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sEMG Analysis of Muscle Fatigue During Prolonged Driving

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ABSTRACT

Prolonged driving is associated with spinal muscle fatigue, a significant contributor to back pain. This study investigates low back muscle (Erector Spinae, Longissimus) fatigue using surface electromyography (sEMG) while driving at backrest inclination angles of 90, 100, and 110 degrees. EMG signals were acquired in actual driving experiments. The signals are filtered and analyzed by calculating mean and median frequency (MNF and MDF) with a 1-second sliding window and 0.5-second increments. Five male healthy subjects participated in the actual driving trials. Each subject completed three trials, resulting in a total of 15 trials. Results indicated a decrease in MDF at 90 and 110 degrees, and a decrease in MNF at 100 degrees. Specifically, the average MDF decreased by an average of 0.36 Hz per hour at 90 degrees and by 0.07 Hz per hour at 110 degrees, while MNF decreased at an average rate of 1.4 Hz per hour at 100 degrees. These findings suggest that muscle fatigue varies significantly with the backrest angle. And further research may shine more light onto the lower back muscle groups response with time.

Keywords: *sEMG; Prolonged driving; Muscle Fatigue*

1.0 INTRODUCTION

Muscle fatigue, defined as an “exercise-induced decrease in the ability to produce force” [1], is a common experience during prolonged driving. Drivers often report pain in the lower back and neck after 1–3 hours, depending on individual routines. The lower back muscle fatigue, specifically within the lumbar spine (L1-L5) and the Erector Spinae muscle group, which are critical for spinal extension and stabilization [2, 3]. Prolonged seated postures can increase strain on these muscles.

Fatigue while driving has been studied from varying perspectives and scenarios. Some studies investigated in actual driving [4-6] and others in simulated scenarios [7-9], and sEMG was implemented to detect and identify muscle fatigue in different exercises including driving.

In [4], Boon-Leng et al. used a wearable device with galvanic skin response (GSR) device and EMG sensors to detect muscle fatigue. They found that specific EMG sensor placements near the palm and fingers, combined with a Support Vector Machine (SVM) classifier, achieved a fatigue detection accuracy of 92%, suggesting that sensor placement significantly impacts the detection of fatigue. While focusing on shoulder muscle fatigue, [5] six machine learning models were used for the classification: Logistic Regression, SVM, Naïve Bayes, k-Nearest Neighbours (kNN), Decision Tree and Random Forest. Both the MDF and MNF are lower when fatigue occurs. The accuracy levels of the classification algorithm were 85.00%, 83.75% and 81.25% for Random Forest, Decision Tree, and kNN respectively. One of the earlier studies, [6] used MNF, MDF, and root mean square (RMS) to identify fatigue and found that MDF decreased 9.5%-18.9% and MNF decreased 11.3%-18.4% from their initial values, and RMS increased from 25.1%-47.7%.

The study in [7] introduces a novel supervised classification method for recognizing three levels of driver lumbar muscle fatigue during extended driving using multi-channel sEMG. Features are derived from Wavelet Packet Transform (WPT) and Continuous Wavelet Transform (CWT) analysis of sEMG signals. The feature space incorporates Shannon entropy, relative energy, and statistical features extracted from instantaneous MDF, MNF, and energy (IE), and RMS values. Employing a Radial Basis Function Support Vector Classifier (RBF-SVC), the average classification accuracy across ten successive tests reached approximately 82.69%, with all areas under the curve (AUC) values exceeding 0.9. [8] explored the effect of using a neck balance system and lumbar support during driving. With twelve healthy male subjects, the RMS of EMG from the erector spinae, semispinalis capitis, and sternocleidomastoid. Using the said devices, it was found to effectively reduce muscle activation in both the neck (during acceleration only) and lower back while driving. This reduction is likely the result of providing a neutral position for the lower back and rebalancing the weight on the neck. Similarly, in [9], sEMG signals of the neck and shoulder muscles while driving on a game simulator was analyzed. The subjects were two sets: professional and non-professional drivers. The 15 minutes duration was enough to cause significant change in the mean power of the fifth level wavelet coefficients of 1st and 15th minutes of all subjects for the left deltoid, bilateral trapezius and splenius capitis muscle groups.

Despite existing research using sEMG to study muscle fatigue during driving [4-9], there is a lack of comprehensive analysis regarding the influence of backrest inclination angles on specific low back muscles. Seat inclination angle directly affects the load distribution on these muscles [10]. This study investigates the changes in MDF and MNF while subjects drive at three different inclination angles (90, 100, and 110 degrees) for three hours. This study aims to address this gap by evaluating how different backrest angles impact muscle fatigue as measured by changes in MDF and MNF.

2.0 METHODOLOGY

Five male healthy subjects (age 26.4 ± 3.93 years, weight 70.2 ± 12.73kg, and height 169.8 ± 5.0cm) volunteered in this study in which they performed actual driving trials with different back seat inclination angles (90, 100 and 110 degrees). This study was approved by International Islamic University Malaysia (IIUM) Research Committee (IREC).

2.1 Experimental Setup

The trials were carried out on a 73.8 KM lap around a section of the Middle Ring Road 2 (MRR2) highway as shown in Figure 1.

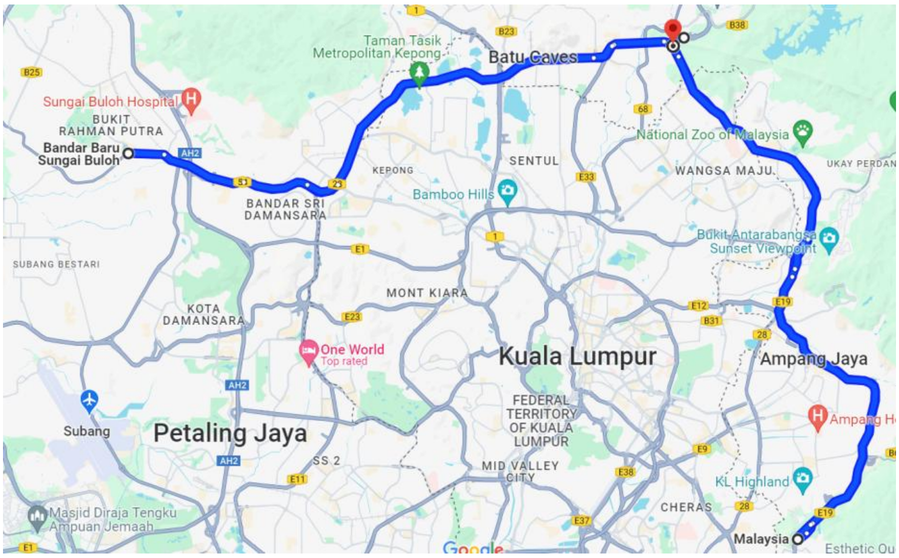

Figure 1: Map of route through MRR2 highway

Starting time was set across all trials at 10:00AM to avoid any circadian cycle effects. The total duration was at least 3 hours for each trial. First, the seat backrest angle was set to 90 degrees. The angle was measured using an inertial measurement unit (IMU) sensor module, MPU6050. The EMG sensor used was Bitalino EMG sensor with Assembled BITalino Core BT/BLE board. The electrodes used were pre-gelled 24 mm Polymer Ag/AgCl with a 20 mm electrode distance. The electrodes were placed vertically oriented at the longissimus subgroup of the erector spinae muscle group. As per SENIAM recommendation, the location is two finger width lateral from the spinous process of L1 as shown in Figure 2 below. The EMG signal was recorded at 1 k·Hz sampling rate.

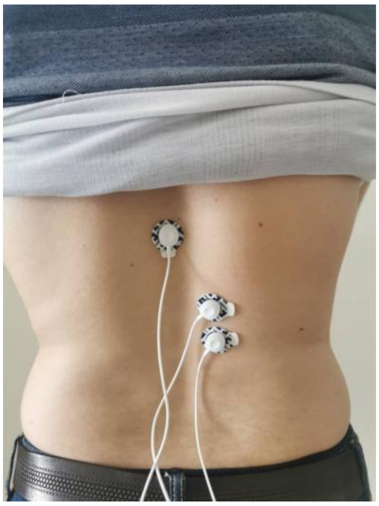
Figure 2: Electrode placement (as per SENIAM recommendation)

2.1 Signal Processing

The EMG signal was analyzed using a 1 s sliding window instead of 30 s that was used in [11] where no clear pattern was observed. The 1 s size was used as recommended by [12] approximately 500 m·s windows for activated muscle and 1 s for low muscle activation exercises.

The signal was first standardized using the Equation (1):

$$x_{standardised} = (x_i - \mu)/\sigma \quad (1)$$

Where x_i is the i^{th} sample value, μ is the mean and σ is the standard deviation of the whole EMG signal. This standardization process is applied to normalize the EMG signals across different subjects and trials, accounting for inter-subject variations in baseline muscle activation.

Windows where the RMS was lower than the mean RMS were not used as such windows represent periods of even lower muscle activation or inactivity. These segments were excluded from further analysis to minimize the influence of baseline noise on the MDF and MNF calculations. This thresholding approach helps to focus the analysis on periods where the muscle was actively engaged, improving the reliability of the fatigue assessment. In each of the selected windows, the signal was filtered using a bandpass filter with cut-off frequencies of 30 Hz and 450 Hz [13] to isolate the relevant EMG signal components and remove low-frequency artifacts (e.g., motion artifacts) and high-frequency noise. With respect to the noise due to power line interference (PLI) observed at harmonics of 50Hz in Malaysia, a series of notch filters centered at 50 Hz, 100 Hz, and 150 Hz were applied. Then, the MNF and MDF were calculated for each window based on the Equation (2) and Equation (3) respectively:

$$MNF = \sum_{j=1}^{M} f_j P_j \, / \sum_{j=1}^{M} P_j \quad (2)$$

$$MDF = \sum_{j=1}^{MDF} P_j = \sum_{j=MDF}^{M} P_j = \frac{1}{2}\sum_{j=1}^{M} P_j \quad (3)$$

where f_j and P_j are the frequency and power values of EMG power spectrum at frequency j, respectively, and M is the length of frequency array. The best-fit lines were drawn for MNF and MDF to compare among trials. Slopes of the best lines were indicative of the rate at which the feature was changing.

3.0 RESULTS AND DISCUSSION

The rate of change of MDF for each trial and the average rate of change of MDF across the 5 subjects are presented in Figure 3. At a back seat inclination angle of 90 degrees, the MDF shows the greatest decrease in MDF with only subject 5, S5 showing an increase. For 100 degrees inclination angle, the MDF increases except for subject 1, S1 who shows a decrease. For 110 degrees inclination angle, the MDF decreases except for subject 3, S3. Both 90 degrees and 110 degrees angles show predominantly decreasing trends of MDF while at 100 angles the MDF tends to increase.

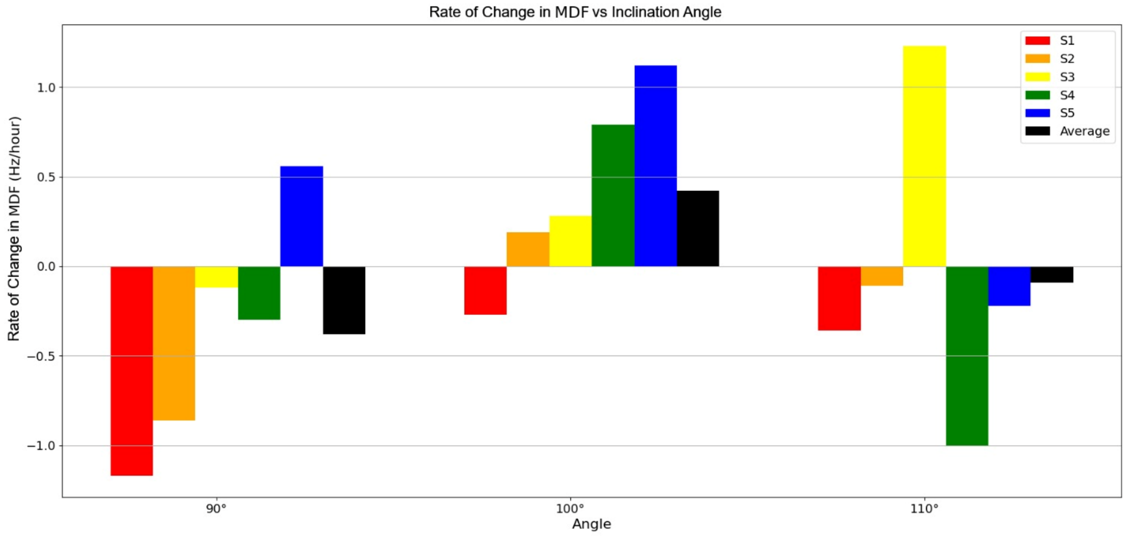

Figure 3: Rate of change in MDF for trials at each inclination angle

The rate of change of MNF for each trial and the average rate of change of MNF across the 5 subjects are shown in Figure 4. At a back seat inclination angle of 90 degrees, the MNF shows the greatest increase in MNF with only subject 5, S5 exhibiting a decrease. For 100 degrees inclination angle, the MNF decreases except for subject 1, S1 who shows an increase. There is no trend of MNF at the 110-degree inclination angle. Overall, the 100-degree angle shows a predominantly decreasing trend of MNF.

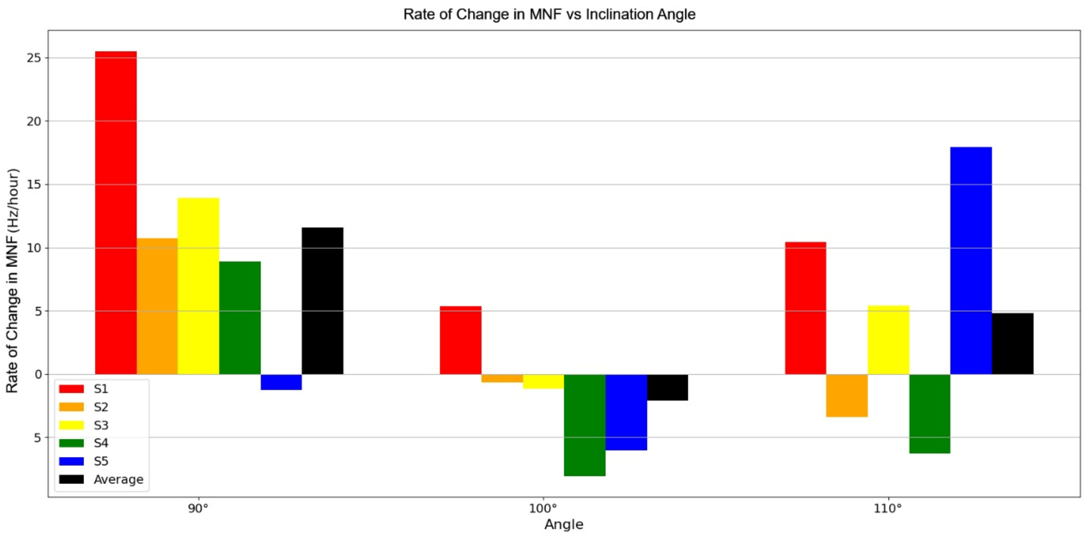

Figure 4: Rate of change in MNF for trials in each inclination angle

Based on the results in Figures 3 and 4, the 90-degree and 110-degree trends are mixed. Fatigue is more consistently observed at the 100-degree inclination angle where the rate of MDF value increases while the rate of MNF value decreases. Only one subject out of five showed a consistent trend for both MDF and MNF across the three angles which is Subject 1, S1 who showed decreasing MDF and increasing MNF in all trials. However, this pattern does not indicate fatigue.

Results indicated a decrease in the average rate of change of MDF at 90 and 110 degrees, and a decrease in the average rate of change of MNF at 100 degrees. Specifically, the average MDF decreased by an average of 0.36 Hz per hour at 90 degrees and by 0.07 Hz per hour at 110 degrees, while MNF decreased by an average rate of 1.4Hz per hour at 100 degrees.

The observed changes in MDF and MNF provide insights into the underlying physiological processes associated with muscle fatigue. MDF represents the frequency at which the full power spectrum of the EMG signal is divided into two equal halves. A decrease in MDF is generally interpreted as an indicator of muscle fatigue, as it reflects a shift towards lower frequencies in the EMG signal due to the reduced conduction velocity of muscle fibers [13]. MNF, on the other hand, represents the average frequency of the power spectrum. While a decrease in MNF is also often associated with muscle fatigue, it can be influenced by various factors, including changes in motor unit recruitment patterns and signal noise.

The results indicate that there are complex shifts in the spectral content of the EMG signals which may be influenced by factors other than muscle fatigue. Among such factors are fiber recruitment patterns and signal noise. In this study, a high pass cut off frequency of 30 Hz was used to reduce noise due to electrocardiography (ECG) contamination. Also, PLI was filtered using a notch filter at harmonics of 50 Hz which could also affect the EMG signal information. In future work, techniques for reducing the ECG and PLI effects will be studied and implemented to obtain a less noisy EMG signal. Overall, the results show that the erector spinae do not show any signs of fatigue during prolonged driving.

4.0 CONCLUSION

The contradictory results between MDF and MNF in this study suggest that the relationship between these parameters and muscle fatigue during prolonged driving is complex and may be influenced by individual differences in muscle activation strategies as well as the presence of confounding factors such as ECG contamination and power line interference.

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CONFLICT OF INTEREST

Competing interests: No relevant disclosures.

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