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Quantifying thigh muscle co-activation during isometric knee extension contractions: Within- and between-session reliability

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Introduction

Co-activation is the simultaneous activation of agonist and antagonist muscle groups around a joint which contributes to joint stability, homogeneous load distribution [Baratta et al, 1988], control of bone displacements [Solomonow et al, 1987] and movement efficiency [Levine et al, 1952]. Co-activation of knee joint muscles has been extensively studied over the past two decades due to its importance during ambulation and balance [Baratta et al, 1988; Seyedali et al, 2012]. Opposing muscle groups such as the quadriceps and the hamstrings function as synergists to provide stability and stiffness to the knee joint [Ait-Haddou et al, 2000]. Selected joint pathologies, central or peripheral nervous system disorders can induce abnormal levels of co-activation [Busse et al, 2005]. Inappropriate co-activation levels produce movement dysfunction, which in turn can lead to joint injury [Baratta et al, 1988; Busse et al., 2005; Macaluso et al, 2002]. Reliable and meaningful measures are needed that accurately assess co-activation levels by calculation of the co-activation index (CI). Such a CI will permit comparisons between studies and serve as an outcome measure for rehabilitation interventions.

There are a number of parameters that may affect the reliability and validity of the CI calculation. Parameters that are related to the data collection include the number of muscles or muscle segments sampled, pennation angle, the inclusion of monoarticular or multiarticular muscles, type of contraction, joint position, and electrode placement. Parameters that are related to data analysis include the selection of the time unit (window) and the smoothing approach applied to the electromyographic (EMG) signal, as well as the equation/method for the quantification of the CI. While most data collection parameters have inherent and inevitable limitations that affect comparison among studies, parameters that are related to data analysis can be controlled and standardized.

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There are four commonly utilized methods for the quantification of the CI. The first two rudimentary methods were the semi-quantitative estimates of EMG magnitude [Frost et al, 1997] and the agonist-to-antagonist ratio of EMG activity utilizing millivolts of electrical activity [Damiano et al, 2000; Fung et al, 1989]. The limitations of these two methods led to the adoption of more robust techniques that normalized the EMG amplitude for each of the agonist and antagonist muscle groups to the respective maximum voluntary contraction values (MVC; [Ervilha et al, 2012; Knutson et al, 1994]). The last and more recent method for the calculation of the CI quantified the antagonist moment using mathematical modeling of the EMG/joint torque relationship, but with controversial applicability due to changes in the slope attributable to evolution of the firing frequency and recruitment across the range of muscle activation [Merletti et al, 2004].

Normalization methods have been widely adopted but there are many inconsistencies with respect to window size and smoothing techniques utilized to estimate muscle activation. These inconsistencies reduce the comparability of calculated CIs between studies. Researchers have used peak EMG amplitude [Yang et al, 1984], average EMG [Kellis et al, 2011], integrated EMG [Kubo et al, 2004], root mean square [Hortobágyi et al, 2005] and envelope EMG [Frost et al, 1997] of various window sizes among other filtering and smoothing techniques. Besides the peak amplitude technique, which estimates muscle activation from a single value, the other techniques calculate an average value over a selected segment of data (window). Signal processing using RMS requires fewer steps in the data reduction process and minimizes signal distortion [Cram et al, 1998]. The second important issue is the selection of the optimum window size. Utilizing a small window or even choosing a single value (e.g., peak amplitude) can be affected by artifacts or outliers. A larger EMG window that is temporally associated with the highest joint torque produced during the MVC may be more representative of the muscle's activation. On the contrary, an excessively large window size may distort estimates by including segments of submaximal muscle activation. It still remains unanswered which data smoothing method and window size can generate the most reliable and meaningful CI.

Replication of electrode placement can be a limiting factor in between-day reproducibility. Electrode placement on the belly of an agonistic muscle during MVC has produced very reliable between-day estimates of maximal muscle EMG [Larsson et al, 2003; McKenzie et al, 2010]. However, when assessing co-activation, the antagonist muscle group undergoes a submaximal contraction. During submaximal contractions, a slight shift in electrode placement between sessions could capture different EMG activity or increase the variability of the signal [Van Dijk et al, 2009] due to changes in spatial summation of the signals.

Therefore, the purpose of the present study was to assess thigh muscle CI during isometric contractions by comparing the results from commonly employed signal processing techniques and to determine within- and between-session CI reliability. It was hypothesized that RMS EMG of a window size around the peak torque during a maximum voluntary isometric contraction would produce more reliable estimates of CI. Additionally, we hypothesized that within-session reliability would be higher than between-session values.

Methods

Informed consent

The study was approved by the Creighton University Institutional Review Board and conducted in accordance with the Declaration of Helsinki. All subject volunteers read, understood and signed the informed consent document prior to participation.

Subjects

Ten healthy young adults (5 males, 5 females; 27.5±4yrs, 174.6±12cm; 77.8±12kg) volunteered from a convenience sample of young, healthy university students that were enrolled in health sciences graduate curricula. None of the volunteers were currently participating in formal collegiate sports teams. To minimize practice/learning effects, subjects were required to have prior experience with an isokinetic dynamometer. Subjects were free of musculoskeletal and/or neurological problems that may have affected their ability to generate maximal knee flexion and extension torque in the dominant (preferred kicking) leg.

Torque recordings

Subjects were positioned and restrained on a Biodex System 3 isokinetic dynamometer per manufacturer recommendations for knee assessment (Biodex Medical Systems Inc., Shirley, NY, USA) with joint angles of 60° at both the hip and knee using zero-neutral measurements. The lateral epicondyle was aligned with the dynamometer's power shaft and the inferior edge of the ankle cuff was placed 2 cm above the lateral malleolus. The subject's arms were kept folded in front of their chest during testing. The Biodex raw torque signal was digitally sampled at 1 kHz and stored on a PC (Windows XP) running DataPac 2K2 (Run Technologies, Mission Viejo, CA, USA). The raw torque data were low pass filtered at 10Hz, converted to Newton-meters and corrected for gravity offset.

EMG recordings

The skin over the vastus lateralis, rectus femoris, vastus medialis, lateral and medial hamstrings was shaved and wiped with alcohol on the dominant side. Pairs of surface EMG electrodes (BIOPAC Systems, Inc., Goleta, CA, USA) were attached over each muscle following the SENIAM guidelines maintaining a 2 cm center-to-center interelectrode spacing [Hermens et al, 2000]. A ground electrode was placed on the ipsilateral lateral malleolus. Electrodes were attached by short, shielded wires to on-site preamplifiers. The differential amplifiers had a gain of 1000–2000, an input impedance of 100kMΩ, and a common mode rejection ratio of 100dB (Motion Lab Systems MA-300™). EMG signals and the Biodex torque signal were digitally sampled at 1 kHz per and stored on a PC using DataPac 2K2. Using a custom-written Matlab program (Matlab 2012b; Mathworks Inc., Natick, MA, USA), EMG signals were band-pass filtered (10 to 450 Hz), notch filtered at 60 Hz, corrected for zero offset, full-wave rectified and stored for further analysis.

Experimental design

After measuring and recording stature and body mass, each subject warmed-up by walking on a treadmill for 5 min at 5–6 km/h. Prior to testing, each subject was allowed to become accustomed to the dynamometer by performing brief submaximal and maximal contractions for both knee extension and flexion.. Three to four familiarization contractions lasting 2–3 sec were allowed in each direction. A computer monitor provided feedback of their performance by displaying the torque curve in real time while a horizontal line demarcated the subject's highest torque.

In the first session, each subject performed one knee flexion MVIC, rested for two minutes then four successive maximum knee extension MVICs. They were instructed to flex or extend the knee as hard as possible for 2–3 seconds. One minute of rest was allowed between each extension contraction. Upon successful completion of the first session, all subjects were retested with a minimum of 3 days between sessions.

Data processing

EMG activity was used to quantify CI during quadriceps maximum voluntary isometric contraction (MVIC). The CI was calculated using 7 different approaches that were based on a) a single value (the peak amplitude of the raw EMG signal), b) an interval around the peak torque [20msec (RMS_{20PT}), 50msec (RMS_{50PT}) and 500msec (RMS_{500PT}); Figure 1], and c) throughout the entire period of extensor torque production [for the entire burst (RMS_{BURST}), and in moving windows of 20 (RMS_{20}) and 50msec (RMS_{50})]. Except for the single value method, the EMG activity of each muscle segment was normalized to the corresponding MVIC. Equation 1 provides the CI calculation where the numerator and the denominator represent the corresponding values from any of the measured parameters for each of the 7 approaches. The CI was calculated for each of the four knee extension MVICs.

$$\mathrm{CI}=\frac{\frac{\text{EMG activity of the hamstrings (medial+lateral)}}{2}}{\frac{\text{EMG activity of the quadriceps (3 heads)}}{3}}\times 100 \quad (1)$$

Statistical analysis was performed using SPSS software (SPSS 20; SPSS Inc., Chicago, IL, USA). Descriptive statistics including means, standard deviations (SD) and coefficient of variation (CV) were calculated for each parameter. A paired t-test was performed between sessions to test for learning effect. Intraclass correlation coefficient (ICC) analysis was used to assess the consistency, or conformity, of measurements made by applying the different calculation approaches. According to Sleivert and Wenger [1994], ICC values below 0.6 indicates poor reproducibility, while it has been suggested that ICC values greater than 0.8 are acceptable for clinical work [Currier, 1984]. To give more practical meaning to the ICC measurements, minimum detectable change (MDC) was calculated by multiplying the standard error of measurement (SEM) by 2.77 [Weir, 2005]. MDC indicates the level of change in a parameter attributed with 95% certainty to a true change in a subject's condition, as differentiated from changes due to test-retest errors. In other words, any retest measurement should exceed the MDC to indicate a real change. In addition, inter-session

CV and intra-session CV values were calculated to determine between- and within-sessions variability, respectively.

Results

CI values ranged from 11.28 to 13.17. The calculations employing peak amplitude across the entire burst gave the highest CI values and methods using RMS around the PT produced the lowest CI values (Table 1). A paired t-test revealed no statistical difference in CI between sessions therefore mean CI was calculatedfor each subject by taking the average of all contractions across sessions.

Within-session

ICC analysis of the CI quantification approaches revealed that within-session coefficients were greater than 0.80 with most values greater than 0.90, while MDC values ranged from 3.5% to 7.7% (Table 2). The low MDC value with the 500ms-window approach indicated that changes exceeding 3.7 units of the CI were beyond measurement error. CV values ranged from 9.0% to 21.8% (Table 1). Within-session results indicated that CI was highly reliable for all approaches. The 500ms-window and the entire burst window yielded the most reliable and the least variable CI.

Between-session

ICC values for the between-session analysis ranged from 0.376 to 0.680, while MDC ranged from 11.9% to 20% (Table 2). CV values ranged from 24.2% to 34% (Table 1). Between-session ICC values were consistently lower than within-session ICC values across all CI approaches. The opposite holds true for CV values with between-session values consistently higher than within-session CV values.

Discussion

The primary goal of the present study was to determine within- and between-session CI reliability from commonly employed signal processing techniques. We found notable differences in the reliability and variability measures among the signal processing techniques, which partially supported our first hypothesis. In support of our second hypothesis, within-session measures of reliability displayed higher reliability and lower variability than between-session measures. Nonetheless, the average value of CI across the different approaches fluctuated between 11.28 to 13.17, which lies within the range reported in the literature [Grabiner et al, 1989].

Within-session measures

Overall, ICC analysis indicated that CI quantification was highly reliable within the same session independent of the approach employed (ICC >0.86). To our knowledge, only one study has reported test-retest measures by calculating reliability coefficients [Falconer et al, 1985]. However, the study by Falconer et al [1985] examined the co-activation of the ankle muscles during gait. As such, interpretation of the ICC of the present study should be restricted to the reliability of CI calculations based on EMG from isometric knee extension

contractions, since the CI equation depends solely on the ratio between the EMG activity of the antagonist and agonist muscles (equation 1). It is well established that EMG activity of knee flexors and extensors is highly reliable during contractions close to MVIC (ICC>80; McKenzie et al, 2010; Viitasalo et al, 1975], while there is a moderate to high reliability for MH and LH at intensities 10–30% MVIC (0.73 < ICC < 0.96; Kellis et al, 2008]. The high within-session reliability values in the present study are consistent with these reports.

In support of our first hypothesis, the RMS_{500PT} approach yielded the most reliable and least variable CI values. The other two approaches, which utilized a much smaller window size around the PT, yielded a twofold increase in variability and a proportional increase in the MDC values. The primary reason for this discrepancy was that PT does not necessarily coincide with the highest electrical activation of the muscles involved due to electromechanical delay. Therefore a 20 or 50ms window around PT may not have captured the maximum activity of the EMG signal. Another influential factor was the number of data points analysed. A small window size could be affected more by artifacts or unusual EMG activity due to motion or other electrical interference, whereas a larger window size could be less influenced by such activity. The same phenomenon could influence the other two approaches that utilized a moving window of 20 and 50ms. In addition, the two latter approaches could have been influenced by the fact that antagonist EMG activity was greater at the beginning and end of a movement [Baratta et al, 1988; Kellis, 1998]. The increases in antagonist activity at the beginning and end of contraction would have been accentuated during dynamic or isokinetic contractions in which there was acceleration and deceleration of the limb at the initial and final parts of the movement, respectively [Hagood et al, 1990; Kellis et al, 1996; Osternig et al, 1984]. However, when assessing isometric contractions using the Biodex apparatus, there was a small limb movement that occurred between rest and full exertion due to compression of the cushioning material of the restraining straps and chair. These small movements could have created artifacts or higher activations of the antagonist muscle group.

The RMS_{BURST} approach yielded results that were comparable to the RMS_{500PT} approach. This was expected since the average PT during a MVIC was close to 95% of the knee extensors' PT. The initial and final parts of the contraction were only a small fraction of the entire burst and may not have affected the calculated outcome. However, when analysing longer contractions that contain submaximal effort or contractions in which the PT lasts only a few milliseconds, selection of an appropriate window size must be carefully considered. Further research is required to investigate the CI calculation algorithms for submaximal muscle contractions similar to the conditions occurring during functional activities and if CI calculations are affected by studying populations with impaired motor control..

Between-session measures

ICC values fluctuated between 0.376 and 0.680 indicating a poor to moderate between-session reliability. In support of our second hypothesis, these values were much lower than the corresponding within-session values. Also, MDC and CV were much higher between- than within-session. The PT normalized to body weight (1.69±0.3) for knee extension was highly reliable within- and between-sessions (ICC 0.982 and 0.958, respectively). Subjects

were able to reproduce their PT at both sessions, which implied that any variations in CI values were not due to variations in torque output. The underlying cause of this discrepancy may have been the placement of the electrodes and/or the spatial pattern of motor unit recruitment within the muscle.

Surface electrodes captured the EMG activity of motor units within a limited distance from the skin placement. During a knee extension MVIC – ideally – all or most of the agonist muscle motor units within a detectable range from the pickup electrodes were fully active, thus a slight shift in electrode placement between sessions would not have markedly affected the measured EMG activity. On the contrary, the antagonist muscle group underwent a submaximal contraction and therefore fewer motor units within pickup range were active during an agonist MVIC. The consistency of our torque measurements implied that the net tension generated by the motor units was not altered within the same session. However, poor to moderate between-session reliability in antagonist EMG could be attributed to between-session differences in the subpopulation of recruited motor units and/or small changes in electrode placement that affected the net spatial summation of the motor unit electrical activity within detection range. In addition, reliability outcomes can be affected by the analytical approach employed for the quantification of the CI.

Our review of literature on CI quantification showed that the most commonly employed approaches utilized a moving window of a small size or the peak amplitude of the raw EMG signal. Our current findings clearly showed that a small sampling window elicited the lowest ICC values ($0.376 < ICC < 0.441$). ICC values lower than 0.6 are considered to have poor reproducibility [Sleivert et al, 1994] and may produce inconclusive or misleading results. However, the RMS_{500} and RMS_{BURST} approaches generated the highest reliability and lowest variability values, which correspond to moderate reproducibility in general. Nonetheless, these approaches may be reliable and effective as a relative measure among conditions and are best utilized for within-session experimental designs.

Limitations

Potential limitations of our current study were the inherent issues with bipolar surface electrodes, such as cross-talk and placement. Advancement in EMG acquisition systems and proper preparation has minimized cross-talk [Hof, 1984; Winter et al, 1994], while careful measurements following electrode placement standards such as SENIAM can minimize this source of placement error. Other researchers have applied indelible ink skin markings during the first testing to improve accuracy of electrode application during subsequent sessions [Zech et al, 2008]. Nonetheless, neither of the above methods can control for subtle variations in motor unit recruitment patterns, which in turn varies spatial summation of the electrical signal. Perhaps spatial summation variations in submaximal contractions could be addressed by employing high-density surface EMG electrode arrays rather than standard bipolar arrangement [Drost et al, 2006].

Another potential limitation was the difference in the execution strategy between subjects. The knee joint is a complex articular structure controlled by a number of interacting elements including muscles, tendons and ligaments. Subjects could produce similar torque output by using subtly different limb positions and muscle activation patterns. For instance,

changes in femoral or tibial rotation along with ankle dorsi/plantar flexion may have affected the recruitment and subsequent EMG motor unit activation of the hamstring and gastrocnemius muscles during a knee extensor's MVIC. While we were largely able to control the femur and hip joint position, the tibia was able to rotate in the transverse plane and ankle was able to freely move in all planes during the testing procedures. We intentionally focused on well-controlled isometric knee extension contractions to minimize differences in execution strategies. Studies that utilize inherently more variable dynamic contractions impose complex demands such as variations in angular velocity, changes in internal and external torque and muscle length/tension changes. We speculate that CI reliability would decrease for dynamic contractions and would reflect within- and between-subject variations in execution strategy.

Lastly, a potential limitation can be the equation utilized to calculate the CI. Following the overwhelming majority of the current literature, the CI calculation was based on the average EMG activity of three quadriceps and the average EMG activity of two hamstrings muscle segments for a specific time window. However, due to anatomic and fiber-type differences between muscle segments, each segment does not contribute equally to muscle tension. Future studies should focus on weighting the EMG of each segment to account for its overall contribution to forcegeneration, as well as examining the effect of the combination of muscle segments used in the calculation.

Conclusion

Interpretation of the present findings can offer insight into the technique that is more appropriate for the quantification of CI during isometric knee extension at 60° of flexion. A selection of a large window size around the PT appeared to deliver more reliable and less variable results. However, the lower values of between-session reliability accompanied with more than a twofold increase in variability suggests caution in calculation and interpretation of CI values across sessions and between research reports.

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Biographies

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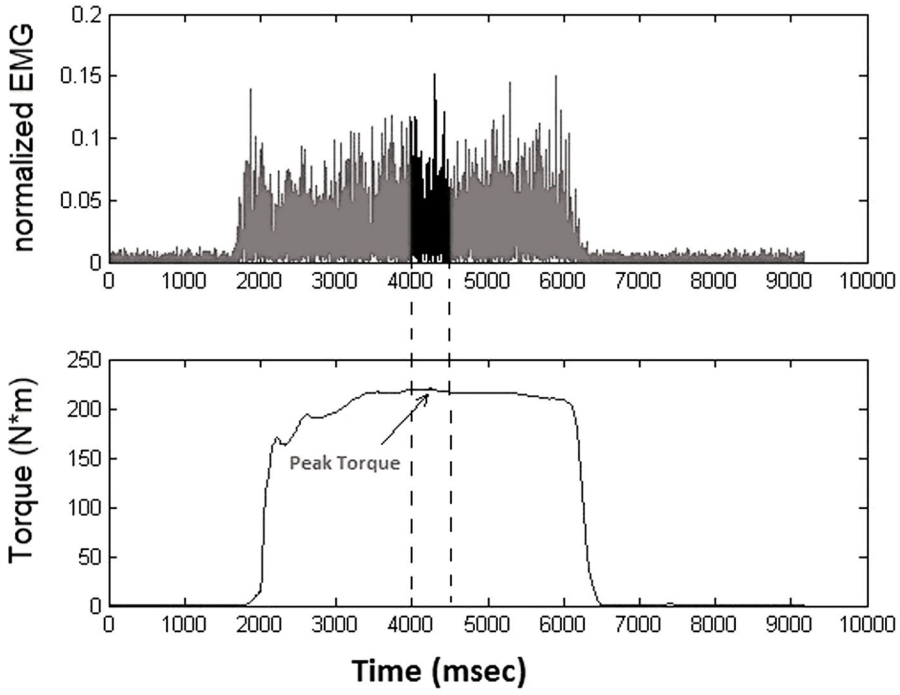

Figure 1.

The top portion of the figure displays the normalized activation from one of the lateral hamstring muscle group. In this example, the Matlab algorithm calculated the RMS in a window of 500ms around the peak torque, which is one of the 7 analytical approaches utilized for the quantification of the CI.

Table 1

Within- and between-session variability for the different quantification methods of the coactivation index.

	CI (mean±SD)	Within-session	Between-session
		CV (%)	CV (%)
Amplitude	13.17 (8.2)	14.2%	34.0%
RMS_{20PT}	11.54 (7.4)	21.8%	30.2%
RMS_{50PT}	11.50 (7.6)	19.9%	30.3%
RMS_{500PT}	11.28 (7.8)	10.5%	24.9%
RMS_{20}	12.68 (8.5)	13.8%	31.9%
RMS_{50}	12.78 (9.1)	12.6%	31.1%
RMS_{BURST}	11.64 (7.6)	9.0%	24.2%

CI = Coactivation Index; CV = Coefficient of Variation.

Table 2

Within- and between-session reliability indices for the different quantification methods of the coactivation index.

	Within-session		Between-session	
	ICC	**MDC**	**ICC**	**MDC**
Amplitude	0.910	6.7	0.461	14.1
RMS_{20PT}	0.861	7.3	0.578	11.7
RMS_{50PT}	0.896	6.6	0.645	11.4
RMS_{500PT}	0.971	3.7	0.664	11.4
RMS_{20}	0.900	7.2	0.431	15.0
RMS_{50}	0.904	7.7	0.376	16.6
RMS_{BURST}	0.979	3.5	0.680	10.9

MDC = Minimum Detectable Change; ICC = Intraclass Correlation Coefficient.