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ORIGINAL ARTICLE

Differences in electromechanical delay components induced by sex, age and physical activity level: new insights from a combined electromyographic, mechanomyographic and force approach

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Abstract

Background Electromyographic (EMG), mechanomyographic (MMG) and force (F) signals combined analysis represents an interesting approach to partition the electrochemical and mechanical events contributing to total electromechanical delay, i.e., the time lag existing between the muscle activation and the onset of force generation.

Aim The study sought to assess the differences in electromechanical delay due to sex, age, and physical activity level.

Methods Electromechanical components were assessed on vastus lateralis muscle during a maximum voluntary contraction and electrically evoked contractions in 180 participants. During each contraction, the EMG, MMG and F signals were recorded. Electromechanical delays and its two components (Δt EMG-MMG, mainly electrochemical component; and Δt MMG-F, mainly mechanical component) were computed. Measurements' reliability (intraclass correlation coefficient, ICC) and sensitivity (minimum detectable changes at 95% confidence as a percentage, $MDC_{95\%}$) were also calculated.

Results ICC spanned from 0.89 to 0.97 with a percentage change of the standard error of the measurement (SEM%) ranging from 1.6 to 4.9%. $MDC_{95\%}$ values ranged between 3.1 and 9.8%. Longer electromechanical delay values were observed in: (1) women compared to men; (2) 40–45 years old compared to 30–35 years and 20–25 years; and (3) sedentary than active participants. Differences were accompanied by increments in Δt MMG-F but not in Δt EMG-MMG values.

Conclusions The alterations in the whole electromechanical delay induced by sex, age, and physical activity level could be ascribed to the difference in the duration of the mechanical events included in the electromechanical delay, possibly due to modifications in the muscle–tendon unit characteristics.

Keywords Maximum voluntary contraction · Electrically evoked contraction · Electromyography · Mechanomyography

Abbreviations

EMG	Surface electromyographic
MMG	Mechanomyographic signal
Δt	Time latency
Stim	Stimulation current
S	Sedentary
A	Active
M	Men
W	Women
MVC	Maximum voluntary contraction
pT	Peak torque
EMD	Electromechanical delay during voluntary contraction
$Delay_{TOT}$	Electromechanical delay during electrically evoked contraction

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Introduction

Electromechanical delay is usually defined as the time lag between the beginning of muscle activation and the force generation [1]. Within this period, several electrochemical and mechanical events are enclosed [1]. The electrochemical events include the excitation–contraction coupling, i.e., the transmission of the action potential generation at the sarcolemmal level to the cross-bridges formation, while the mechanical events include the time taken for tensioning the in-series elastic elements. Interestingly, the duration of the electromechanical delay could be influenced by physio-pathological phenomena. To date, differences between sex [2–4], age [2, 4, 5], training regime [6–10], or neuromuscular disorders [11, 12] have been documented. However, the role of electrochemical and/or mechanical events in explaining such differences still remains to be clarified.

Typically, the electromechanical delay is calculated as the time difference between the onset of the surface electromyographic (EMG) and force signals [1, 10]. In the last years, an approach including the mechanomyographic (MMG) signals permitted to discriminate the duration of the electrochemical and the mechanical events [13–15]. Indeed, MMG may be considered the first mechanical manifestation of acto–myosin interaction which provides changes in the muscle fibres geometry, mirrored by the muscle surface displacement, before the tensioning of the in-series elastic elements generating the active torque at the bone segment [14]. Therefore, this parameter permitted to partition the electromechanical delay into a first component (Δ EMG-MMG), mainly including the early electromechanical coupling events, and a second component (Δ MMG-F) mainly including the mechanical transmission of the tension impulse through the passive components of the muscle model [9, 13]. This combined EMG, MMG and force approach has been recently utilized to highlight alterations of the electrochemical and mechanical events during muscle contraction induced by fatigue [7, 16], muscle temperature manipulation [17], muscle stretching [8, 18, 19] and neuromuscular disorders, such as type 1 myotonic dystrophy [11, 12].

On these bases, the present study aimed to elucidate, by a combined EMG, MMG, and force approach, the role of the electrochemical and mechanical events in explaining possible differences in the electromechanical delay induced by sex, age and training level.

Methods

Recruitment

Candidates of both sexes and aged between 20 and 45 years were evaluated to participate in the study. Exclusion criteria were the presence of neurological, musculoskeletal or vascular impairments at lower limb level, or the presence of other pathologies that preclude the maximal activation of the leg extensor muscles under voluntary or electrically evoked conditions. Physical activity level was evaluated by fulfilling the International Physical Activity Questionnaire [20]. The candidates involved in at least three session/weeks of resistance training involving lower limb muscles (> 2520 METs) were considered as active (A), whereas candidates with a low level of physical activity (< 700 METs, as indicated by International Physical Activity Questionnaire score) were considered as sedentary (S). Two hundred and fifty-three candidates were evaluated and 180 were enrolled in the study. Participants were advised on the study aim and experimental design. After signing an informed consent, the participants were allocated in different groups based on their age (20–25 years; 30–35 years; and 40–45 years); sex (man, M; woman, W) and physical activity level (A; S). A sample of fifteen participants for each group was obtained (Table 1).

The study was conducted in accordance with the principles of the latest Helsinki Declaration upon receiving necessary approval from Milan University Ethics Committee.

Experimental design

The measurements were performed in a temperature- (22 ± 1 °C) and relative humidity-controlled room (50 ± 5%). The participants underwent three visits: the first was devoted to familiarization purposes. In this occasion, the participants experienced maximum voluntary (MVC) and electrically evoked contraction procedures. A map with some skin reference points (e.g., moles, scars, angiomas) together with the position of linear array for EMG signal detection and of the mono-axial accelerometer for MMG signal detection was drawn on a transparency sheet. This procedure was followed to permit the re-positioning of the instruments in the same muscle area during the second and third visit.

During the second and third visit, the participants were asked to avoid caffeine or similar substances in the 24 h preceding the tests and to avoid any resistance workout involving lower limb muscles in the 72 h prior to the tests. In both visits, the participants performed three leg extensors MVC of the dominant limb, interspersed by at least

Table 1 Anthropometric characteristics of the participants

	N	Age (years)	Body mass (kg)	Stature (m)
20–25 years_M_S	15	22 ± 1	72 ± 2	1.77 ± 0.05
20–25 years_M_A	15	21 ± 2	74 ± 2	1.78 ± 0.06
30–35 years_M_S	15	33 ± 1	75 ± 4	1.76 ± 0.04
30–35 years_M_A	15	32 ± 2	73 ± 4	1.77 ± 0.05
40–45 years_M_S	15	42 ± 2	76 ± 3	1.76 ± 0.06
40–45 years_M_A	15	42 ± 1	75 ± 5	1.77 ± 0.05
20–25 years_W_S	15	23 ± 1	61 ± 4	1.67 ± 0.04
20–25 years_W_A	15	22 ± 2	59 ± 5	1.69 ± 0.05
30–35 years_W_S	15	33 ± 1	62 ± 5	1.68 ± 0.04
30–35 years_W_A	15	32 ± 1	60 ± 4	1.67 ± 0.06
40–45 years_W_S	15	42 ± 2	62 ± 4	1.68 ± 0.04
40–45 years_W_A	15	42 ± 1	61 ± 3	1.67 ± 0.05

Data are expressed as mean ± standard deviation

5 min of passive recovery. Thereafter, vastus lateralis muscle was electrically stimulated for three times (5 min of recovery in between) with a supramaximal tetanic stimulation, at the proximal motor point level. The motor point position was identified by means of a pen-shaped stimulating electrode, finding on the muscle surface the point generating a visible contraction with the lowest stimulation amplitude.

Measurements and data analysis

Maximum voluntary contraction

The participants were seated on a purposely made ergometer for the knee extensors force assessment, with an angle of 90° at the hip, knee, and ankle level. The distal third of the dominant leg was secured to the metal support by Velcro® straps. The whole apparatus had a resonant frequency > 200 Hz. The metal plate and the leg support were both connected to a calibrated load cell (SM-2000N; operating linearly between 0 and 2000 N; Interface, Crowthorne, UK) for the detection of force signal, which was then amplified (gain: × 200; mod. UM150, Biopac System, Santa Barbara, CA, USA) and successively driven to the auxiliary input of the EMG amplifier, where it was acquired at a sample frequency of 10,240 Hz. MVC was identified as the highest force recorded during the three maximum contractions of 3 s.

Electrically evoked tetanic contraction

Neuromuscular electrical stimulation was delivered to vastus lateralis in the monopolar technique by an electrical stimulator (mod. St-Pro Multichannel Programmable Neuromuscular Stimulator, LISiN, Turin, Italy). A receiving electrode was positioned at the third proximal of the thigh and the stimulating electrode was placed over the most proximal motor point of the muscle. A set of brief 2-Hz stimulations of increasing amplitude was provided to determine the maximum compound motor unit stimulus. After the stimulus eliciting the maximal peak-to-peak M-wave was identified, the participants rested for 5 min. Then, a set of three tetanic stimulations, consisting of a train of pulses (wave shape: biphasic; pulse duration: 304 µs; stimulation frequency: 50 Hz; current amplitude: 110% of the maximum compound motor unit stimulus; duration: 2 s) was delivered. During the stimulations, the participants were instructed to maintain the lower limb muscles as relaxed as possible. Further, the vastus lateralis electrically evoked peak torque (pT) was assessed as the highest torque recorded during the three electrically evoked contractions.

EMG and MMG signals

The surface EMG and MMG signals were detected during the contractions and acquired by a multichannel amplifier (mod. EMG-USB, OtBioelettronica, Turin, Italy; input impedance: > 90 MΩ; CMRR: > 96 dB; EMG and MMG bandwidth: 10–500 and 4–120 Hz, respectively; gain: × 1000 and × 2 for EMG and MMG, respectively), with a sampling rate of 10,240 Hz. The EMG signal was detected by a linear array of four electrodes (mod. ELSCH004, OtBioelettronica, Turin, Italy; probe 45 mm × 20 mm; electrode length 2 mm; inter-electrode distance 10 mm), fixed to the skin by dual-adhesive foam (mod. AD004, OtBioelettronica, Turin, Italy) and filled with conductive gel (Cogel, Comedical, Trento, Italy). The skin area under the EMG electrodes was cleaned with ethyl alcohol, abraded gently with fine sand paper and prepared with a conductive cream (Nuprep, Weaver and Co., Aurora, USA) to achieve an inter-electrode impedance below 2000 Ω. The third electrode of the EMG array was removed and replaced by a mono-axial accelerometer (mod. ADXL103, Analog Devices, Norwood, MA, USA; device weight: < 1.0 g; sensitivity: 1000 mV/g; measure range: ± 1.7 g) for MMG detection from the same muscle area as EMG. The EMG array was placed over the vastus lateralis muscle belly near the point of maximum skin displacement during contraction, along the direction of the muscle fibres, with the EMG electrodes positioned perpendicular to the major axes of the fibres between the tendon and the motor point, in accordance with the European recommendations for surface EMG [21]. A supplementary EMG linear array (mod. ELSCH004, 45 mm × 20 mm; electrode 2 mm × 1 mm; inter-electrode distance 10 mm; OtBioelettronica, Turin, Italy) monitored the lack of activation of the antagonist muscles (biceps femoris muscle) during contractions.

Electromechanical delays determination

Delays' identification in MVC and in the electrically elicited contraction in a representative participant is given in Fig. 1.

The signals acquired (EMG, MMG, and force) were analysed offline by a custom-built routine of a commercially available software (Labview 7.1. National Instruments, Austin, TX, USA).

The criteria used to identify the reference points for delays components have been fully reported in previous investigations [7, 11, 13, 17, 18].

During MVC, a condition of three standard deviations from the mean baseline noise measured over a time window of 100 ms was set for detecting the onset of each signal. Δt EMG-MMG (mainly electrochemical component) was calculated from the onset of EMG to the onset of the MMG signal. Similarly, Δt MMG-F (mainly mechanical component) was calculated as the time lag between MMG signal

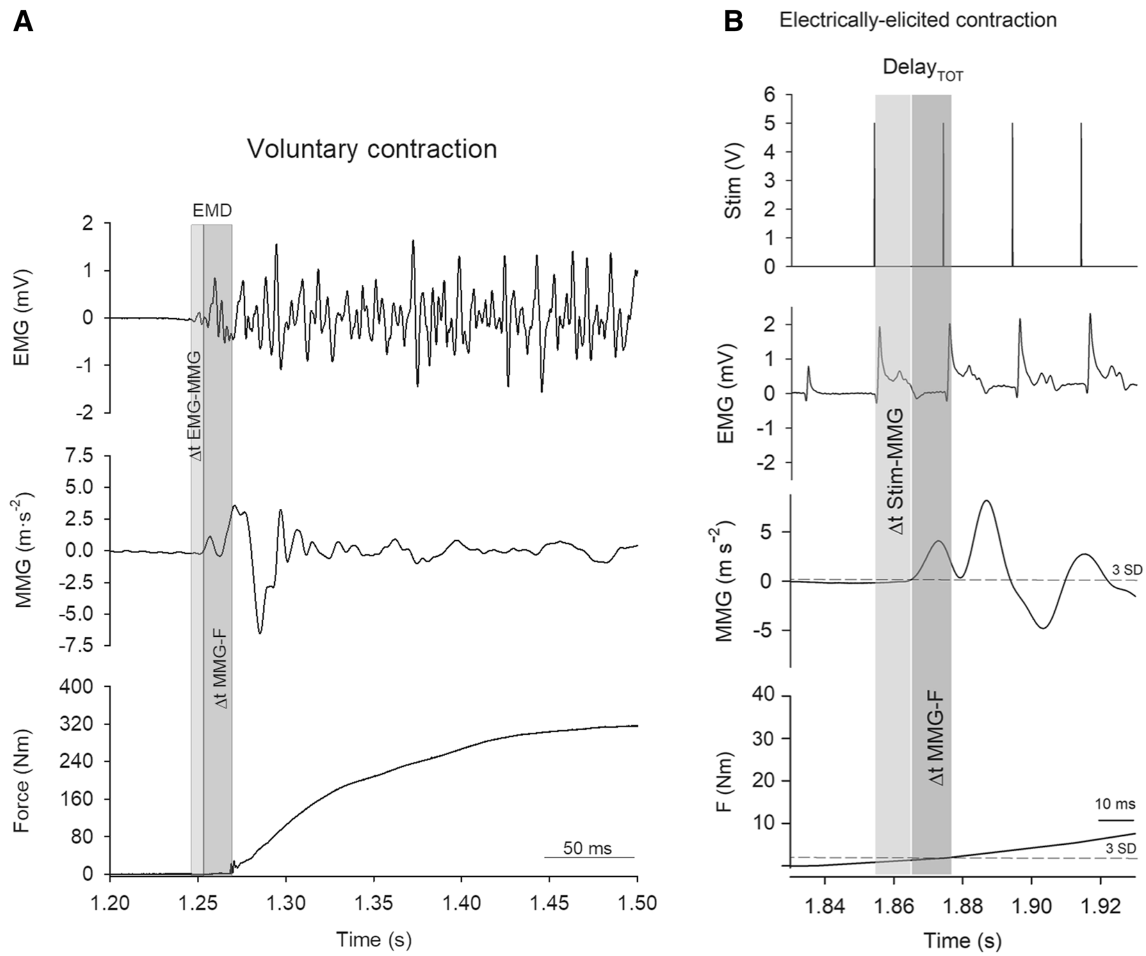

Fig. 1 **a** Voluntary contraction: electromyographic (EMG), mechanomyographic (MMG), force signals, and delays in a representative participant. Light-grey bar represents Δ*t* EMG-MMG, while grey bar represents Δ*t* MMG-F, respectively. The sum of the two components identifies the electromechanical delay (EMD). **b** Electrically elicited contraction: stimulation (Stim), EMG, MMG and force signals and delays in a representative participant. Light-grey bar represents Δ*t* Stim-MMG, and grey bar represents Δ*t* MMG-F, respectively. The sum of the two components identifies the total electromechanical delay ($Delay_{TOT}$). The dashed lines represent three standard deviations (SD) from the mean baseline noise measured over a time window of 100 ms. For delays explanation, see "Methods"

onset and the force signal. The sum of the components is defined as electromechanical delay during voluntary contraction (EMD).

Since the M-wave onset is not clearly identifiable during the electrically evoked contraction, the mainly electrochemical component was calculated from the first positive peak of the recorded Stim signal to the onset of the MMG signal (three standard deviations from the mean baseline noise measured over a time window of 100 ms, for three consecutive points). This component was called Δ*t* Stim-MMG. Δ*t* MMG-F was calculated as the time lag between MMG signal onset and the force signal (three standard deviations from the mean baseline noise measured over a time window of 100 ms, for three consecutive points). The sum of the two components is identified as electromechanical delay during electrically evoked contraction ($Delay_{TOT}$). As reported in previous studies [11, 13], acquiring the EMG, MMG and force signals at 10,240 Hz permits to highlight possible significant differences between delay components that otherwise would be missed.

Statistical analysis

Raw data were analysed using a statistical software package (IBM SPSS Statistics v. 24, Armonk, NY, USA). To check the normal distribution of the sampling, a Shapiro–Wilk's test was applied. Based on previous investigations [11, 12, 16], a sample size of 15 participants was selected to ensure a statistical power higher than 0.80, with a type 1 error <0.05. A three-way (sex × age × activity level) analysis of variance (ANOVA) was used to determine possible differences among groups. The location of possible differences was assessed by a Bonferroni post hoc test. The level of significance was set at $\alpha<0.05$. To determine inter-sessions reliability between values, the intraclass correlation coefficient (ICC) and percentage change of the standard error of the measurement (SEM%) were calculated. The ICC was interpreted as follows: > 0.90: very high; 0.89–0.70: high; 0.69–0.50: moderate [22]. The sensitivity in detecting the effects of fatigue in the two groups was checked by calculating the minimum detectable change at 95% confidence as a percentage ($MDC_{95\%}$). Unless otherwise stated, the results are expressed as mean ± standard deviation (SD).

Results

Reliability and sensitivity

Table 2 reports the inter-sessions reliability and level of sensitivity in the different electromechanical delays' components and in force. ICC spanned from 0.890 to 0.961 with SEM% ranging from 1.9 to 4.9%. $MDC_{95\%}$ values were comprised between 3.6 and 9.8%.

Maximum voluntary contraction

Mean MVC values for the different groups and between-group comparisons are reported in Fig. 2, upper panels.

ANOVA disclosed the significant main effects in MVC for sex ($F = 6778$; $P < 0.001$), age ($F = 79.57$; $P < 0.001$), and activity level ($F = 771$; $P < 0.001$). Interactions were found between sex and age ($F = 21.54$; $P < 0.001$); sex and activity level ($F = 33.82$; $P < 0.001$); and age and activity level ($F = 3.97$; $P = 0.021$). Interactions between sex, age and activity level were also retrieved ($F = 5.14$; $P = 0.007$).

Peak torque

The mean values for pT in the different groups and between-group comparisons are represented in Fig. 3, lower panels. Main effects in pT were observed for sex ($F = 5747$; $P < 0.001$); age ($F = 92.62$; $P < 0.001$); and activity level ($F = 1152$; $P < 0.001$). Interactions between sex and age ($F = 30.52$; $P < 0.001$) and sex and activity level ($F = 9.72$; $P < 0.001$) were found. An interaction between sex, age and activity level was also retrieved ($F = 15.15$; $P < 0.001$).

Electromechanical delay components during voluntary contraction

Figure 4 demonstrates the electromechanical delay components calculated during MVC.

Main effects for sex and age were found in Δt EMG-MMG ($F = 6.80$; $P = 0.010$ and $F = 3.14$; $P = 0.046$, respectively). Main effects for sex ($F = 57.56$; $P < 0.001$), age ($F = 4.24$; $P = 0.016$), and activity level ($F = 24.73$; $P < 0.001$) were found in Δt MMG-F. Significant interactions between sex and age ($F = 3.16$; $P = 0.045$), sex and activity level ($F = 6.80$; $P = 0.010$), and age and activity level ($F = 3.54$; $P = 0.031$) were also retrieved.

ANOVA disclosed significant main effects for sex ($F = 26.64$; $P < 0.001$); age ($F = 3.70$; $P = 0.027$); and activity level ($F = 11.09$; $P = 0.001$) in EMD. A significant interaction between sex and activity level was also retrieved ($F = 7.03$; $P = 0.009$).

Electromechanical delay components during electrically evoked contraction

Electromechanical delay components calculated in electrically evoked contractions in the different groups are shown in Fig. 4.

Main effects for sex ($F = 4.66$; $P = 0.034$) and activity level ($F = 12.97$; $P < 0.001$) were found in Δt Stim-MMG. Main effects for sex ($F = 180$; $P < 0.001$); age ($F = 18.82$; $P < 0.001$); and activity level ($F = 70.21$; $P < 0.001$), together with an interaction between sex and activity level ($F = 11.04$; $P = 0.001$) were disclosed in Δt MMG-F. $Delay_{TOT}$ evidenced main effects for sex ($F = 65.55$; $P < 0.001$), age ($F = 9.00$; $P < 0.001$), and activity level ($F = 37.64$; $P < 0.001$).

Table 2 Intersession reliability and sensitivity of measurements

	Session 1 (m ± DS)	Session 2 (m ± DS)	ICC	SEM%	$MDC_{95\%}$
Δt Stim-MMG (ms)	9.33 ± 0.94	9.22 ± 1.02	0.961	1.9	3.6
Δt MMG-F (ms)	12.62 ± 1.11	12.75 ± 1.73	0.921	3.2	6.2
$Delay_{TOT}$ (ms)	21.95 ± 2.53	21.97 ± 2.13	0.890	3.5	7.0
Δt EMG-MMG (ms)	7.94 ± 1.13	8.07 ± 1.02	0.911	4.0	7.9
Δt MMG-F (ms)	13.69 ± 1.18	13.83 ± 1.41	0.925	2.6	5.1
EMD (ms)	21.63 ± 2.31	21.89 ± 2.34	0.892	3.5	7.0
MVC (Nm)	268 ± 11	270 ± 74	0.901	4.9	9.8
pT (Nm)	72 ± 19	72 ± 3	0.923	4.1	8.2

ICC intraclass correlation coefficient, *SEM%* percentage standard error of measurement, $MDC_{95\%}$ minimum detectable change at 95% of coefficient interval

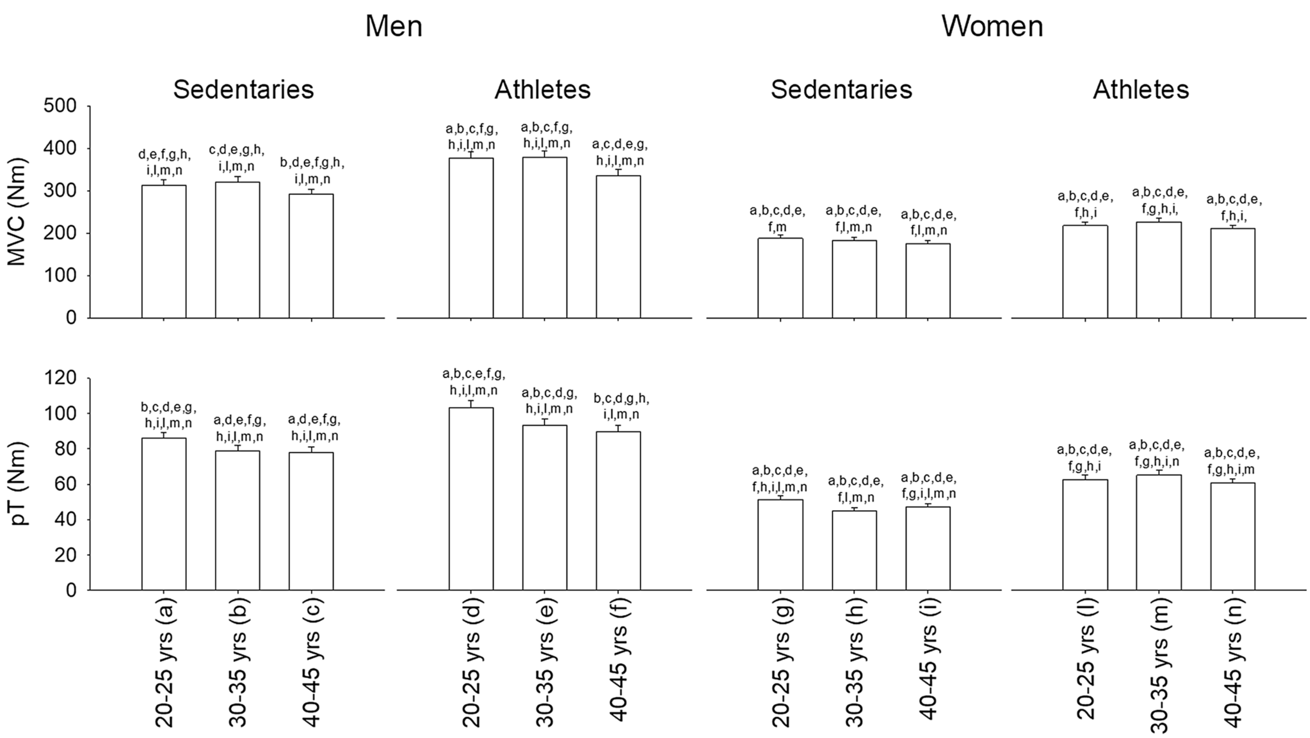

Fig. 2 Maximum voluntary contraction (MVC, upper panels), and peak torque (pT, lower panels) in the different groups. Data are expressed as mean ± standard deviation. Letters indicate the differences between groups with $P<0.05$

Discussion

The aim of this study was to investigate the possible contribution of the electrochemical and mechanical events involved in modulating the differences in electromechanical delay induced by sex, age and physical activity level. As expected, differences between groups induced by sex, age, and activity level were retrieved. Specifically, EMD and $\text{Delay}_{\text{TOT}}$ were (1) longer in women than men; (2) increased with age, and (3) decreased in individuals chronically involved in lower limb resistance training. These differences seem to be due to alterations in the mechanical rather than electrochemical components of the electromechanical delay. Together with the differences in EMD and $\text{Delay}_{\text{TOT}}$, differences in MVC and pT were also described for sex, age and physical activity level.

Methodological considerations

In the current study, the electromechanical delay components were evaluated during MVC and electrically evoked contractions. The purpose of investigating both contraction regimes was to assess the influence of central/cortical activation on both mainly electrochemical and mainly mechanical components of the total electromechanical delay. Indeed, it is known from the literature that both age and physical activity can influence the cortical activation level [23], and, in turn, the ability of activating a muscle: on one hand, lower level of cortical activations was reported in elderly compared to young people in non-fatigued and fatigued conditions [24]. On the other hand, a larger and prompt cortical activation was found in master athletes compared to sedentary people [25]. Differences in cortical activation level might influence the duration of the electrochemical events of electromechanical delay. Possible differences in Δt EMG-MMG could have been observed between MVC and electrically evoked contraction. However, when examining the EMD and $\text{Delay}_{\text{TOT}}$, the present findings did not highlight such a difference between the two modalities. This might suggest that the cortical activation may have played only a marginal role.

The previous investigations that have focused on the $\text{Delay}_{\text{TOT}}$ divided it into three components: (1) from Stim to EMG signal onset (Δt Stim-EMG), (2) from EMG to MMG onset (Δt EMG-MMG), and (3) from MMG to force signal onset (Δt MMG-F) [11, 13, 16]. The authors considered these components as suggestive of the duration of the mechanisms occurring at synaptic, electrochemical, and mechanical levels, respectively [11, 13, 16]. In particular, the marker used to identify the Δt Stim-EMG and Δt EMG-MMG in the EMG signal was the beginning of the stimulation artefact. Acknowledging that this marker could not properly identify the onset of the EMG activity, and given

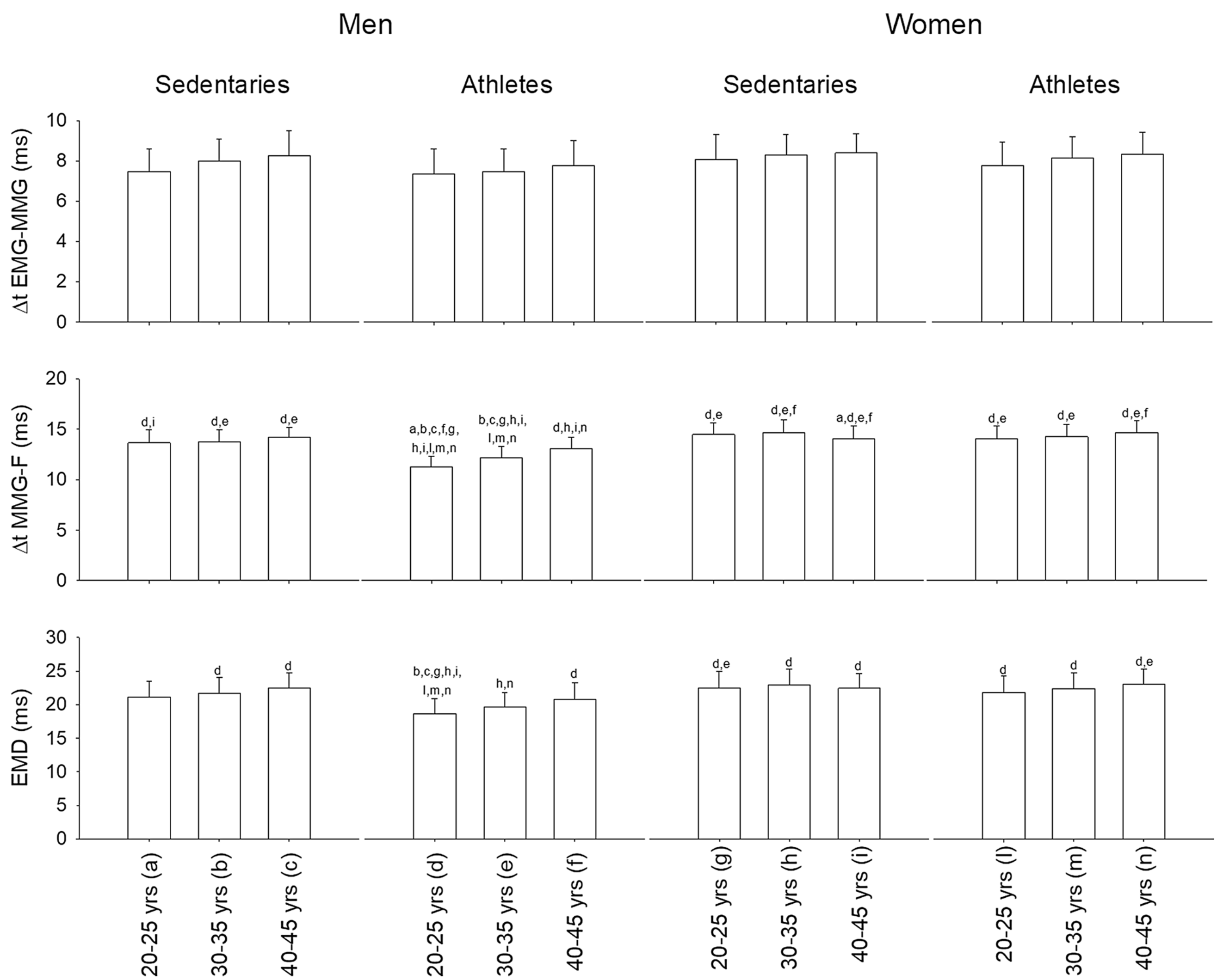

Fig. 3 Electromechanical delays components calculated during maximum voluntary contraction. Data are reported as mean ± standard deviation. Letters indicate the differences between groups with $P < 0.05$

that the M-wave very often overlapped with the stimulation artefact, the stimulation artefact was used because it was the only one that was detectable accurately in each subject (see Fig. 1b). The overlapping of the stimulation artefact with the M-wave was probably due to the very short distance between the EMG and stimulation electrodes. It should be taken into account that, within this experimental setup, the placement of the EMG electrodes and of the accelerometer for the MMG signal must meet two criteria: (1) being in a portion of the muscle between the motor point and the tendon, parallel to muscle fibers direction; (2) being on the part of the muscle showing the highest displacement to optimize the detection of the MMG signal [11, 13, 16]. This often leads to not permitting a clear distinction of the M-wave from the stimulation artefact. On the contrary, positioning the accelerometer in a position not reflecting the maximum displacement during the contraction could seriously affect the detection of the onset of muscle activation, thus not permitting the determination of the main electrochemical components of $Delay_{TOT}$. Our first intent in this study was measuring both Δt Stim-EMG and Δt EMG-MMG separately. However, since no between-group difference emerged in both parameters, and acknowledging the limitations imposed by the use of the stimulation artefact, we decided to combine all components into the Δt Stim-MMG, i.e., from stimulation signal to the onset on the MMG signal. This indicates the whole period taken by the electrochemical mechanisms involved in muscle activation.

Reliability and sensitivity

As previously reported, the combined EMG, MMG and force approach presented adequate levels of sensitivity in detecting the differences induced by sex, age, and physical

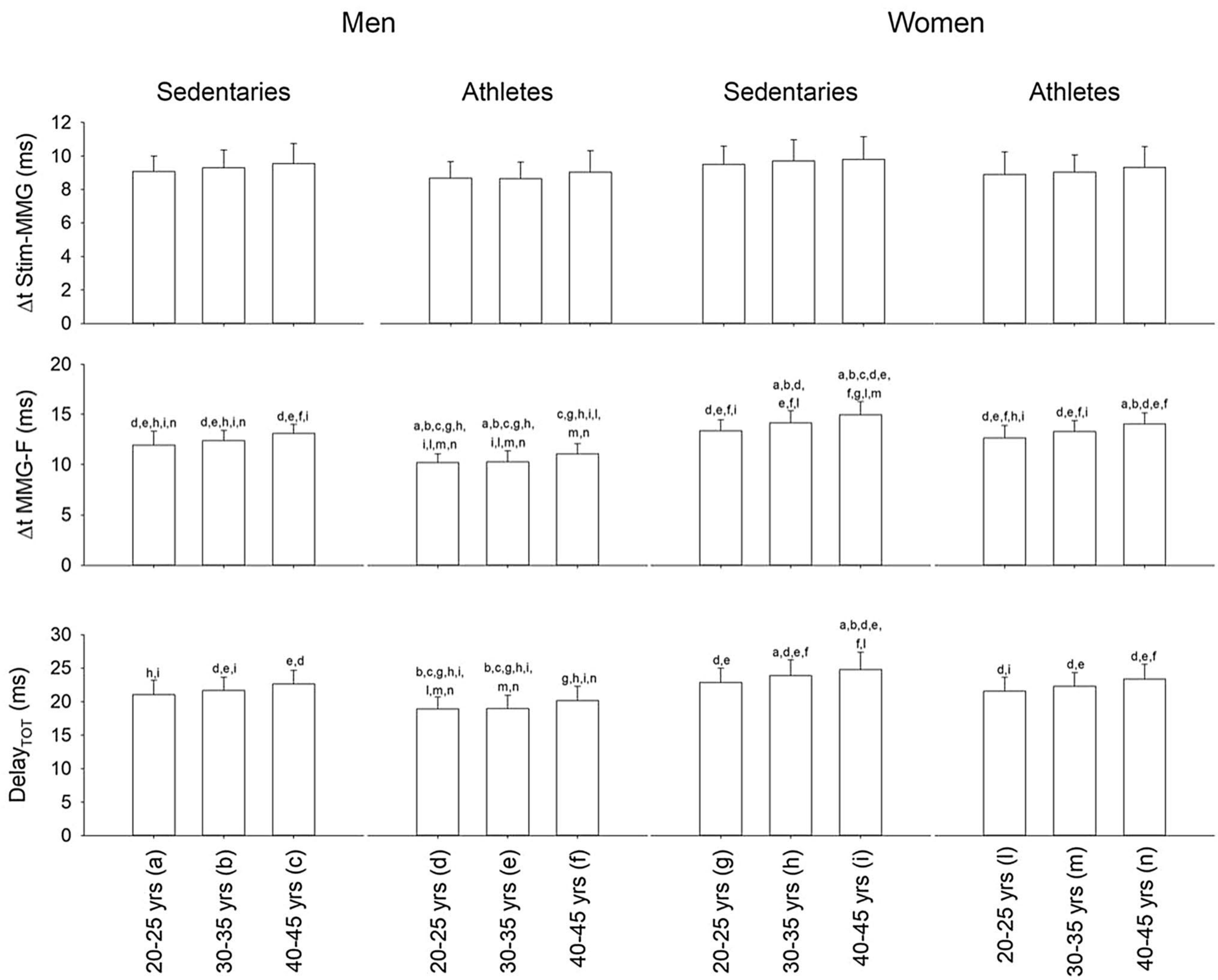

Fig. 4 Electromechanical delays components calculated during the electrically evoked contraction. Data are given as mean ± standard deviation. Letters indicate the differences between groups with $P < 0.05$

activity level [6, 11, 13, 18]. Indeed, this approach could be utilized as a biomarker to monitor the effects of a specific training in ameliorating the ability of activate a muscle, or to follow the effect of specific interventions in maintaining an adequate level of muscle activation with age.

Effect of sex on electromechanical delay components

EMD and $\text{Delay}_{\text{TOT}}$ were longer in women than men. Such a difference is mainly supported by an increase in Δt MMG-F (mainly mechanical component of the electromechanical delay) rather than an increase in Δt EMG-MMG or Δt Stim-MMG (mainly electrochemical component of the electromechanical delay during voluntary and electrically elicited contraction, respectively). These findings suggest that the force transmission in women could be slower than in men, probably due to differences in muscle–tendon unit stiffness. Indeed, the estrogenic hormones have an important role in the collagen synthesis, which might directly lead to a reduction in the tendon stiffness [3, 26]. This may support a slower force transmission at tendon insertion point, and, therefore, the lengthening in the mechanical component of electromechanical delay. Additionally, in a previous investigation [18], strong negative correlations between Δt EMG-MMG and force level and Δt MMG-F and force level were reported, supporting the notion that the level of force output could influence the duration of the electrochemical and mechanical events included in the electromechanical delay.

The present results partially agree with some studies [27–29], but contrast with other previous investigations, in which no significant difference was retrieved in the whole duration of the electromechanical delay between sexes [4, 30–32]. The different sample size and methods used to identify the electromechanical delay duration could explain the differences to the present study.

Effect of age on electromechanical delay components

Irrespective of the contraction regime, the present study evidenced a significant lengthening in the electromechanical delay with the increase in age. The main contributors to such a lengthening seem to be an elongation of the mechanical rather than the electrochemical component. These results are in agreement with a previous investigation, in which a longer electromechanical delay was found in 41–60 years compared to 18–40-year-old people [4]. In literature, it is reported that the electromechanical delay increases with age, especially above 40 years of age [4]. Changes in the muscle–tendon unit junction's viscoelastic properties together with a slackening of the tendon may occur with age, thus potentially lengthening the electromechanical delay [4]. Evidence of a remodulation in the muscle–tendon unit properties came from a previous work, reporting that the tendon structure in older people is less stiffer than young people [2]. Interestingly, a study including children and teenagers denoted that the electromechanical delays start to decrease from the age of 7–11 reaching the minimum at 16–18 years [5]. Thereafter, an increase in force transmission capacity at tendon insertion point seems to increase with age until about 18 years old [5]. This seems to be followed by a possible plateau close to 25 years; then, if not adequately counteracted with physical activity, a physiological decrease in force transmission efficiency occurs [4].

Influence of physical activity level on the electromechanical delay components

The present findings evidenced shorter electromechanical delays in active compared to sedentary people. Specifically, such differences seem to be ascribed to an increased efficiency in force transmission as witnessed by a shorter Δt MMG-F observed in active participants. In line, a stiffening of the muscle–tendon unit has been described in literature after several models of resistance training [33]. Results agree with some previous studies [3, 34] but are in contrast with other previous investigations [35, 36]. Differences in the participants' activity level and numerosity as well as in methods used to detect the electromechanical delay make an accurate comparison challenging.

Study limitations

The present work comes with some known limitations. First, the age range considered included only youth and adulthood but not elderly. This could limit the generalizability of the outcomes coming from this study. Nevertheless, the EMG, MMG and force combined approach presented a sensitivity high enough to disclose differences in electromechanical delay components even in a limited age range. Second, although a sample of 15 participants per group satisfied the prerequisite of a statistical power > 0.80, a larger sample size could have better clarified the interpretation of some results, such as Δt EMG-MMG, in which only sporadic and non-systematic differences between groups have been retrieved. Lastly, when stimulating on the motor point, it is difficult to clearly distinguish the onset of the M-wave, since this often overlapped with the stimulation artefact. The EMG onset assessment may influence the exact duration of the first component of $\mathrm{Delay}_{\mathrm{TOT}}$ (Δt Stim-MMG). In the present study, we chose to selectively stimulate only the vastus lateralis muscle instead of the femoral nerve, to reduce possible bias on the force signal onset induced by the activation of the other knee extensor muscles.

It was decided not to remove the stimulation artefact to avoid partial EMG signal cancellation due to artefact suppression [37]. Therefore, the choice of the stimulation artefact as the beginning of muscle electrical activation, which was adopted also in many other investigations [38–41] may slightly anticipate the EMG onset and, therefore, introducing an overestimation of the duration of Δt Stim-MMG.

Conclusions

The present outcomes suggest that the differences in the whole electromechanical delay induced by sex, age, and physical activity level could be due to the alterations in the duration of the mechanical events involved in this process, possibly depending on specific modifications in the muscle–tendon unit characteristics. In the light of these findings and of the very high level of reliability, the combined EMG, MMG and force approach could represent a valid tool to detect possible changes in the muscle mechanical behaviour induced by a specific stimulus, for example, resistance training.

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Compliance with ethical standards

Conflict of interest The authors declare no conflicts of interest.

Ethical approval The study was conducted in accordance with the principles of the latest Helsinki Declaration upon receiving necessary approval from Milan University Ethics Committee (CE 27/17 11-07-2017).

Informed consent Informed consent was obtained from all individual participants included in the study.

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