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Surface EMG Verification of Oblique Muscle Activation During a Forcemeter-Based Unilateral Isometric Pulling Task

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Abstract

Muscle imbalances in trunk rotators may impair function and contribute to injury risk, yet practical field-based assessment methods remain limited. This pilot methodological study explored whether a unilateral isometric pulling task performed with a digital forcemeter elicits measurable trunk rotator activation and whether force output is associated with muscle activation. Fifteen physically active adults performed standardized unilateral isometric pulling under stable sitting support while surface EMG recorded activation of the external and internal oblique muscles. Force output was recorded continuously with a digital forcemeter. Within-subject associations between synchronized force and sEMG signals during the pulling task were evaluated using Spearman correlation coefficients, calculated separately for each participant, hand, and muscle, and then summarized as median values with interquartile ranges. Median within-subject correlations were observed between pulling force and ipsilateral external oblique (EO) and internal oblique (IO) activation (EO: R=0.67, IQR=0.36; IO: R=0.64, IQR=0.31), as well as for contralateral muscles (EO: R=0.69, IQR=0.28; IO: R=0.63, IQR=0.34). These findings suggest that the task elicits measurable oblique muscle activation and that force and sEMG signals covary within trials under the tested conditions. Further studies are required to determine test–retest reliability, validity, and potential diagnostic utility.

Keywords: *electromyography, muscle strength, dynamometer, athletic injuries, physical examination*

Introduction

Muscle imbalances in the trunk, particularly among rotator muscles such as the obliquus externus and internus abdominis, may be associated with significant functional deficits. These deficits may manifest in asymmetrical movement patterns, limited trunk mobility, or chronic discomfort and pain (Page et al., 2010; Zemková et al., 2019). The importance of early detection and correction of these imbalances is emphasized in sports performance optimization as well as in clinical rehabilitation settings. In practice, detecting such imbalances non-invasively, reliably, and efficiently is a persistent challenge (Page, Frank, & Lardner, 2010).

Trunk stability and trunk strength are distinct but related qualities that contribute to load distribution and postural control. Spinal stability depends on passive structures, active muscular support, and neural control. Trunk muscles contribute by increasing intra-abdominal pressure and generating stabilizing torque around the spine (Kolář, 2009).

Muscles involved in trunk rotation, such as the obliquus externus and internus abdominis, as well as the deep back muscles (erector spinae), contribute to rotational control and spinal support (Čihák, Grim, Druga, & Fejfar, 2004). Their coordinated action enables both segmental stability and dynamic motion.

Deficits in trunk control have been associated with injury risk in athletic populations, including trunk injuries, lower-extremity injuries, and altered shoulder loading in overhead athletes (De Blaiser et al., 2018; Oyama et al., 2014; Pontillo et al., 2021). Because rotational control is an important component of trunk function, practical methods that describe trunk rotator involvement during force-producing tasks may be relevant for future screening and rehabilitation research.

Muscle strength is commonly assessed using force-measuring devices, such as handheld dynamometers or digital force gauges, whereas muscle activity is evaluated from electrical signals recorded during contraction using electromyography (EMG) (Kamen & Gabriel, 2010). Digital force gauges have shown acceptable reliability in selected trunk and upper-limb strength assessments and offer a portable option for recording external force during isometric tasks (Karthikbabu & Chakrapani, 2017; Wałecka et al., 2023).

The relationship between EMG and muscle force has been a focal point of research since the mid-20th century. Establishing a reliable correlation would allow EMG to serve as a non-invasive proxy for strength measurement. However, challenges persist, particularly in dynamic movements like concentric or eccentric contractions, where artifacts and angle variability compromise signal integrity (Kuriki et al., 2011).

Consequently, most studies focus on isometric contractions, where muscle length and joint angles remain constant, allowing for clearer signal interpretation. Current research suggests a quasi-linear model: EMG activity increases linearly at low intensities but shows an exponential curve at higher intensities (Kamen & Gabriel, 2010). This transition reflects motor unit recruitment patterns governed by Henneman's size principle—motor units are activated from smallest to largest based on force demand (Cacek, 2014; Heckman & Enoka, 2012).

This hierarchical recruitment explains the apparent nonlinearity: slow-twitch type I fibers are engaged at lower intensities and produce lower electrical activity, while fast-twitch type II fibers—activated at higher intensities—produce stronger signals, leading to an accelerated rise in EMG output (Kupa, Roy, Kandarian, & De Luca, 1995). Thus, while EMG can indicate relative activation levels, interpreting it as a direct measure of force remains context-dependent.

Surface EMG (sEMG) further enhances the practicality of this approach. By placing electrodes on the skin above the muscle belly, sEMG allows for simultaneous measurement of multiple muscles during dynamic tasks. This makes it valuable in sports and rehabilitation for assessing recruitment order, muscle synergies, and fatigue (Krobot & Kolářová, 2011).

Nevertheless, accurate sEMG readings require attention to external factors such as electrode placement, skin preparation, and noise reduction. Electrodes should align with the muscle fibers, placed over the thickest part of the muscle belly with minimal spacing (typically 20 mm) to reduce cross-talk from neighboring muscles (Kamen & Gabriel, 2010).

While EMG and muscle force are closely related, their correlation depends on various factors such as contraction type and measurement conditions. Combining force measurement with sEMG may therefore help describe how external force output relates to trunk muscle activation under standardized conditions. This approach may be particularly useful in rotationally demanding tasks, where objective information on force production and neuromuscular involvement remains limited.

This pilot methodological study aimed to examine whether a unilateral isometric pulling task performed with a digital forcemeter elicits measurable activation of the external and internal oblique muscles, and whether force output covaries with sEMG amplitude under the tested conditions. The findings are intended to provide preliminary physiological verification of the task and to inform further methodological studies.

Methods

Ethical considerations

This study was approved in advance by Masaryk University Research Ethics Committee (EKV-2025-134). Each participant voluntarily provided written informed consent before participating.

Participants

Fifteen volunteers (n=15), both male and female, aged 22 to 28 years, with a height of 165 to 189 cm and a weight of 62.3 to 112.2 kg, participated in the study. All of them were healthy and without any movement impairments. All participants were right-handed, physically active individuals who engaged in moderate recreational sports but had no history of professional athletic involvement or unilateral physical overuse.

Measurement instruments

Muscle activity was recorded using the Trigno Avanti Wireless EMG System (Delsys, Boston, MA, USA), which combines EMG sensors with integrated accelerometers with a sampling frequency of 2048 Hz. The system, operating via Bluetooth, was managed through EMGworks® Acquisition and Analysis software. Sensor fixation was achieved using specialized adhesive patches provided by the manufacturer.

Isometric pulling force was measured with a Digital Force Gauge (Insize 1000/0.1 N, Type B, Suzhou, China), a high-precision device featuring an external sensor and real-time data logging capabilities (sampling frequency 10 Hz). The device, accurate to ±0.2% by the manufacturer, was capable of exporting force-time data to Excel via dedicated software.

Experimental protocol

All measurements were conducted in a laboratory setting. Subjects were seated on an aerobic step with hips flexed to approximately 90°, and feet placed at hip-width apart on a solid surface. The tested upper limb held a handle connected to the forcemeter, with the elbow flexed at 90° and the shoulder slightly flexed, ensuring the pulling motion occurred in the horizontal plane. The contralateral arm remained relaxed. Position is displayed in Figure 1.

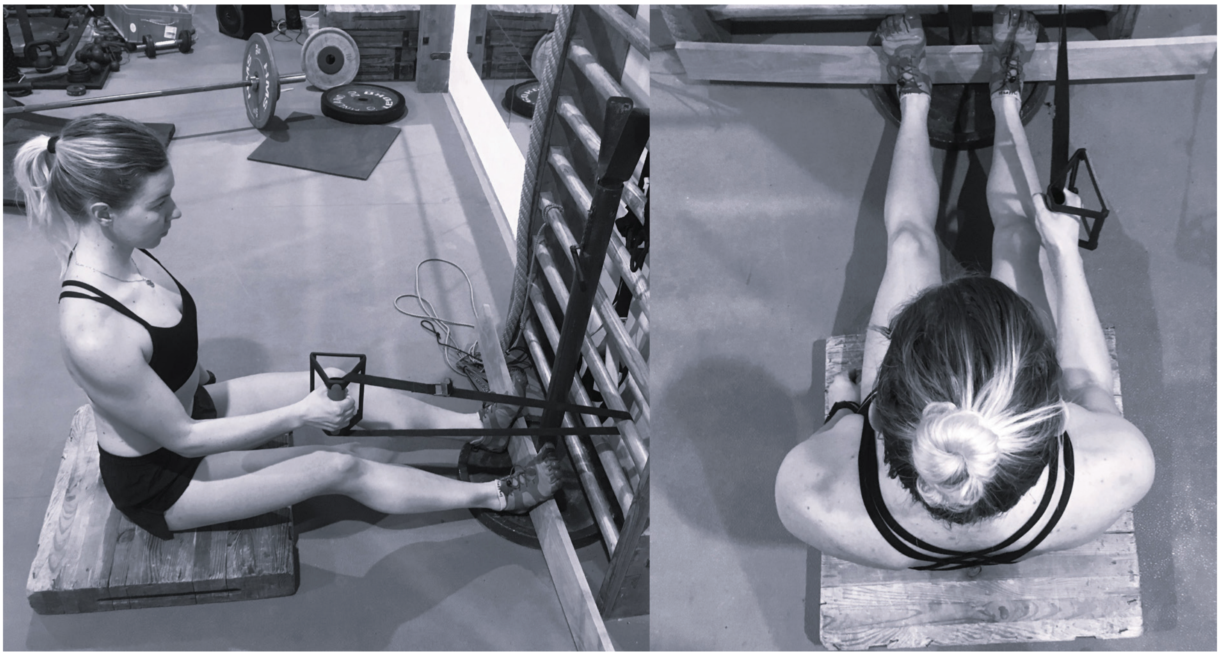

Figure 1. Experimental setup with seated position and force gauge handle

Prior to testing, each participant received instructions and performed a trial to familiarize themselves with the procedure. Electrode sites were prepared according to SENIAM guidelines — this included shaving and cleansing the skin with disinfectants. EMG electrodes were placed bilaterally over the external and internal oblique muscles (Figure 2), following anatomical landmarks described in previous studies (Bigolin et al., 2020; Huebner et al., 2015; Southwell et al., 2016). The external oblique electrodes were positioned over the lateral abdominal wall and aligned inferomedially with the presumed fibre direction. The internal oblique electrodes were positioned inferior and medial to the anterior superior iliac spine, above the inguinal region, and aligned superomedially with the presumed fibre direction. Inter-electrode distance was 10 mm (Trigno Avanti Sensor). Because the internal oblique is deeper and partly overlapped by adjacent abdominal muscles, its surface EMG signal was interpreted as regional activation rather than isolated muscle activity.

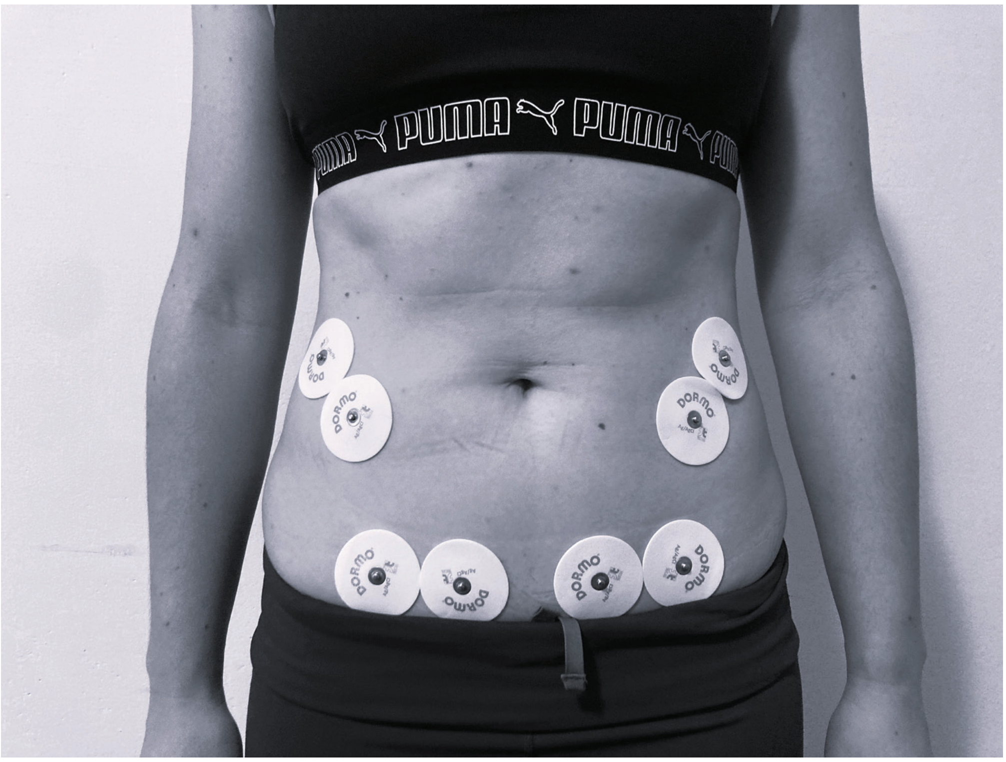

Figure 2. Illustrative example of EMG electrode placement over the external and internal oblique muscles.

Participants performed a static unilateral isometric pulling task in the seated position described above. They were instructed to apply maximal pulling force with one arm for approximately 3–5 seconds while minimizing any trunk movement. The procedure was then repeated with the other arm. Each trial was monitored by two experimenters—one operated the EMG system while the other managed the force meter and provided verbal instructions. After the final test, the electrodes were removed, and the setup was reset for the next participant.

Data processing

EMG data were initially processed in EMGworks® software. The raw signals underwent bandpass filtering (20–500 Hz) to remove movement artifacts and high-frequency noise, followed by notch filtering at 50 and 60 Hz to eliminate electrical interference. Subsequently, rectification was performed to convert all negative values to positive, and smoothing was applied using the root mean square (RMS) method. The RMS window length was approximately 136 ms, resulting in a post-processing signal sampling frequency of 15 Hz. The processed EMG data were exported to Excel and later visualized in SignalPlant 1.2. Here, an additional averaging (function averaging in SignalPlant 1.2) was applied for further smoothing without reducing the data resolution. Trials containing significant signal artifacts or loss were excluded from analysis (subjects 3, 9, and 11). Therefore, complete data from twelve subjects (n=12) were available for statistical evaluation.

Force data were directly exported from the force meter's software into Excel format. Upon inspection in SignalPlan 1.2, the data were determined to be artifact-free and required no additional preprocessing. Combined data of the first proband's pull force and muscle activity are displayed in Figure 3.

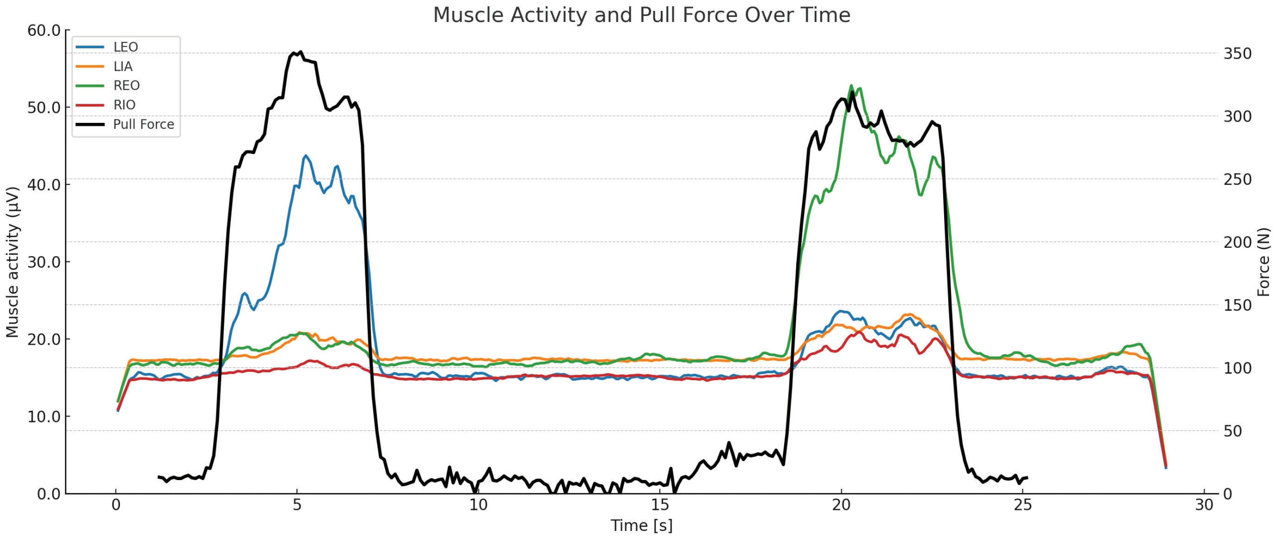

Figure 3. Graph of recorded muscle activity via EMG and pull force via Forcemeter

Statistical analysis

Statistical analysis was conducted using Statistica 14 software. Spearman's rank correlation coefficient was used to assess within-trial associations between pulling force and sEMG amplitude. Correlations were calculated separately for each participant, hand, and muscle using paired force–sEMG time-series values from the active pulling phase of each 3–5 s trial. Because force and processed sEMG signals had different sampling frequencies, both signals were manually time-aligned according to the onset of the active pulling phase and interpolated to a common frequency of 30 Hz before analysis. Participant-level correlation coefficients were then summarized using median and interquartile range values for ipsilateral and contralateral external and internal oblique muscles.

Results

To explore within-trial associations between pulling force and ipsilateral oblique muscle activity, Spearman correlation coefficients were calculated separately for each participant, hand, and muscle. Average values of pull force and muscle activity are shown in Table 1 for descriptive purposes. Median Spearman correlation coefficients were R=0.67 (IQR=0.36) for ipsilateral external oblique activity and R=0.64 (IQR=0.31) for ipsilateral internal oblique activity. These values indicate moderate-to-strong positive within-trial associations. Individual correlation coefficients are shown in Table 2.

Table 1. Average values of pull force and muscle activity

Proband	Right hand					Left hand				
	Pull force (N)	**LEO (μV)**	**LIO (μV)**	**REO (μV)**	**RIO (μV)**	**Pull force (N)**	**LEO (μV)**	**LIO (μV)**	**REO (μV)**	**RIO (μV)**
1	238.60	20.62	20.77	40.03	18.73	244.60	20.72	20.83	40.49	18.80
2	194.00	36.99	33.97	21.86	43.25	202.10	23.48	67.41	50.16	47.50
4	312.80	27.60	19.68	18.49	21.67	344.00	21.23	32.87	81.09	25.44
5	225.50	80.64	31.33	95.59	95.99	-	-	-	-	-
6	263.90	44.70	32.81	19.51	110.92	246.40	20.60	74.09	37.09	68.05
7	410.90	51.57	19.50	21.27	26.51	413.30	19.29	24.73	35.09	22.78
8	146.80	40.77	30.14	29.77	26.61	125.50	29.03	30.72	27.38	18.11
10	381.10	36.08	26.69	89.45	30.22	368.20	52.65	24.37	39.59	24.46
12	163.40	60.14	90.54	89.94	47.86	169.50	44.52	82.71	32.70	67.30
13	254.30	34.43	46.92	320.06	19.91	229.30	16.86	-	466.00	27.20
14	213.80	27.91	261.95	38.89	19.55	161.50	31.86	-	26.43	24.24
15	126.30	18.15	42.66	25.96	23.06	176.60	31.99	-	22.55	34.11

Note. LEO: Left external oblique; LIO: Left internal oblique; REO: Right external oblique; RIO: Right internal oblique.

Table 1. Average values of pull force and muscle activity

Proband	Hand	Spearman correlation coefficient			
		Ipsi EO	**Ipsi IO**	**Contra EO**	**Contra IO**
1	R	0.800	0.683	0.559	0.730
2	R	0.816	0.924	0.834	0.426
4	R	0.414	0.449	0.746	0.683
5	R	0.956	0.960	0.806	0.313
6	R	0.768	0.889	0.785	0.691
7	R	0.809	0.442	0.921	0.771
8	R	0.276	0.202	0.221	-0.010
10	R	0.312	0.560	-0.045	0.738
12	R	0.613	0.839	0.691	0.584
13	R	-0.099	0.493	0.964	0.934
14	R	0.919	0.500	0.935	-0.096
15	R	0.925	0.623	0.619	0.632
1	L	0.742	0.530	0.811	0.682
2	L	0.709	0.819	0.583	-0.029
4	L	0.854	0.771	0.874	0.172
6	L	0.626	0.823	0.953	0.663
7	L	0.817	0.940	0.913	0.799
8	L	0.358	0.597	0.639	0.425
10	L	0.669	0.655	0.660	0.111
12	L	0.634	0.594	0.122	0.851
13	L	0.418	-	0.226	0.591
14	L	0.489	-	0.594	0.520
15	L	0.513	-	0.449	0.635
Median (IQR)	-	0.67 (0.36)	0.64 (0.31)	0.69 (0.28)	0.63 (0.34)

Note. Ipsi: Ipsilateral; Contra: Contralateral, EO: External oblique; IO: Internal oblique.

For contralateral oblique muscles, median correlations were R=0.69 (IQR=0.28) for the external oblique and R=0.63 (IQR=0.34) for the internal oblique. These results similarly suggest moderate-to-strong within-trial associations between pulling force and contralateral oblique muscle activity. All results are displayed in Figure 4.

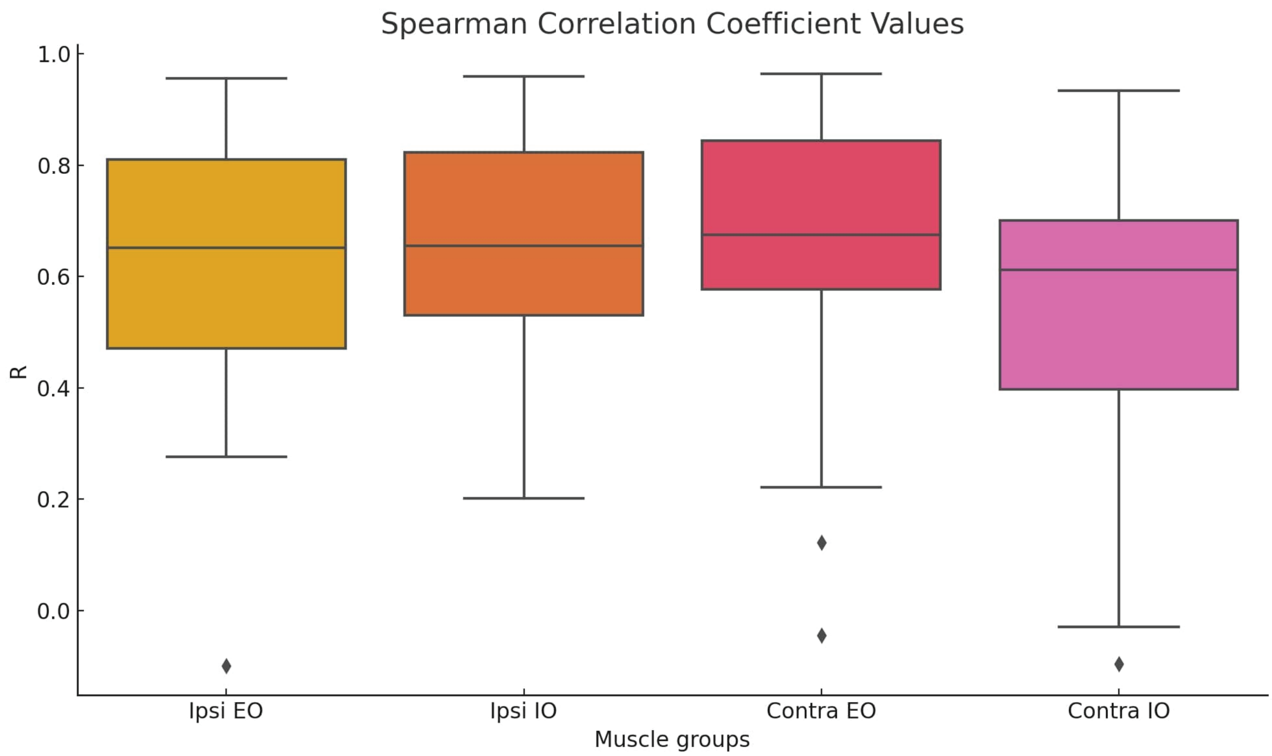

Figure 4. Graph of average correlation coefficients

Discussion

The main aim of this pilot methodological study was to examine whether a forcemeter-based unilateral isometric pulling task elicits measurable activation of the external and internal oblique muscles and whether force output covaries with sEMG amplitude during the task. The results suggest moderate-to-strong within-trial associations between pulling force and oblique muscle activation under the tested conditions. These findings provide preliminary physiological support for the task, but they should not be interpreted as validation of a trunk rotator strength test.

To assess the purpose, based on a review of the literature, an experiment was conducted on fifteen volunteers of both sexes. The subjects were introduced to the test protocol described in the methodological part of this work. During the test, the pulling force of each upper limb was measured in a defined position, and the simultaneous activation of the trunk stabilizers was measured. For clarity, the correlation rate was evaluated separately for ipsilateral and contralateral muscles.

The results showed that for the activity of the ipsilateral oblique abdominal muscles and the pulling force of the hands, the correlation is assessed as strong (R=0.67 (0.36) for external and R=0.64 (0.31) for internal). For the contralateral oblique muscles, the correlation is assessed as strong (R=0.69 (0.28) for external and R=0.63 (0.34) for internal). The slightly lower correlation rate in the case of the internal oblique muscle could be due to lower involvement of this muscle, or its deeper location and therefore more problematic recording with superficial electrodes.

The use of a forcemeter to assess muscle strength has recently been a frequent topic, due to the significantly lower acquisition and operating costs compared to a dynamometer and higher accuracy compared to field tests. This use was investigated, for example, in the work of Wałecka et al. (2023), where a forcemeter (sometimes referred to as an isometric dynamometer) was used to assess the strength of the shoulder girdle. The forcemeter proved to be a suitable tool for this measurement with high intra-rater reliability (r=0.66–0.93) and inter-rater reliability (r=0.72–0.91).

Similar reliability results were also demonstrated by the research of Karthikbabu and Chakrapani (2017). This research focused on the assessment of trunk muscles and showed an overall intra-rater (0.88–0.98) and inter-rater (0.84–0.96) reliability of trunk muscle strength testing using a hand-held forcemeter. In the case of rotators, the study showed an even higher reliability value (0.89–0.96).

These studies mainly tested the reliability and replicability of the tests, compared to our work, which aims to demonstrate the involvement of the given muscles, in this case the trunk rotators, with sEMG.

To use EMG for this purpose, it was necessary to take into account that the relationship between muscle strength and activity is quasi-linear, especially at higher muscle contraction intensity. However, the results showed a moderate to strong linear correlation between the dynamometer and sEMG. This result is probably caused by the selected muscles and the measurement method. Thanks to the sitting position and the pull always generated by one hand, the measured muscles function isometrically as stabilizers. The maximum effort and intensity are performed by the measured limb, due to the direction of pull mainly by the m. latissimus dorsi. Other trunk muscles are therefore involved later within the muscle chains (Page et al., 2010). Thanks to this chaining, the observed muscles will not reach the maximum possible contraction. As a result, motor units will not be recruited to activate the largest ones, which are mainly composed of type II fibres, and which cause an exponential increase in muscle activity. This clearly fits into the current description of the relationship between muscle strength and activity as quasi-linear (Kamen & Gabriel, 2010).

Another factor that could have caused a smaller activation of the trunk rotators is the fact that the pulling force is generated primarily by one muscle (with the help of other synergists). In contrast, the four oblique abdominal muscles are actively strengthened to stabilize the trunk in rotation, and the deep back muscles and the rectus abdominis will also play their role. Also, due to this distribution, the maximum activation of the measured muscles may not be reached.

The fact that, due to muscle chaining, the maximum activation of the trunk muscles does not occur brings a surprising advantage of this measurement technique. The investigated relationship is therefore more linear and allows for more accurate conclusions to be drawn for movements associated with the upper limb. A limitation of this measurement could occur in the case of large or rapid forces acting on the body (collisions in games or combat sports). In this case, a response would likely occur at the spinal level and the recruitment of motor units would be immediate and virtually maximal. The main difficulty of the whole work is the above-mentioned problem with artifacts and errors in EMG readings in general. This problem caused some probands to have to be excluded from the correlation table because their muscle activity was not recorded or could not be processed. The problem occurred despite compliance with all the principles according to the SENIAM project (Surface Electromyography for the Non-Invasive Assessment of Muscles). Based on this, it is assumed that the errors were not caused by any easily influenced factor.

One source of measurement error may have been Bluetooth signal interference caused by multiple nearby devices; using a wired EMG system could mitigate this by reducing signal disruption, offering more stable control, and avoiding electrode interference with clothing. Additionally, real-time signal monitoring and faster activation would allow pre-measurement signal checks without extending the experiment. Another refinement could involve measuring maximum voluntary contraction (MVC) before testing, enabling assessment of each muscle's activation relative to its maximum capacity. While this could clarify the nature of the force–activity relationship, accurately determining MVC during trunk rotations is challenging due to the simultaneous engagement of multiple trunk muscles.

Since there were no problems with the forcemeter or the data from it and the results were positive, it appears to be a suitable tool for further research. The present findings should be interpreted as methodological evidence that the proposed forcemeter-based task engages trunk rotators and yields consistent force–activation coupling; however, reliability (test–retest) and diagnostic performance for detecting imbalance were not evaluated and require dedicated validation studies.

Conclusions

This pilot methodological study suggests that a forcemeter-based unilateral isometric pulling task elicits measurable activation of the external and internal oblique muscles and that force and sEMG signals show within-trial association under the tested conditions. These findings provide preliminary physiological support for the proposed task. The protocol provides a practical foundation for future research assessing test–retest reliability and evaluating its potential utility for imbalance screening in athletes.

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Conflict of interest

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

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