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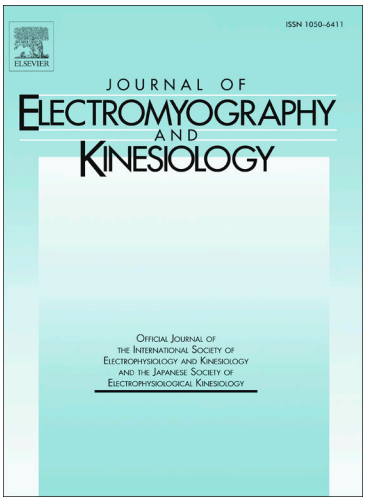

EMG gait data from indwelling electrodes is attenuated over time and changes independent of any experimental effect

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PII: S1050-6411(20)30103-6
DOI: https://doi.org/10.1016/j.jelekin.2020.102461
Reference: JJEK 102461

To appear in: *Journal of Electromyography and Kinesiology*

Received Date: 1 April 2020
Revised Date: 3 August 2020
Accepted Date: 25 August 2020

Please cite this article as: J. Reeves, C. Starbuck, C. Nester, EMG gait data from indwelling electrodes is attenuated over time and changes independent of any experimental effect, *Journal of Electromyography and Kinesiology* (2020), doi: https://doi.org/10.1016/j.jelekin.2020.102461

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EMG gait data from indwelling electrodes is attenuated over time and changes independent of any experimental effect.

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Abstract

The effect of time on the validity of electromyography (EMG) signals from indwelling fine-wire electrodes has not been explored. This is important because experiments using intramuscular electrodes are often long and biochemical and mechanical factors, may impair measurement accuracy over time. Measures over extended periods might therefore be erroneous. Twelve healthy participants (age=33±8 years) walked for 50 minutes at a controlled speed. Fine-wire electrodes were inserted into tibialis anterior and a surface EMG sensor attached near the fine-wire insertion site. EMG signals progressively and significantly decreased with time with the fine-wire electrode, but not the surface electrode. For the fine-wire electrode, after 25 minutes mean amplitude had reduced by 11% ($p<0.001$) and after 50 minutes by 16% ($p<0.001$), and peak amplitude reduced 22% at 20 minutes ($p=0.006$) and 37% at 50 minutes ($p<0.001$). Reduced amplitude with indwelling EMG without concurrent changes in surface EMG signal suggests an important inconsistency in data from fine-wire EMG electrodes. Changes in EMG signal will occur over time independent of the experimental condition and this questions their use in experiments of more than 30 minutes. These results should impact on experimental study design. They also invite reinterpretation of prior literature and sensor innovation to improve measurement performance.

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Introduction

With deep or small muscles that are located in close proximity to other muscles, it is necessary to insert an electrode directly into the muscle to record electromyography (EMG). Fine-wire electrodes are used for indwelling EMG (Konrad, 2005) and the technique remains relatively challenging compared to use of EMG electrodes on the skin surface (O'Connor and Hamill, 2004, Semple et al. , 2009, Stacoff et al. , 2007). As such, fine wire EMG is used far less often than surface EMG, and there is relatively limited knowledge of methodological issues concerned with its use. Indeed, from a recent expert consensus process there was a call for more empirical evidence to improve use of indwelling EMG (Besomi et al. , 2019). In the consensus project caution was advised when using fine-wire EMG to study dynamic contractions (Besomi, Hodges, 2019). Nonetheless fine-wire EMG is currently used to study muscle activity in dynamic tasks like walking (Akuzawa et al. , 2016, Allison et al. , 2018, Barn et al. , 2014, Farris et al. , 2019, Kelly et al. , 2015, Kelly et al. , 2016, Lawrenson et al. , 2019, Maharaj et al. , 2016, Maharaj et al. , 2017, 2018, Mazzoli et al. , 2018, Murley et al. , 2009, Murley et al. , 2010, Murley et al. , 2014, Zacharias et al. , 2019).

Whilst shank muscle activation patterns and onset times recorded using indwelling EMG and surface EMG sensors have been shown to be similar (Bogey et al. , 2003, Bogey et al. , 2000, Chimera et al. , 2009, Peter et al. , 2019), there are important theoretical differences that could affect the comparability of the signals over time. Implanting a foreign metal object into a muscle can result in an accumulation of fluid around the electrode surface, which increases the distance to the active tissue of chemical conductivity (Geddes and Roeder, 2003, Polikov et al. , 2005) as well as having possible changes in the electrode capacitance itself (Lykken, 1959), both of which could affect the amplitude of recordings. Anecdotal experience within our group suggests there is a reduction in signal amplitude after 20-40 mins of walking with fine-wire electrodes in situ. This is at odds with recent protocols that involved fine-wire electrodes being in place for 1-1.75 hours (Kingston and Acker, 2018) and 2-4 hours (Raven et al. , 2018). Protocols with five insertion sites would likely involve a time consuming setup, leaving electrodes in situ for long periods before data collection begins (Alenabi et al. , 2018). Measurements of a dependent variable must be reliable over the entire time of data collection if an independent variable is to be confidently investigated. The aim of this study was to compare EMG signals from fine-wire and surface EMG electrodes over extended periods,

using the electrical activity of tibialis anterior (TA) during the commonly studied task of walking as the basis of the comparison.

Methods

Participants

Twelve healthy participants (age=33 ± 8 years, height= 1.73 ± 0.09 m, mass= 75.5 ± 13.8 kg, mean ± SD) walked shod along a 6-10 m walkway in the laboratory for 50 minutes. Exclusion criteria were: 1) any recent lower limb injury; 2) any cardiovascular, musculoskeletal or neurological conditions or disease, immune deficiency or haemophilia; 3) the use of anti- biotic medication, anti-coagulant medication, and anti–platelet therapy; 4) walking with use of an aid. All participants were classified as right leg dominant based on their response to the question "which leg would you kick a ball with?" and provided informed consent. The study was approved by the University's research ethics board.

Procedure

Participants were randomly selected to perform either ten walking trials of five minutes each (n=7) interspersed with one minute breaks, or five walking trials of 10 minutes each (n=5), interspersed with two minute breaks. The two protocols were chosen to explore if the consistency of the EMG signal over time was affected by the frequency of breaks in walking. It was thought that one-two minutes would be sufficient rest to negate fatigue and similar to the time between conditions in a typical gait study. Self-selected walking speed was monitored using timing gates (Brower Timing Systems, UT, USA), and participants were asked to adjust their walking speed if they deviated beyond ±5% of the mean self-selected walking speed established prior to insertion of the fine-wire electrodes.

EMG was recorded from the right leg. A fine-wire electrode was inserted into the TA at a third of the distance from the tip of the fibula and the tip of the medial malleolus, as per guidelines for the surface EMG approach (SENIAM) (Hermens et al. , 2000). Insertion was performed with the participant sitting or lying supine with legs relaxed and extended. Prior to insertion the skin was cleaned with an isopropyl alcohol wipe and nitrile gloves were worn. Insertion depth was 1-1.5 cm which typically corresponds to the superficial uni-pennate half of the TA (Maganaris, 2001). The commercially available bipolar fine-wire electrodes (0.051 mm diameter, paired-hook wires, Teflon-coated stainless-steel wire) were inserted with an unused, sterile, hypodermic needle (50 mm long and 25 gauge, Chalgren Enterprises Inc., CA, USA). The bent uninsulated recording tips of the electrode were 2 mm in length and staggered so as

to not be in contact. The electrode wires are not supplied glued together. The brand of electrodes used in this study have been used in previous research (Alenabi, Whittaker, 2018, Barn et al. , 2012, Farris, Kelly, 2019, Kelly et al. , 2012, Kelly, Lichtwark, 2015, Kelly, Lichtwark, 2016, Maharaj, Cresswell, 2017, 2018). Participants were asked to dorsiflex their ankle after insertion to encourage the electrode to fixate in the muscle (© Motion Lab System, 2018). A loop was made in the protruding wires and this was attached to the skin with tape to minimize artefact associated with wire movement (© Motion Lab System, 2018). The bare wire terminations were connected to a spring contact sensor (Delsys, Inc., Figure 1).

The sensing head of a Delsys Trigno™ Mini sensor (25 mm x 12 mm x 7 mm) was attached near the fine-wire insertion site. The Delsys Trigno™ Mini sensor has a 200 mm long cable which connects the sensing head to a main sensor, which serves as a stabilizing reference. The skin preparation (as per guidelines from Delsys) included shaving excessive hair, wiping the skin with isopropyl alcohol and allowing the skin to dry before application. A standard Delsys Trigno™ surface sensor with an inbuilt triaxial accelerometer (27 mm x 37 mm x 15 mm, EMG + IMU) was attached to the distal aspect of the right shin to determine foot contact. The sensors have four 1 mm x 5 mm parallel bars (contacts), of 99.9% silver with a fixed inter-electrode spacing of 10 mm. Double sided stickers were used to affix the sensors. Apertures in the adhesive tape were aligned with the electrode contacts to maintain skin contact (Roy et al. , 2007). The cable connecting the sensing head of the Trigno™ Mini sensor to the main sensor was taped to the skin to reduce the potential for movement artefact. EMG signals were collected at 2000 Hz and acceleration data at 150 Hz. Data was collected continuously throughout each walking trial.

Data analysis

Data was processed in MATLAB (R2017b, Mathworks Inc., MA, USA). The DC offset in the EMG data was adjusted for by subtracting the mean of each signal, the data was then bandpass filtered between 10-950 Hz and 10-500 Hz, for indwelling and surface data respectively, and then rectified. A linear envelope was created per gait cycle using a lowpass Butterworth filter with a cutoff of 10 Hz. Every tenth gait cycle was automatically sampled for each five minute interval, providing ten data sets per participant. From each five minute block a minimum of six good gait cycles were selected (mean: 21 ± 5). The peak and mean activity from each selected gait cycle was calculated and averaged. Signal amplitude was normalized to the mean of the peak from each gait cycle from the first five minutes of recording, as such signal amplitude at each subsequent five minute walking bout was relative to baseline. Vertical acceleration data

was filtered using a lowpass Butterworth filter with a 60 Hz cutoff. EMG was time normalized to the gait cycle based on identifying foot contact using the peaks in tibial acceleration.

Statistics

Statistical analysis was performed with SPSS (IBM SPSS Statistics 26) and Excel (Microsoft Office Excel 2013). Visual inspection of the data concluded the two protocols were similar, so the data from the two protocols were pooled. Participants were identified as outliers and excluded if their peak EMG value was 1.5 times the IQR beyond the upper or lower quartiles in the signal distribution. A repeated measures ANOVA was performed on the normalized mean and peak activity over the gait cycle and data were tested for sphericity using Mauchly's test and corrected using a Huynh–Feldt adjustment if required. A Holm-Bonferroni correction was applied for comparisons relative to the first trial. The normalized peak activity for surface EMG were non-normally distributed, so a Friedman's test was performed. The alpha significance level was set at 0.05 (two tailed).

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Results

Indwelling and surface TA EMG during the first five minutes and at 15, 30 and 50 minutes are plotted over the gait cycle as a mean of the group in Figure 2 and for individual participants in Figure 3. From Figure 2 it can be seen that the mean indwelling signal amplitude had reduced in early stance beyond one standard deviation of the mean from the first five minutes by 30 minutes and again by 50 minutes. In Figure 3 indwelling EMG amplitude from individual participants can be seen to progressively reduce over time throughout the gait cycle (in Figure 3 g data is missing for 50 minutes due to hardware issues).

Mean and peak signal values expressed as a percentage of the mean of the peak from each gait cycle from the first five minutes are shown in Table 1. Mean TA EMG amplitude significantly decreased with time for indwelling EMG ($p<0.001$), but not surface EMG. In the first five minutes mean amplitude was 38% ±9% and reduced to 31% ±11% ($p=0.004$) by 15 minutes and 22% ±8% by 50 minutes ($p<0.001$) (Table 1). Three participants were identified as outliers in the surface data (Figure 3. e, f and i) and excluded from subsequent statistical analysis of the surface data. There was subsequently no significant effect of time on mean TA surface EMG ($p=0.218$). Peak TA amplitude recorded with indwelling EMG also significantly reduced over time ($p<0.001$). Peak amplitude of indwelling EMG was reduced by 22% at 20 minutes ($p=0.006$) and 37% at 50 minutes ($p<0.001$) (Table 1). There was no significant difference in peak amplitude of the surface EMG signal over time ($p=0.654$).

Discussion

The aim of this study was to compare changes in EMG signal over time when recorded from TA using fine-wire and surface electrodes. When recording with fine-wire electrodes mean TA EMG amplitude decreased by 11% after 25 minutes ($p<0.001$) and peak amplitude decreased by 26% after 25 minutes ($p=0.001$) and both mean and peak further decreased by 50 minutes. There was no significant reduction in TA EMG amplitude using surface EMG. The duration of a protocol is therefore an important consideration when using indwelling EMG electrodes to study gait.

Potential mechanisms

We hypothesise that the cause of the signal reduction was not physiological and is a measurement error. Firstly, there was a relative reduction in amplitude in the indwelling EMG signal without a corresponding change in the surface EMG signal, and a change in both would

be expected if muscle activation changed over time. Secondly, data was collected on healthy participants, with no neurological disorders, walking at a self-selected speed for less than an hour, and we therefore we do not expect participants to have fatigued. In any case, this would have been identified with both techniques and fatigue leads to increased EMG amplitude (Konrad, 2005), rather than the decrease we observed.

The decrease in signal amplitude with indwelling EMG electrodes could be due to processes that are known to affect other clinical implants and electrodes. The presence of the fine-wire electrode and/or the mechanical trauma of insertion likely causes localised oedema around the electrode, which is fluid build-up from the introduction of a foreign body (Polikov, Tresco, 2005). Fluid build-up around a recording electrode distances the electrode from the active tissue, reducing signal amplitude (Geddes and Roeder, 2003). The foreign body response is similar to wound healing in which hemostasis, which stops bleeding, occurs in a matter of minutes to hours (Wisniewski and Reichert, 2000). During hemostasis blood-borne proteins can adhere to a sensor surface (Lotti et al. , 2017, Wisniewski and Reichert, 2000). The adhesion of biomolecules like proteins as a result of hydrophobic, hydrophilic, and electrostatic reactions can lead to biofouling of a sensor membrane or electrode fouling/passivation (Campuzano et al. , 2019). The biomolecules on the electrode create an insulating barrier that affects the reliability of the recordings (Hanssen et al. , 2016, Harreither et al. , 2016). It is conceivable that biofouling of the recording tips of fine-wire electrode can occur over time and reduce signal amplitude. Additionally, current flow within the muscle could lead to the attachment of ions to the electrode, effectively creating a capacitor that could affect the electrical characteristics of the electrode (Joynt, 1994). Increased impedance of stainless steel needle electrodes has been observed after exposure to saline, which mimics tissue, however this was after 5-48 hours, which is longer than it took to observe the reduction in amplitude in walking in the current study (Kalvøy et al. , 2010).

Conversely the decrease in signal amplitude with indwelling EMG over time could be due to mechanical processes in a contracting muscle. For instance the electrode could be damaged by shear forces during movement (Helton et al. , 2011). Furthermore, despite the bent electrode tips helping a fine-wire electrode to stay in place in a contracting muscle, the ends may not maintain a fixed relationship to the contracting muscle fibres as they change their length and pennation angle during contraction (Besomi, Hodges, 2019). When using fine-wire EMG, caution was advised not just in recording from dynamic contractions, but also if using maximum voluntary contractions (MVCs) for the purposes of normalisation, because such

forceful contractions might move the position and/or orientation of recording tips within the muscle and may damage the wire (Besomi, Hodges, 2019). Further work is necessary to establish the mechanisms responsible for the potential for indwelling EMG to reduce amplitude over time in order to mitigate the problem. For instance, future study designs could involve a comparison of the indwelling EMG signal over time from isometric and dynamic contractions, with and without prior performance of MVCs.

Difference between surface and indwelling EMG signal in early stance

The steeper slope of the TA EMG pattern in early stance in the surface EMG compared to the indwelling EMG was unexpected, given a similar shape between the two signals in previous literature (r=0.88, (Chimera, Benoit, 2009)), (Peter, Andersson, 2019)). Differences between surface and indwelling EMG may occur because fine-wire electrodes record from a smaller number of motor units than surface electrodes, so indwelling EMG may be less representative of whole muscle activation (Besomi, Hodges, 2019). Furthermore, skin and adipose tissue have low pass filter effects on the surface signal that will not affect the indwelling electrode signal (Besomi, Hodges, 2019, Konrad, 2005). Crosstalk with plantar flexor muscles could explain muscle activation recorded in mid-late stance in the surface, but not fine-wire signal for some participants (Peter, Andersson, 2019).

Individual responses

The reduction in amplitude of the indwelling signal compared to the surface signal was more notable in some participants than others, however it is not clear why this was so. One participant (Figure 3 c) experienced a dull ache in the muscle which grew progressively worse. For three participants (Figure 3 e, g and l) the insertion had to be performed twice, for instance because bleeding occurred on the first attempt and the electrode proved uncomfortable. As the insertion of the electrode itself causes micro damage (Polikov, Tresco, 2005) it could be that by performing the insertion twice more damage occurred than usual at the outset, even before the first trial started, and thereafter there might have been less change in the signal over time. However neither discomfort nor number of insertions appeared to visibly affect the reduction in signal amplitude over time. Further research into the mechanisms of potential signal attenuation over time with indwelling EMG might explain why amplitude typically reduces but not necessarily to the same extent in all cases.

For two of the three participants, who were excluded from the statistical analysis of the surface signal, there was a large drop in the surface signal after the first trial but not thereafter. It is suspected that this was an artefact in the surface data for those two participants, perhaps due to

reduced adhesion of the surface sensor as a consequence of sweating and walking (Besomi, Hodges, 2019). A small sensing head was chosen for the surface sensor to place the detection area as close as possible to the fine-wire sensing site. However, reducing sensor size also reduces the surface area for the double-sided tape in contact with the skin, and thus likely adhesion (Roy, De Luca, 2007). Artefact in the surface sensor signal clearly occurred in the third participant excluded from surface data (Figure 3 f) as the signal became increasing noisy over time. Additionally, surface sensor adhesion was an issue in a further participant (Figure 3 d) and the sensor fell off and needed to be reattached during data collection.

Limitations

It was not possible to determine the effect of the duration of walking trial or walking speed. Data was pooled over the two walking protocols because after visual inspection the results were similar across participants. There were seven participants who performed five minute walking bouts and five who performed ten minute walking bouts. With a greater sample size in each group we may have been able to detect differences in signal reduction due to speed and walking bout. Participants walked at self-selected speed and were asked to adjust their speed accordingly if they were deviating beyond 5% of their pre-established average self-selected speed. As EMG data was collected continuously we were not able to exclude specific gait cycles that deviated from this range. However faster or slower lengths of the walkway would likely not influence the mean of gait cycles sampled throughout a walking bout of at least five minutes, therefore it is unlikely that a slower walking speed would account for the reduction in amplitude of the indwelling EMG signal. The timing of the TA EMG pattern across the gait cycle could be a little offset from true foot contact as we were not able to use the gold standard of ground reaction force to determine foot contact and instead used acceleration from the distal aspect of the shin. Nonetheless shank mounted accelerometers can be used for accurate and reliable detection of gait events (Sinclair et al. , 2013) and shank acceleration was sufficient for the purposes of this study as it was concerned with relative changes in EMG amplitude over time and not the timing of muscle activation.

Ultrasound guidance was not used in this study for the insertion of the fine-wire electrodes as the TA is an easy to locate, superficial muscle, that does not have a finite safety-window in which to insert, unlike other muscles (Won et al. , 2011). Consequently variability in electrode placement between individuals may have occurred, which could explain the difference in the slope of the overall activation pattern recorded with the fine-wire electrode compared to previous work (Peter, Andersson, 2019). However, a reduction in signal amplitude over time

recorded from dynamically contracting muscle was demonstrated, irrespective of the location of the electrode itself.

Conclusion

Reduced amplitude with indwelling EMG and not surface EMG suggests poor reliability of data from fine-wire electrodes over time. This effect should be considered when designing protocols of long duration and interpreting the findings from such studies.

Conflict of interest

The authors have no conflicts of interest to declare.

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Conflict of interest

The authors have no conflicts of interest to declare.

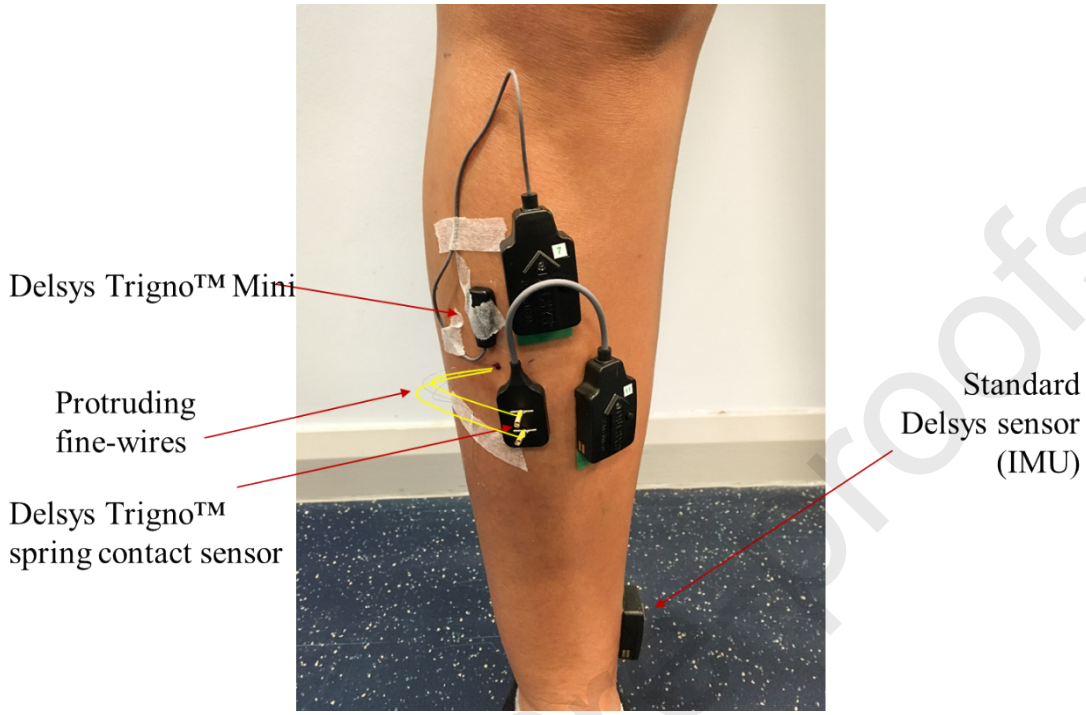

Figure 1. Set up with standard, mini and spring contact Delsys sensors

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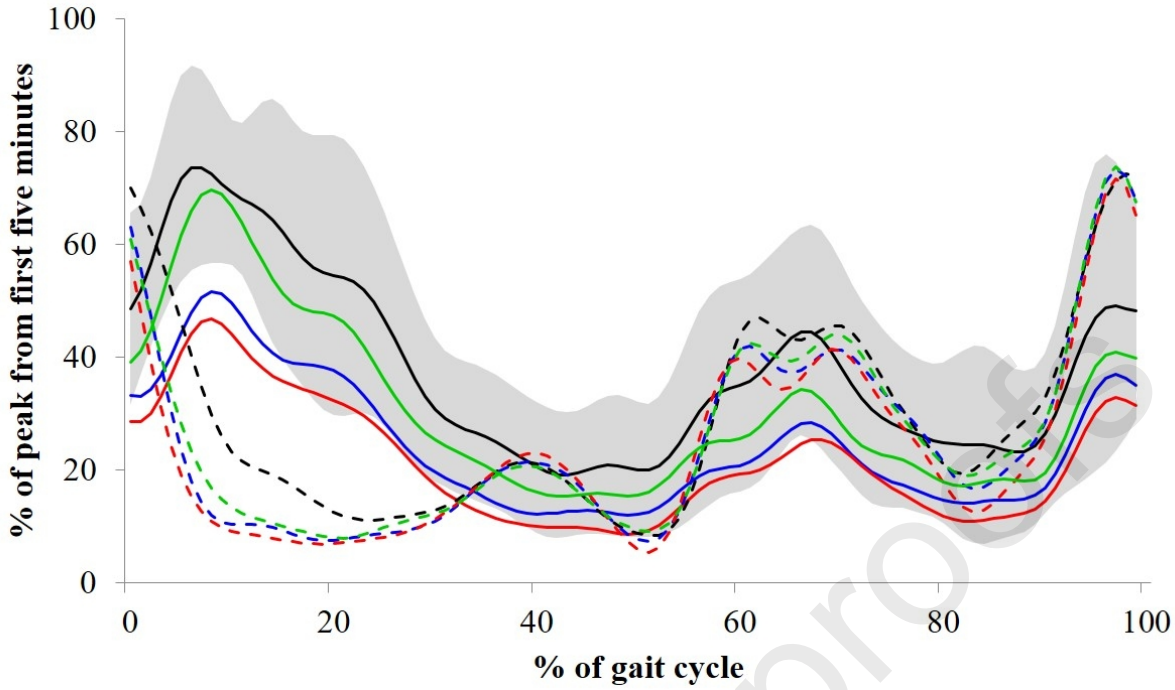

Figure 2. Group mean tibialis anterior EMG over the gait cycle recorded with fine-wire electrodes (solid lines) and surface electrodes (dashed lines) Blackline= 0-5 minutes; green lines= 10-15 minutes; blue lines= 25-30 minutes; redline= 45-50 minutes. Shaded grey area= standard deviation from first 0-5 minutes

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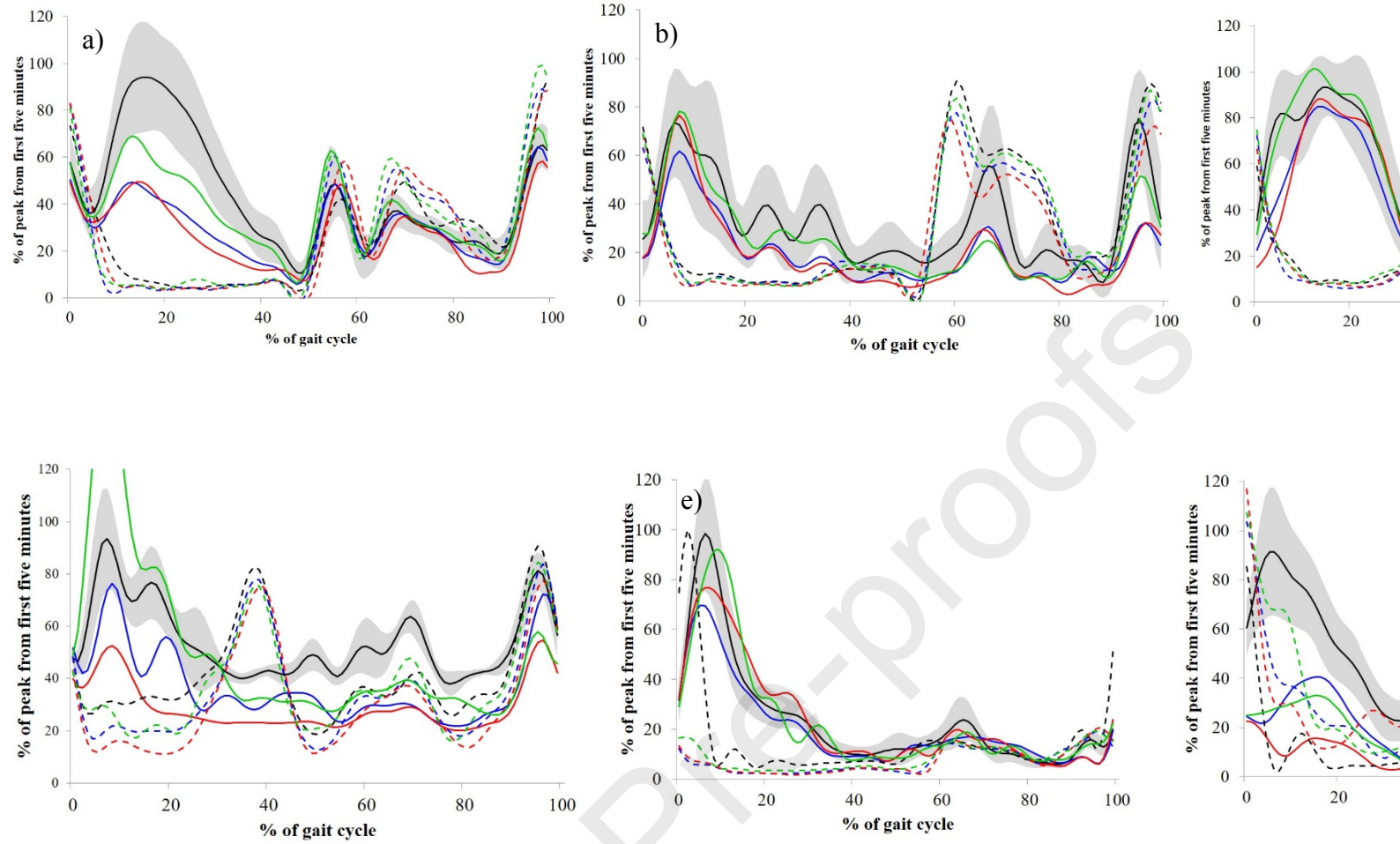

Figure 3. Mean tibialis anterior EMG over the gait cycle recorded with fine-wire electrodes (solid lines) and surf individual participants from the five minute walking protocol. Blackline= 0-5 minutes; green lines= 10-15 minutes; bl 45-50 minutes. Shaded grey area= standard deviation from first 0-5 minutes

Journal Pre-proofs

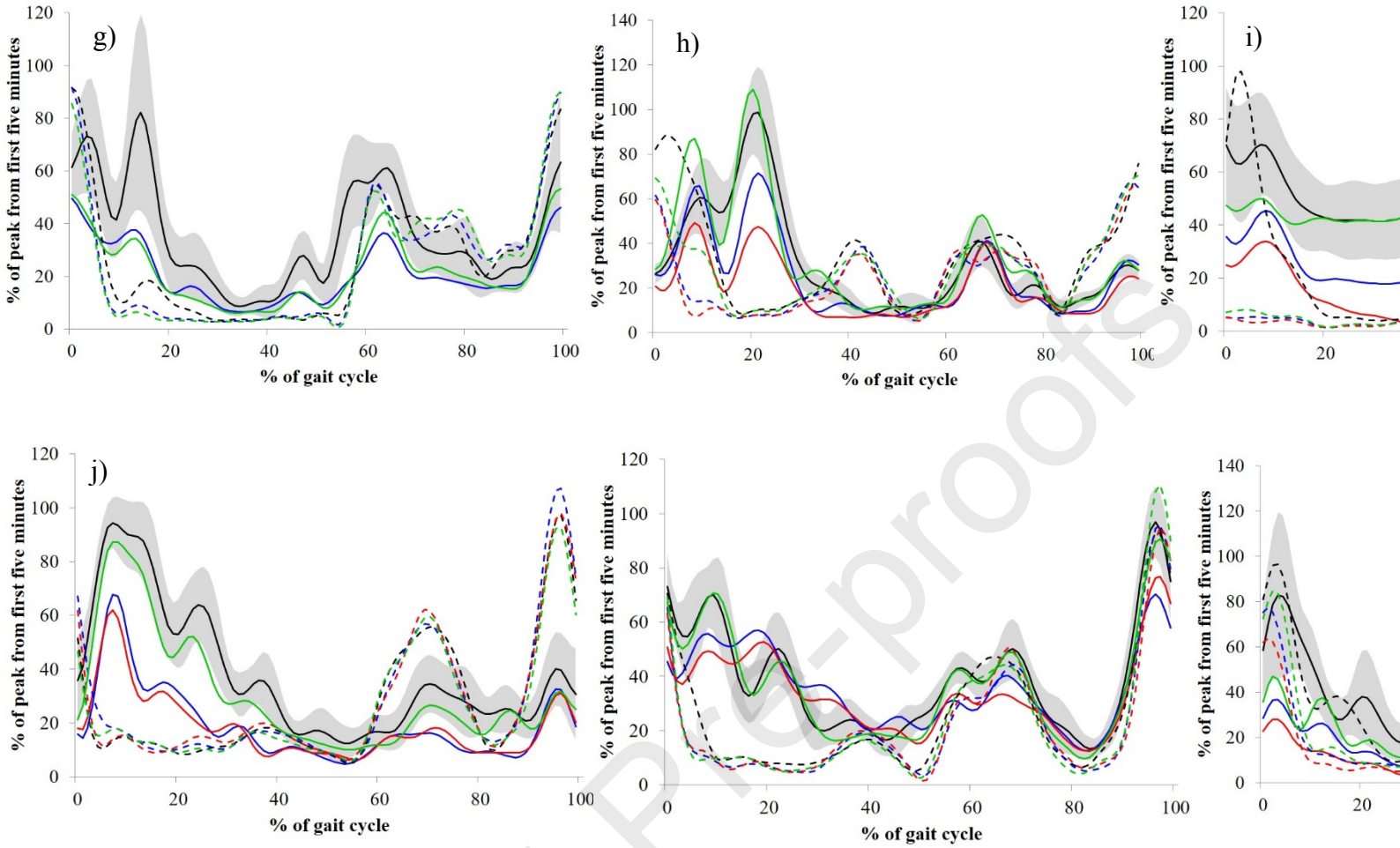

Figure 3 continued… Mean tibialis anterior EMG over the gait cycle recorded with fine-wire electrodes (solid lines) a lines) for individual participants from g) the five minute protocol and h-l) the ten minute walking protocol. Blackline 10-15 minutes; blue lines= 25-30 minutes; redline= 45-50 minutes. Shaded grey area= standard deviation from first 0

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Jo completed her PhD in biomechanics at the University of Salford, UK on the effect of foot orthoses on muscle activity and morphology, foot biomechanics and skin sensitivity. During the latter part of her PhD Jo was based at the University of Guelph, Canada. In that time she was involved in a collaborative project with the University of Salford investigating how mechanoreceptors on the sole of the foot respond to loads similar to those experienced in walking and another study on the influence of texture on joint position sense. Fine-wire EMG

of the tibialis posterior was a fundamental part of Jo's PhD thesis and she has been continuing to work on methodological issues with fine-wire electrodes during her time as a postdoctoral fellow with Dr. McLean at the University of Ottawa.

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Table 1. Mean and peak values expressed as a percentage of the mean of the peak from each gait cycle from the first five minutes

Variable	5		10		15		20		25		30		35	
Fine-wire (n=12)														
Mean	38	±9	33	±9	31	±11*	29	±12*	27	±9*	25	±8*	24	±7
Peak	100	±0	96	±27	88	±33	78	±23*	74	±19*	68	±15*	65	±1
Surface (n=9)														
Mean	29	±5	27	±4	28	±5	28	±5	28	±5	27	±4	27	±4
Peak	100	±0	96	±7	98	±12	97	±10	94	±14	96	±12	96	±1

** denotes p<0.05*

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