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The Effect of Background Muscle Activity on Computerized Detection of sEMG Onset and Offset

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Abstract

The performance of two computerized algorithms for the detection of muscle onset and offset was compared. Standard deviation (SD) method, a commonly used algorithm, and approximated generalized likelihood ratio (AGLR) method, a more recently developed algorithm, were evaluated at different levels of background surface EMG (sEMG) activity. For this purpose, the amplitude ratio between the period of muscle inactivity and activity was varied from 0.125 to 1 in artificially assembled sEMG traces. In addition, 1230 real sEMG signals, obtained from various trunk muscles, were raised to a power of 3 to change the relative amplitude ratio. As the relative level of background activity increased, both the SD and AGLR methods produced longer latencies and detected fewer muscle responses, suggesting that a detection artifact can be introduced if the subject populations being compared have different levels of background muscle activity. Of the two methods, AGLR appears to be the least affected by background activity. However, above the ratio 0.8, results from AGLR are also unreliable particularly in detecting offsets. Average latency artifacts near this ratio were 8 ms for AGLR and 46 ms for SD.

1. Introduction

Muscle reflex response is often used to study pathology such as low back pain (LBP). A number of studies have reported that reflex responses are delayed in people with LBP (Wilder *et al.*, 1996; Radebold *et al.*, 2000; Radebold *et al.*, 2001; Reeves *et al.*, 2005) and those predisposed to injury (Cholewicki *et al.*, 2005) as compared to healthy controls. These studies also suggest that LBP patients respond with fewer muscles than healthy controls to equivalent perturbations. Other studies have reported that LBP patients also have increased co-activation of trunk muscles (Marras *et al.*, 2001; Lariviere *et al.*, 2000; Van Dieen *et al.*, 2003). Is it possible that higher levels of background muscle activity resulted in the longer reflex delays and fewer muscles responding following a perturbation? There is evidence that higher level of background activity causes detection artifact in terms of longer latencies for muscle onsets (Staude *et al.*, 2001; Allison, 2003). To our knowledge, no studies have investigated the effects of background activity on muscle offsets. Given that the differences between LBP patients and healthy controls were primarily with muscle offset, the effects of background activity on latencies and number of muscles responding requires investigation to ensure that findings reported in the literature did not arise solely from detection artifact.

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Consequently, the goal of this paper was to determine if methods used for detecting muscle onset, offset, and number of muscles responding, are susceptible to changes in background activity. The two most common automated detection methods used in past studies were standard deviation (SD) method (Hodges and Bui, 1996) and approximated generalized likelihood ratio (AGLR) method (Staude and Wolf, 1999, Staude *et al*. 2000). Staude *et al.* 2001 found that the SD method was more susceptible to background activity than the AGLR method for muscle onset detection. We hypothesized that the muscle onset and offset latencies will be longer and the number of muscles responding will be smaller in both methods as background activity increases, but of the two, the AGLR method would be the least affected.

2. Method

2.1 Signal

One hundred artificial signals with obvious onset were generated using relatively steady sections of surface EMG (sEMG) signals. Segments were randomly selected from both inactive and active trunk muscles recorded from 303 individuals in a previous study (Cholewicki *et al.*, 2005) (see the next paragraph for a detailed description of electrode placement and EMG collection process). The segments were repeated such that the 1st half and the 2nd half were identical. Initially, the first half of the signal amplitude was decreased by half and the second half was increased by a factor 4, resulting in a signal with an onset at 312.5ms (Figure 1a). Artificial offsets were created by mirroring the artificial onset signals. Next, the amplitude of the 1st half of the onset signal and the 2nd half of the offset signal were multiplied by 1 to 8 with increments of 0.5 until the 1st and 2nd halves of the signals were equal in amplitude, yielding amplitude ratios between the response and background from 0.125 to 1. Figures 1a–c represents three levels of background activity. These signals will be referred to as artificial signals throughout the text.

Additionally, sEMG signals, representing muscle responses from an experimental trunk perturbation study, were used. Forty-one quick release trials from each of 3 directions (flexion, extension, and left lateral bending) were randomly selected from data collected previously (Cholewicki *et al.*, 2005). Briefly, subjects were placed in a semi-seated position in the testing apparatus for exerting isometric contraction in trunk flexion, extension and left lateral bending. Isometric force was applied through a cable attached to a chest harness at approximately T5 and was held with an electromagnet. The quick release of this electromagnet produced sudden unloading resulting in displacement of the trunk, and initiating reflex responses in the trunk muscles. sEMG signals were collected bilaterally from rectus abdominis (RA) (3 cm lateral to the umbilicus), external oblique (EO) (approximately 15 cm lateral to the umbilicus), internal oblique (IO) (approximately midway between anterior superior iliac spine and symphysis pubis, above inguinal ligament), lumbar erector (LE) (3 cm lateral to L4 spinous process), and thoracic eerector (TE) (5 cm lateral to T9 spinous process) muscles. Signals were band-pass filtered between 20 and 420 Hz, differentially amplified (input impedance = 100GΩ, CMRR>140dB) and A/D converted at a rate of 1600Hz. Signals were then cubed (raised to the power of 3) to accentuate the relative amplitude difference. Both the original and cubed sEMG signals were later used for analysis and will be referred to as real signals throughout the text.

2.2 Algorithm

For the SD method, the sEMG signals were first rectified and low pass filtered (4th order Butterworth, 50Hz cutoff). For both real and artificial signals, a fixed baseline window of 50ms was used, and then the mean and standard deviation was calculated. The mean plus (onsets) or minus (offsets) 2 times the standard deviation was used as a threshold in the detection window. If the mean of the 25ms wide sliding detection window exceeded the threshold, the event was

detected and the latency was reported. The filter's cutoff frequency and the width of the sliding window were set based on Hodges and Bui's recommendation for high background activity (Hodges and Bui, 1996). The same parameters were used for the detection of sEMG offsets.

With the AGLR method developed by Staude et al. (2000), segments of signals were compared to see the likelihood of them being statistically different using a pre-defined likelihood threshold. Detection was performed in 2 steps. First, a log-likelihood ratio was calculated for the sliding window with length L. If this ratio exceeded a pre-set threshold, h, an alarm time was given $t_a(i)$. In the second step, the maximum likelihood ratios were calculated from the previously detected event $t_0(i-1)$ to the alarm time $t_a(i)$+delta, and the time at the largest maximum likelihood ratio was reported as $t_0(i)$. For each direction, the parameters (h, L and delta) were optimized to a separate set of latencies found from visual inspection (Table 1). Staude and Wolf (1999) suggested that a pre-whitening filter be used to negate any spectral distortions introduced by the EMG collection process. To be consistent between the detection methods, all the signals were processed with a pre-whitening filter for both algorithms.

For artificial signals, the detection latencies were restricted from 50ms before to 150ms after the actual event (onset or offset). For real signals, the detection latencies were limited from 15ms to 150ms after the perturbation to ensure that only reflexive responses were considered and not voluntary activation.

2.3 Statistical analysis

Artificial signals were compared using 2 (onsets and offsets) 1-factor (background activity level), repeated measures ANOVA. Real signals were grouped into onset and offset for each direction (flexion, extension, and lateral bending). Mean latencies and number of responses from original and cubed signals were compared with 12 (3 directions × 4 dependent variables) T-tests using SD and AGLR to determine if these methods are susceptible to increased background activity. Next, original and cubed signals of the 10 individual muscles were compared with 30 (3 direction × 1 dependent variables × 10 muscles) paired T-tests using SD and AGLR. Since the goal was to observe trends in susceptibility, no correction for multiple comparisons was made. The critical level for significance of $p<0.05$ was used for all comparisons.

3. Results

3.1. Artificial signals

The higher level of background sEMG activity resulted in significantly longer onset latencies detected by both methods ($p < 0.001$) (Figure 2a). AGLR performed significantly better than SD in its ability to detect the onsets of muscle activity accurately ($p<0.001$). The higher level of background sEMG activity resulted in detection of significantly longer offset latencies as well ($p < 0.001$) (Figure 2b). Like onsets, AGLR produced significantly less error than SD ($p<0.001$). These findings suggest that AGLR was less susceptible to changes in background muscle activity than SD when used to detect the sEMG onsets and offsets.

3.2. Real signals

For the real signals, there was a general trend for SD to be affected more by level of background activity than AGLR. As expected, average muscle onset and offset latencies, detected with SD were significantly longer for the original sEMG signal as compared to the cubed signal (Table 2). However, with AGLR, only onset latencies detected for original signals were significantly longer than for cubed sEMG signals (Table 2). The numbers of detected offset responses were significantly increased regardless of the direction of pull with SD, whereas with AGLR, only

the extension direction was affected. As for the number of detected onsets, only AGLR in lateral bending was significantly affected by cubing of signals (Table 2).

When considering the response latencies of individual muscles rather than the average of all onset and offset responses, a similar trend appeared in that AGLR was less susceptible to changes in background sEMG activity than SD in flexion and extension. Among 10 muscles monitored in these 2 trunk exertion directions (20 cases), AGLR detected significantly different response latencies between the original and cubed signals in 4 cases (Table 3). In contrast, SD detected significantly different response latencies in 11 cases (Table 4). In lateral bending, both methods generated shorter onset latencies after cubing the signal, possibly due to larger antagonistic muscle activity in this direction (Table 3, 4) (McGill, 1992).

4. Discussion

Our first hypothesis that the level of background activity would affect the number of muscle responses and their onset and offset latencies was supported by the findings in this study. As background activity increased, the error in detecting muscle onset and offset increased. With more background activity, the detected muscle onset and offset latencies became longer. This has implications for investigating differences in muscle responses between LBP patients and healthy controls, since LBP patients tend to have higher levels of background activity from trunk muscle co-activation (Marras *et al.*, 2001; Lariviere *et al.*, 2000; Van Dieen *et al.*, 2003). Consequently, the findings reported in the literature regarding delayed muscle responses in the LBP population should be interpreted with caution.

The second hypothesis that the AGLR method would be less susceptible to background activity than the SD method was also supported by the current findings, which are in line with the Staude *et al.* 2001 study. However, the accuracy of detection for both AGLR and SD methods degrades rapidly at the higher EMG ratios (Figure 2b). Regardless of the method, whether it be visual or computerized, they are limited when high background EMG activity is present. Hence, latencies detected when background and changed signal ratios are close to 1 should be handled with care.

The analyses of artificial and real signals possess unique advantages and disadvantages. It is easy to manipulate the background activity in artificial signals, but their temporal characteristics may not match real signals. On the other hand, cubing the real signals is a non-linear operation and changes their probability density and spectral characteristics. For this reason, the analyses were performed with both artificial and real signals. The results are similar in both cases, confirming the validity of our conclusions.

The underlying concept of both the AGLR and SD methods is to detect deviation from baseline. The variance of baseline, which is small in onsets and large in offsets, is greatly affected by background activity. Because it is easier to detect changes from small variance to large variance than the converse, one would expect better performance from both algorithms in onset than in offset detection, which is supported by our findings (Figure 2).

In conclusion, the AGLR method appears to be more robust than the SD method for investigating differences between subject groups with possible differences in background muscle activity. Given that some studies reporting delayed muscle responses in people predisposed to LBP employed the AGLR method (Cholewicki *et al.*, 2005), it is possible that this effect is real and not induced by detection artifact. More work is required to assess the level of background activity in LBP and healthy control populations to determine if these differences could possible lead to erroneous findings in muscle reflex latencies and the number of muscles responding.

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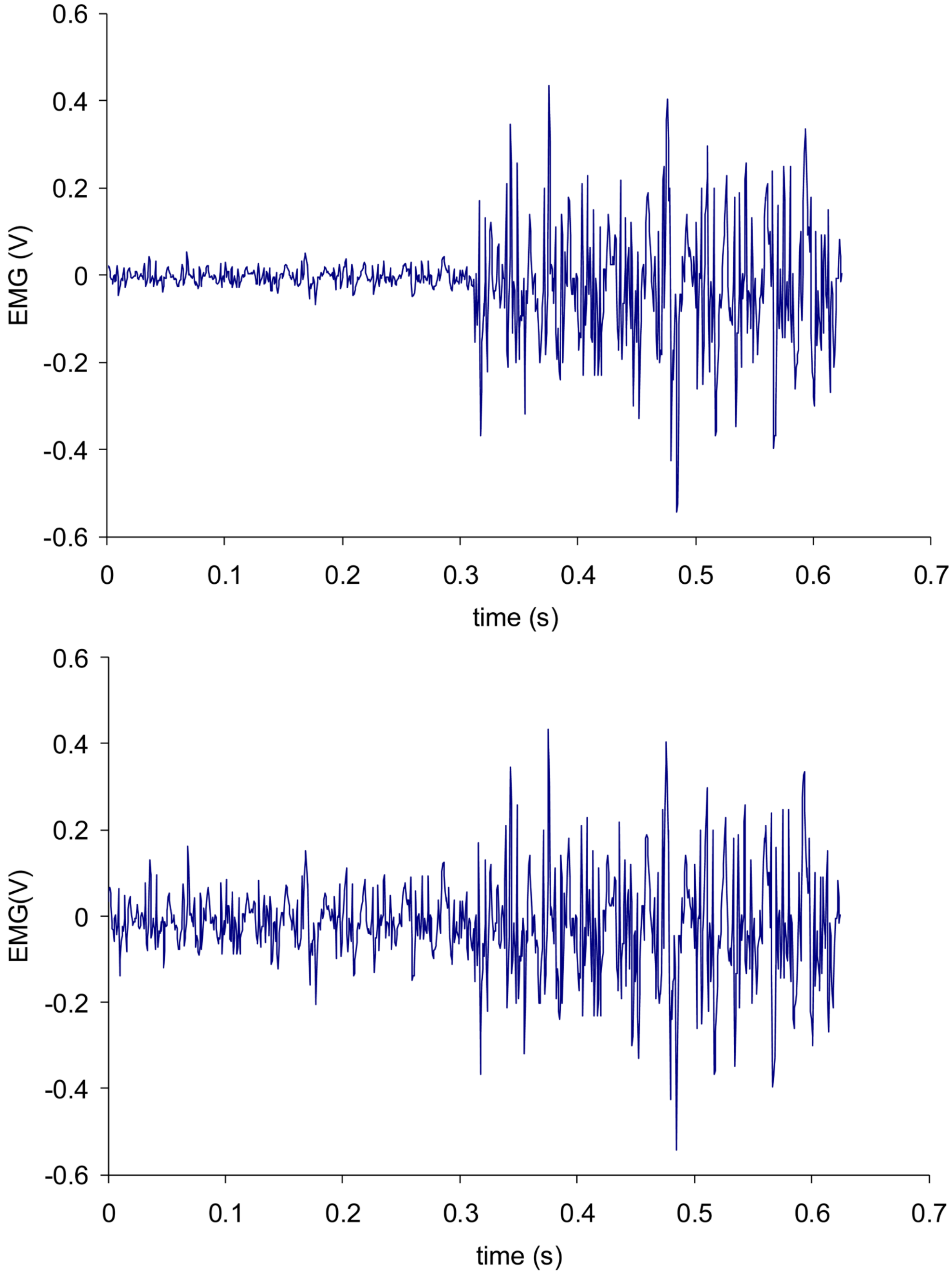
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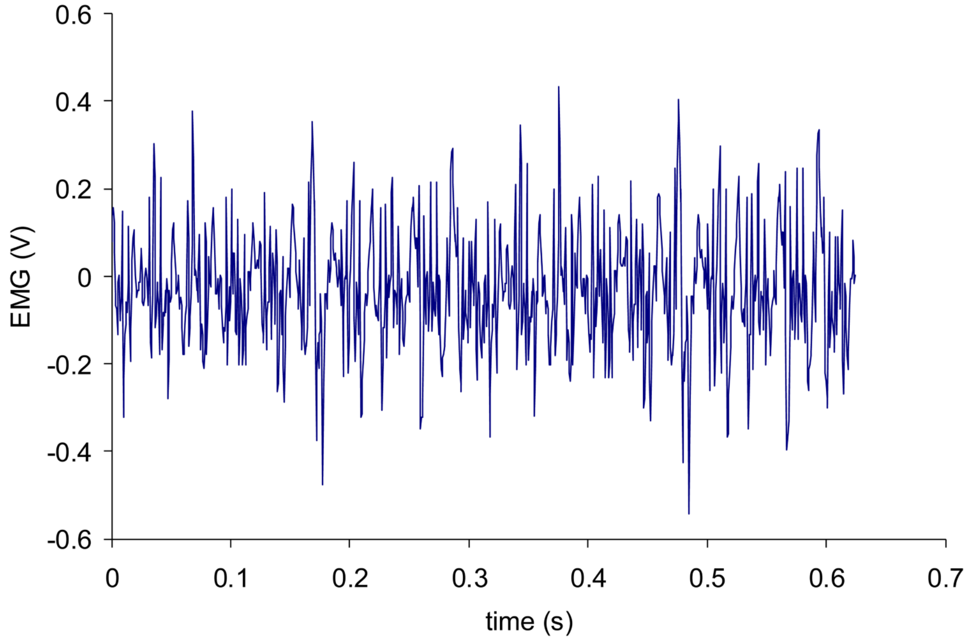
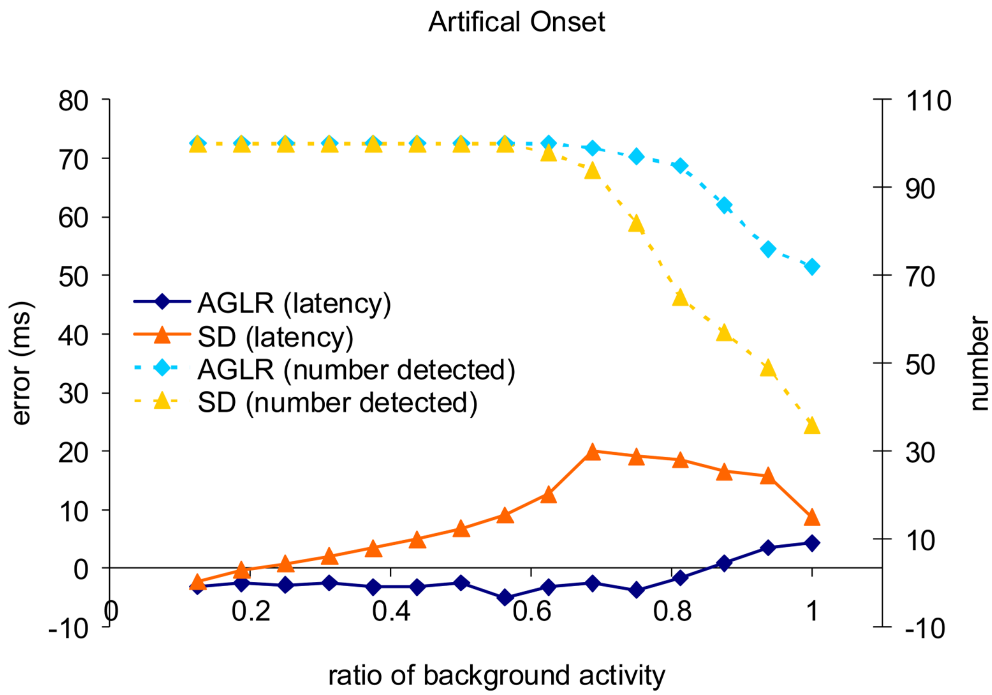


Figure 1.
a–c: An artificial onset signal is created using steady section of sEMG at ratio of 0.125(a), 0.375(b), and 0.875(c).



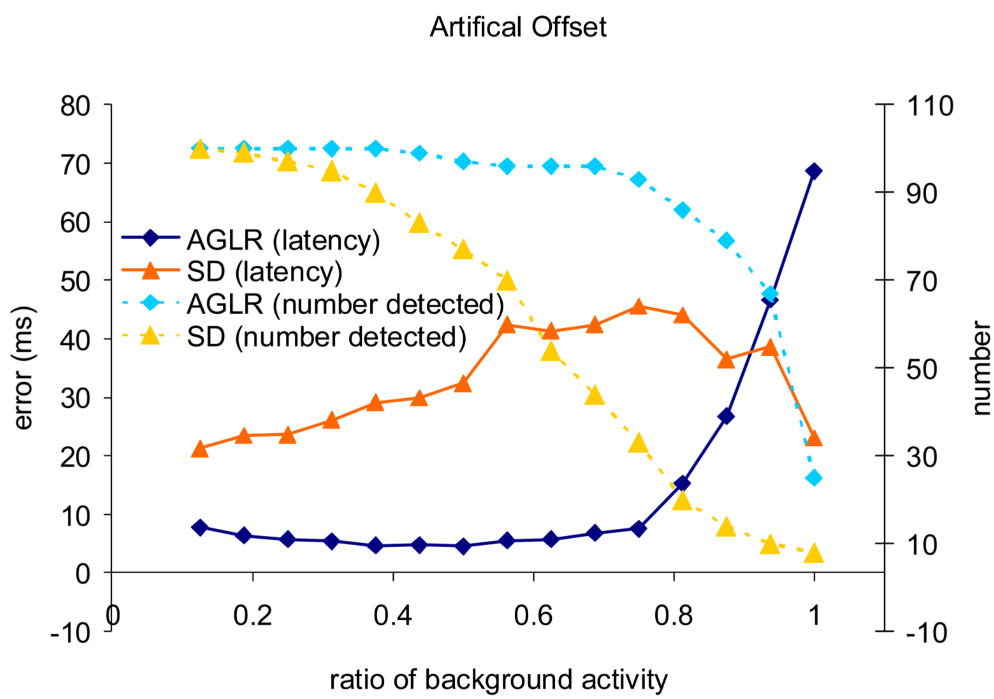


Figure 2.
a–b: Detection error for artificial onset (a), and offset (b). Positive error indicates longer latency.

Table 1

Parameters used for AGLR. Each parameter was optimized to a set of latencies found from visual inspection.

		Threshold (h)	Window length (L)	Delta
Extension	Onset	6.3	30	17
	Offset	1.0	310	20
Flexion	Onset	5.1	34	7
	Offset	6.0	201	128
Lateral Bending	Onset	10.5	16	16
	Offset	1.0	205	65

Table 2

Mean (std) number of detected trunk muscle responses and their latencies separated by exertion direction, and onset and offset of activity. Maximum possible number of onset muscles for extension, flexion and lateral bending were 6, 4, and 5 respectively and for offset muscles, they were 4, 6, and 5. Bold typeface indicates significant differences between original and cubed EMG signals ($p<0.05$).

		AGLR			SD		
		Extension	**Flexion**	**Lateral Bending**	**Extension**	**Flexion**	**Lateral Bending**
# of onset muscle	Original	6.0(0)	4.0(0.2)	**4.3(1.0)**	5.7(0.7)	3.9(0.5)	4.7(0.8)
	Cubed	6.0(0)	4.0(0)	**5.0(0.2)**	6.0(0.2)	4.0(0.2)	4.9(0.4)
# of offset muscle	Original	**2.2(1.2)**	2.2(1.7)	1.6(1.3)	**1.4(1.2)**	**1.7(1.5)**	**0.9(1.0)**
	Cubed	**1.5(1.1)**	2.3(1.4)	1.1(1.0)	**3.5(0.7)**	**4.3(1.4)**	**2.8(1.4)**
Onset latency (ms)	Original	**54(9)**	**58(12)**	**51(18)**	**54(14)**	**51(24)**	**56(19)**
	Cubed	**51(8)**	**52(10)**	**35(11)**	**40(12)**	**35(17)**	**45(16)**
Offset latency (ms)	Original	43(28)	82(37)	93(36)	**60(23)**	**67(22)**	**51(18)**
	Cubed	35(16)	74(39)	105(38)	**33(13)**	**41(16)**	**41(18)**

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Table 3

Mean (std) trunk muscle response latencies (ms) detected using AGLR. Bold typeface indicates significant differences between original and cubed EMG signals ($p<0.05$).

	Extension		Flexion		Lateral Bending	
	Original	**Cubed**	**Original**	**Cubed**	**Original**	**Cubed**
RRA	54(15)	49(15)	46(33)	44(37)	**56(27)**	**32(12)**
REO	**53(9)**	**46(14)**	66(46)	74(46)	**42(22)**	**28(10)**
RIO	53(17)	51(15)	106(35)	104(42)	**51(27)**	**37(17)**
RTE	**35(10)**	**23(6)**	**60(21)**	**51(14)**	**46(20)**	**32(13)**
RLE	38(20)	48(35)	54(20)	52(16)	**58(34)**	**42(24)**
LRA	54(12)	50(18)	74(41)	62(47)	**136(4)**	**142(4)**
LEO	55(15)	55(11)	104(42)	93(51)	105(34)	106(32)
LIO	57(25)	51(16)	93(44)	91(42)	101(43)	112(27)
LTE	38(18)	34(16)	**61(19)**	**54(18)**	77(56)	106(59)
LLE	33(18)	45(38)	56(12)	54(17)	74(49)	93(47)

Table 4

Mean (std) trunk muscle response latencies (ms) detected using SD. Bold entries indicate significant differences between original and cubed EMG signals ($p<0.05$). For LRA in lateral bending, only one response latency was detected. Bold typeface indicates significant differences between original and cubed EMG signals ($p<0.05$).

	Extension		Flexion		Lateral Bending	
	Original	**Cubed**	**Original**	**Cubed**	**Original**	**Cubed**
RRA	**56(22)**	**41(18)**	**66(21)**	**48(32)**	**62(29)**	**52(29)**
REO	**30(15)**	**43(20)**	56(24)	15(0)	**53(20)**	**45(19)**
RIO	**39(24)**	**32(23)**	70(31)	33(13)	**45(29)**	**36(24)**
RTE	48(13)	40(24)	**65(32)**	**39(22)**	**71(30)**	**48(29)**
RLE	67(22)	36(22)	**43(27)**	**29(20)**	**49(42)**	**33(29)**
LRA	**61(20)**	**41(19)**	55(21)	38(26)	98(−)	74(−)
LEO	**57(24)**	**40(28)**	71(22)	38(23)	51(4)	41(22)
LIO	**46(27)**	**40(19)**	48(31)	38(26)	57(23)	52(36)
LTE	47(16)	38(18)	**52(35)**	**38(25)**	54(16)	43(24)
LLE	65(26)	30(21)	**45(27)**	**30(20)**	48(15)	43(36)