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Low-cost electromyography – Validation against a commercial system using both manual and automated activation timing thresholds

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1. Introduction
Clinical utility of objective measurement tools relates to a number of factors including time, portability, equipment, training and cost (Tyson, 2009). Despite its potential benefits in clinical practice, the incorporation of electromyography (EMG) into non-laboratory settings may be limited by cost (Botelho et al., 2013; Supuk et al., 2014; Trask et al., 2007). Previously the creation of low-cost, custom EMG systems was achievable (Supuk et al., 2014) but the need to build most of the system from separate components may be a potential barrier. Recently electronic microchip EMG systems became available which perform the backend electrode connection, signal amplification and analog filtering (Bhat and Gupta, 2016; Zhao et al., 2015). The flexibility to interface these low-cost microchips to a range of data acquisition systems could result in greater uptake of electromyography into non-laboratory settings, as complete EMG systems could be created for under USD$100.

However, validation of prototype low-cost EMG system data is limited. Lower limb muscle activation patterns in gait from low-cost EMG systems correlate moderately with EMG data reported in the literature (Supuk et al., 2014) but it remains unclear if low-cost systems provide adequate data quality to be comparable to a criterion reference. To date, no single study has investigated the agreement between low-cost EMG systems and commercially available EMG systems to our knowledge. The primary aim of this study was to assess the concurrent validity of a custom built, low-cost EMG microchip system compared to a commercial EMG system during isometric and functional exercises of varying intensity. Our hypothesis was that there could be issues with noise in the low-cost system, therefore the study also examined two data processing methods to determine the onset and end of a muscle contraction. We compared standard manual extraction of data using cursors on the
enveloped signal to the Teager-Kaiser energy operator (TKEO) technique, which factors in the mean and variation of the background noise of the signal and has been shown to be effective for improving contraction timing estimation (Li et al 2007; Solnik et al., 2008). This method is particularly appealing as it is mathematically simple, and therefore has great potential for implementation on low-cost embedded systems such as Arduino where processor power and memory may be restricted. This resulted in the secondary study aim, which was to determine if incorporating this mathematically simple TKEO thresholding technique may facilitate EMG data extraction with a resultant improved inter-rater reliability.

2. Methods

2.1 Participants

10 recreationally active women without any neuromotor impairments or limitations to exercise volunteered to participate in the study (age 28.1±6.8 yrs, 162.1±6.8 cm, 60.3±10.2 kg). All participants agreed to and signed the study’s written informed consent documents which were approved by the Human Research Ethics Committee and attended one session over a ten day testing period.

2.2 Equipment

A commercial wireless EMG system (Telemyo DTS, Noraxon, Arizona, USA) was used as the criterion reference. The commercial system pre-amplified the signal (gain 500) and transmitted data wirelessly with a 64ms delay to a base station outputting analog data to a National Instruments CompactDAQ with a 16-bit analog input module (9215 BNC). The custom system consisted of a Myoware muscle sensor chip with electrode connectors (Advancer Technologies, North Carolina,
USA). Shielding was added externally to reduce electromagnetic interference with liquid electrical insulation (Star brite, Florida, USA) and epoxy putty (Knead-it, Selleys, New South Wales, Australia). The custom system also incorporated an Analog Devices AD8236 operational amplifier with a common-mode rejection ratio of 110dB at gain ≥100. This was powered by 5V. Raw data were outputted to a low-cost 14-bit data acquisition device (USB-6001, National Instruments, Texas, USA). Both systems were sampled at 2000Hz using custom LabVIEW software (National Instruments, Texas, USA). These systems were used concurrently on each participant during maximal voluntary isometric contraction (MVC) and during exercises (Table 1).

2.3 Electrodes, preparation and placement

Independent electrodes (Duo-Trode, Myotronics, USA; Ag/AgCl; disc diameter 12.5mm) were used for each of the two systems. Anatomical references for electrode placement for Vastus Lateralis (VL) of the right leg were modified to allow placement of two sets of electrodes from previous recommendations (Barbaro and Rainaldi, 2012; Hermens et al 1999) (Figure 1). The system placement position (proximal or distal) was randomized for each participant with the inter-electrode distance 19mm in the commercial system, and 30mm for the custom system. The reference electrode was fixed for the commercial system on the back of the wireless transmitter and for the custom system was placed on the iliotibial band.

2.5 MVC Assessment

Maximal Voluntary Contraction (MVC) for quadriceps occurred in a seated position with the knee in 80 degrees of flexion and resisted torque recorded using a wireless
hand-held dynamometer (Model 01163, Layfayette Instrument Company, Indiana, USA). All participants previously experienced MVC testing and completed a short MVC review training prior to testing. Participants were instructed to push as hard as possible and hold an isometric contraction for five seconds. Two maximal contractions were completed following two warm-up submaximal contractions.

2.6 Exercises

Exercises included squatting at slow and fast speeds, stepping up, isometric and concentric active knee extension and jumping (further detail of each exercise can be found in Table 1). Prior to testing, the same instructor first demonstrated, then practiced each exercise with the participant. Audible feedback was provided for those exercises with pre-determined speeds using a metronome set at 60 beats per minute. If the instructor determined the participant completed the exercise out of time with the metronome, that trial was marked on a data collection sheet to be discarded and the exercise was repeated.

2.7 Data processing:

For both systems, the raw EMG signals were first digitally passband filtered (Butterworth 20–500 Hz, 12 poles, zero-phase shift), with a notch filter included to remove powerline noise (Butterworth 45-55 Hz, 12 poles, zero-phase shift). An additional notch filter for the custom system addressed significant noise present centred around 150Hz (Butterworth 140–160 Hz, 12 poles, zero-phase shift) in all signals. Data were then converted to absolute values (i.e. rectified) before a linear envelope was applied (5Hz lowpass Butterworth, 12 poles, zero-phase shift). Figure 2 shows an example of the raw data, a spectrogram generated using a short-time Fourier transform (time step = 10 samples, Hanning window length = 64 samples,
Frequency Bins = 1024) to display the time-frequency content of the signal, and the linear envelope pre- and post-filtering for each system. All analyses and filtering were performed using LabVIEW (National Instruments, U.S.A.).

Inter-tester reliability for identifying the exercises appropriately, and processing the EMG data (including the outcomes of peak and mean muscle activity and duration of contraction) was determined for four testers. Two of the testers were Physiotherapists with more than 10 years of clinical experience and two of the testers were undergraduate students studying Exercise Science. All testers held limited experience analyzing visual EMG traces in the preceding 12 months. Each tester was required to load the EMG data files, determine which trials were to be analysed and to mark the start and end of two of the exercises (resisted isometric knee extension and the fast squat) that each participant completed for both the commercial system data and the custom system data. Two methods of EMG data processing were implemented, a manual technique and an automatic thresholding technique. The manual method required the tester to manually select the start and end of the contraction (using a cursor) based on the point they determined as being a significant departure from the resting baseline. This method displayed the linear envelope trace graphically, with the ability to zoom in to as fine a precision as preferred by the tester. The TKEO method can be used to measure the instantaneous energy changes occurring in a surface EMG signal (Li et al., 2007), which allows for the attenuation of random noise and amplification of real signal. This dramatically improves the signal to noise ratio (SNR) and accuracy of activation onset detection, with Solnik et al. (2010) reporting an almost 30 times improvement in SNR and >50% reduction in onset detection error during gait trials.
This analysis predominantly followed the methodology of Solnik et al. (2010), and included the following steps:

1) The TKEO method was applied to the bandpass filtered data prior to the linear envelope.

2) The tester extracted a section of the data that they determined to be a rest period for the participant with no contraction, with only background noise of the system present in the signal.

3) Based on a preliminary visual evaluation of the data prior to this study, the threshold was set as six standard deviations above the mean resting level. These values were extracted from this rest period.

4) The tester then set the cursors at least 1 second before and after each of the contractions, and an automated algorithm was used to detect when the enveloped data first and last crossed this threshold for a minimum of 100ms. This was deemed the onset and offset time points respectively. These time points were used to define the contraction.

Peak and mean muscle activation (mV) during each contraction for the fast squat and the isometric knee extension were normalized and expressed as a percentage of the highest value obtained during the isometric MVC (%MVC). The contraction duration was expressed in milliseconds (ms). Each tester was blinded to the results of the other testers.

2.8 Statistical Analyses
The peak and mean muscle activation (%MVC) and duration of muscle activation (ms) were identified in each exercise trial using both the manual and the TKEO method, and these discrete variables averaged for each exercise repetition (table 1).

For the concurrent validity, a range of correlation statistics explored the association between the two EMG systems for all the exercises. Relative agreement was computed using Spearman’s correlation coefficient (IBM SPSS Statistics, Version 23.0, USA) and Intra-class Correlation Coefficient ICC(2,1). Estimates of correlation were interpreted as excellent (0.75-1), modest (0.4-0.74), or poor (0-0.39) (Fleiss, 1986). Absolute agreement was explored using Bland Altman plots (Bland and Altman, 1986) (Microsoft Excel, 2007) with mean difference and variability between the two systems calculated. Linear regression (R²) between systems for the average results for each exercise were also calculated.

Inter-tester reliability for the isometric resisted knee extension and fast squat data were calculated using a single measure two-way random model with absolute agreement (IBM SPSS, Version 23.0, USA), (Shrout and Fleiss 1979; McGraw and Wong 1996) for the four testers and the two different EMG data extraction methods (TKEO and manual).

3. Results

3.1 Concurrent validity

Concurrent validity for peak muscle activity (%MVC) (Table 2; Figure 3a) demonstrated modest to excellent relative agreement between the commercial and the custom system across all the exercises, on average ICC 0.84 (range 0.77-0.96) and Spearman’s correlation coefficient 0.70 (range 0.5-0.87). The mean difference
in muscle activity between the systems was on average 6% (range 3-13%) with the commercial system recording slightly higher values (Supplemental material 1).

Mean muscle activity (%MVC) (Table 3; Figure 3b) relative agreement (ICC) values were modest to excellent, on average 0.86 (range 0.68-0.95) and Spearman’s correlation coefficient on average 0.72 (range 0.47-0.82). The mean difference between the two systems was 1% (Supplemental material 1).

For the duration of contraction (ms) (Table 4; Figure 3c) during the exercises the relative agreement between the two systems was modest to excellent with ICC on average 0.86 (range 0.65-0.99) and Spearman’s correlation coefficient values were on average 0.78 (range 0.50-0.99). The mean difference in duration of muscle activation between the two systems was on average 288ms across all exercises (Supplemental material 1).

Both the manual and automatic TKEO methods of data extraction were evaluated for concurrent validity, with only the TKEO method presented in the tables as the inter-tester analysis (3.2) revealed that it offered more reliable and less variable outcomes in the fast squat data.

3.2 Inter-tester reliability

Inter-tester reliability for peak muscle activity was excellent for the four testers (ICC>0.99) for both the TKEO and the manual method in the isometric contractions and the fast squats. The results for the inter-tester reliability of the four testers for mean activation and duration of contraction are presented in Table 5. ICC values for mean contraction across both systems were stronger and less variable using the TKEO method than manual extraction for isometric contractions. The average ICC of the isometric contractions for the TKEO method was 0.92 (range 0.69-0.99) and
for the manual method was 0.81 (range; 0.44-0.97). Similar ICCs were found for each of the two methods for the mean contraction activity during squats with the TKEO method averaging 0.88 (range 0.77-0.95) and 0.92 (range 0.84-0.99) for the manual method. Regarding the duration of contraction, greater inter-tester reliability occurred in the isometric contraction duration compared to the squat duration across both systems, although the TKEO method again had stronger and less variable agreement between testers for the isometric contractions. The average ICC for isometric contraction duration was 0.85 (range 0.72-0.91) for the TKEO method and 0.67 (range 0.43-0.84) for the manual method. ICC for squat duration ranged from poor to modest for both the TKEO method (average ICC=0.49, range 0.28-0.65), and the manual method (average ICC=0.46, range 0.20-0.77).

4. Discussion

This study found that a low-cost, custom-built EMG system using off-the-shelf hardware holds potential for assessing muscle activation levels and duration of contraction during specific isometric and dynamic exercises. The relative agreement between the commercial EMG system and the custom system varied from modest to excellent for peak contraction, mean contraction and contraction duration. This indicates that while custom systems can provide valid muscle activation information, the results are not interchangeable with the commercial system. Inter-tester reliability was typically higher using the automated TKEO method for determining mean muscle activity and contraction duration in comparison with the manual technique. This may be because, despite specific instructions on defining of the onset and cessation of muscle activity, the manual method is still influenced by subjectivity for the tester. Subjective decisions are required to define when there is a significant departure from resting baseline activity to set the cursors at each end of the
contraction. This is influenced by the degree to which the tester magnifies the trace using a zooming function. The association between the commercial EMG system and the custom system varied from modest to excellent. The variation in the association between the two systems may be due to:

a) Positioning of electrodes. Potentially the major source of difference in signal intensity between the two systems was the need to position the electrodes on different parts of the muscle during the same trial. The relative alignment and movement of muscle fibres, location of the innervation zone, and cross-talk between muscles is known to influence EMG signal intensity and quality (Basmajian and DeLuca, 1985; Beck et al., 2010; Farina et al., 2002; Rainoldi et al., 2004). Varying the position of the electrodes can change the recorded Vastus Lateralis muscle activity by 28-44% during weighted squats and jumping (Earp et al., 2017). Normalisation of muscle activity values, as performed in this study, attenuates issues with the positioning of electrodes and location of the innervation zone in EMG recording of Vastus Lateralis activity (Beck et al., 2008). Inter-electrode distance will also influence the recording of muscle activity (Barbaro and Rainoldi, 2012; Basmajian and DeLuca, 1985). The concurrent recording of data with each system’s electrodes positioned on different anatomical parts of the muscle will influence results, and is a study limitation. For example, recent research shows that the shape of the detected motor unit activation potential is heavily influenced by the distance that the electrode is from the innervation zone (Del Vecchio et al., 2018). Although system electrode location placement (proximal or distal) randomisation partially addressed this placement concern, the electrodes were still recording different muscle areas with varied available motor units. Additionally, the varied position
of the reference electrode, which could not be changed, will have influenced the signals.

b) Data acquisition and processing. The usage of a 16- and 14-bit Analog-to-Digital Convertor (ADC) for the commercial and custom system respectively may have contributed to some of the difference in results between systems. However, given that 14-bits still provides 16384 independent data points across the ±10V range of the ADC, and that the data is rectified and smoothed heavily during EMG processing regardless of the system used, this difference is unlikely to have more than a minor influence on the results.

c) Gain settings. More likely to have impacted the findings are the non-standardised gain settings on the custom microchip, which is controlled using an adjustable potentiometer and is therefore not defined easily. Prior to testing we adjusted the gain to optimize signal amplitude without clipping, with a bias towards reduced amplitude. This resulted in no clipping observed in any of the data collected, but data precision may have been influenced due to this gain adjustment.

d) Participant numbers. Small subject numbers potentially influenced the comparison between the two EMG systems and is a study limitation.

Few studies have investigated inter-tester agreement in EMG data extraction and analysis in lower limb exercises (Gupta et al., 2014). Peak muscle activity in this study demonstrated excellent agreement between testers as it does not require thresholding. Inter-tester agreement for determining the onset of muscle activity is acknowledged to be lower than intra-tester agreement (DeFabio et al., 1986; Bolgla et al., 2012). Visual inspection of EMG onset is subjective and based on perceptions
of the signal (Tenan et al., 2017). These perceptions of muscle onset and also turn-off points probably influenced both the mean and duration inter-tester agreement. We produced inter-tester absolute agreement ICC values similar to, although slightly more variable, than gastrocnemius onset activity in hopping (Gupta et al., 2014). The variability may be due to the comparison between a greater number of testers in this study and some testers with limited EMG analysis experience which potentially impacted signal perception as either noise or anticipatory muscle activity, or exercise-related muscle activity. The TKEO thresholding technique consistently outperformed the manual method of determining the muscle activity duration, with stronger absolute agreement between testers. This result agrees with previous research (Li et al., 2007) with the TKEO thresholding attenuating testers’ subjective interpretation of muscle activity onset or cessation. Given the simplicity of the TKEO technique, future studies interested in assessing either muscle activation duration or activation levels relative to timing are recommended to utilise this thresholding technique.

Future directions for low-cost surface electromyography may include expanding the measurement parameters away from global EMG features. Global measures include variables such as whole muscle peak amplitude or activation onset, which are often considered as proxy measures of activity of all active motor units in a muscle group. While useful, traditional surface EMG assessment using one channel per muscle group has known validity issues with respect to determining mechanistic factors, such as the underlying activity of the individual motor units (Del Vechio et al., 2017; Farina et al., 2008; Farina et al., 2010; Keenan et al., 2006). Low-cost, widely available EMG-on-a-chip hardware such as the Myoware used in this study allows for the creation of multichannel, composite systems consisting of multiple electrodes.
placed on the same muscle group. Making this technique less cost-prohibitive would allow for the more widespread assessment of the data that these systems provide, for example the ability to examine individual motor unit firing patterns and estimate conduction velocity.

Finally, this study focused on examining the results of discrete variables calculated from the EMG trace, namely the peak intensity, mean intensity and duration of contraction during different exercises. We did not include pattern matching analysis (eg. cross-correlation between traces), as we wanted to focus on outcome measures that reflect those typically reported in the literature.

5. Conclusion

With respect to practical applications, low cost EMG microchip systems hold promise for clinical implementation. It is possible to integrate these chips with low cost microprocessors to create complete EMG systems for under USD$100. When combined with custom software there are multiple options for both clinical research and biofeedback application. However, despite our positive findings indicating the custom-built EMG unit could be of value for clinical biofeedback, this interpretation does not include any assessment of the low-cost systems reliability or durability. However, anecdotally, experimentation with low-cost systems across clinical situations finds them to be robust. Next, multi-channel EMG custom systems need to be built and tested and long-term, empirical trialling will determine their reliability.
6. Acknowledgements

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References


**Figure 1.** An example of the electrode placement of the two systems along the fibres of Vastus Lateralis.
**Figure 2.** An example of the filtered (left) and unfiltered (right) data from the custom system and the commercial system. This example shows a range of muscle contraction intensities, with all charts time synchronized.

A) The spectrogram obtained from a short-term Fourier transform of the data. Note the similarity of the intensity levels post-filtering, with red representing high signal intensity and purple low intensity. The y-axis scale is linear from 0Hz at the intercept up to 500Hz.

B) The pre-envelope trace in mV. Note the much higher signal to noise ratio of the custom system pre-filtering.

C) The linear envelope of the commercial system (black line) and the custom system (shaded grey). The similarity is high post-filtering, however pre-filtering the custom system was heavily contaminated by noise.
Figure 3 Linear regression comparing the two EMG systems for the average results for each exercise; A. Peak EMG B. Mean EMG and C. Contraction Duration

%MVC: percentage Maximal Voluntary Contraction; KExt: Knee extension; SSqS: Single Squat Slow; SSqF: Single Squat Fast; TSqS: Triple Squat Slow; TSqF: Triple Squat Fast; SStep: Single Step Up; TStep: Triple Step Up;
<table>
<thead>
<tr>
<th>Exercise</th>
<th>Instructions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Squats</td>
<td>Slow speed (three seconds down to approximately 90 degrees of knee flexion and three seconds back to the starting position) Fast speed (one second down to 90 degrees knee flexion and one second to return).</td>
</tr>
<tr>
<td>Step up (35 cm high)</td>
<td>Comfortable</td>
</tr>
<tr>
<td>Concentric active knee extension (90 degrees to 0 degrees)</td>
<td>Comfortable.</td>
</tr>
<tr>
<td>Isometric knee extension</td>
<td>Resisted isometric knee extension (in 80 degrees of knee flexion) in sitting</td>
</tr>
<tr>
<td>Counter-movement jump</td>
<td>As high and as fast as possible</td>
</tr>
</tbody>
</table>
Table 2
Concurrent validity: Relative agreement (ICC), Spearman’s correlation coefficient and mean difference between the Commercial EMG system and the Custom EMG system for peak Vastus Lateralis activity.

<table>
<thead>
<tr>
<th>Exercise</th>
<th>COM Peak (%MVC)</th>
<th>CUS Peak (%MVC)</th>
<th>Relative agreement (ICC)</th>
<th>rs</th>
<th>MD (%)</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single Squat Slow</td>
<td>81 ±35%</td>
<td>76 ±29%</td>
<td>0.85</td>
<td>0.62</td>
<td>-6</td>
<td>23</td>
</tr>
<tr>
<td>Triple Squat Slow</td>
<td>87 ±38%</td>
<td>81 ±25%</td>
<td>0.77</td>
<td>0.60</td>
<td>-6</td>
<td>28</td>
</tr>
<tr>
<td>Single Squat Fast</td>
<td>81 ±34%</td>
<td>78 ±29%</td>
<td>0.80</td>
<td>0.64</td>
<td>-3</td>
<td>26</td>
</tr>
<tr>
<td>Triple Squat Fast</td>
<td>110 ±46%</td>
<td>105 ±33%</td>
<td>0.81</td>
<td>0.73</td>
<td>-5</td>
<td>31</td>
</tr>
<tr>
<td>Single Step Up</td>
<td>154 ±82%</td>
<td>150 ±60%</td>
<td>0.89</td>
<td>0.87</td>
<td>-4</td>
<td>45</td>
</tr>
<tr>
<td>Triple Step Up</td>
<td>204 ±97%</td>
<td>191 ±77%</td>
<td>0.85</td>
<td>0.79</td>
<td>-13</td>
<td>63</td>
</tr>
<tr>
<td>Knee extension</td>
<td>54 ±35%</td>
<td>51 ±42%</td>
<td>0.96</td>
<td>0.82</td>
<td>-3</td>
<td>16</td>
</tr>
<tr>
<td>Jump</td>
<td>189 ±115%</td>
<td>180 ±69%</td>
<td>0.81</td>
<td>0.50</td>
<td>-9</td>
<td>76</td>
</tr>
</tbody>
</table>

Correlation Coefficient, rs, Spearman’s correlation coefficient, MD: Mean score difference (CUS-COM); SD: standard deviation
Table 3 Concurrent validity: Relative and absolute agreement (ICC), Spearman’s correlation coefficient and mean difference between the Commercial EMG system and the Custom EMG system for mean Vastus Lateralis activity.

<table>
<thead>
<tr>
<th>Exercise</th>
<th>COM Mean (%MVC)</th>
<th>CUS Mean (%MVC)</th>
<th>Relative agreement (ICC)</th>
<th>rs</th>
<th>MD (%)</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single Squat Slow</td>
<td>30 ± 13%</td>
<td>28 ± 8%</td>
<td>0.80</td>
<td>0.71</td>
<td>-2</td>
<td>9</td>
</tr>
<tr>
<td>Triple Squat Slow</td>
<td>28 ± 11%</td>
<td>26 ± 6%</td>
<td>0.76</td>
<td>0.59</td>
<td>-2</td>
<td>8</td>
</tr>
<tr>
<td>Single Squat Fast</td>
<td>28 ± 10%</td>
<td>27 ± 7%</td>
<td>0.68</td>
<td>0.47</td>
<td>-1</td>
<td>9</td>
</tr>
<tr>
<td>Triple Squat Fast</td>
<td>36 ± 14%</td>
<td>34 ± 11%</td>
<td>0.86</td>
<td>0.77</td>
<td>-2</td>
<td>9</td>
</tr>
<tr>
<td>Single Step Up</td>
<td>39 ± 19%</td>
<td>39 ± 17%</td>
<td>0.95</td>
<td>0.81</td>
<td>0</td>
<td>8</td>
</tr>
<tr>
<td>Triple Step Up</td>
<td>45 ± 22%</td>
<td>44 ± 19%</td>
<td>0.95</td>
<td>0.82</td>
<td>-1</td>
<td>9</td>
</tr>
<tr>
<td>Knee extension</td>
<td>22 ± 12%</td>
<td>22 ± 15%</td>
<td>0.93</td>
<td>0.81</td>
<td>0</td>
<td>7</td>
</tr>
<tr>
<td>Jump</td>
<td>54 ± 29%</td>
<td>56 ± 23%</td>
<td>0.91</td>
<td>0.81</td>
<td>2</td>
<td>15</td>
</tr>
</tbody>
</table>

COM: Commercial EMG system; CUS: Custom-made EMG system; ICC: Intraclass Correlation Coefficient, rs: Spearman’s correlation coefficient, MD: Mean score difference (CUS-COM); SD: standard deviation
Table 4 Concurrent validity: Relative and absolute agreement (ICC), Spearman’s correlation coefficient and mean difference between the Commercial EMG system and the Custom EMG system for duration of Vastus Lateralis activity.

<table>
<thead>
<tr>
<th>Exercise</th>
<th>COM Mean (ms)</th>
<th>CUS Mean (ms)</th>
<th>Relative agreement (ICC)</th>
<th>( r_s )</th>
<th>MD (ms)</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single Squat Slow</td>
<td>12426 ± 1726</td>
<td>12112 ± 1621</td>
<td>0.76</td>
<td>0.50</td>
<td>-313</td>
<td>1463</td>
</tr>
<tr>
<td>Triple Squat Slow</td>
<td>36116 ± 2033</td>
<td>35595 ± 1397</td>
<td>0.75</td>
<td>0.50</td>
<td>-521</td>
<td>1569</td>
</tr>
<tr>
<td>Single Squat Fast</td>
<td>5769 ± 996</td>
<td>5737 ± 870</td>
<td>0.87</td>
<td>0.69</td>
<td>-32</td>
<td>641</td>
</tr>
<tr>
<td>Triple Squat Fast</td>
<td>13402 ± 1527</td>
<td>13129 ± 1586</td>
<td>0.93</td>
<td>0.76</td>
<td>-273</td>
<td>818</td>
</tr>
<tr>
<td>Single Step Up</td>
<td>5918 ± 836</td>
<td>5602 ± 700</td>
<td>0.97</td>
<td>0.92</td>
<td>-316</td>
<td>271</td>
</tr>
<tr>
<td>Triple Step Up</td>
<td>14696 ± 2398</td>
<td>14286 ± 2277</td>
<td>0.99</td>
<td>0.99</td>
<td>-410</td>
<td>422</td>
</tr>
<tr>
<td>Knee extension</td>
<td>6265 ± 1272</td>
<td>5881 ± 1286</td>
<td>0.99</td>
<td>0.94</td>
<td>-384</td>
<td>286</td>
</tr>
<tr>
<td>Jump</td>
<td>5277 ± 1316</td>
<td>5334 ± 793</td>
<td>0.65</td>
<td>0.99</td>
<td>58</td>
<td>1385</td>
</tr>
</tbody>
</table>

COM: Commercial EMG system; CUS: Custom-made EMG system; ms: milliseconds; ICC: Intraclass Correlation Coefficient, \( r_s \): Spearman’s correlation coefficient; MD: Mean score difference (CUS-COM), SD: standard deviation;
Table 5
Inter-tester reliability: ICC and 95% confidence interval for mean contraction and duration of contraction using the TKEO method and the manual method

<table>
<thead>
<tr>
<th>Exercise:</th>
<th>System</th>
<th>Mean Contraction TKEO method</th>
<th>Mean contraction Manual method</th>
<th>Duration of contraction TKEO method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Isometric knee extension (estimated percentage resistance) or Squat</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>100 (MVC)</td>
<td>COM</td>
<td>0.69 (0.36, 0.90)</td>
<td>0.44 (0.14, 0.77)</td>
<td>0.72 (0.43, 0.98)</td>
</tr>
<tr>
<td>100 (MVC)</td>
<td>CUS</td>
<td>0.88 (0.69, 0.96)</td>
<td>0.57 (0.19, 0.85)</td>
<td>0.80 (0.56, 0.97)</td>
</tr>
<tr>
<td>75</td>
<td>COM</td>
<td>0.89 (0.71, 0.97)</td>
<td>0.75 (0.33, 0.93)</td>
<td>0.84 (0.59, 0.97)</td>
</tr>
<tr>
<td>75</td>
<td>CUS</td>
<td>0.97 (0.91, 0.99)</td>
<td>0.87 (0.56, 0.97)</td>
<td>0.90 (0.74, 0.97)</td>
</tr>
<tr>
<td>50</td>
<td>COM</td>
<td>0.88 (0.71, 0.91)</td>
<td>0.84 (0.45, 0.96)</td>
<td>0.82 (0.58, 0.97)</td>
</tr>
<tr>
<td>50</td>
<td>CUS</td>
<td>0.97 (0.91, 0.99)</td>
<td>0.88 (0.61, 0.97)</td>
<td>0.89 (0.71, 0.97)</td>
</tr>
<tr>
<td>25</td>
<td>COM</td>
<td>0.94 (0.84, 0.98)</td>
<td>0.88 (0.68, 0.97)</td>
<td>0.88 (0.71, 0.97)</td>
</tr>
<tr>
<td>25</td>
<td>CUS</td>
<td>0.99 (0.98, &gt;0.99)</td>
<td>0.91 (0.76, 0.98)</td>
<td>0.87 (0.69, 0.97)</td>
</tr>
<tr>
<td>10</td>
<td>COM</td>
<td>0.99 (0.96, &gt;0.99)</td>
<td>0.95 (0.86, 0.99)</td>
<td>0.91 (0.79, 0.98)</td>
</tr>
<tr>
<td>10</td>
<td>CUS</td>
<td>0.99 (0.99, &gt;0.99)</td>
<td>0.97 (0.88, 0.99)</td>
<td>0.91 (0.78, 0.97)</td>
</tr>
<tr>
<td>Single Squat fast</td>
<td>COM</td>
<td>0.85 (0.55, 0.96)</td>
<td>0.88 (0.63, 0.97)</td>
<td>0.36 (0.67, 0.71)</td>
</tr>
<tr>
<td>Single Squat fast</td>
<td>CUS</td>
<td>0.77 (0.42, 0.93)</td>
<td>0.84 (0.44, 0.96)</td>
<td>0.28 (0.03, 0.69)</td>
</tr>
<tr>
<td>Triple Squat fast</td>
<td>COM</td>
<td>0.98 (0.92, 0.99)</td>
<td>0.99 (0.94, &gt;0.99)</td>
<td>0.65 (0.27, 0.79)</td>
</tr>
<tr>
<td>Triple Squat fast</td>
<td>CUS</td>
<td>0.95 (0.86, 0.98)</td>
<td>0.98 (0.89, &gt;0.99)</td>
<td>0.65 (0.28, 0.79)</td>
</tr>
</tbody>
</table>

COM: Commercial EMG system; CUS: Custom-made EMG system; MVC: Maximum Voluntary Contraction; TKEO: Teager Kaiser energy operator
Sophie Heywood holds a B Physiotherapy (Hons) and Masters (Sports Physiotherapy) from the University of Melbourne. Sophie works at St Vincent’s Hospital Melbourne, the Melbourne Sports Medicine Centre, the University of Melbourne and Monash University and is also a PhD candidate at the University of the Sunshine Coast. She was awarded a Winston Churchill Memorial Trust Fellowship in aquatic physiotherapy and chronic disease management. Sophie has published and teaches both in Australia and overseas in aquatic physiotherapy and exercise prescription for musculoskeletal conditions and has a special interest in the management of knee osteoarthritis.