Inherent in this classification are both voluntary and evoked modes of muscle contraction. Typically, the maximal muscle strength can be estimated during isometric contraction whereas muscle dynamic action is known to describe functional activities (Kroemer, 1999).

#### 3.1. Application of MMG during voluntary muscle contraction activities

Accordingly, while the signal may have appeared as a good valuation tool in voluntary contraction, the extent of its limitation should not be taken lightly. Posatskiy and Chau (2012) have verified the insensitivity of the MMG signal to the motion artifact during dynamic muscle contraction by demonstrating an improved signal-to-noise ratio when a condensed microphone was used for the signal acquisition as opposed to an accelerometer. Some authors have justified that MMG could even be obtained in a noisy environment and thus handle artifact efficiently if acquired with a differential probe (Beck, 2010; Gregori et al., 2003). Tables 1 through 3 summarize the key studies that reflected the application of MMG signal in voluntary contractions. Based on the reviewed articles, MMG has been shown to correlate with the muscle torque output and assess muscle fatigue during isometric and dynamic muscle activities. This supports the validity of the application of MMG as a mechanical counterpart to the EMG.

#### 3.2. Application of MMG during electrically evoked muscle contraction activities

Generally, electrical stimulation of the nerves especially at supramaximal level, influences the muscle to behave as a big artificial motor unit (Partridge, 1965) making its contraction to couple with dimensional changes due to transverse diameters (Gobbo et al., 2006; Partridge and Benton, 1981) leading to muscle activation. Thus, the acquired MMG signal represents “the summation of single mechanical
activities of the fibers belonging to motor unit recruited by stimuli” (Oriizio et al., 1999b). Despite this justification on validity of the signal and available literature on MMG variables and muscle function assessments, as evaluated in Table 4, it could be observed that little research attention was paid to the protocol validation, signal acquisition and processing despite the significant quantitative assessment of the signal’s outcome. Only studies that recruited human subjects with detailed stimulation protocols, clear data analysis and presentation were included in Table 4.

It should however be stated that the dearth of data on assessment of muscle dynamic contraction with MMG signal is a significant knowledge gap. According to Stokes (1992), MMG signal could only monitor muscle force during functional activity if its relationship is considered with force during dynamic muscle action. By inference, practical gait enhancement via electrical stimulation induced gait could only be made possible with muscle dynamic assessment. However, a few numbers of research trials were documented in the literature on the application of MMG signal to assess evoked muscle contractions. Many of such studies even recruited insufficient sample size of spinal cord injury (SCI) volunteers with control group thus no sufficient evidence to generalize the observed trends. However, a desirable strength of the MMG signal is that the concurrent acquisition of the signal during electrical stimulation of the nerves does not warrant stimulation artifact circuit as with EMG. This potentially makes a compact circuit arrangement with prospects in sport and rehabilitation medicine during the clinical management of individuals with SCI.

### Table 1

<table>
<thead>
<tr>
<th>Area of application</th>
<th>Methods</th>
<th>Statistical analysis and findings</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Examination of the neuromuscular disorder</td>
<td>20 subjects with the neuromuscular disorder and 17 healthy volunteers were recruited. Carotid phonoangiography microphone was used to detect muscle sound from medial biceps when the microphone was pressed against stationary object.</td>
<td>Significant difference found between patients and healthy volunteers ($P &lt; .05$). Data from subject under 60 years generated better significant result ($P &lt; .01$).</td>
<td>Rhatigan et al. (1986)</td>
</tr>
<tr>
<td>Examination of the spastic cerebral palsy (CP) between patient and healthy volunteers</td>
<td>6 moderate spastic and 8 healthy volunteers were recruited. Piezoelectric sensor was used to obtain the acoustic myogram from biceps brachii (BB). X-ray tomography was used to determine muscle cross-sectional area (CSA).</td>
<td>CF group had smaller CSA and MVC of BB (CSA: 10.3 plus or minus 3.9 cm$^2$; MVC: 22.4 plus or minus 8.6 Nm) than healthy volunteers’ (CSA: 15.5 plus or minus 22.2 cm$^2$; MVC: 53.3 plus or minus 11.9 Nm). Reduced muscular force was partly due to the deterioration in contractile properties of CP.</td>
<td>Akataki et al. (1996)</td>
</tr>
<tr>
<td>To monitor the myotonic dystrophy (MD) non-invasively</td>
<td>Strain gauge was used to measure elbow flexion force. 10 MD patients and 10 healthy volunteers (HV) were recruited. MMG recorded during isometric contraction at 20, 40, and 60% MVCs, from forearm and finger flexors.</td>
<td>MMC was 30% lower in MD than HV ($P &lt; .05$) for elbow flexor, but 74% lower for finger flexor ($P &lt; .001$). MMC-RMS for MD was significantly lower than (HV) ($P &lt; .05$).</td>
<td>Oriizio et al., (1997)</td>
</tr>
<tr>
<td>To determine the cause of low back pain muscle fatigue.</td>
<td>8 healthy volunteers were recruited. Each subject performed isometric back extension for 60 s at 15° with reference to horizontal plain. Bipolar Ag/AgCl electrode and microphone were used to obtain EMG and MMG respectively from erector spinae. NIRS obtained to determine the level muscle blood volume and oxygenation.</td>
<td>During sustained back extension; RMS of EMG increases significantly up to 36 s ($P &lt; .01$) and falls thereafter ($P &lt; .05$), and MPF of EMG decreases significantly as a function of time ($P &lt; .01$). RMS of MMG increases up to 20 s ($P &lt; .05$) and decreased subsequently ($P &lt; .01$). MPF of MMG remains unchanged at the end of fatiguing contraction ($P &gt; .05$). Intramuscular mechanical pressure was identified as a factor underlying fatigue in low back pain.</td>
<td>Yoshitake et al. (2001)</td>
</tr>
<tr>
<td>To assess the musculoskeletal pain</td>
<td>12 healthy volunteers were recruited. MVC of elbow was obtained as the highest elbow flexion force recorded in three trials. Before pain, static (0, 10, 30, 50, 70) % MVC and ramp isometric contraction were conducted randomly. Experimental muscular pain was induced. Force transducer was used to acquire force output while piezoelectric accelerometer was used to MMG from biceps brachii.</td>
<td>RMS of MMG increased significantly after induction of pain ($F = 7.8; P = 0.019$). MMG indicated decrease firing rate and increase twitch force due to muscle pain.</td>
<td>Madeleine and Arendt-Nielsen, (2005)</td>
</tr>
<tr>
<td>To analyze muscular characteristics after stroke</td>
<td>5 healthy and 8 stroke survivors with pure spastic hemiplegia but stable condition were recruited. 2 surface EMG electrodes were attached above the belly of biceps brachii and reference placed on olecranon while accelerometer sensor was used to obtain MMG while placed in between the 2 EMG electrodes. Each subject performs MVC of elbow flexion for 5 min and maximum torque set to contraction level of 100% MVC. Each subject then performed 20%, 40%, 60% and 80% MVCs after 5 min rest of MVC.</td>
<td>MPF and high frequency-rate of survivors were significantly lower than healthy volunteers ($P &lt; .05$) at high contraction level 80% MVC. MMG complemented EMG to analyze properties of paretic muscle due to atrophy. RMS of MMG and EMG of healthy volunteers were significantly higher than patients ($P &lt; .05$) at 60% &amp; 80% MVCs. These could be due to atrophy of fast twitch fibers and reduction of neural input in affected muscles of patients.</td>
<td>Hu et al. (2007)</td>
</tr>
<tr>
<td>Determination of forearm muscle activities</td>
<td>12 healthy volunteers were recruited. Accelerometer was used to obtain MMG from flexor and extensor carpi radialis muscle. Data acquired over five sessions with 10 min of interval between sessions.</td>
<td>Average classification accuracy of the five sessions was 89 plus or minus 2% with no decline in accuracy over time. MMG potentially demonstrated as a multifunction control signal for muscle–interface machine (MMI).</td>
<td>Alves and Chau (2010a)</td>
</tr>
<tr>
<td>Detection of spasticity in hamstring muscle</td>
<td>4 spinal cord injured (SCI) individuals were recruited. Subjects were divided into two groups. Accelerometer was used to obtain MMG from RF, VL and VM.</td>
<td>The significance of each muscles are; RF ($P &lt; .004$), VL ($P &lt; .001$) and VM ($P &lt; .007$). Spasticity in hamstring was assessed using antagonist muscle group (quadriceps) Positive correction was obtained between invasive needle EMG and MMG for static jaw opened by 30 mm ($P = 0.000, r^2 = 0.725$), by 40 mm ($P = 0.000, r^2 = 0.725$), by 50 mm ($P = 0.000, r^2 = 0.725$) MMG could be used to monitor the activity of lateral Pterygoid muscle.</td>
<td>Krueger et al. (2012)</td>
</tr>
<tr>
<td>Non-invasive evaluation of the craniofacial disorder.</td>
<td>2 healthy volunteers participated. Condensed microphone installed in earplug was used to measure MMG from external auditory canal, while needle EMG electrode used to obtain for inferior head of lateral Pterygoid.</td>
<td></td>
<td>Kawakami et al. (2012)</td>
</tr>
</tbody>
</table>

Abbreviation: NIRS — near infrared spectroscopy, RF — radius femoris, VL — vastus lateralis, VM — vastus medialis, MPF — mean power frequency.
### Table 2
Summary of the key studies on application of MMG signals for the muscle output/response assessment.

<table>
<thead>
<tr>
<th>Area of applications</th>
<th>Methods</th>
<th>Statistical analysis and findings</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Non-invasive monitoring of the motor unit fatigue in vivo.</td>
<td>5 healthy volunteers were recruited Subjects were instructed to hold weights (0, 2.27, 4.54, 6.80, and 9.07) Kg with forearm parallel to the ground. Each weight held for 20 s with 10 s between weights. Phonocardiograph was used to measure MMG from biceps brachii during sustained isometric contraction.</td>
<td>Qualitative data was not presented. RMS of MMG increased with increasing contraction. MMG characteristics followed declined in force due to fatigue. Since MMG could resolve individual motor unit, further analysis may have prognosis/diagnosis information.</td>
<td>Bary et al. (1985)</td>
</tr>
<tr>
<td>To investigate the muscle fatigue noninvasively in quadriceps</td>
<td>5 healthy volunteers were recruited Force of MVC was defined as maximum force maintained for ≥1 s. Submaximal force was calculated at 10, 25, 50, 60 and 75% MVCs when target force visual feedback was shown to subjects. Intermittent isometric contraction at 75% MVC was repeated until 40% MVC was sustained for three consecutive contractions to induce fatigue.</td>
<td>Linear relationship observed between force and MMG in fresh (r = 0.98 plus or minus 0.001) and fatigue muscle (r = 0.97 plus or minus 0.002) and regression line was shown to be similar (P &gt; .05). MMG validated changes in force of isometric quadriceps contraction when fresh and fatigued.</td>
<td>Stokes and Dalton (1991)</td>
</tr>
<tr>
<td>To investigate the force accompanying muscle contraction.</td>
<td>31 healthy volunteers were recruited Inductive force transducer and microphone were used to obtain isometric contraction force and MMG of flexor and extensor muscles. PEAK force of spherical and trapezoid contraction recorded for 10 N, 20 N, and 30 N. Target force was set in range of 5–15 N, 15–30 N, and 30–40 N respectively. The recorded force of contraction was 5 N–50 N and was in range of 4.0–40.2% MVC.</td>
<td>Within both triangular and trapezoid contractions, differences between mean values of MMG amplitude at 3 different force levels were significant (P = .01). Between MVC intensity levels at 30% MVC level within triangular contraction were significantly different (P &lt; .01 or .05), while for trapezoid contraction, only differences between 10 N and 20 N and 40 N were significant (P &lt; .05). Amplitude/intensity level of MMG was shown to correlate with change in force of voluntary contraction.</td>
<td>Celichowski et al. (1998)</td>
</tr>
<tr>
<td>To obtain the complementary information about the electrical and mechanical activities of the muscle</td>
<td>13 healthy volunteers were recruited. MVCs were estimated for each subject. Bipolar EMG electrode and piezoelectric accelerometer were used to acquire EMG &amp; MMG respectively from first dorsal interosseus muscle during slow concentric, isometric and eccentric (CIE) contraction at 0, 25, 50, 75, and 100% MVCs.</td>
<td>During voluntary CIE, non-linear significant difference was observed in EMG/force &amp; MMG/motor-unit relationship (P &lt; .001). RMS of EMG significantly increased from 0 to 50% MVC during concentric &amp; isometric, and up to 75% MVC during eccentric contractions (P &lt; .05). RMS of MMG increased significantly from 0 to 50% MVC during concentric contraction (P &lt; .05).</td>
<td>Madeleine et al., 2001</td>
</tr>
<tr>
<td>To obtain detailed information about the MU activation strategy underlying MMG/force relationship.</td>
<td>10 healthy volunteers were recruited MVC was obtained from maximal force of three repeated trials. Each subject exert submaximal isometric contraction from 5% to 80% MVC with 10% increment. Accelerometer and strain gauge were used to obtain MMG and elbow flexion force respectively from biceps brachii.</td>
<td>Complementary information about MU recruitment. At low force, RMS of MMG increased slowly with reduction between 12 and 20%, and a progressive increase that decreased beyond 60% MVC. For MPF, a rapid increase up to 30% MVC followed by slow increment between 30 and 50% MVC, then slight reduction at 50% followed by increment above 60% was observed.</td>
<td>Akataki et al. (2001)</td>
</tr>
<tr>
<td>Estimation of the force/torque output during fatiguing dynamic contraction.</td>
<td>10 healthy volunteers were recruited. Isokinetic testing was performed at velocities 60°/s, 180°/s, and 300°/s and range of motion standardized from 90° to 180° knee flexion for each subject. Piezoelectric was used to obtain MMG from quadriceps femoris.</td>
<td>RMS of EMG increased progressively as contraction develops. MF regression slopes of the signal between male and female differed between isometric and eccentric contraction. MF amplitude and %PT was significantly different (F = 46.5, P = .001).</td>
<td>Perry-Rana et al. (2002)</td>
</tr>
<tr>
<td>Fatigue assessment of the upper trapezius muscle</td>
<td>11 healthy volunteers were recruited. Non-fatiguing and fatiguing contractions were conducted. MMG was obtained from right trapezius muscle with piezoelectric accelerometer and rate of perceived exertion in five arm positions (0° abductor &amp; flexion, 45° &amp; 90° flexion &amp; 90° abduction) with different bilateral arm loads for 3 s for non-fatiguing contraction and 3 m for fatiguing contraction.</td>
<td>For non-fatiguing contraction, significant difference in the mean of MMG's ARV and RMS among different arm positions (F = 6.79, P = .001). For fatigue contraction, mean rates of perceived exertion were significantly different (F = 46.5, P = .001) except for one arm position. MMG could be used to monitor fatigue development with relevant in ergonomics.</td>
<td>Madeleine et al., 2002</td>
</tr>
<tr>
<td>To quantify fatigue development in isometric exercise.</td>
<td>18 healthy volunteers (male, N = 9, female, N = 9) of close age range were recruited. MVC was estimated. Two accelerometers were used to obtain MMG signal on the biceps &amp; brachioradialis during sustained isometric contraction at 23% MVC.</td>
<td>RMS of MMG increased with increased fatigue, and MF of PSD decreased progressively as contraction develops. MF regression slopes of the signal between male and female were shown to be closer (biceps, 1.55 and brachioradialis, 13.2%) MMG could be used as an indicator of muscle activation and fatigue.</td>
<td>Tarata (2003)</td>
</tr>
<tr>
<td>To reflect changes in the MUAS due to fatigue.</td>
<td>10 healthy sedentary volunteers were recruited MVC was obtained for each subject. Accelerometer was used to measure MMG from dominant arm biceps brachii during isometric contraction while load cell was used to measure elbow flexor force output. RMS and MF-PSD of MMG were obtained. Linear mixed effect model was used to compare fresh and fatigue muscle of considered parameters against MVC relationship.</td>
<td>In fresh muscle, at 20% to 65% MVC effort range, MMG-RMS progressively increased and decreased between 65% and 86% MVC range. While in fatigued muscle, MMG-RMS did not increase with increasing effort but rather decrease from low to high effort intensity with exception to 30% and 35% MVC effort. MMG could be used to study the interaction between recruitment and firing rate of motor unit activation strategy.</td>
<td>Orizio et al. (2003)</td>
</tr>
<tr>
<td>To describe motor unit control strategy that modulates torque production.</td>
<td>12 volunteers were recruited. Isokinetic testing was conducted following orientation session and warm up. Peak torque (PT) was also obtained.</td>
<td>RMS of MMG against PTs were best fitted with linear model (r² = 0.984). No significant relationship between MFP of MMG and SPT (P &gt; .05). RMS of EMG against PTs was best fitted with linear model (r² = 0.988). No significant relationship between MFP of EMG and SPT (P &gt; .05).</td>
<td>Beck et al. (2004)</td>
</tr>
</tbody>
</table>
### 4. Discussion

#### 4.1. A recap of the challenges of MMG signal

Once the MMG signal limitation is established prior to acquisition and application, the chance for erroneous conclusion will be minimized. Experienced investigators have identified some isolated cases of challenges when MMG signal is applied, presumably due to the limited available knowledge leading to procedural limitations (Antonelli et al., 2009). Recent experiment conducted by Al-Zahrani et al. (2009) to examine within and between day reliability of MMG signal in quadriceps isometric muscle fatigue assessment suggested that within day reliability was high but between day reliability was low. Investigators recommended the further verification of their findings.

During the analysis of dynamic muscle action which is considered to be more relevant in clinical applications (Stokes, 1993), a number of factors have been identified to affect the response of the signal including the variation in muscle length and consequently the muscle torque, the motion artifacts, the ambient temperature and the effect of the volume conductor due to the thickness between the muscle of interest and the MMG transducers (Farina et al., 2008; Jaskólska et al., 2004; Stokes, 1993). Sensor selection has equally been suggested to affect the interpretation of the pattern of the motor control strategies. There is need to continue to investigate the pattern of response between sensors’ (accelerometer, piezoelectric contact sensor, condenser microphone, laser distance sensor etc.) responses during different muscle activities including voluntary and evoked isokinetic contraction.

It has also been observed during the established application of MMG as a control signal in prosthesis that the artifact due to the limb movement is seldom encountered (Silva et al., 2003) and efforts to manage this noise thus remain a significant research gap. There is insufficient evidence to ascertain the effectiveness of the MMG in multifunction control signal in individual with severe disability (Table 3). Therefore the demonstration and validation of the technology in the target population will be enlightening.

Lack of sufficient evidence and threat to internal and external validity may be likely in most available studies on MMG signal. For that reason we may not be able to generalize most of the observed trends in a specific area of muscle research. As practical implementation and clinical deployment should not be underestimated, investigators should be aware of the factors that could influence MMG during application in muscle research, such as muscle mass, stiffness, temperature, intramuscular pressure, viscosity of intracellular and extracellular fluid media, and firing rates of active motor units (Marchetti et al., 1992; Orizio et al., 2003a).

So far, the signal has not been sufficiently validated to assess muscle function especially during electrically evoked contraction of SCI individuals. Based on the current literature, there is insufficient evidence based research with adequate sample size of individuals with neurological impairment, such as SCI, stroke, or multiple sclerosis, and the control group under electrical evoked contraction. In addition, few reported information on muscle contraction. A significant correlation ($r = 0.83$) for MPF at fatigue threshold versus RCP was obtained. Result could be used to demarcate different exercise intensity domain. normalized RMS error (0.131 plus or minus 0.018), and cross correlation coefficient (0.892 plus or minus 0.033) of MMG signal. Estimated force and measured force were demonstrated to almost coincide from ANN model. MMG combined with EMG was verified to yield complementary information on muscle contraction.

Dynamic force generation may be modulated by recruitment than increase in global MUFR. Mean of electromechanical efficiency i.e. ratio of ARV of MMG to ARV of EMG (that shows how motor unit action potential convert to muscle contraction) of post-fatigue was lower than pre-fatigue significantly at 10% ($P < .01$) and at 20% MVC ($P < .05$). Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P < .05$).

Cubic decreased in torque at 60°/s ($r^2 = 0.972$), 300°/s ($r^2 = 0.931$) could be linked with decline in torque by 59 plus or minus 24% and 53 plus or minus 11%. RMS and MPF of MMG responses showed that each muscle of quadriceps femoris uniquely contributes to the force output control across the 50 repetitions.

Normalized RMS error (0.131 plus or minus 0.018), and cross correlation coefficient (0.892 plus or minus 0.033) of MMG signal. Estimated force and measured force were demonstrated to almost coincide from ANN model. MMG combined with EMG was verified to yield complementary information on muscle contraction.

### Table 2 (continued)

<table>
<thead>
<tr>
<th>Area of applications</th>
<th>Methods</th>
<th>Statistical analysis and findings</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>To estimate the elbow flexion force</td>
<td>7 healthy volunteers were recruited. Each participant performed isometric contraction biceps brachii while accelerometer was used to obtain MMG signal. Two layer model NN was used to estimate contraction strength. The algorithm was compared with MMG-RMS linear mapping. Evaluation conducted under constant posture and contraction at 0.5 Hz, 0.25 Hz and 0.125 Hz.</td>
<td>A significant correlation ($r = 0.83$) for MPF at fatigue threshold versus RCP was obtained. Result could be used to demarcate different exercise intensity domain.</td>
<td>Zuniga et al. (2010)</td>
</tr>
<tr>
<td>Determination of fatigue threshold during cycle ergometry</td>
<td>5 moderately trained volunteers were recruited. Each volunteer participated in maximal cycle ergometer test after orientation session. Accelerometer was used to obtain MMG from VL.</td>
<td>Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P &lt; .05$). At increasing contraction, RMS of MMG increased monotonously, while VAR decreased. Frequency variance parameter of MMG was used to estimate the contractile strength of muscle.</td>
<td>(Lei et al., 2013)</td>
</tr>
<tr>
<td>To estimate the muscle contraction strength</td>
<td>17 healthy volunteers were recruited. Each subject performed 50 maximum concentric muscle action on Biodex dynamometer at 60°/s and 300°/s. Bipolar surface electrode and piezoelectric sensor were used to acquire EMG and MMG respectively from quadriceps during dynamic contraction.</td>
<td>Estimated force and measured force were demonstrated to almost coincide from ANN model. MMG combined with EMG was verified to yield complementary information on muscle contraction.</td>
<td>Ebersole et al. (2006)</td>
</tr>
<tr>
<td>Assessment of fatiguing dynamic muscle action.</td>
<td>16 healthy volunteers were recruited. Each subject performed pre- and post-fatigue trials at 0, 5, 10, and 20% MVCs for 20 s, with 5 min recovery period allowed only during pre-fatigue trials.</td>
<td>Dynamic force generation may be modulated by recruitment than increase in global MUFR. Mean of electromechanical efficiency i.e. ratio of ARV of MMG to ARV of EMG (that shows how motor unit action potential convert to muscle contraction) of post-fatigue was lower than pre-fatigue significantly at 10% ($P &lt; .01$) and at 20% MVC ($P &lt; .05$). Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P &lt; .05$).</td>
<td>Itoi et al. (2006)</td>
</tr>
<tr>
<td>To investigate masseter muscle fatigue noninvasively</td>
<td>7 healthy volunteers were recruited. Accelerometer and bipolar electrode were used to obtain MMG and EMG signal respectively. MCV of isometric elbow flexion was estimated with 3 trials with 5 minute rest between trials and maximum value set as 100% MVC. Level of target force set at 10, 20, 30 &amp; 40% MVCs and ramp muscle contraction at 10, 20 &amp; 30% MVCs to avoid muscle fatigue. MAV and zero crossing (ZC) were input to ANN and forces measured by force sensor were target values.</td>
<td>Dynamic force generation may be modulated by recruitment than increase in global MUFR. Mean of electromechanical efficiency i.e. ratio of ARV of MMG to ARV of EMG (that shows how motor unit action potential convert to muscle contraction) of post-fatigue was lower than pre-fatigue significantly at 10% ($P &lt; .01$) and at 20% MVC ($P &lt; .05$). Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P &lt; .05$).</td>
<td>(2010)</td>
</tr>
<tr>
<td></td>
<td>16 healthy volunteers were recruited.</td>
<td>Dynamic force generation may be modulated by recruitment than increase in global MUFR. Mean of electromechanical efficiency i.e. ratio of ARV of MMG to ARV of EMG (that shows how motor unit action potential convert to muscle contraction) of post-fatigue was lower than pre-fatigue significantly at 10% ($P &lt; .01$) and at 20% MVC ($P &lt; .05$). Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P &lt; .05$).</td>
<td>(2010)</td>
</tr>
<tr>
<td></td>
<td>17 healthy volunteers were recruited. Each subject performed 50 maximum concentric muscle action on Biodex dynamometer at 60°/s and 300°/s. Bipolar surface electrode and piezoelectric sensor were used to acquire EMG and MMG respectively from quadriceps during dynamic contraction.</td>
<td>Dynamic force generation may be modulated by recruitment than increase in global MUFR. Mean of electromechanical efficiency i.e. ratio of ARV of MMG to ARV of EMG (that shows how motor unit action potential convert to muscle contraction) of post-fatigue was lower than pre-fatigue significantly at 10% ($P &lt; .01$) and at 20% MVC ($P &lt; .05$). Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P &lt; .05$).</td>
<td>(2010)</td>
</tr>
<tr>
<td></td>
<td>17 healthy volunteers were recruited.</td>
<td>Dynamic force generation may be modulated by recruitment than increase in global MUFR. Mean of electromechanical efficiency i.e. ratio of ARV of MMG to ARV of EMG (that shows how motor unit action potential convert to muscle contraction) of post-fatigue was lower than pre-fatigue significantly at 10% ($P &lt; .01$) and at 20% MVC ($P &lt; .05$). Mean ARVs of EMG of post-fatigue were also higher at 10 and at 20% MVC ($P &lt; .05$).</td>
<td>(2010)</td>
</tr>
</tbody>
</table>

Abbreviation: MVC — maximum voluntary contraction, MUFR — motor unit firing rate, RMS — root mean square amplitude, MF — median frequency, MPF — mean power frequency, ARV — average rectified value, MAV — mean average value, MUAS — motor unit activation strategy, ANN — artificial neural networks, RCP — respiratory compensation point, MAE — mean absolute error, VAR — frequency variance.
insensitive to the body tremor and muscle artifact and able to record MMG signal in physical units to ease the comparison between trials. Thus, theoretical and experimental studies in these areas are clearly warranted. However, the limitations highlighted here could still be further verified, as future experimental validation may prove otherwise.

4.2. Future prospects

The quantitative muscle function assessment used to prescribe the effective rehabilitation program is often found to be marred with confounding data due to inconsistencies and the limited representation capability of muscle function assessment signals. Consequently, maintenance of healthy muscle physiology in able population and individuals with disability remain limited and continue to constitute a wide knowledge gap. The complementary alternative envisioned by MMG modality might be enlightening considering several research efforts on the validation of the procedure of acquisition and analysis of the signal. Currently, MMG is gaining awareness among exercise physiologist in the field of its potential applications. A breakthrough in the reliability validation of MMG sensor technology continues to enhance the relevance of the signal in rehabilitation science as advances were achieved in terms of ease of use and detection accuracy, signal processing techniques, as well as its verification as an alternative mechanical modulating signal to assistive technologies’ control systems especially in areas where electrical artifact is more pronounced. This includes multifunction control signal in prosthesis (Silva et al., 2005) and in muscle machine interfaces (Zeng et al., 2009).

It is evident from the literature that MMG could be applied to discriminate different muscle fiber types and quantify and monitor the effects of muscle training through assessment of muscle power, endurance and strength for application in exercise training and as a biofeedback signal. In order for MMG application to remain relevant, the investigators have continuously demonstrated the reliability and the validity of the technique in different fields of its potential applications.

The mechanomyographic signal being inherently mechanical facilitates the signal acquisition task. There are continuous research efforts on development of detection sensors that will not require skin preparation and contact. These strengths could be well exploited during signal acquisition while the muscle contraction is due to the influence of electrical stimulation and also in situations where electrical noise could contaminate the generated muscle data. Hence, importance of further procedural and evaluation of the signal modality is warranted in order to fully justify its application.

MMG has been particularly shown to be relevant in assessing muscle force when force is difficult to obtain in vivo from a specific muscle group

Abbreviation: RMS — root mean square amplitude, PSD — power spectrum density, PCA — principal component analysis.
or due to the close proximity of other muscles. The signal is equally relevant during electrical stimulation when torque and fatigue development are not easily detectable (Gobbo et al., 2006). Furthermore, the signal has also been shown to reflect the contractile properties of muscle and contribution of individual motor unit to muscle contraction (Yoshitake et al., 2002) thus, the signal application can no longer be considered premature in various areas of its relevances.

5. Conclusions

It is evident that MMG is an indispensable, non-invasive indicator of muscle function of experimental and clinical importance and has been demonstrated to be applicable to all population i.e. pediatrics to geriatrics; irrespective of muscle training level i.e. sedentary or trained athletes, and to all form of muscle activities; static and dynamic muscle contractions. However, it is particularly important for future studies to continue to explore the amplitude and frequency content of the signal especially during dynamic muscle contractions for optimal application. Inference from the literature cited reliably justified that the signal is a viable complementary alternative to the conventional muscle assessment tool, such as EMG and also generate mechanical information about physiological aspect of muscle activities. Evidence has been presented to show that this area of research remains a rich source of further studies with the major aim to improve the potential application of the MMG signal modality in experimental and clinical practices.

Conflict of interest

On behalf of the authors of the manuscript “Mechanomyography and muscle function assessment: A review of current state and prospects” I hereby declare that we do not have any financial or personal relationship with other people or organization that could have inappropriately influenced this review.
Acknowledgments

This study was fully supported by the Ministry of Higher Education, Malaysia and University of Malaya through HIR Grant No. UM.C/625/1/HIR/ MOHE/ENG/39. The authors would like to acknowledge Dr Ivan Djordjevic of University of Malaya for his contribution in editing.

References


