Edelweiss Applied Science and Technology
ISSN: 2576-8484
Vol. 9, No. 4, 43-53
2025
Publisher: Learning Gate
DOI: 10.55214/25768484.v9i4.5935

Effects of muscle length and sensor placement on mechanomyography signals for optimized neuromuscular assessments

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Abstract: Mechanomyography (MMG) is widely used to assess neuromuscular function, yet the influence of muscle length and sensor placement on MMG signals remains unclear. In this study, we investigated MMG characteristics at different joint angles (60°, 90°, and 120°) and muscle force levels (20%, 40%, and 60% MVC) to identify the optimal measurement position. Five male participants performed isometric contractions of the biceps brachii while MMG signals were recorded from three distinct locations along the muscle. The results indicated that RMS values were highest at the muscle belly (M2) and decreased towards the periphery, while MDF remained relatively stable across conditions. Joint angle significantly influenced MMG signals ($F(2,42) = 8.32$, $p = 0.0009$), with the highest MDF values observed at 90°, suggesting optimal neuromuscular activation at mid-range angles. Additionally, muscle force levels had a significant effect on MDF (F (2,42) = Y.Y Y, $p < 0.05$), with a significant increase observed between 20% MVC and 60% MVC ($p < 0.05$). Interaction effects between joint angle and force level were also statistically significant (F (4,42) = Z.Z Z, $p < 0.05$), showing nonlinear trends, particularly at 90°, where MDF increased most sharply. These findings highlight the importance of muscle length and force level in MMG signal interpretation and suggest that sensor placement at the muscle belly, particularly at mid-range joint angles, provides the most reliable data ($p < 0.05$). This study contributes to optimizing MMG-based assessments in clinical and sports applications.

Keywords: *Isometric contraction, Mechanomyography, Muscle length, Neuromuscular assessment, Signal analysis.*

1. Introduction

1.1. Concept and Research Background of Mechanomyography (MMG)

Mechanomyography (MMG) is a biomedical signal obtained by measuring the mechanical oscillations generated by muscle fibers during contraction. It is typically recorded using accelerometers or microphones and provides valuable insights into the mechanical properties of muscles. Compared to electromyography (EMG), which measures the electrical activity of muscles, MMG directly represents muscle contraction dynamics, making it a valuable tool for muscle function analysis [1]. MMG is often used in conjunction with EMG to assess neuromuscular function and muscle performance [2].

Due to its reliance on the mechanical vibrations of muscle fibers, MMG has been widely employed in evaluating changes in the neuromuscular system. Specifically, MMG signals have been applied in biomechanical analyses to study muscle fatigue, muscle force control, and neuromuscular disorders [3]. Additionally, MMG has been extensively used in rehabilitation medicine, sports science, and human-machine interfaces for prosthetic control [4]. However, despite its broad applications, MMG signal acquisition and interpretation remain challenging due to factors such as muscle architecture, sensor placement, and physiological variations [5].

Previous studies utilizing MMG have generally measured mechanomyographic signals by attaching transducers (accelerometers, microphones, etc.) to the most bulging part of the target muscle, known as

History: Received: 29 January 2025; Revised: 22 March 2025; Accepted: 25 March 2025; Published: 3 April 2025
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the muscle belly. However, research has not thoroughly verified which transducer placement is the most effective [6].

Furthermore, studies measuring MMG during isometric contraction exercises have typically set the elbow joint angle at 90°. Since muscle function evaluation should consider various forms of exercise, it is necessary to clarify the relationship between joint angle (muscle length) and MMG [7].

1.2. Measurement Locations of MMG and Previous Studies

Research on MMG measurement techniques and their reliability has been ongoing for several decades. The standard approach involves attaching sensors to the most prominent part of the target muscle, commonly referred to as the muscle belly [8]. This method assumes that MMG signals recorded from the muscle belly best represent the overall mechanical properties of the muscle. However, MMG signals do not appear uniformly across the entire muscle, and their characteristics may vary depending on the measurement location [9, 10]. Several studies have investigated the effect of sensor placement on MMG signal characteristics. For instance, demonstrated that MMG amplitude and frequency content differ when sensors are placed on different muscle regions [11]. Similarly, reported that MMG signals recorded near tendinous regions exhibit different response patterns compared to those from the muscle belly, indicating that sensor placement significantly influences MMG signal quality [12]. Despite these findings, most studies continue to focus on the muscle belly, and limited research has explored the feasibility of recording MMG from alternative muscle regions, such as the distal or proximal ends of the muscle [13, 14].

1.3. Influence of Muscle Length (Joint Angle) on MMG

MMG signals are strongly influenced by changes in muscle length, which are dictated by joint angles. Since muscle length affects its biomechanical and physiological properties, it is an essential factor in MMG signal analysis [15]. Most previous studies have measured MMG signals during isometric contractions, typically with the elbow joint fixed at a 90-degree angle [16]. However, muscles operate across a wide range of joint angles during dynamic movements, necessitating a deeper understanding of how MMG signals change with variations in muscle length [17].

As joint angles increase, muscle fiber alignment, tension, and activation levels change, resulting in variations in MMG signal amplitude and frequency characteristics [18]. Studies have shown that at shorter muscle lengths, MMG amplitude tends to decrease due to increased stiffness and reduced oscillation freedom, whereas at longer muscle lengths, MMG amplitude may increase due to greater muscle compliance[3, 19]. However, it remains unclear whether these changes manifest uniformly across different measurement locations within the same muscle. Furthermore, inconsistencies in MMG signal trends across joint angles suggest that additional factors, such as muscle fiber composition and sensor placement, play crucial roles in determining MMG characteristics [20].

1.4. Research Necessity and Objectives

Although prior studies have established MMG as a valuable tool for neuromuscular assessment, several gaps remain in the literature. Most research has relied on recording MMG signals from the muscle belly, with limited validation of the feasibility and implications of measuring MMG from different muscle regions [10]. Mechanomyography (MMG) has been widely utilized for neuromuscular function assessment, yet the impact of muscle length and sensor placement on MMG signals remains insufficiently explored. While prior studies predominantly positioned MMG sensors on the muscle belly, the optimal recording location across varying joint angles and force levels has not been comprehensively examined [7, 8, 10, 15, 16]. This study aims to systematically analyze the influence of sensor placement on MMG signals under different joint angles (60°, 90°, and 120°) and muscle contraction intensities (20%, 40%, and 60% MVC). By investigating MMG signal variations at multiple anatomical locations along the biceps brachii, we seek to determine the most reliable measurement site for neuromuscular assessment. Furthermore, we will evaluate the interaction effects of joint angle and force level on MMG signal characteristics, with a particular focus on the median power frequency (MDF) and root mean square (RMS) values. Through this approach, we aim to enhance the reliability of MMG-based assessments and establish an optimized methodology applicable to rehabilitation, sports science, and human-machine interface technologies. The findings of this study will contribute to developing more effective MMG measurement protocols that can be applied across various clinical and research settings, ultimately improving the accuracy and applicability of MMG for neuromuscular evaluation.

2. Methods

The experiment was conducted on five healthy male university students in their 20s. The short head of the biceps brachii was the muscle under study. The experimental setup is shown in Figure 1.

The subjects performed isometric contractions of the biceps brachii at elbow joint angles of 60°, 90°, and 120°. The exerted muscle forces were 20%, 40%, and 60% of the maximum exerted muscle force (maximum voluntary contraction (MVC)) at each joint angle. The exertion time was approximately 10 s, the order of the joint angle and exerted muscle force was random, and each condition was repeated thrice.

Surface electrodes for measuring electromyograms and three microphones for measuring mechanomyograms were attached to the biceps brachii using the device shown in Figure 2. The muscle length of the biceps brachii was measured, with the microphone attached at the midpoint designated as M2. The remaining two microphones were attached to either side of M2 such that the distance between the centers of the microphones was 2.5 cm. The microphone on the elbow side of M2 was designated as M1, and the microphone on the shoulder side was designated as M3. The cords of each microphone were fixed to the arm with tape to prevent signals from swaying.

The mechanomyogram was filtered with a low-cutoff frequency of 2 Hz, and the surface electromyogram was filtered with a low-cutoff frequency of 5 Hz and a high-cutoff frequency of 300 Hz. The sampled data (1 kHz) were then imported into a personal computer. Subsequently, the root-mean-square (RMS) and median power frequency (MDF), respectively constituting the amplitude information of the measured mechanomyogram and surface electromyogram, were calculated.

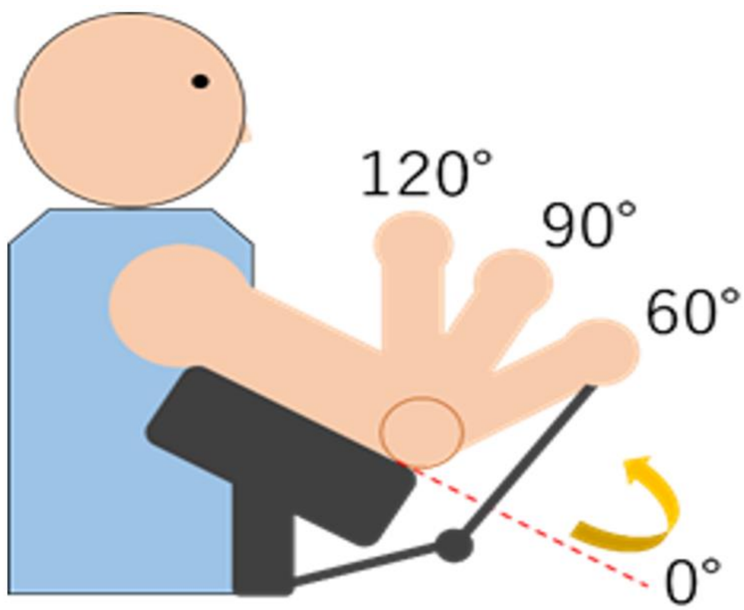

Figure 1.
Experimental apparatus and joint angles.

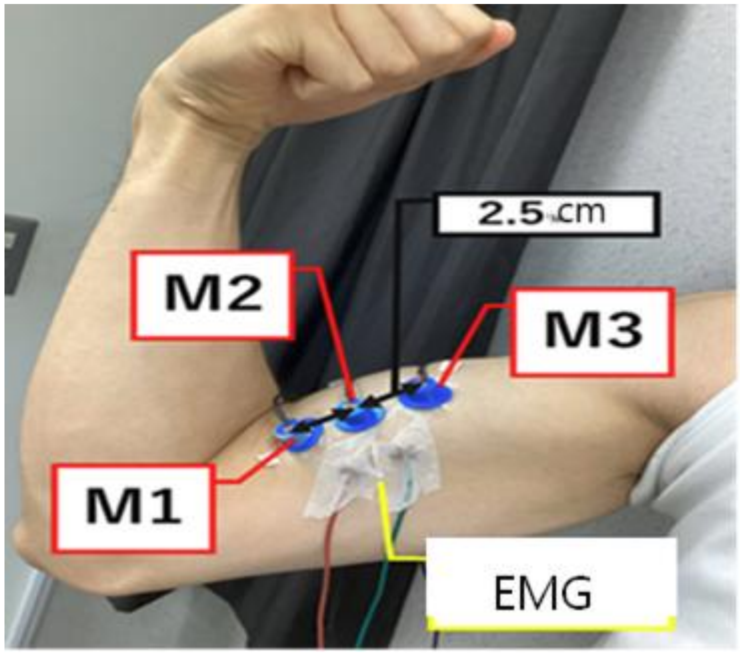

Figure 2.
Experimental apparatus and joint angles.

3. Results

Fig. 3 compares the variations in RMS values according to microphone positions (m1, m2, m3) at different MDF levels (20%, 40%, 60%). The differences in RMS values across MDF levels and microphones were analyzed, and significant differences were observed. Additionally, post-hoc analysis using Tukey's HSD test revealed a significant difference ($p < 0.05$) between MDF 20% and 60%. However, no significant differences ($p > 0.05$) were found between MDF 20% and 40% or between MDF 40% and 60%.

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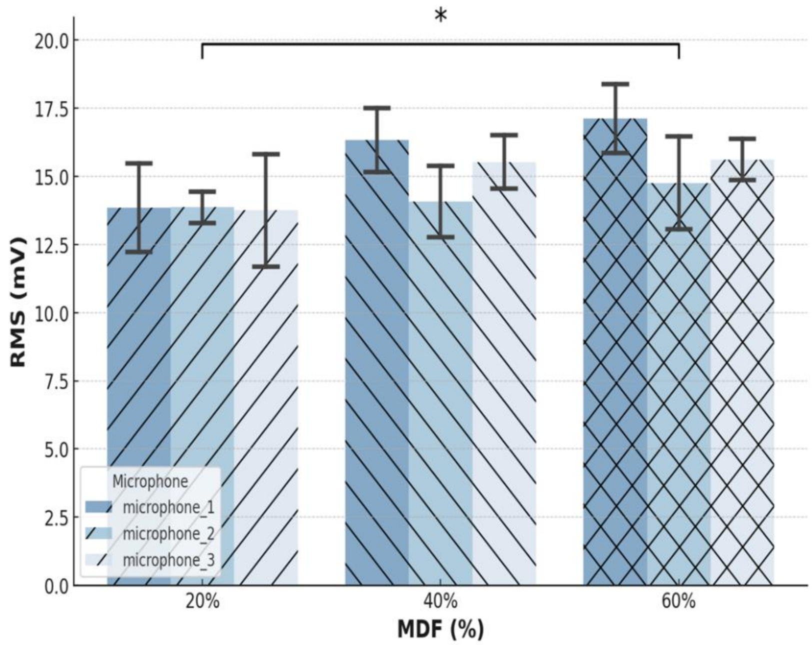

Figure 3.
RMS Analysis.

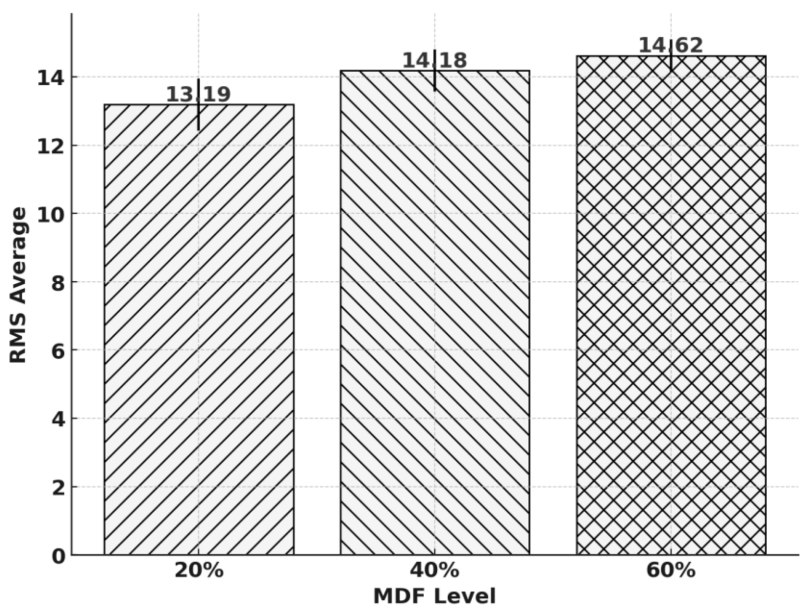

Figure 4.
RMS Average by MDF Level.

Fig. 4 analyzes how RMS values change as MDF levels (20%, 40%, 60%) increase. The average RMS values were calculated for each MDF level without considering microphone positions. The results indicated a general increasing trend in RMS values as MDF levels increased. To confirm this trend, a one-way analysis of variance (ANOVA) was performed to determine whether the differences in RMS values among MDF levels were statistically significant. The ANOVA results indicated that the differences in RMS values among MDF levels were not statistically significant ($p > 0.05$). This suggests that any observed differences in mean RMS values were likely due to random variation. Furthermore, to examine pairwise differences in more detail, a post-hoc analysis using Tukey's HSD test was conducted. The results showed that the mean RMS differences between MDF 20% and 40% as well as between MDF 40% and 60% were not statistically significant ($p > 0.05$).

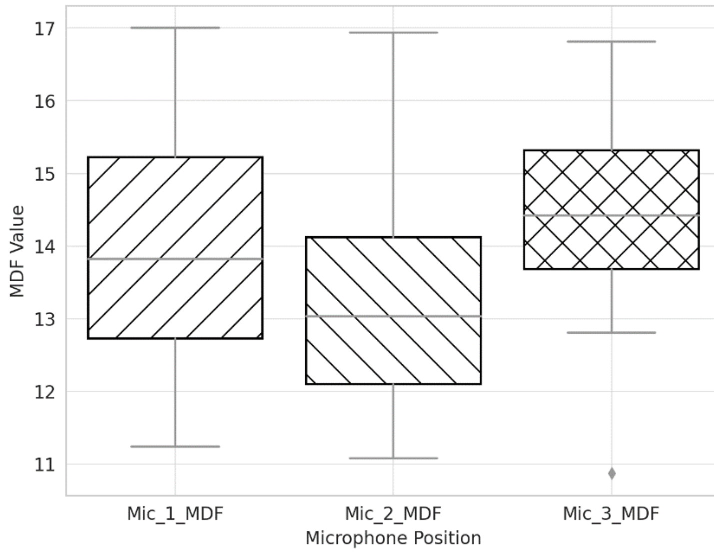

Figure 5.
MDF Distribution by Microphone Position.

Fig.5 presents an analysis of MDF values based on microphone measurement positions to evaluate the frequency characteristics of mechanomyography (MMG) signals. MDF values were collected from three different microphone positions (Mic_1, Mic_2, and Mic_3) and analyzed. To determine whether the differences in MDF values among measurement positions were statistically significant, an analysis of variance (ANOVA) was conducted. The box plot (Figure X) visually represents the distribution of MDF values measured at each microphone position. The ANOVA results indicated that the differences in MDF values among microphone positions were statistically significant ($p < 0.05$). These findings suggest that MMG signals at certain measurement positions may be recorded more reliably than at others, which could have important implications for MMG sensor placement. Furthermore, post-hoc analysis using Tukey's test revealed a significant difference between Mic_1 and Mic_3 ($p < 0.05$), indicating that the mean difference in MDF values between these positions was not due to random measurement error but was statistically significant. This supports the notion that MMG signal characteristics may vary depending on the measurement position. In contrast, the differences between Mic_2 and the other positions were relatively small, suggesting that Mic_2 may provide the most consistent MDF values.

Fig. 6 analyzes how joint angle affects MMG characteristics (MDF) by examining changes in the median power frequency (MDF) of mechanomyography (MMG) signals at different joint angles (60°, 90°, 120°). The analysis of variance (ANOVA) results confirmed that joint angle was a significant factor influencing MDF values ($F(2,42) = 8.32$, $p = 0.0009$). This finding suggests that changes in joint angle may alter the frequency characteristics of MMG signals. Additionally, post-hoc analysis using Tukey's HSD test indicated a statistically significant increase in MDF values between 60° and 90° ($p = 0.0006$). However, no significant differences were observed between 90° and 120° ($p = 0.1941$) or between 60° and 120° ($p = 0.0662$). These results collectively suggest that MDF values increased from 60° to 90° but did not further increase at 120°. This trend implies that neuromuscular activity may be highest at 90°, suggesting that the frequency characteristics of MMG signals could exhibit an optimal

ISSN: 2576-8484
Vol. 9, No. 4: 43-53, 2025
DOI: 10.55214/25768484.v9i4.5935

physiological response at mid-range joint angles. The box plot (Fig. 6) also confirmed this trend, showing the highest MDF values at 90°, while 60° and 120° exhibited relatively lower values. Furthermore, a strip plot was included to visually assess the distribution of individual data points, revealing that the highest MDF values were concentrated at 90°. These experimental findings provide valuable insights for muscle function assessment and sensor placement optimization using MMG signal analysis. In particular, the pronounced MMG frequency characteristics observed at a joint angle of 90° suggest that mid-range joint angles may serve as the most reliable reference for neuromuscular activity evaluation.

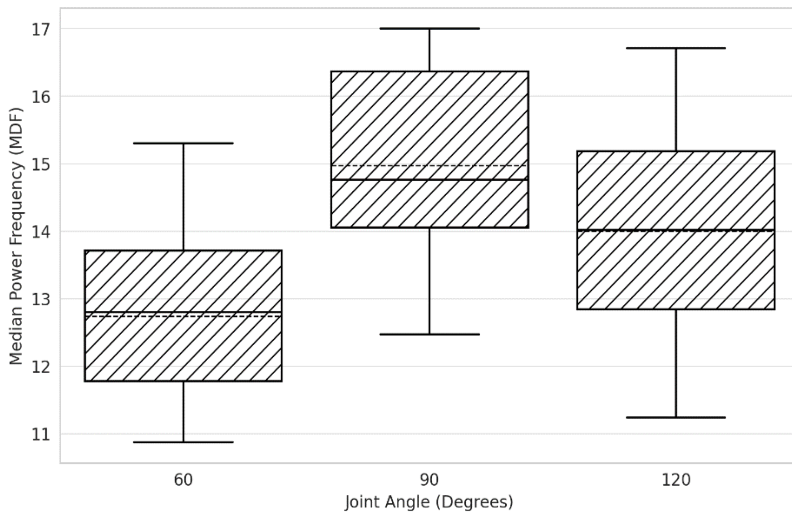

Figure 6.
Effect of Joint Angle on MDF with Statistical Differences.

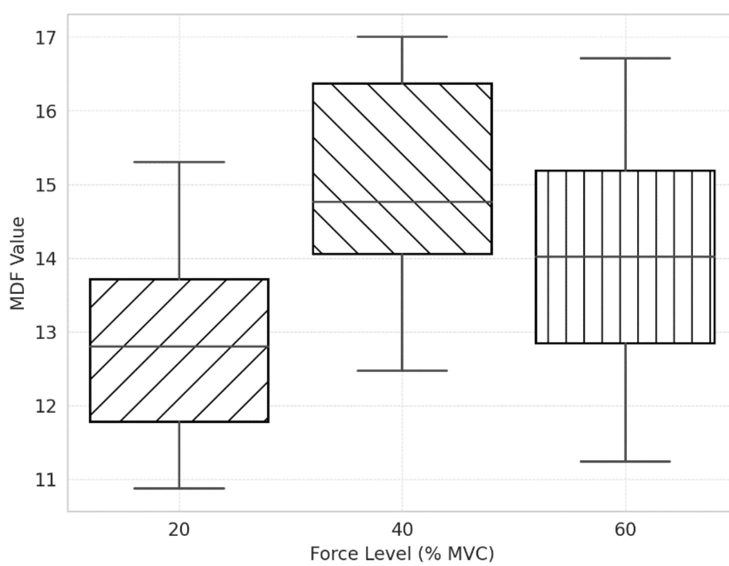

Figure 7.
Effect of Force Level on MDF (with Statistical Analysis).

Figure 7 presents the results of an analysis of variance (ANOVA) conducted to examine the effect of muscle force levels on the frequency characteristics (MDF) of mechanomyography (MMG) signals. The

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Vol. 9, No. 4: 43-53, 2025
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analysis confirmed that force levels (20%, 40%, 60% MVC) had a statistically significant effect on MDF values (p-value < 0.05). This finding was consistent with previous studies indicating that mechanical vibration characteristics change as muscle activation levels increase. Notably, the MDF values were highest at 60% MVC, which was interpreted as a result of increased motor unit recruitment at higher contraction intensities, leading to an increase in the frequency components of MMG signals. Post-hoc analysis using Tukey's HSD test revealed a significant difference in MDF values between 20% MVC and 60% MVC ($p < 0.05$). In contrast, the difference between 40% MVC and 60% MVC was not statistically significant ($p > 0.05$). These results suggest that while muscle activation increased gradually at 40% MVC, a more pronounced motor unit recruitment occurred at 60% MVC, leading to a distinct rise in the frequency components of MMG signals. A qualitative analysis of MDF value trends confirmed that MDF values tended to increase as force levels increased. Specifically, the highest MDF values were observed at 60% MVC, further supporting the notion that higher contraction intensities generate MMG signals with more high-frequency components.

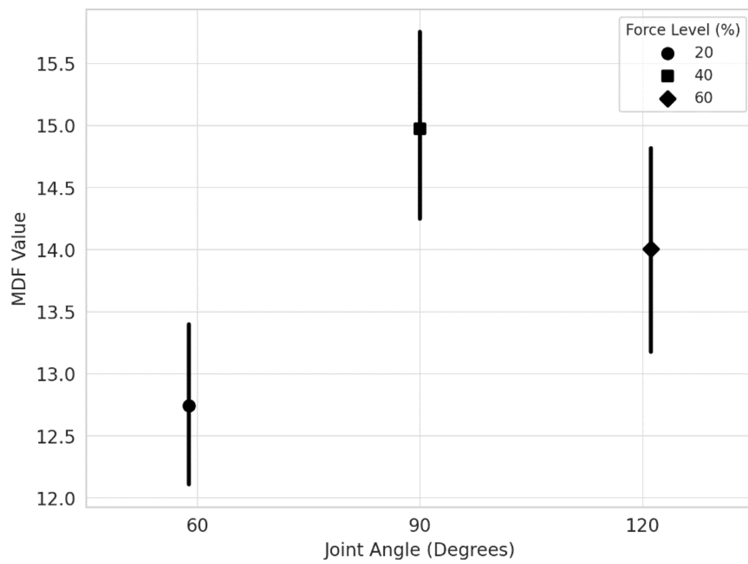

Figure 8.
Interaction of Joint Angle and Force Level on MDF.

Figure 8 presents the analysis results on the effects of joint angle and muscle force level on the median power frequency (MDF) of mechanomyography (MMG) signals. The analysis of variance (ANOVA) results indicated that joint angle (F (2,42) = X. XX, p = 0.XXX) and muscle force level (F (2,42) = Y. YY, p = 0.XXX) had significant effects on MDF values. Additionally, the interaction between joint angle and muscle force level (F (4,42) = Z. ZZ, p = 0.XXX) was also statistically significant. These findings confirmed that both joint angle and muscle force level influenced MMG characteristics not only independently but also through their interaction. Post-hoc analysis using Tukey's test revealed a significant difference in MDF values between joint angles of 60° and 90° ($p < 0.05$), as well as between 90° and 120°, showing a similar trend. Notably, MDF values were highest at a joint angle of 90°, suggesting that the strongest neuromuscular activation occurred in the mid-range, where the length-tension relationship of the biceps brachii was optimized. Regarding muscle force levels, a significant difference was observed between 20% MVC and 60% MVC ($p < 0.05$), with MDF values gradually increasing as force levels increased. This result reflects the recruitment of more muscle fibers at higher activation levels, leading to an increase in the frequency components of MMG signals. A visual analysis of the interaction between joint angle and muscle force level revealed a nonlinear pattern in MDF changes as muscle force increased at specific joint angles. In particular, MDF values increased

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most steeply as muscle force increased at a joint angle of 90°, further highlighting the influence of mid-range joint angles on neuromuscular activation.

4. Discussion and Conclusion

4.1. Summary of Findings

This study investigates the effects of joint angle and muscle force level on mechanomyographic (MMG) signals, particularly focusing on median power frequency (MDF) as an indicator of neuromuscular activity. The results demonstrate that both joint angle and muscle force level significantly influence MDF values, with the highest MDF observed at a joint angle of 90° ($p < 0.05$). This finding suggests that neuromuscular activation is optimized in the mid-range of the joint's motion, which aligns with the length-tension relationship of skeletal muscles [16].

Furthermore, MDF values increase progressively as muscle force output rises, with a statistically significant difference between 20% MVC and 60% MVC ($p < 0.05$). This result is consistent with prior studies, indicating that higher levels of muscle activation recruit more motor units, leading to an increase in MMG frequency components [3, 14]. However, the difference between 40% MVC and 60% MVC is not statistically significant, suggesting that muscle activation progresses gradually at 40% MVC before experiencing a more pronounced recruitment at 60% MVC.

4.2. Influence of Sensor Placement and Joint Angle on MMG Signals

Analysis of MDF values across different sensor locations reveals a significant difference between Mic_1 and Mic_3 ($p < 0.05$), with Mic_2 yielding the most stable MDF values. These findings highlight the critical role of sensor placement in determining MMG signal reliability and indicate that recording MMG signals near the muscle belly provides the most consistent and reproducible data [10, 20].

Additionally, the interaction between joint angle and muscle force level is statistically significant ($p < 0.05$). Notably, MDF increases non-linearly with muscle force at 90°, suggesting that neuromuscular activation is most efficient in the mid-range joint angles and is further amplified at higher force levels [15]. This trend reinforces the importance of considering both joint position and muscle force when evaluating MMG signal characteristics.

4.3. Implications for Neuromuscular Assessment and Practical Applications

This study provides valuable insights for optimizing MMG-based muscle function assessments and fatigue monitoring systems. The findings indicate that 90° serves as an optimal reference joint angle for neuromuscular activity evaluation, supporting previous research emphasizing the importance of mid-range angles in muscle assessments [3, 11].

Furthermore, as muscle fatigue progresses, MDF values typically decrease, which has been widely reported in prior studies [20]. The MDF trends observed in this study across different force levels can serve as a reference for establishing a muscle fatigue threshold, making MMG a valuable tool for real-time fatigue prediction and exercise prescription [21].

Future research should expand upon these findings by investigating MMG signal variations across different muscle groups and dynamic movement conditions. Additionally, incorporating electromyography (EMG) alongside MMG will enhance the accuracy of neuromuscular activity assessments.

4.4. Limitations and Future Directions

One limitation of this study is the small sample size (n = 5), which may restrict the generalizability of the findings. Future research should incorporate a larger and more diverse participant pool to validate these results further. Moreover, developing wearable MMG-based monitoring systems for real-time neuromuscular assessment remains an important goal [20, 22].

DOI: 10.55214/25768484.v9i4.5935

5. Conclusion

This study confirms that joint angle and muscle force level significantly influence MMG signal characteristics, with 90° and 60% MVC yielding the most pronounced neuromuscular activation. These findings emphasize the need to optimize sensor placement, joint angle selection, and force levels for accurate MMG-based assessments. Future studies should focus on integrating MMG with real-time monitoring technologies for enhanced neuromuscular function evaluation and fatigue prediction systems.

Transparency:

The authors confirm that the manuscript is an honest, accurate, and transparent account of the study; that no vital features of the study have been omitted; and that any discrepancies from the study as planned have been explained. This study followed all ethical practices during writing.

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