MICRO ELECTRO MECHANICAL SYSTEMS BASED SENSOR FOR MECHANOMYOGRAPHY

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Abstract

The purpose of this work was to design a micro electrical mechanical sensor (MEMS) based system to measure the mechanomyogram of electrically stimulated muscle. The measuring device for the mechanomyography (MMG) system consisted of a dual axis accelerometer and a signal conditioning circuit designed specifically to enhance raw MMG signals . Currently electromyography (EMG) is the standard tool for measuring muscle contraction. During electrical stimulation however, EMG measurements are corrupted with a large stimulus artefact. This obscures any contributions from the much smaller electrical activity of the muscle tissue itself for the first 10ms to 12ms, before disappearing. MMG, being based on kinetic measurements, offers an alternative in such cases. We illustrate this with a dual modality EMG/MMG simultaneous acquisition for a Hoffman-reflex study.

1 Introduction

1.1 Mechanomyography

MMG is a technique for recording and interpreting mechanical activity in contracting muscle. This mechanical activity has a range of a few hertz to approximately 45Hz, [1], [2], [3], [4], [5]. This MMG vibration is produced by lateral oscillations of muscle fibres which occur at the resonant frequency of the muscle [6]. MMG signals can be measured using devices such as accelerometers, microphones, piezoelectric sensors and laser distal sensors. Analysis of MMG signals have allowed examination of various aspects of muscle function such as neuromuscular fatigue [7], neuromuscular disorders in adult and paediatric populations [8], to control external prostheses and measure the effectiveness of anaesthesia [9]. This study demonstrates the usefulness of a MEM's based sensor in the detection of an MMG. MMG signals typically have better signal to noise ratio (SNR) than EMG signals and are higher in magnitude, also noise such as unwanted motion artefacts can be easily detected and removed. MEMS accelerometers used in electrical muscle stimulation experiments are less susceptible to electrical interference. We choose to compare how both systems reacted while eliciting the well known Hoffman response.

1.2 Electromyography

Electromyography (EMG) is a technique used to record and interpret graphic records of the electrical activity of contracting muscles. It is measured either by using surface electrodes on the skin (surface EMG) or by invasive needle electrodes which are inserted directly into the muscle. When a muscle contracts, nerve impulses travel down the motor neurons causing an action potential to sweep down the length of the muscle fibre [10]. When a muscle fibre is artificially stimulated by an electrical impulse the resulting electrical response recording by EMG is initially overwhelmed and an accurate observation cannot be elicited for the first 10 – 12ms, due to the presence of the stimulus artefact, as will be shown later. This difficulty in using the EMG to measure direct muscle response to electrical stimulation is a serious limitation in its application, particularly when considering therapeutic processes such as

functional electrical stimulation used in the clinical environment.

1.3 Hoffmann Reflex

The Hoffman reflex (H-reflex) is an electrically elicited response of a monosynaptic stretch reflex, which provides a non-invasive method of monitoring the integrity and functionality of the central nervous system, particularly information about the monosynaptic pathway [11]**.** The H-reflex bypasses the muscle spindle and, therefore is useful for assessing modulation of monosynaptic reflex activity in the spinal cord [12]. This measurement can be used to assess the response of the nervous system to various neurological conditions, musculoskeletal injuries, and application of therapeutic modalities, pain, exercise training and performance of motor tasks [12].

The electrical stimulus is transmitted by afferent, Ia sensory fibres to synapse on the alpha motor neurons (αMNs) in the spinal cord. This results in an action potential, generated by the αMNs, travelling along efferent fibres until they reach the neuromuscular junction, resulting in a synchronised twitch in the muscle [12]. This contraction can be monitored using EMG, which measures the summation of the muscle fibre action potentials. A typical response is shown in Figure 1a consisting of a stimulus artefact, a component of short latency and a second component of longer latency. The first response (M-response or M-wave) is due to direct activation of the αMNs and has a latency of about 5ms, however the stimulus artefact overlaps the initial phase of the M-wave.

As the stimulus artefact dissipates at around 10ms to 12ms the M-wave becomes apparent. This is a direct motor response elicited at much stronger stimulation levels, and increases as stimulation levels are increased. Figure 1a was measured at 25% of maximum stimulation intensity. The later response, at around 35ms post-stimulus, is elicited at much lower stimulus levels and is the H-reflex. The initial low intensity stimulus set up action potentials that first recruit the larger axon sensory fibres; they travel both orthodromically and antidromically. Those action potentials travelling orthodromically have the greater potential and activate some of the muscle fibres, which are recorded as the H-reflex by electrodes placed on the soleus muscle. As the stimulation intensity increases thinner axions of the αMNs are recruited, travelling directly towards the muscle and recorded as the M-wave. At the same time, action potentials travelling antidromically in the αMNs towards the spinal cord, collide with action potentials of the evoked reflex response thereby resulting in partial cancellation of the reflex response [12].

This eventually results in total cancellation of the H reflex as the antidromic activity is equal to or larger than the afferent activity. The H-reflex appears at much lower stimulation levels, it's magnitude decreasing as the M-wave appears. At mid-levels of intensity both responses are displayed simultaneously, as shown in the recruitment curve in Figure 1b. However as the stimulation level increases the M-wave increases to a maximum (Mmax), a level that plateaus while the H-reflex disappears. The recruitment curve are loci of peak response amplitude versus stimulus intensity and can vary in character as a function of muscle temperature, state of contraction, age etc [13].In this study, the stimulation level was gradually increased and the corresponding EMG responses recorded to produce the recruitment curve shown in Figure 1b. This shows the H-response beginning to appear at 10% stimulation, peaking at around 17% stimulation, which is recorded as Hmax. At around 15% stimulation intensity the M-wave begins to appear and increasing it beyond 17% continues to rise to Mmax at around 40%. Hmax is a measure of maximal reflex activation or alternatively an estimate of the number of αMNs one is capable of activating in a given state. Mmax represents activation of an entire αMN pool and therefore maximum muscle activation [12] . The H-reflex was used for this study because it has been comprehensively studied using EMG and it offered a reliable and repeatable source of electrically elicited muscle and nerve responses.

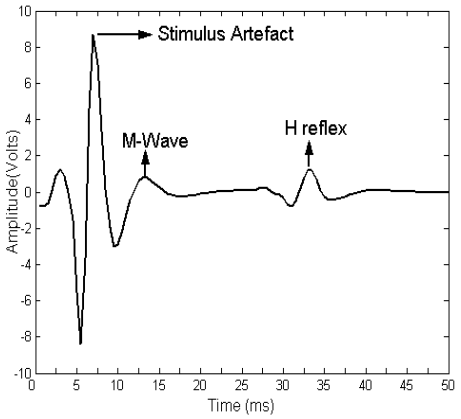

Fig 1.a) EMG signal, with stimulus artefact

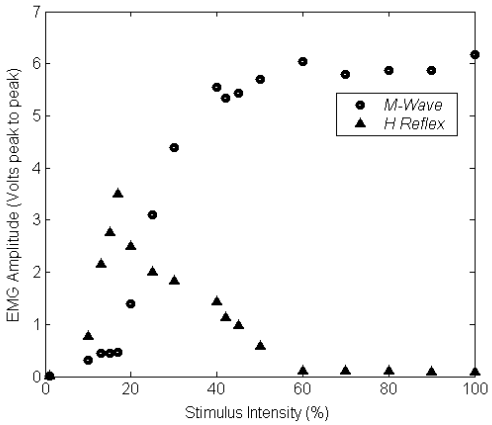

Fig1:b) EMG recruitment curve

2 Methodology

This section describes how a MEMS accelerometer can be configured to measure MMG signals. The advantages of accelerometer use in MMG studies is discussed. The need for a MMG signal conditioning circuit and its theory of operation are explained. The experimental setup and procedure for a simultaneous EMG and MMG H-reflex experiment are explained also.

2.1 MEMS Sensor

MEMS based accelerometer sensors have advantages over EMG electrodes as they can be placed directly onto the muscle without any skin preparation. To take an EMG measurement requires three electrode areas to be prepared. Accelerometers are widely used as the sensor of choice in MMG systems [14], [15] . The sensor required a resolution capable of measuring muscle vibrations/sounds and needed to be physically small enough to be placed with ease on the soleus (calf) muscle. The typical noise floor is 110 μg/√Hz where g is gravity, allowing signal resolution below 1 mg (0.06° of inclination) to be resolved at low bandwidths (<60Hz). Lowering the bandwidth reduces noise and increases the signal to noise ratio of the accelerometer. The output bandwidth for each axis was set to 70Hz. Since no information was available on H-reflex experiments measured by MMG it was decided to allow for signals above 45Hz and below 70Hz in order to avoid filtering out possible MMG information.
At a bandwidth (BW) of 70 Hz, the noise present in a signal is

$$rmsNoise = (110\mu g / \sqrt{Hz})\, x \sqrt{(BWx1.6)} = 1.16mg$$

2.2 Signal Conditioning Circuit

The aim of the signal conditioning circuit was to provide specific amplification and filtering for MMG signals. The accelerometer outputs a static signal (gravity) as direct current (DC) and a dynamic signal (vibration) as alternating current (AC). The static DC signal is of no use and is filtered out using a high pass filter (0.323Hz cut-off) leaving only vibration signals. MMG signals measured by the accelerometer are too low in amplitude initially (typically 100mVpp) and they contain Gaussian white noise which is stochastic in nature and present across all frequencies. Therefore a signal conditioning circuit, shown in Figure 3, is required to amplify and further filter the MMG signals.

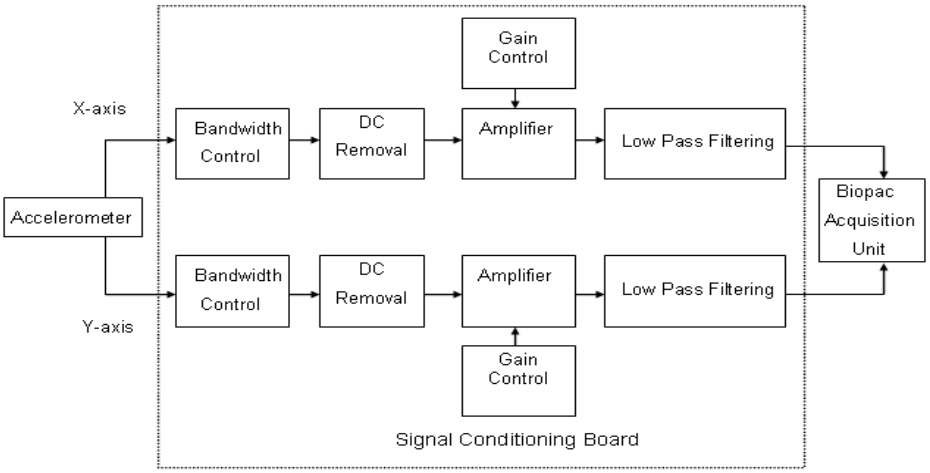

Fig 3. Signal Conditioning circuit block diagram.

Figure 4 (A) shows the response of the MMG sensor to a vibration signal with a broad frequency content. The other three Figure 4 (B,C,D) demonstrate the functions being performed by the signal conditioning circuit. The bandwidth was set using capacitors on both axes along with an internal resistor to create a 70Hz low pass filter , which effectively band limits the accelerometer to measure signals 70Hz or less. This is effective in reducing unwanted elements or noise as MMG signals of interest do not occur beyond 70Hz. When the accelerometer measures a signal it is passed on to the conditioning circuit where the DC component is removed, this is shown in Figure 4 (B). The signal is then amplified, shown in Figure 4(C) . The gain can be easily adjusted using a digital potentiometer according to how low the signals amplitude is. Once amplified the signal was further filtered in order to create the cleanest signal possible. A 5 pole Butterworth low pass filter was used because it provides maximally flat magnitude response in the pass-band. Butterworth filters have a better pulse response than Chebyshev filters and a better rate of attenuation than Bessel filters which makes them more suitable for MMG applications. A filtered signal is shown in Figure 4 (D). Figure 5 shows the filters frequency response. Once filtered the signals were acquired by a Biopac data acquisition unit for further processing and analysis.

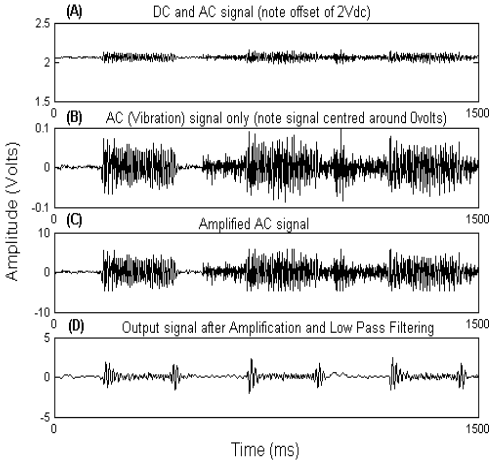

Fig 4. Processing a vibration signal (125Hz)

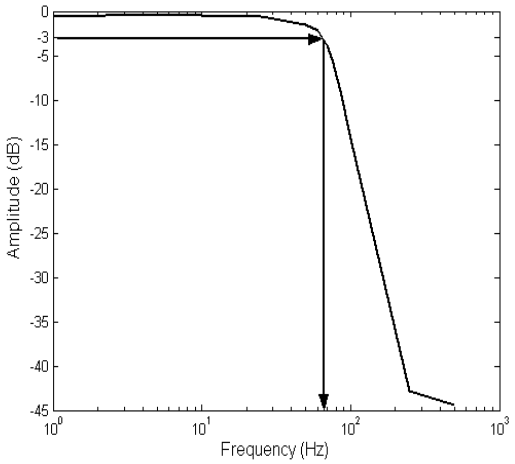

Fig 5. Frequency Response of circuit

2.3 Experimental Setup

A simultaneous EMG and MMG Hoffmann experiment was performed. To measure a Hoffmann response the subject was instructed to lie face down on a bed with their right leg

slightly elevated. The subject had to keep the leg as still as possible for the duration of the experiment. Electrical stimulation was applied to the poplitial fossa area at the rear of the knee just over the tibial nerve. The EMG sensors were placed at the top and bottom of the belly of muscle and on the ankle. The MMG sensor was placed in between the two EMG electrodes over the centre of the muscle, Figure 6. Three datasets were recorded for each level of stimulation.

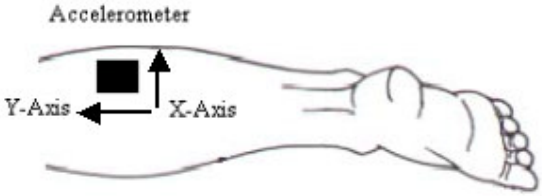

Fig 6. Shows the direction in which the sensor measures vibrations at the surface of the skin over the muscle region.

The MMG signal was collected using Analog Devices ADXL203 dual axis accelerometer. The accelerometer was soldered to printed circuit boards (PCB), which was as thin and light as possible to maintain accuracy and reduce vibration damping. The accelerometer was placed over the soleus muscle between the positive and negative EMG electrodes; to ensure both the MMG and EMG picked up signals from the same muscle contraction regions. The accelerometer was adhered to the skin using double-sided adhesive tape. The x-axis was placed perpendicular to the length of the muscle in order to measure lateral muscle contractions. The system was designed to be as light as possible as over 5 grams of mass can substantially attenuate the MMG signal [17]. The overall mass of the accelerometer and PCB is less than 2 grams.

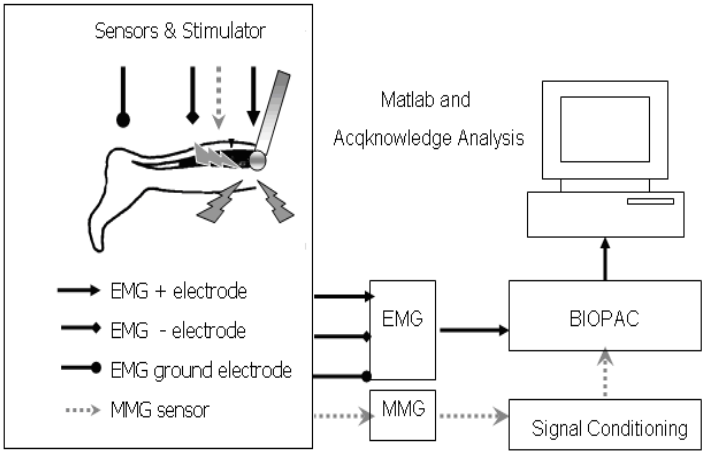

Fig 7. Experimental Setup showing how the Hoffmann experiment was performed

A square wave signal with a pulse width of 500usec was used to stimulate the tibial nerve. Simultaneous EMG and MMG readings were taken over this range of stimulation. As shown in Figure 7, a Biopac was used to read in the EMG and conditioned MMG signals. The MMG and EMG signals were read in on different channels at 2000 samples per second. The Biopac high pass filter and low pass filter were set to 1Hz and 5000Hz respectively. It was decided to measure signals for 150msec as all physiological events of interest occurred during this timeframe. The trigger event occurred when the negative edge of the EMG signal was over -0.08volts. Once this trigger event occurred the EMG and MMG signals were recorded

simultaneously and displayed in real time.

3 Results

This section shows how MMG is not susceptible to stimulus artefacts. We see how EMG and MMG both have the same delay between their respective M and H responses which suggests they share the same origins. It is also shown that the sensors y-axis is capable of measuring important information that supports the sensors x-axis information.

3.1 Stimulus Artefact

Figure 8 shows a simultaneous EMG and MMG H-reflex measurement. The EMG signals clearly show a stimulus artefact. This overlaps and obscures useful information at the beginning of the trace. It is clear from Figure 8 that there is no indication of a mechanical analogue to the electrical stimulus artefact on the mechanomyograph shown. Other studies have shown that accelerometers are capable of measuring mechanical activity with no electrical interference during evoked electrical stimulation [16].

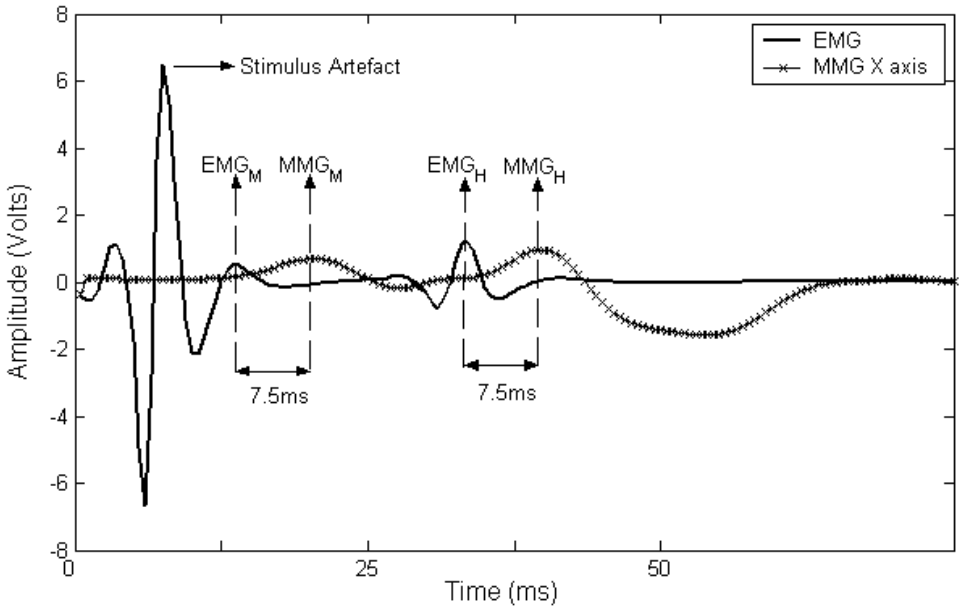

Fig 8: Simultaneous EMG and MMG measurement showing no mechanical artefact.

3.2 MMG Hoffman Response

EMG signals originate from motor unit action potentials propagating along motor units. This creates a mechanical response that follows this action potential. Investigating the x-axis MMG response shows that the mechanical response is not instantaneous, it has to build up as the muscle fibre depolarises. This explains the latency between the EMG and MMG responses [18]. Thus, the electrical phenomena relating to muscle contraction appears to be closely related to the mechanical phenomena. A close comparison of the EMG and MMG waveforms indicates that the distinctive M and H responses are clearly present, this is shown in Figure 8.

The presence of M and H responses in the MMG data is inferred from the fact that the delay measured between two distinct features on the MMG response is the same as those responses on the EMG trace. Also there is a high correlation (94.5%) between the EMG M-

wave amplitude behaviour over the range of stimulation and MMG M response amplitude behaviour . This would be expected as they both are measured from the same region of muscle and occur due to the same phenomena. At 30% to 40% stimulation the MMG M Response reached its maximum showing that the muscle had reached its maximum contraction level by direct electrical stimulation. Also the EMG trace reaches a plateau around 30% to 40% stimulation intensity further showing similarities in the generation of these signals. It is still not fully understood where exactly the MMG signal comes from but it can be seen that the EMG M-wave and MMG M-response share the same origins.

3.3 MMG Y axis information

Information taken from the MMG sensor in the y-axis is shown in Figures 9,10 and 11. The y-axis appears less sensitive to muscle vibration than the x-axis. The y-axis was more sensitive to limb movements due to the ankle and calf moving upward and forward as a result of the H-reflex, causing vibrations to travel in the y direction. Whereas the x-axis measured the lateral oscillations of the soleus muscle and was less sensitive to limb movements.

However the y-axis did still measure important information such as latencies in the onset of muscle contractions based on the y-axis distance away from the muscle being directly stimulated. This is shown in Figure 9.

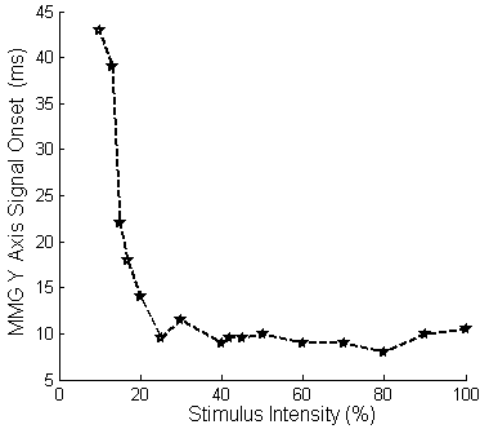

Fig 9.Y-axis latency

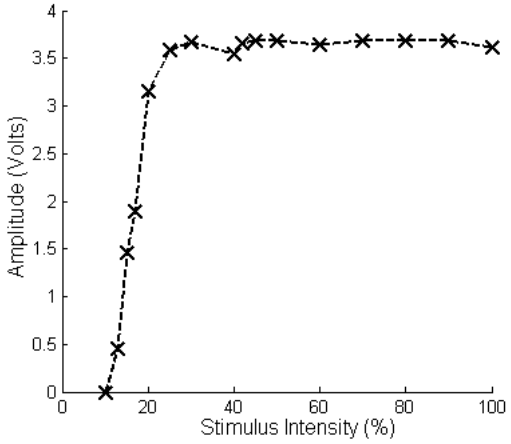

Fig 10. Y-axis Amplitudes

It can be deduced from Figure 9 that at lower stimulation intensities the mechanical response from the part of the muscle being directly stimulated takes longer to traverse the underlying tissue. This could be because the muscle being stimulated at low intensities lies further away from the MMG sensor, hence the vibration takes longer to affect the sensor. As the stimulus is increased more of the muscle is directly stimulated (M-wave) that lies closer to the sensor and so appears earlier in the trace. There is high degree of correlation between the EMG M-wave amplitude behaviour and the MMG y-axis amplitude behaviour.

Another explanation could be that as stimulus intensities rise the responses are stronger and hence traverse the underlying tissue faster. Once 25% stimulation intensity has been reached almost all of the muscle is contracting (The EMG M max occurs at 30% stimulation) so the latency and amplitude stabilises and begins to plateau. The y-axis response amplitudes also plateau as maximum stimulation is reached around 25%, which is shown in Figure 10.

Figure 11 shows a simultaneous EMG and MMG measurement. The y-axis measures a large positive acceleration as a result of the M response at 8.5ms. The H response overlaps with the M response on the y-axis. This can be observed by the fact the y-axis measures a prolonged positive acceleration for the duration of the M and H responses from 8.5ms to 30ms. After the H-response has peaked both axis amplitudes immediately start to decrease

which suggests they measured the same vibrations.

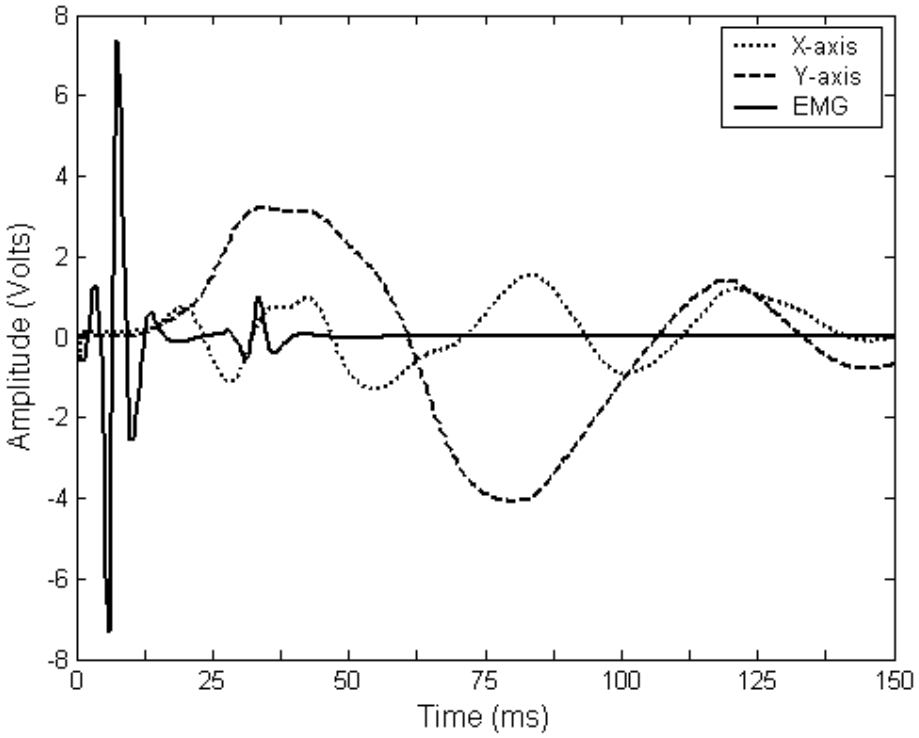

Fig11. Y–axis response to the mechanical M and H response's at 25% stimulation

4 Conclusion

This work demonstrated the use of an MEMs accelerometer in the acquisition of an MMG signal. Work was completed to extract the muscle response from the accelerometer signal. The MMG response to a Hoffman experiment has been presented. It is clear that our aim to design and build a MEMs based MMG sensor system has been achieved as demonstrated.
During the course of the study, data collected from the MMG sensor is shown to be devoid of artefact caused by the electrical stimulation. By repeating this Hoffman experiment a number of times it has been demonstrated that the MEMs based MMG is an accurate and easier way to measure muscle vibration activity related to electrically stimulated muscle contraction. Furthermore it was shown that valuable information can be measured on both axes.

The latencies of the MMG analogs of the H and M responses were highly consistent with their electrical counterparts suggesting that the responses share a similar origin. There is evidence that the patterns contain recruitment characteristics which are a prerequisite for further clinical analysis.

In conclusion it was found that the MEMS-based dual axis accelerometer system was capable of useful physiological measurement, at least in the context of H-reflex studies and should provide a complementary signal source for more in depth analysis of neurological function at the spinal level, when used in conjunction with EMG.

4.1 Future Work

The next phase of this study is to investigate simultaneous EMG and MMG H-reflex studies using tri-axial accelerometers. This will allow for a more in depth analysis of the neuromuscular responses to the electrically stimulated H-reflex. Also muscle response to

functional electrical stimulation (such as off the shelf muscle toning devices) will be investigated using the same MMG system as above to examine MMG's capability to measure and monitor electrically stimulated muscle activity.

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