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Original scientific paper

IMPACT OF SKINFOLD THICKNESS ON WAVELET-BASED MECHANOMYOGRAPHIC SIGNAL

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Eddy Krueger[1,2], Eduardo M. Scheeren[3],
Carla Daniele Pacheco Rinaldin[3], André E. Lazzaretti[2],
Eduardo Borba Neves[2], Guilherme Nunes Nogueira-Neto[3],
Percy Nohama[2,3]

[1]Neural Engineering and Rehabilitation Laboratory, Anatomy department,
State University of Londrina, Londrina, Brazil
[2]Rehabilitation Engineering Laboratory/CPGEI/PPGEB,
Federal Technological University of Paraná (UTFPR), Curitiba-PR, Brazil
[3]Rehabilitation Engineering Laboratory/PPGTS,
Pontifical Catholic University of Paraná (PUCPR), Curitiba-PR, Brazil

Abstract. *Surface mechanography (MMG) is a non-invasive technique that captures signs of low-frequency vibrations of skeletal muscles through the skin. However, subcutaneous structures may interfere with the acquisition of MMG signals. The objective of this study was to verify the influence of skinfold thickness (ST) on the MMG wavelet-based signal in the rectus femoris muscle during maximal voluntary contraction in two groups of individuals: group I (n = 10, ST <10 mm) and group II (n = 10, ST equal to or> 20 mm). Negative correlation was observed between the 19 Hz, 28 Hz and 39 Hz frequency bands with ST. There was a statistical difference in almost all frequency bands, especially in the X and Y axes. All MMG axes in group II presented higher magnitudes in frequency bands 2 and 6 Hz (like low-pass filter). Thus, these results can be applied to calibrate MMG responses as biofeedback systems.*

Key words: *Skinfold thickness, Wavelet, Mechanomyography*

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Corresponding author: Carla Daniele Pacheco Rinaldin
Rehabilitation Engineering Laboratory/PPGTS, Pontifical Catholic University of Paraná, Curitiba-PR, Brazil
E-mail: rinaldin99@gmail.com

1. INTRODUCTION

Mechanomyography (MMG) is a noninvasive technique that can be also used for monitoring muscle biofeedback as, for instance, can be apply in the control of myoelectrical [1-3] and neural prostheses [4] and to support physical therapy sessions [5, 6]. The MMG records vibrations [7] of detectable muscle fibers on the surface of the skin. The vibrations are related: to the rate of activation of the motor units of global form; to contractile properties; to the time of contraction and relaxation [8, 9]. However, the detectable vibrations on the skin surface and the frequency of the MMG signal can be affected by the adipose tissue that attenuates the spectral and temporal characteristics of the MMG, compromising the acquisition and processing of the signals. In a previously study [10] we found a negative correlation between skinfold thickness (ST) and MMG mean frequency, showing the ST can be considered as a natural selective filter [11]. However, when it is applied Fast Fourier Transform (FFT) to extract the mean frequency [10] it becomes unclear which frequencies are affected by ST. Differing from transforms as FFT that uses only basis functions as sine and cosine to process the frequency, wavelet transform presents the spectral content and temporal space in specific band of frequencies through basis functions called mother wavelet [12, 13]. Peñailillo et al. [14] state that wavelet transform provides information of the frequency changes in electromyography (EMG) that is not detected by FFT. The Cauchy wavelet [15] (CaW), originally developed for the analysis of EMG signals, may be used with MMG technique adjusting the processing to particularities such as time and frequency resolution requirements of the MMG signals [16].

Therefore, the objective of this study is to evaluate the influence of ST on wavelet-based mechanomyographic signal. Based on previous studies, our hypothesis is that all frequency bands present a decrease in magnitude of energy in each frequency band with greater ST.

2. METHODS

2.1. Participants

This investigation was performed according to principles of the Declaration of Helsinki and was approved by Pontifical Catholic University of Paraná’s (PUCPR) Human Research Ethics Committee under register n. 2416/08. Twenty able-bodied volunteers (22.15±6.43 yo.) participated in this study. During the period of tests, the volunteers did not use any drug that could change their motor condition. Anthropometric data such as weight, height, body mass index (BMI) and the quadriceps skinfold were collected. The volunteers were divided in two groups: group I (N= 10) with skinfold below 10 mm and group II (N= 10) with ST equal or above 20 mm.

2.2 Sensors and Data Acquisition

The developed MMG instrumentation used Freescale MMA7260Q MEMS triaxial accelerometer with sensitivity equal to 800 mV/G at 1.5 G (G: gravitational acceleration). Electronic circuits allowed 10× amplification. A load cell (100 kg, 2.0±0.1 mV/V) was used to acquire the quadriceps torque information. A LabVIEW™ program was coded to acquire and the display signals. The acquisition system contained a DT300 series Data

Translation™ board working at 1 kHz sampling rate. The data were saved into European Data Format (EDF) files.

2.3 Research design

The volunteers performed a standard muscular and warm-up stretching before the experimental protocol. They were seated on a bench with the hip and knee angles set to 70º [17]. After trichotomy and skin cleaning, superficial MMG sensors were positioned over the belly of *rectus femoris* muscle and attached with double-sided tape. The anterior ankle joint was positioned in a brace (foam coated) positioned in 60° of flexion from total knee extension (0^o) as shows Fig. 1. From three initial knee flexion repetitions, the highest torque value was obtained to measure the maximal isometric voluntary contraction. All participants were verbally instructed to provide their maximum effort and held it for 5 s. In order to minimize a possible muscular fatigue, 2 min rest was performed between contractions [18].

Fig. 1 Scheme of the experimental setup. The MMG sensor (triaxial accelerometer) was placed over the *rectus femoris* muscle belly and knee angle was positioned in 60º of flexion with a load cell

2.4 Signal processing and statistical analysis

Inside the 5 s of maximal torque, 0.5 s before and 0.5 s after the peak torque was selected visually through the software BioProc2© version 2.4 totalizing 1 s with 1000 points (1 kHz sampling rate) was computed to characterize the performance of analysis. The points were processed by the software MatLab® version R2008a. Ten epochs of 0.1 s [19, 20] was computed inside 1 s. A third-order Butterworth filter was selected with bandpass of 5-100

Hz [7, 21-28]. The MMG signal was processed in eleven bands (2 – 119 Hz) of frequency with CaW [15] to each participant. To each frequency band was computed the RMS value to characterize the energy of wavelet band.

Spearman (ρ) correlation coefficient was applied to check the relation between MMG features and ST. Mann-Whitney U test was applied in order to check the differences between the two groups.

3. RESULTS

Table 1 shows the anthropometric data of the participants. The BMI classification of the group I was normal and the group II was considered overweight. The ST between groups I (7.6±1.13 mm) and group II (36.57±10.33 mm) was statistically significant ($p < 0.01$).

Table 1 Anthropometric data BMI: Body Max Index. I: group with skinfold below 10 mm; II: group with skinfold equal or above 20 mm.

Group	Volunteer	Age (yo)	Weight (kg)	Height (m)	BMI (kg/m^2)	Skinfold (mm)
I	1	19	60.0	1.80	18.52	6.8
	2	25	59.6	1.68	21.12	8.0
	3	19	61.6	1.78	19.44	6.9
	4	19	66.3	1.75	21.65	6.0
	5	19	63.0	1.75	20.57	8.5
	6	20	60.0	1.70	20.76	7.5
	7	36	70.0	1.74	23.12	9.0
	8	21	72.2	1.75	23.58	8.0
	9	19	70.6	1.72	23.86	9.4
	10	18	59.9	1.76	19.34	6.0
II	11	36	90.0	1.78	28.41	36.3
	12	23	110.8	1.79	34.58	52.0
	13	23	106.0	1.76	34.22	51.0
	14	36	88.0	1.84	25.99	30.0
	15	21	90.0	1.92	24.41	21.0
	16	19	82.5	1.73	27.57	43.5
	17	19	90.0	1.90	24.93	20.0
	18	18	79.5	1.77	25.38	40.0
	19	23	76.8	1.85	22.44	37.0
	20	18	80.1	1.84	23.66	35.0

Table 2 shows the MMG axes mean values and standard deviation split by frequency band to each group. The frequency band to 2 Hz and above 65 Hz presented smallest values to all axes (Z, X and Y).

Table 2 Mean±S.D to MMG wavelet bands split by axes to each group

Group	Wavelet Band (Hz)	Axis		
		Z (mV$_{\text{RMS}}$)	X (mV$_{\text{RMS}}$)	Y (mV$_{\text{RMS}}$)
I	2	4.81 ± 8.10	5.93 ± 8.65	4.01 ± 5.12
	6	19.20 ± 12.59	47.06 ± 33.49	20.92 ± 14.34
	11	101.91 ± 58.39	225.02 ±143.48	94.09 ± 65.02
	19	92.86 ± 50.74	167.52 ± 87.83	82.72 ± 45.65
	28	64.31 ± 34.44	72.28 ± 36.63	52.53 ± 25.33
	39	54.00 ± 31.29	55.40 ± 28.95	26.62 ± 10.96
	51	29.82 ± 16.69	23.94 ± 14.41	12.33 ± 7.00
	65	12.97 ± 11.58	9.82 ± 8.55	4.78 ± 4.61
	81	3.93 ± 4.49	4.93 ± 5.63	2.26 ± 2.98
	99	1.78 ± 3.14	2.53 ± 3.39	1.07 ± 1.82
	119	1.27 ± 2.73	1.84 ± 3.34	0.73 ± 1.83
II	2	6.03 ± 7.92	19.73 ± 30.60	8.71 ± 8.76
	6	34.02 ± 25.40	108.65 ± 89.53	74.36 ± 65.76
	11	93.60 ± 59.02	291.54 ±193.07	190.64 ±146.51
	19	56.50 ± 32.75	121.34 ± 69.51	84.43 ± 53.05
	28	52.37 ± 30.24	74.86 ± 51.81	42.19 ± 32.71
	39	43.01 ± 24.37	44.26 ± 31.70	23.63 ± 13.69
	51	28.06 ± 17.66	28.06 ± 24.99	14.69 ± 8.25
	65	17.98 ± 14.12	16.88 ± 14.65	11.12 ± 8.98
	81	8.57 ± 6.32	10.02 ± 8.08	7.07 ± 6.29
	99	4.84 ± 4.23	5.34 ± 3.95	4.94 ± 4.71
	119	2.73 ± 2.61	2.96 ± 3.13	3.17 ± 4.06

Table 3 shows the Spearman (ρ) coefficients among MMG axes and ST split to frequency bands. Only the frequency band to 51 Hz do not show $p > 0.05$ to any axis. Frequencies of 19 Hz, 28 Hz and 39 Hz show a negative correlation, indicating that these frequencies are influenced by the ST. The lower frequencies have positive and moderate correlations, possibly due to the attenuation of higher frequencies spreading the signal energy to lower frequency bands.

Table 3 Spearman coefficient (ρ) between MMG features and skinfold

Wavelet Band (Hz)	Axis		
	Z (ρ)	X (ρ)	Y (ρ)
2	0.289^{**}	0.517^{**}	0.344^{**}
6	0.390^{**}	0.421^{**}	0.487^{**}
11	-0.017	0.228^{**}	0.352^{**}
19	-0.257^{**}	-0.268^{**}	0.065
28	-0.09	-0.053	-0.190^{**}
39	-0.204^{**}	-0.220^{**}	-0.158^{*}
51	-0.135	0.007	0.126
65	0.148^{*}	0.217^{**}	0.304^{**}
81	0.249^{**}	0.288^{**}	0.337^{**}
99	0.284^{**}	0.323^{**}	0.440^{**}
119	0.332^{**}	0.352^{**}	0.386^{**}

*: correlation is significant at the 0.05 level (2-tailed)

**: correlation is significant at the 0.01 level (2-tailed)

Figures 2, 3 and 4 show the RMS average to each frequency band across the participants. Almost all bands showed statistical significance, mainly to Y and X axes that present greater averages to 11 Hz in group II. Regarding to 2 and 6 Hz frequency bands we found a pattern in all axes where the values to group II were always greater ($p < 0.01$) than group I.

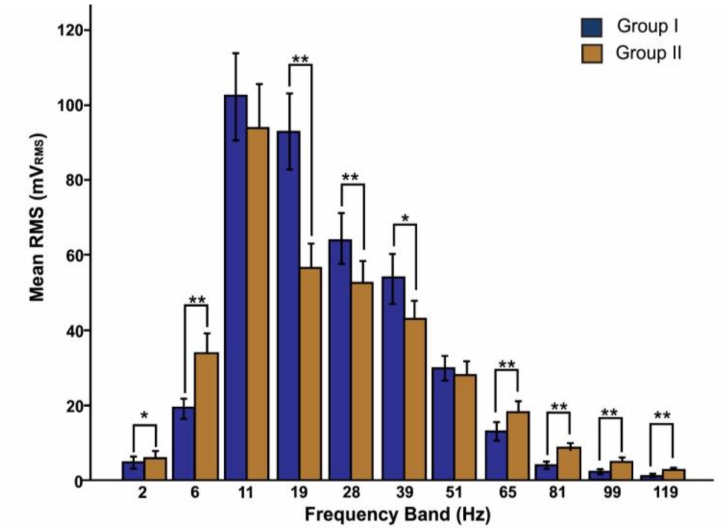

Fig. 2 Mean results (across participants) to Z-axis
*: $p < 0.05$ (2-tailed); **: $p < 0.01$(2-tailed)

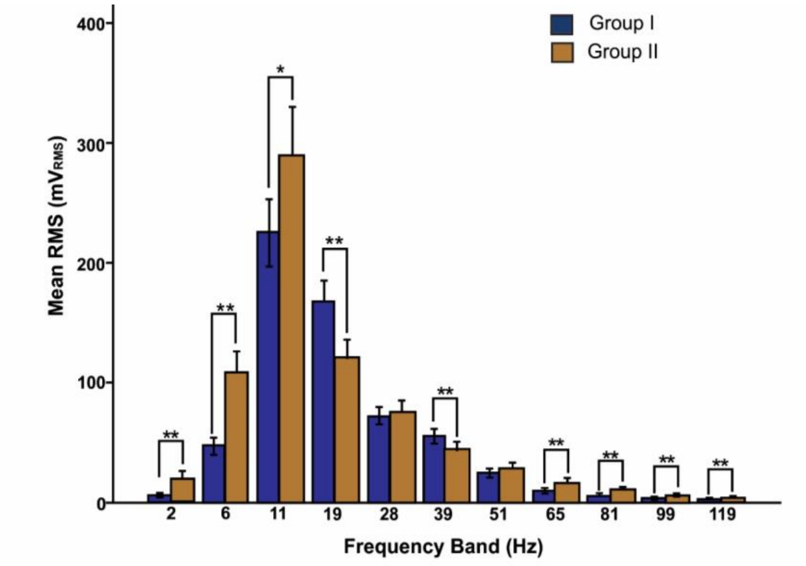

Fig. 3 Mean results (across participants) to X-axis
*: $p < 0.05$ (2-tailed); **: $p < 0.01$ (2-tailed)

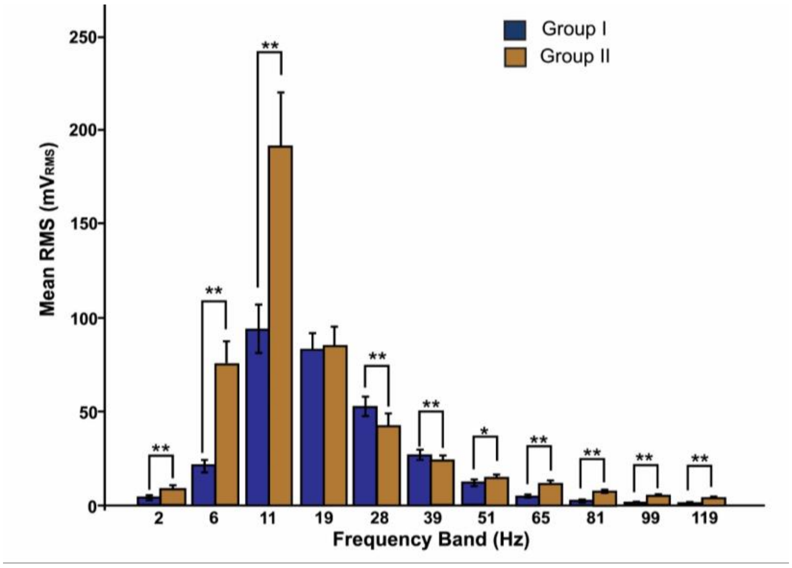

Fig. 4 Mean results (across participants) to Y-axis
*: $p < 0.05$ (2-tailed); **: $p < 0.01$ (2-tailed)

4. DISCUSSION AND CONCLUSIONS

Originally we hypothesized and found that the ST attenuates the MMG frequency bands as a low-pass filter [10]. Similar response was found by Jaskólska et al. [29] that suggested that skinfold works like a low-pass filter, which was confirmed by a negative correlation between the ST and median and peak frequencies in their study. However, we found that to lowest frequency bands (<11 Hz) the magnitude is greater to group II (St> 20 mm).

Cooper et al. [30] observed a negative correlation between skin fold thickness and MMG signal amplitude in gross lateral movements during muscle contractions performed by electrical stimulation in the *rectus femoris*, suggesting that a thick layer of adipose tissue interferes with acquisition of the amplitude of the signal of depolarization of the motor units, consequently decreasing the signal of muscular vibration. The potential of the signal amplitude of muscle activity depends on the density of muscle fibers attached to a motor neuron, which is associated with the type of fiber [31]. However, the decrease in the acquisition of muscle vibration is caused independently of the muscle fiber type, as observed by Herda et al. [11]. They obtained a lower amplitude of the MMG signal of muscle strength in type I and II fibers for the *vastus lateralis* muscle in isometric contractions in individuals with greater subcutaneous fat thickness, suggesting that the adipose tissue attenuates the muscular physiological signals, being a low pass filter acting in a natural way. Cescon et al, [32], suggests that this effect may also be influenced, depending on the location of the accelerometer on the muscle, caused by the distance between the muscle tissue and the MMG sensor, especially in anatomical regions with a thicker layer of adipose tissue.

Regarding to Table 2, wavelet bands to 2 Hz and above 65 Hz presented smallest values ($p < 0.05$) in all axes due are located outside the frequency range of physiological contraction [31, 33].

According to Maggi et al. [34], the fat density (0.95 g cm^{-3}) is smaller than muscle density (1.04 $g.cm^{-3}$) and skin density (1.20 $g.cm^{-3}$). Thus, it can be assumed that this fact leads to attenuation of high frequencies. Moreover, the fat presents lower acoustic impedance than skin and muscle. It allows that an increase in the fat layer does not generate significantly change in the total energy of signal. Polato et al. [35], using biaxial MMG found that the ST raises and MMG values decrease significantly (r = -0.3935, p = 0.0099), therefore without the specification of frequencies that are influenced by ST.

Figures 2, 3 and 4 show the RMS average to each frequency band across the participants. Axes Y and X presented greater averages to 11 Hz in group II. Concerning to of 2 Hz and 6 Hz frequencies band we found a standard in all axes where the values to group II were always greater ($p < 0.01$) than group I. In general, frequencies of 19 Hz, 28 Hz and 39 Hz present a negative correlation and frequencies of 6 Hz and 11 Hz have a increment in RMS values due the increase in ST. These results indicate that ST of 36.57±10.33 mm works like low-pass filter when compared with subjects with skinfolds of the order of 7.60, to frequencies around 19 Hz transferring the MMG signal energy to frequencies below this band. Our results are divergent to those found by Zuniga et al. [36] who investigated the effect of ST at four locations over the *vastus lateralis* muscle during incremental cycle ergometry. They found that the ST over *vastus lateralis* did not affect MMG temporal and spectral features. Probably because the average difference between skin folds studied by Zuniga et al. [36] was of the order of 8.0 mm and, in the present study, reaches the order of 29.0 mm (average group II less average group I).

In summary, individuals with a ST > 20 mm obtained greater amplitudes of the MMG signal at low frequencies in all axes, suggesting that the adipose tissue is a natural low-pass filter, which can attenuate high frequencies and the higher its thickness the lower the bandwidth. For the Y axis this event occurs at 11 Hz of the frequency band, however, the Y-axis (longitudinal), presents low influence for the recognition of muscle contraction [37]. In this sense, the amplitude of higher frequency bands (> 11 Hz) is shifted to lower frequencies due to the subcutaneous tissue. Our results suggest that this change in magnitude of frequency bands provides important information for the calibration of MMG systems when incorporating closed-loop control systems such as neuroprostheses.

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