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Maximal contraction methods influence the magnitude and reliability of global electromyographic signal characteristics

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ABSTRACT

Objective: The purpose was two-fold: (1) to examine differences in maximal voluntary isometric torque (MVIT) production, and electromyographic signal amplitude (EMG_{AMP}) and mean power frequency (EMG_{MPF}) values obtained during traditional ($MVIC_{TRAD}$), rapid ($MVIC_{RAPID}$), and ramp ($MVIC_{RAMP}$) maximal voluntary isometric contractions, and (2) to determine if there were differences in the reliability of MVIT, EMG_{AMP} and EMG_{MPF} among the three MVIC types.
Approach: Twenty-two young males and females completed $MVIC_{TRAD}$, $MVIC_{RAPID}$, and $MVIC_{RAMP}$ muscle actions on two separate visits separated by 48 h. During all MVICs, MVIT and EMG_{AMP} and EMG_{MPF} of the vastus lateralis (VL) and rectus femoris (RF) were quantified.
Main results: MVIT was greater during $MVIC_{TRAD}$ and $MVIC_{RAPID}$ than during $MVIT_{RAMP}$ (both $p < 0.001$). VL and RF EMG_{AMP} were greater during $MVIC_{RAMP}$ than during $MVIC_{RAPID}$ ($p = 0.02$ and 0.004). For EMG_{MPF}, there were no significant differences among MVIC types. Although all MVIC types generally resulted in reliable measurements of MVIT and EMG_{AMP}, reliability was stronger for EMG_{MPF} quantified during the $MVIC_{RAMP}$.
Significance: Investigators may choose MVIC type based on preference or equipment availability. However, investigators should note that $MVIC_{RAMP}$ contractions will likely yield the greatest EMG_{AMP} values and more reliable measurements of VL and RF EMG_{MPF}.

1. Introduction

Researchers commonly evaluate changes in electromyographic signal amplitude (EMG_{AMP}) and mean power frequency (EMG_{MPF}) in response to exercise interventions (Hill et al., 2018, Jenkins et al., 2016), changes in joint torque production (Bampouras et al., 2017, Herda et al., 2015), or fatigue (Bilodeau et al., 2003, Burnley et al., 2012). EMG_{AMP} is thought to contain information regarding the number of active motor units and their discharge rates (Muddle et al., 2018, Sterczala et al., 2018, Yao et al., 2000) and is roughly related to the force that is exerted by the underlying muscle. Consequently, as muscle force production increases, EMG_{AMP} also increases in a linear or non-linear manner (Beck and Housh, 2008). Therefore, EMG_{AMP} is often used to help assess the degree of muscle activation in active (Fuglevand et al., 1993, Trevino et al., 2019) or passive states (Herda et al., 2008, Palmer et al., 2014), the degree of co-activation among muscles (Carr et al., 2018, Contessa et al., 2018), and/or to assess changes in muscle activation in response to an intervention (Herda et al., 2011, Hill et al., 2018, Jenkins et al., 2017). The analysis of the EMG power density spectrum is commonly performed by quantifying EMG_{MPF} (Kamen and Caldwell, 1996). It has been suggested that EMG_{MPF} contains information regarding muscle fiber conduction velocity and motor unit discharge rates (Beck et al., 2004). However, due to the fact that the EMG signal is influenced by intracellular action potential shapes, the number of active motor units, muscle fiber shortening, and the volume conductor, interpretation of EMG_{AMP} and EMG_{MPF} values is not straight-forward, although collection of signals under controlled conditions (e.g., isometric, stationary signals) and proper normalization may enhance interpretability (Beck and Housh, 2008, Farina et al., 2004, Kamen and Caldwell, 1996).

Accordingly, the International Society of Electrophysiology and Kinesiology (ISEK) has endorsed standards for the reporting of EMG data (Merletti, 1999). These standards state that both torque and EMG data should be normalized relative to the values obtained during a maximal voluntary isometric contraction (MVIC). Consequently, previous studies which report EMG data have frequently normalized to the

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values obtained during a maximum voluntary isometric contraction (MVIC) and report relative EMG_{AMP} and EMG_{MPF} values (Booghs et al., 2012, da Silva et al., 2017, Jenkins et al., 2015, Jenkins et al., 2016, Maluf et al., 2005). However, as also noted in the standards for reporting of EMG data, obtaining the best estimates of maximum voluntary isometric torque (MVIT) and global EMG signal characteristics using MVICs requires familiarization, and can be performed in different conditions and with slightly different instructions. For example, it is not uncommon for participants to be instructed to kick out "as hard and as fast as possible" (Maluf et al., 2005), "as hard as possible" (Smith et al., 2016, Smith et al., 2017), "as fast as possible" (Plautard et al., 2015, Sahay et al., 2001), or to kick out while slowly increasing the amount of force they are producing until they are contracting as hard as possible (Park and Hopkins, 2012, Tomko et al. 2018a).

In a recent investigation, Tomko et al. (2018a) examined whether completing maximum isometric contractions while slowly ramping torque to a maximal level ($MVIC_{RAMP}$) resulted in the attainment of different EMG_{AMP} and EMG_{MPF} values than when maximal isometric contractions were simply performed with torque produced 'as hard and as fast as possible' ($MVIC_{TRAD}$). The authors (Tomko et al., 2018a,b) reported that EMG_{AMP} values were in fact greater during the $MVIC_{RAMP}$ than $MVIC_{TRAD}$ and suggested $MVIC_{RAMP}$'s may be the best method for estimating maximal EMG_{AMP}. However, it is unclear whether $MVIC_{RAMP}$ contractions result in more reliable assessments of the global EMG characteristics (e.g., EMG_{AMP} and EMG_{MPF}). An understanding of potential differences in reliability of these variables is critical when choosing the best MVIC method for normalization purposes, especially for studies using a repeated-measures design. Moreover, some investigators have also used maximal isometric contractions in which they instructed their participants to contract "as fast as possible" ($MVIC_{RAPID}$) (Plautard et al., 2015, Sahay et al., 2001). Therefore, the purpose of this investigation was two-fold: (1) to examine differences in MVIT production, and EMG_{AMP} and EMG_{MPF} values obtained during $MVIC_{TRAD}$, $MVIC_{RAPID}$, and $MVIC_{RAMP}$ muscle actions, and (2) to determine if there were differences in the reliability of MVIT, EMG_{AMP} and EMG_{MPF} measurements among these three MVIC types. Based on the study by Tomko, Colquhoun (2018a), we hypothesized that there would be no differences in MVIT among the MVIC types, but that EMG_{AMP} would be greater during the $MVIC_{RAMP}$ than during $MVIC_{TRAD}$ or $MVIC_{RAPID}$. Moreover, we hypothesized that the reliability of the global EMG characteristics would be dependent on the MVIC type, with those obtained during the $MVIC_{RAMP}$ exhibiting the greatest reliability and those obtained during the $MVIC_{RAPID}$ exhibiting the lowest reliability.

2. Materials and methods

2.1. Participants

Twenty-two young, healthy males (n = 10) and females (n = 12) (mean ± SD, age = 21 ± 9 y; weight = 75.5 ± 16 kg; height = 169.6 ± 10 cm, BMI = 25.1 ± 6 $kg{\cdot}m^{-2}$) volunteered to participate in this study. To be eligible for this study, subjects must have reported no history of neurological illness, nor any current or ongoing musculoskeletal injuries of the lower extremities. Prior to enrollment, subjects signed an informed consent form and completed a health and exercise history questionnaire. The subjects reported performing 3.1 ± 3 $h{\cdot}wk^{-1}$ and 3.6 ± 2 $h{\cdot}wk^{-1}$ of aerobic and resistance exercise, respectively. In addition, subjects reported having performed aerobic and resistance exercise for 5.5 ± 4 y and 5.7 ± 4 y, respectively. This investigation was approved by and carried out in accordance with university's Institutional Review Board for the protection of human subjects (IRB Approval# ED-18-51, Approved on June 6, 2018).

2.2. Experimental design

Subjects visited the laboratory on two occasions, separated by 48 h. During visit one, subjects' height and weight were recorded. Subjects then completed a standardized warm-up, were familiarized with the testing procedures, completed two MVICs to obtain a target torque value for subsequent MVIC testing, and completed three each of the following muscle actions in random order: (1) traditional maximal isometric contractions ($MVIC_{TRAD}$), (2) rapid maximal isometric contractions ($MVIC_{RAPID}$), and (3) maximal isometric ramp contractions ($MVIC_{RAMP}$). During all muscle actions, leg extensor torque and EMG signals from the vastus lateralis (VL) and rectus femoris (RF) were recorded. Visit two was a replication of visit one, except for the familiarization procedures. All subjects were instructed to refrain from caffeine for 24 h and from engaging in any lower body exercise for 48 h prior to testing.

2.3. Isometric strength testing

For isometric strength testing, subjects were seated in an isokinetic dynamometer (Biodex System 4; Biodex Medical Systems, Inc. Shirley, NY, USA) with straps securing the trunk and pelvis, the lateral condyle of the femur aligned with the input axis of the dynamometer, and the pad of the dynamometer's lever arm positioned 3–4 cm above the medial malleolus of the participant's right leg. All isometric testing was performed with the right knee placed at a 90° joint angle. Knee (e.g., leg) extension torque (Nm) was measured through the lever arm of the isokinetic dynamometer. The torque signal was displayed in real-time on an external computer monitor for visual feedback to ensure maximal effort and accurate torque trajectory replication.

Subjects completed 2–3 warm-up isometric knee extension muscle contractions at 25, 50, and 75% of their perceived maximal effort with 30 s of rest in between each muscle contraction. Following the warmup, the subjects performed two MVICs which were used to identify the target torque for the $MVIC_{RAMP}$ contractions. The greatest torque achieved during a one second epoch from these two MVICs was identified as the target torque for the $MVIC_{RAMP}$ muscle actions. After five minutes of rest, participants performed knee extension contractions by kicking out as "hard as possible" for the traditional maximal isometric strength assessment ($MVIC_{TRAD}$), by kicking out as "fast as possible" for the rapid maximal isometric strength assessment ($MVIC_{RAPID}$), and by slowly increasing torque production up to maximum torque output while tracking the target torque trajectory on a computer screen for the ramp maximal isometric strength assessment ($MVIC_{RAMP}$). The $MVIC_{RAMP}$ muscle actions were performed by tracing a trapezoidal-shaped torque trajectory that increased linearly at 10% $MVIC{\cdot}s^{-1}$, plateaued and was held at MVIC for six seconds, and then decreased linearly at 10% $MVIC{\cdot}s^{-1}$ back to baseline. Two minutes of rest was provided between attempts for a given contraction type, while five minutes of rest was provided between contraction types. During all voluntary isometric strength testing, strong verbal encouragement was provided.

2.4. Electromyography

During each muscle contraction, EMG signals were collected through a 16-channel Bagnoli acquisition unit (Delsys Inc., Natick, MA, USA). Pre-amplified, parallel-bar surface EMG sensors (Delsys Bagnoli Surface EMG Sensor, contact dimensions = 10 × 1 mm, inter-electrode distance = 10 mm, detection area = 100 mm^2, CMRR (0–500 Hz) = 92 dB, Input Impedance = > 10^{15} Ω//0.2 pF) were placed and secured on the VL and RF muscles of the right thigh with hypoallergenic tape. For the VL, the center of the EMG bipolar electrode was placed at 66% of the distance between the anterior superior iliac spine (ASIS) and the lateral superior boarder of the patella (Hermens et al., 1999). The parallel sensor bars were arranged parallel to the angle of pennation of the VL fibers ~20%; (Fukunaga et al., 1997,

Table 1
$ICC_{2,k}$, 95% CI: intraclass correlation coefficient (model 2,k) and 95% confidence interval; SEM: standard error of measurement; CV: coefficient of variation; *p*-value: type 1 error rate for the one-way repeated-measures ANOVAs used to assess systematic variability.

	MVIT	Grand Mean (Nm)	$ICC_{2,k}$, 95% CI	SEM (Nm)	CV (%)	*p*-Value
	$MVIC_{TRAD}$	228.4	0.986, 0.923–0.987	28.8	12.6	0.97
	$MVIC_{RAPID}$	220.5	0.991, 0.978–0.996	15.5	7.1	0.76
	$MVIC_{RAMP}$	193.9	0.983, 0.959–0.993	18.2	9.4	0.11
	EMG AMP	Grand Mean (mV)	$ICC_{2,k}$, 95% CI	SEM (mV)	CV (%)	*p*-Value
VL	$MVIC_{TRAD}$	175.6 mV	0.930, 0.831–0.971	33.5	19.1	0.15
	$MVIC_{RAPID}$	162.4 mV	0.906, 0.772–0.961	33.1	20.1	0.75
	$MVIC_{RAMP}$	188.2 mV	0.858, 0.664–0.941	49.0	26.0	0.18
RF	$MVIC_{TRAD}$	156.1 mV	0.749, 0.398–0.896	44.9	28.8	*0.04**
	$MVIC_{RAPID}$	144.7 mV	0.785, 0.490–0.91	41.7	28.8	0.31
	$MVIC_{RAMP}$	178.4 mV	0.883, 0.716–0.952	38.0	21.3	0.09
	EMG MPF	Grand Mean (Hz)	$ICC_{2,k}$, 95% CI	SEM (Hz)	CV (%)	*p*-Value
VL	$MVIC_{TRAD}$	83.6 Hz	0.562, 0.08–0.82**	16.3	19.6	0.79
	$MVIC_{RAPID}$	87.3 Hz	0.578, 0.028–0.882	14.8	17.0	0.15
	$MVIC_{RAMP}$	82.0 Hz	0.643, 0.123–0.853	11.8	14.4	0.78
RF	$MVIC_{TRAD}$	87.5 Hz	0.726, 0.331–0.887	16.1	18.4	0.74
	$MVIC_{RAPID}$	86.9 Hz	0.631, 0.097–0.848	17.6	20.4	0.66
	$MVIC_{RAMP}$	82.6 Hz	0.882, 0.714–0.951	11.6	14.1	0.91

** Denotes a 95% CI including zero.
* Denotes significant difference between visits.

Lieber and Friden, 2000). For the RF, the center of the EMG bipolar electrode pair was placed at 50% of the distance between the ASIS and the medical superior border of the patella. A single re-usable, self-adhering neurostimulation electrode (Ultrastim, Axelgaarad Manufacturing Co., Ltd. Fallbrook, CA, USA) was secured on the process of the C7 vertebrae with hypoallergenic tape to serve as the reference electrode. To reduce the inter-electrode impedance and increase the signal-to-noise ratio (Beck and Housh, 2008), local areas of the skin were shaved, abraded, and cleaned with isopropyl alcohol prior to the placement of the electrodes.

2.5. Signal processing

Electromyographic and torque signals were recorded during all isometric testing and were sampled simultaneously with a Delsys Bagnoli Desktop data acquisition system (Deslys, Inc., Natick, MA, USA), recorded on a computer, and processed off-line with custom written (NDMJ) software (Labview v. 16.0, National Instruments, Austin, TX, USA). The EMG signals were amplified (×1000), zero-meaned, and digitally band-pass filtered with a zero-phase shift, 4th order Butterworth filter using a band-pass of 10–499 Hz. The torque signals were low-pass filtered with zero-phase shift, 4th order Butterworth filter using a 15 Hz cutoff frequency. All subsequent analyses were completed on the filtered and scaled signals.

During the MVICs, MVIT, EMG_{AMP}, and EMG_{MPF} were calculated from a 500 ms epoch corresponding to the highest average torque value that occurred during the MVIC plateau (Tomko et al., 2018a). The peak rate of torque development (pRTD) was also quantified as the peak of the first derivative torque signal to verify differences in the 'speed' of contraction. The time domain of the EMG signals (i.e., EMG_{AMP}) were expressed as the root mean square value in mV. To characterize the frequency domain of the EMG signal (i.e., EMG_{MPF}), each 500 ms signal epoch was processed with a Hamming window and a discrete Fourier transformation (DFT) based on the recommendations of (Diemont et al., 1988) and calculated the mean power frequency as described by (Kwatny et al., 1970). Consequently, EMG_{MPF} was expressed in Hz.

2.6. Statistical analysis

For each participant's dependent variables used in reliability analyses, the average value was calculated across the two MVIC attempts that resulted in the most similar (e.g., < 10%) MVIT values at each visit. One-way repeated-measures analyses of variance (ANOVAs) were then used to compare the mean MVIT, EMG_{AMP}, and EMG_{MPF} between visits for systematic variability. The test–retest reliability for MVIT, EMG_{AMP}, and EMG_{MPF} during each MVIC type were examined by calculating intraclass correlation coefficients (ICC) using model "2, k", because this model can be generalized to outside testers and laboratories (Weir, 2005). The 95% confidence interval (CI) for each $ICC_{2,k}$ was also calculated as described previously (Shrout and Fleiss, 1979) and was used to test the null hypothesis that each ICC was equal to zero (Jenkins et al., 2015, 2017; Tomko et al., 2018b). For measures of absolute reliability, the standard error of the measurement (SEM) was calculated as the square root of the mean square error term from the ANOVA table (Weir, 2005). The coefficient of variation (CV) was calculated as a normalized measure of the SEM by expressing the SEM relative to the grand mean (Hopkins, 2000). All data were analyzed using IBM SPSS Statistics v. 23 (IBM, Armonk, NY, USA) and a custom written spreadsheet (Microsoft Excel, Microsoft Corporation, Redmond, WA).

A one-way within-subjects ANOVAs (MVIC type [$MVIC_{TRAD}$ vs. $MVIC_{RAPID}$ vs. $MVIC_{RAMP}$]) was used to examine the influence of MVIC type on MVIT. Two separate, two-way within-subjects ANOVAs (MVIC type [$MVIC_{TRAD}$ vs. $MVIC_{RAPID}$ vs. $MVIC_{RAMP}$]) × muscle [VL vs. RF]) were used to examine the influence of MVIC type on EMG_{AMP} and EMG_{MPF}. Partial-eta squared effect sizes (η_p^2) were calculated for each ANOVA. When appropriate, follow-up analyses included Bonferroni-corrected dependent samples *t*-tests on the simple or marginal means (collapsed across the factor un-involved in the main effect). Cohen's *d* effect sizes were calculated for the post-hoc dependent samples t-tests and were corrected for dependence as described by Morris and DeShon (2002). All statistical analyses were completed using IBM SPSS Statistics (v. 22; Armonk, NY) and a type-I error rate was set *a priori* at 5%.

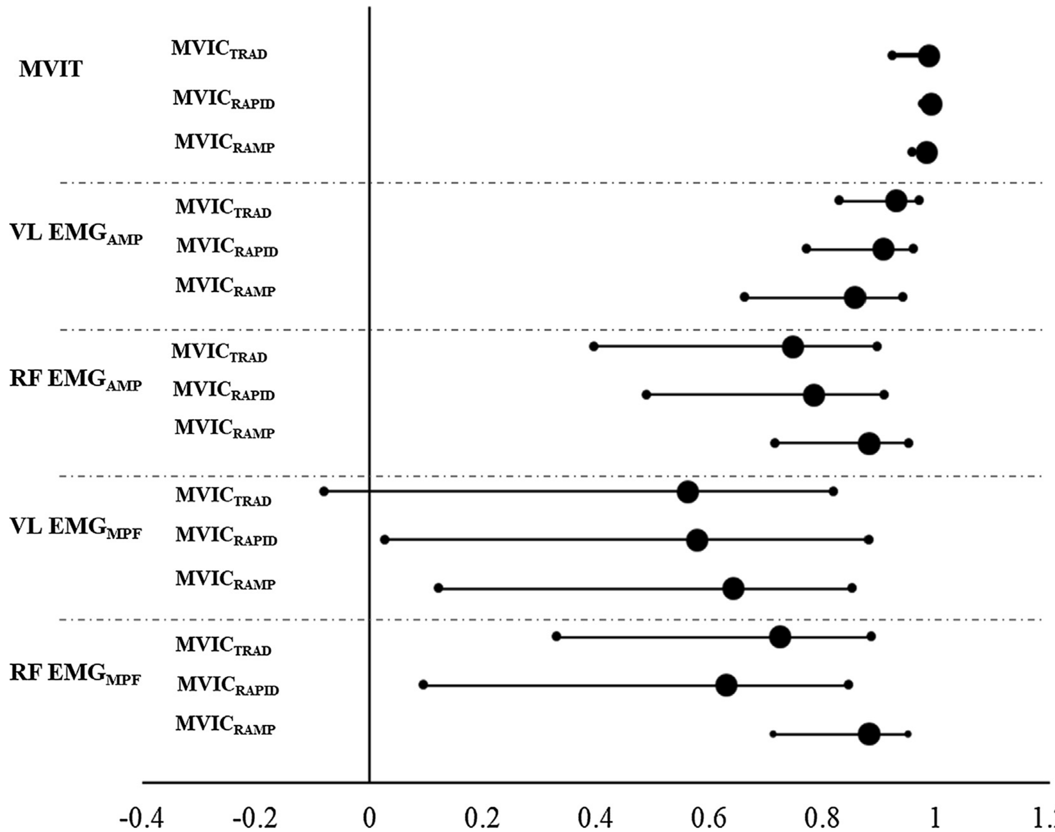

Fig. 1. The intraclass correlation coefficients (± 95% confidence intervals) for maximal voluntary torque (MVIT), and electromyographic signal amplitude (EMG_{AMP}) and mean power frequency (EMG_{MPF}) from the vastus lateralis (VL) and rectus femoris (RF) during the traditional ($MVIC_{TRAD}$), rapid ($MVIC_{RAPID}$), and ramp ($MVIC_{RAMP}$) maximal voluntary isometric contractions.

3. Results

3.1. Reliability

The grand means and test-retest reliability statistics of MVIC, EMG_{AMP}, and EMG_{MPF} are displayed in Table 1. There was no systematic variability between visits for any of the dependent variables except for RF EMG_{AMP} during $MVIC_{TRAD}$ ($p = 0.04$). The $ICC_{2,k}$ for each dependent variable was greater than zero ($p \leq 0.05$), except for VL EMG_{MPF} during $MVIC_{TRAD}$ (Fig. 1). The CVs for MVIT were 12.6% for $MVIC_{TRAD}$, 7.1% for $MVIC_{RAPID}$, and 9.4% $MVIC_{RAMP}$. For VL EMG_{AMP}, the CVs were 19.1% for $MVIC_{TRAD}$, 20.1% for $MVIC_{RAPID}$, and 26.0% for $MVIC_{RAMP}$. The RF EMG_{AMP} CVs were 28.8% during $MVIC_{TRAD}$, 28.8% for $MVIC_{RAPID}$, and 21.3% during $MVIC_{RAMP}$. Finally, the VL EMG_{MPF} CVs were 19.6%, 17.0%, and 14.4%, while the RF EMG_{MPF} CVs were 18.4%, 20.4%, and 14.1% for $MVIC_{TRAD}$, $MVIC_{RAPID}$, and $MVIC_{RAMP}$, respectively. Thus, qualitatively, the reliabilities of MVIT and the EMG signal characteristics did appear to be influenced by the MVIC type, with the best overall reliability displayed during the $MVIC_{RAMP}$ muscle actions.

3.2. MVIT and pRTD

There was a significant effect of MVIC type on MVIT ($F_{2, 86} = 25.4$, $p < 0.001$, $\eta_p^2 = 0.37$). Post hoc analyses indicated that MVIT was significantly greater during the $MVIC_{TRAD}$ and $MVIC_{RAPID}$ than during $MVIT_{RAMP}$ (both $p_{BC} < 0.001$; $d = 0.78$ and 0.95, respectively); however, there was no difference in MVIT during $MVIC_{RAPID}$ and $MVIC_{TRAD}$ ($p_{BC} = 0.32$; $d = 0.22$) (Fig. 2).

There was also a significant effect of MVIC type on pRTD ($F_{2, 86} = 191.0$, $p < 0.001$, $\eta_p^2 = 0.82$). The pRTD was greater during the $MVIC_{RAPID}$ than $MVIC_{TRAD}$ (1398.9 ± 582.2 vs. 1104.0 ± 528.7 $Nm{\cdot}s^{-1}$; $p_{BC} > 0.001$) and $MVIC_{RAMP}$ (1398.9 ± 582.2 vs. 159.8 ± 103.1 $Nm{\cdot}s^{-1}$; $p_{BC} > 0.001$), and pRTD during the $MVIC_{TRAD}$ was greater than $MVIC_{RAMP}$ (1104.0 ± 528.7 vs 159.8 ± 103.1 $Nm{\cdot}s^{-1}$; $p_{BC} > 0.001$).

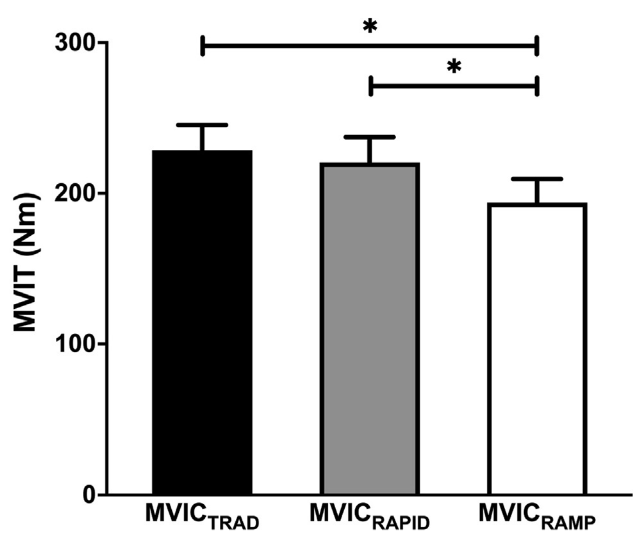

Fig. 2. The means (± standard error) for the maximal voluntary isometric torque (MVIT) (collapsed between days) achieved during the $MVIC_{TRAD}$, $MVIC_{RAPID}$, and $MVIC_{RAMP}$ muscle actions are shown. * indicates a significant difference between contraction types ($p \leq 0.05$).

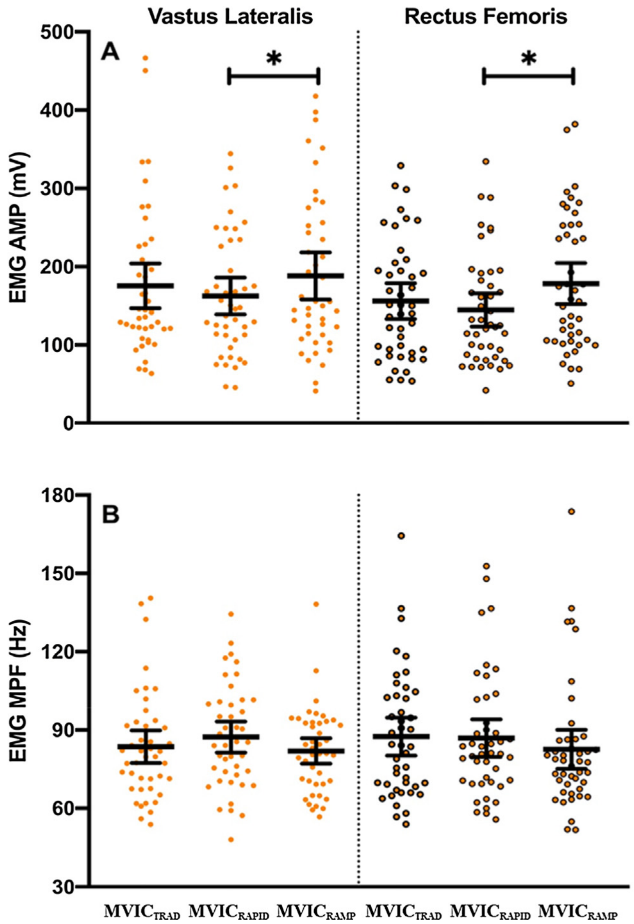

Fig. 3. Mean (± 95% CI) and individual electromyographic signal amplitude (EMG_{AMP}) and mean power frequency (EMG_{MPF}) values during the traditional (TRAD), rapid, and ramp MVICs. *Indicates that EMG_{AMP} was greater during $MVIC_{RAMP}$ than $MVIC_{RAPID}$ ($p \leq 0.05$).

3.3. EMG amplitude

For EMG_{AMP}, there was a MVIC type × muscle interaction ($F_{2, 43} = 7.8$, $p = 0.001$, $\eta_p^2 = 0.15$). Post hoc analyses revealed that VL EMG_{AMP} was significantly greater during $MVIC_{RAMP}$ than during $MVIC_{RAPID}$ (188.2 ± 99.3 mV vs. 162.4 ± 77.5; $p_{BC} = 0.02$, $d = 0.4$). However, VL EMG_{AMP} was not significantly different during $MVIC_{RAPID}$ and $MVIC_{TRAD}$ (162.4 ± 77.5 vs. 175.6 ± 93.7 mV; $p_{BC} = 0.66$, $d = 0.18$), or during $MVIC_{TRAD}$ and $MVIC_{RAMP}$ (175.6 ± 93.7 vs. 188.2 ± 99.3 mV; $p_{BC} = 0.65$, $d = 0.19$). In addition, RF EMG_{AMP} was significantly greater during $MVIC_{RAMP}$ than $MVIC_{RAPID}$ (178.4 ± 86.1 vs. 144.7 ± 69.7 mV; $p_{BC} = 0.004$, $d = 0.5$). However, RF EMG_{AMP} was not significantly different during $MVIC_{TRAD}$ and $MVIC_{RAPID}$ (156.1 ± 75.0 vs. 144.7 ± 69.7 mV; $p_{BC} = 0.88$, $d = 0.16$) or during $MVIC_{TRAD}$ and $MVIC_{RAMP}$ (156.1 ± 75.0 vs. 178.4 ± 86.1 mV; $p_{BC} = 0.10$, $d = 0.32$) (Fig. 3A).

3.4. EMG mean power frequency

For EMG_{MPF}, there was no significant MVIC type × muscle interaction ($F_{2, 43} = 2.0$, $p = 0.14$, $\eta_p^2 = 0.05$), nor main effects for MVIC type ($F_{2, 86} = 0.06$; $p = 0.94$, $\eta_p^2 < 0.01$) or muscle ($F_{1, 43} = 0.44$, $p = 0.51$, $\eta_p^2 = 0.01$) (Fig. 3B).

4. Discussion

Recently, Tomko et al. (2018a) demonstrated that, although MVIT achieved did not differ based on MVIC type, EMG_{AMP} was 18.1% higher and EMG_{MPF} was 6% lower during $MVIC_{RAMP}$ than $MVIC_{TRAD}$. The present study expanded on the results of Tomko et al. (2018a) by further examining the influence of MVIC type on both the magnitude and reliability of EMG_{AMP} and EMG_{MPF}. We also examined the influence of an additional, commonly used MVIC type ($MVIC_{RAPID}$) on MVIT, EMG_{AMP}, and EMG_{MPF}. The results of the present study suggested that, qualitatively, the reliability of the global EMG signal characteristics, but not MVIT, appear to be minimally influenced by the MVIC type. Furthermore, while the MVIT achieved was greatest during $MVIC_{TRAD}$, EMG_{AMP} was greatest during $MVIC_{RAMP}$, whereas EMG_{MPF} did not differ among the MVIC types. Thus, overall, our data suggest that each of the three MVIC methods generally provided reliable measurements of MVIT and EMG_{AMP} and can therefore be used based on investigator preference. However, it should be noted that $MVIC_{RAMP}$ contractions will likely yield the greatest EMG_{AMP} values and more reliable measurements of EMG_{MPF}.

Previously, Tomko et al. (2018a) reported that the EMG_{AMP} and EMG_{MPF} values obtained during a MVIC were dependent on the MVIC type used, with greater EMG_{AMP} and lower EMG_{MPF} values observed during a $MVIC_{RAMP}$ than a $MVIC_{TRAD}$, even though MVIT production was similar. The results of the present study build on those of Tomko et al. (2018a) and suggest that EMG_{AMP} values obtained during $MVIC_{RAMP}$ muscle actions are greater than those obtained during $MVIC_{RAPID}$, but not $MVIC_{TRAD}$ (Fig. 3). Furthermore, EMG_{AMP} was the greatest during the $MVIC_{RAMP}$ despite the fact that MVIT was 12–15% lower during $MVIC_{RAMP}$ than during the $MVIC_{TRAD}$ or $MVIC_{RAPID}$ (Fig. 2). We also observed that EMG_{MPF} was 5.5% lower during the $MVIC_{RAMP}$ compared to $MVIC_{RAPID}$ ($d = 0.23$) and 3.8% lower compared to $MVIC_{TRAD}$ ($d = 0.14$), although these differences were not statistically significant. These data likely suggest differences in motor unit behavior during the MVIC types related to the 'speed' of contraction, since pRTD was dramatically lower (e.g., 86–89% lower) during the $MVIC_{RAMP}$ than $MVIC_{TRAD}$ and $MVIC_{RAPID}$. For example, during slow increases in muscle force, low-threshold motor units do not increase discharge rates even with progressive increases in synaptic input, a phenomenon known as discharge rate saturation (Enoka, 2019, Fuglevand et al., 2014, Gydikov and Kosarov, 1974). Thus, it is possible that for any given torque level during the $MVIC_{RAMP}$ versus $MVIC_{TRAD}$ and $MVIC_{RAPID}$ muscle actions, greater motor unit recruitment was required. This phenomenon may also explain why MVIT was lower during the $MVIC_{RAMP}$ than $MVIC_{RAPID}$ and $MVIC_{TRAD}$ muscle actions in the present study. However, given that EMG_{AMP} is a crude indicator of neural drive and EMG_{MPF} is influenced by factors other than motor unit discharge rates, caution is warranted when interpreting these differences and additional studies are needed to further investigate this hypothesis.

A primary aim of this study was to quantify the test–retest reliability of MVIT, EMG_{AMP}, and EMG_{MPF} during the different MVIC types. Generally, the ICCs were high (0.98–0.99) and the CVs low (7–13%) for MVIT during all MVIC types. The ICCs were also high (0.86–0.93) and the CVs moderate (19–26%) for VL EMG_{AMP}. However, the ICC for VL EMG_{MPF} included zero, and was therefore unreliable during $MVIC_{TRAD}$, whereas it was 0.58 and 0.64 during the $MVIC_{RAPID}$ and $MVIC_{RAMP}$, respectively. There was also systematic variability across days for RF EMG_{AMP} during the $MVIC_{TRAD}$. Despite this, the ICC was 0.75 for $MVIC_{TRAD}$, and was 0.79 and 0.88 during $MVIC_{RAPID}$ and $MVIC_{RAMP}$, respectively. Furthermore, the CVs were 29% during the $MVIC_{TRAD}$ and $MVIC_{RAMP}$, and was 21% during the $MVIC_{RAMP}$. Finally, for RF EMG_{MPF}, the ICC was lowest (0.63) and CV greatest (18%) during the $MVIC_{RAPID}$, whereas the ICC was highest (0.88) and the CV lowest (12%) during the $MVIC_{RAMP}$. Thus, overall, MVIT and VL EMG_{AMP} can be measured with good overall reliability using any of the MVIC types. Given that $MVIC_{RAMP}$ muscle actions require preliminary MVIC trials, an experimental set-up that includes the ability to provide a ramp template for a participant to trace, and because MVIT and EMG_{AMP} were generally reliable among all three MVIC types, our recommendation is that investigators choose an MVIC type based on preference. However, if investigators are also completing MVICs to obtain EMG_{MPF} values for normalization purposes, they should consider using an $MVIC_{RAMP}$ muscle action.

In conclusion, our data suggest that MVIT production was lower during a $MVIC_{RAMP}$ than during a $MVIC_{TRAD}$ or $MVIC_{RAPID}$ muscle action. Despite this, the EMG_{AMP} values obtained during a $MVIC_{RAMP}$ were greater than during a $MVIC_{RAPID}$. It is plausible that these differences exist due to differences in motor unit behavior (e.g., discharge rate saturation) among the contraction types related to the rate of torque development, although future studies are needed to test this hypothesis. The results of the present study also suggest that, qualitatively, the three MVIC methods examined in this study provided reliable measurements of MVIT and $EMG_{AMP.}$ Therefore, investigators may choose MVIC type based on preference or equipment availability. However, investigators should note that $MVIC_{RAMP}$ contractions will likely yield the greatest EMG_{AMP} values and more reliable measurements of VL and RF EMG_{MPF}.

Declaration of Competing Interest

The authors report no conflict of interest.

References

Bampouras, T.M., Reeves, N.D., Baltzopoulos, V., Maganaris, C.N., 2017. The role of agonist and antagonist muscles in explaining isometric knee extension torque variation with hip joint angle. Eur. J. Appl. Physiol. 117, 2039–2045.

Beck, T.W., Housh, T.J., 2008. Use of Electromyography in Studying Human Movement. Routledge, New York, NY, USA.

Beck, T.W., Housh, T.J., Johnson, G.O., Weir, J.P., Cramer, J.T., Coburn, J.W., et al., 2004. Mechanomyographic and electromyographic time and frequency domain responses during submaximal to maximal isokinetic muscle actions of the biceps brachii. Eur. J. Appl. Physiol. 92, 352–359.

Bilodeau, M., Schindler-Ivens, S., Williams, D., Chandran, R., Sharma, S., 2003. EMG frequency content changes with increasing force and during fatigue in the quadriceps femoris muscle of men and women. J. Electromyogr. Kinesiol. 13, 83–92.

Booghs, C., Baudry, S., Enoka, R., Duchateau, J., 2012. Influence of neural adjustments and muscle oxygenation on task failure during sustained isometric contractions with elbow flexor muscles. Exp. Physiol. 97, 918–929.

Burnley, M., Vanhatalo, A., Jones, A.M., 2012. Distinct profiles of neuromuscular fatigue during muscle contractions below and above the critical torque in humans. J. Appl. Physiol. 113, 215–223.

Carr, J.C., Stock, M.S., Hernandez, J.M., Ortegon Jr, J.R., Mota, J.A., 2018. Additional insight into biarticular muscle function: the influence of hip flexor fatigue on rectus femoris activity at the knee. J. Electromyogr. Kinesiol. 42, 36–43.

Contessa, P., Letizi, J., De Luca, G., Kline, J.C., 2018. Contribution from motor unit firing adaptations and muscle coactivation during fatigue. J. Neurophysiol. 119, 2186–2193.

da Silva, J.J., Schoenfeld, B.J., Marchetti, P.N., Pecoraro, S.L., Greve, J.M., Marchetti, P.H., 2017. Muscle activation differs between partial and full back squat exercise with external load equated. J. Strength Conditioning Res. 31, 1688–1693.

Diemont, B., Figini, M.M., Orizio, C., Perin, R., Veicsteinas, A., 1988. Spectral analysis of muscular sound at low and high contraction level. Int. J. Biomed. Comput. 23, 161–175.

Enoka, R.M., 2019. Physiological validation of the decomposition of surface EMG signals. J. Electromyogr. Kinesiol.

Farina, D., Merletti, R., Enoka, R.M., 2004. The extraction of neural strategies from the surface EMG. J. Appl. Physiol. 96, 1486–1495.

Fuglevand, A.J., Lester, R.A., Johns, R.K., 2014. Distinguishing intrinsic from extrinsic factors underlying firing rate saturation in human motor units. J. Neurophysiol. 113, 1310–1322.

Fuglevand, A.J., Zackowski, K., Huey, K., Enoka, R., 1993. Impairment of neuromuscular propagation during human fatiguing contractions at submaximal forces. J. Physiol. 460, 549–572.

Fukunaga, T., Ichinose, Y., Ito, M., Kawakami, Y., Fukashiro, S., 1997. Determination of fascicle length and pennation in a contracting human muscle in vivo. J. Appl. Physiol. 82, 354–358.

Gydikov, A., Kosarov, D., 1974. Some features of different motor units in human biceps brachii. Pflügers Archiv. 347, 75–88.

Herda, T.J., Cramer, J.T., Ryan, E.D., McHugh, M.P., Stout, J.R., 2008. Acute effects of static versus dynamic stretching on isometric peak torque, electromyography, and mechanomyography of the biceps femoris muscle. J. Strength Conditioning Res. 22, 809–817.

Herda, T.J., Walter, A.A., Costa, P.B., Ryan, E.D., Hoge, K.M., Stout, J.R., et al., 2011. Percent voluntary inactivation and peak force predictions with the interpolated twitch technique in individuals with high ability of voluntary activation. Physiol. Meas. 32, 1591.

Herda, T.J., Zuniga, J.M., Ryan, E.D., Camic, C.L., Bergstrom, H.C., Smith, D.B., et al., 2015. The influence of electromyographic recording methods and the innervation zone on the mean power frequency–torque relationships. J. Electromyogr. Kinesiol. 25, 423–430.

Hermens, H.J., Freriks, B., Merletti, R., Stegeman, D., Blok, J., Rau, G., et al., 1999. European recommendations for surface electromyography. Roessingh Res. Dev. 8, 13–54.

Hill, E.C., Housh, T.J., Keller, J.L., Smith, C.M., Schmidt, R.J., Johnson, G.O., 2018. Early phase adaptations in muscle strength and hypertrophy as a result of low-intensity blood flow restriction resistance training. Eur. J. Appl. Physiol. 118, 1831–1843.

Hopkins, W.G., 2000. Measures of reliability in sports medicine and science. Sports Med. 30, 1–15.

Jenkins, N.D., Housh, T.J., Bergstrom, H.C., Cochrane, K.C., Hill, E.C., Smith, C.M., et al., 2015. Muscle activation during three sets to failure at 80 vs. 30% 1RM resistance exercise. Eur. J. Appl. Physiol. 115, 2335–2347.

Jenkins, N.D., Housh, T.J., Buckner, S.L., Bergstrom, H.C., Cochrane, K.C., Hill, E.C., et al., 2016. Neuromuscular adaptations after 2 and 4 weeks of 80% versus 30% 1 repetition maximum resistance training to failure. J. Strength Conditioning Res. 30, 2174–2185.

Jenkins, N.D.M., Miramonti, A.A., Hill, E.C., Smith, C.M., Cochrane-Snyman, K.C., Housh, T.J., Cramer, J.T., 2017. Greater neural adaptations following high- vs. low-load resistance training. Front. Physiol. 8.

Kamen, G., Caldwell, G.E., 1996. Physiology and interpretation of the electromyogram. J. Clin. Neurophysiol. 13, 366–384.

Kwatny, E., Thomas, D.H., Kwatny, H.G., 1970. An application of signal processing techniques to the study of myoelectric signals. IEEE Trans. Biomed. Eng. 303–313.

Lieber, R.L., Friden, J., 2000. Functional and clinical significance of skeletal muscle architecture. Muscle Nerve 23, 1647–1666.

Maluf, K.S., Shinohara, M., Stephenson, J.L., Enoka, R.M., 2005. Muscle activation and time to task failure differ with load type and contraction intensity for a human hand muscle. Exp Brain Res. 167, 165–177.

Merletti, R., 1999. Standards for Reporting EMG Data.

Morris, S.B., DeShon, R.P., 2002. Combining effect size estimates in meta-analysis with repeated measures and independent-groups designs. Psychol. Methods 7, 105.

Muddle, T.W., Colquhoun, R.J., Magrini, M.A., Luera, M.J., DeFreitas, J.M., Jenkins, N.D., 2018. Effects of fatiguing, submaximal high-versus low-torque isometric exercise on motor unit recruitment and firing behavior. Physiol. Rep. 6, e13675.

Palmer, T.B., Jenkins, N.D., Thompson, B.J., Smith, D.B., Cramer, J.T., 2014. The relationship between passive stiffness and muscle power output: influence of muscle cross-sectional area normalization. Muscle Nerve 49, 69–75.

Park, J., Hopkins, J.T., 2012. Within-and between-session reliability of the maximal voluntary knee extension torque and activation. Int. J. Neurosci. 123, 55–59.

Plautard, M., Guilhem, G., Cornu, C., Guével, A., 2015. Time-course of performance changes and underlying mechanisms during and after repetitive moderately weight-loaded knee extensions. J. Electromyogr. Kinesiol. 25, 488–494.

Sahay, R., Vandewalle, H., Driss, T., Monod, H., 2001. Maximal voluntary force and rate of force development in humans – importance of instruction. Eur. J. Appl. Physiol. 85, 345–350.

Shrout, P.E., Fleiss, J.L., 1979. Intraclass correlations: uses in assessing rater reliability. Psychol. Bull. 86, 420.

Smith, C., Housh, T.J., Hill, E., Cochrane, K.C., Jenkins, N., Schmidt, R., et al., 2016. Effects of fatiguing constant versus alternating intensity intermittent isometric muscle actions on maximal torque and neuromuscular responses. J. Musculoskelet. Neuronal Interact. 16, 318.

Smith, C.M., Housh, T.J., Hill, E.C., Keller, J.L., Johnson, G.O., Schmidt, R.J., 2017. Effects of fatigue and recovery on electromechanical delay during isokinetic muscle actions. Physiol. Meas. 38, 1837.

Sterczala, A.J., Miller, J.D., Trevino, M.A., Dimmick, H.L., Herda, T.J., 2018. Differences in the motor unit firing rates and amplitudes in relation to recruitment thresholds during submaximal contractions of the first dorsal interosseous between chronically resistance-trained and physically active men. Appl. Physiol. Nutr. Metab. 43, 759–768.

Tomko, P.M., Colquhoun, R.J., Magrini, M.A., Muddle, T.W., Jenkins, N.D., 2018a. Global electromyographic signal characteristics depend on maximal isometric contraction method in the knee extensors. J. Electromyogr. Kinesiol. 42, 111–116.

Tomko, P.M., Muddle, T.W., Magrini, M.A., Colquhoun, R.J., Luera, M.J., Jenkins, N.D., 2018b. Reliability and differences in quadriceps femoris muscle morphology using ultrasonography: the effects of body position and rest time. Ultrasound 26, 214–221.

Trevino, M.A., Sterczala, A.J., Miller, J.D., Wray, M.E., Dimmick, H.L., Ciccone, A.B., et al., 2019. Sex-related differences in muscle size explained by amplitudes of higher-threshold motor unit action potentials and muscle fibre typing. Acta Physiol. 225, e13151.

Weir, J.P., 2005. Quantifying test-retest reliability using the intraclass correlation coefficient and the SEM. J. Strength Conditioning Res. 19, 231–240.

Yao, W., Fuglevand, R.J., Enoka, R.M., 2000. Motor-unit synchronization increases EMG amplitude and decreases force steadiness of simulated contractions. J. Neurophysiol. 83, 441–452.

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